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Wearable Optical Fiber Sensors for Biomechanical Measurement in Medical Rehabilitation: A Review

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***Abstract*—With the increasing demand for rehabilitation of the elderly population and patients with limb injuries, it is imperative to develop new wearable health monitoring system and assistive rehabilitation equipment. Optical fiber sensing has the advantages of light-weight, anti-electromagnetic interference, higher strain and elastic limit, safety and dependableness. It has been significantly utilized in engineering prospecting, building structure monitoring, and medical and health fields. At present, the application of optical fiber sensing in medical rehabilitation has been developed rapidly, however, the discussion and summary of its key theories and technologies are insufficient. This review details different types of optical fiber sensing mechanisms and devices, as well as the measurement of physical parameters using optical fiber sensors and application examples in the field of medical rehabilitation. This review systematically analyzes the advance of the sensitivity, flexibility, accuracy, and wearability of the optical fiber sensing in the monitoring of human body sign information. This review also discusses the challenges of making a new generation of optical fiber sensors that fit human joints, muscles, and rehabilitation equipment. Finally, this review provides a research direction for the realization of flexible, comfortable, safe, stable monitoring of human biomechanical parameters and real-time precise control of rehabilitation equipment in the future.**

***Index Terms*—Optical fiber sensing technology, health monitoring, medical rehabilitation, physical parameters.**

I. Introduction

On May 11, 2021, the National Bureau of Statistics announced the data of the seventh census. The population of China aged 60 and above is 264 million, accounting for 18.7% of the total population. Compared with the sixth national census in 2010, the population aged 60 and above ratio increased by 5.44 percentage points, and the aging situation is becoming increasingly severe [1]. The aging population's physiological functions degenerate, inflicting health problems like diabetes, stroke, cerebral palsy, spinal cord injury, and Parkinson's disease, which have a great impact on their daily life. On the other hand, the proportion of individuals with limb or joint injuries is also increasing. For the basic living needs of these two groups of people, medical rehabilitation methods are indispensable, while traditional rehabilitation treatment is limited to clinical environments. With the continuous development of sensing technology, wireless communication technology, and signal processing methods, wearable sensor technology can provide patients with more flexible and free dynamic monitoring, and can provide real-time monitoring data for rehabilitation therapists, so that the rehabilitation treatment process can be remote and not limited by the fixed environment.

At present, electronic sensors are still the mainstream sensors for monitoring human biomechanical parameters and assisting exoskeleton control. Typically, potentiometers and encoders are used to monitor position information. However, its structure is bulky and not compact, and it is difficult to align with the wearable robot's joints, making the measurement inaccurate [2]. The small and integrated structure of the inertial measurement unit (IMU) is easy to wear and portable. It has high measurement accuracy, but the IMU-based solution requires frequent calibration because the gyroscope and magnetometer data drift over time after a period of use [3]. The resistance, capacitance and piezoelectric force sensors need to be connected to a signal amplification stage, which increases the difficulty of wearability [2]. It should be noted that these sensors described above are sensitive to electromagnetic fields, which are not friendly in the exoskeleton assistance control system, because the activation of the electric actuator will cause the interference of the sensing system [4]. The anti-electromagnetic advantages of optical fiber sensors make them competitive candidates for wearable sensors. Optical fiber sensors also have the characteristics of small size, lightweight, not sensitive to humidity, straightforward to combine into sensor networks and integrate into the Internet of Things, it is commonly utilized in the field of gyroscopes, underwater acoustic and geo-acoustic detection, oil exploration, fire monitoring, intelligent structure, and building monitoring [5].

Optical fiber sensors have high sensitivity for small deformation monitoring, strong adaptability to complex structures, and have the advantages of being arrayable, low transmission loss over long distances, and easy to wear. By selecting different types of optical fiber sensors, combined with temperature [6] and humidity sensitive [7] or flexible encapsulated materials [8], sensors or sensor arrays that fit human joints and muscles are made to sense the patient's respiratory rate, heart rate [9], and muscle flexion and extension awareness [10] during human rehabilitation training, and obtain the motion range of joint angles and gait parameters [4], which can assist exoskeleton or orthoses to achieve real-time on-demand control, and assist doctors to evaluate exercise ability and

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formulate rehabilitation plans for the patient. Although optical fiber sensors have shown great potential in the field of medical rehabilitation, the manufacturing processes of the sensors are complex, and the interrogators are expensive and not portable, which makes them impossible to realize industrial production and clinical application in various rehabilitation scenarios. For monitoring biomechanical parameters, most solutions based on optical fiber sensing are still in the research and experimental stage.

Due to the increasing importance of optical fiber sensing in medical rehabilitation monitoring, there is currently a lack of systematic analysis and sorting of key technologies for optical fiber sensing in medical rehabilitation applications. To comply with the rapid development of optical fiber sensors and provide a reference for the new technology of optical fiber sensing for lightweight wearable human health monitoring, this paper reviews and discusses the latest technologies of optical fiber sensors in terms of sensitization, flexibility, measurement accuracy, and wearability. Section II mainly reviews the optical fiber sensing mechanism and devices for human body sign information perception. Section III summarizes the biomechanical measurement and application examples of optical fiber sensing of physical quantities in the field of medical rehabilitation. Section IV discusses some challenges and future research directions of realizing wearable optical fiber sensing systems that can monitor human biomechanical parameters, and the last section is the conclusion.

II. Optical Fiber Sensors

The optical fiber sensor can convert the state of the measured object into a measurable light signal. when the external measured parameter interacts with the optical fiber sensor, the optical properties of the light, such as light intensity, wavelength, frequency, phase or polarization state will change, and become a modulated optical signal. Optical fiber sensor based on intensity modulation has low sensitivity, for example, the sensitivity coefficient of optical fiber tactile sensor in reference [11] is only 5.5 mv/N, so it is necessary to consider adding reflection or refraction in the optical path. Different sensitization structures and interrogation or signal processing techniques can improve the sensitivity and measurement accuracy of optical fiber sensors. Such sensors are classified as sensitivity-enhanced optical fiber sensors. In addition, the strain limit of FBG optical sensor made of silica glass is less than 1% [12], and the skin on human feet, waist and joints stretches and contracts repeatedly by as much as 55% [13]. Fabricating or encapsulating sensors with flexible materials can effectively solve the problems that silica glass optical fiber materials are easy to breakage, limited stretching and bending deformation, and cannot fit human joints and muscles when measuring human physical parameters. These sensors are classified as flexible materials fabricated or encapsulated optical fiber sensors.

A. Sensitivity-enhanced Optical Fiber Sensors

The sensitivity of the sensor is one of the performances that need to be focused on. In recent years, many sensitized

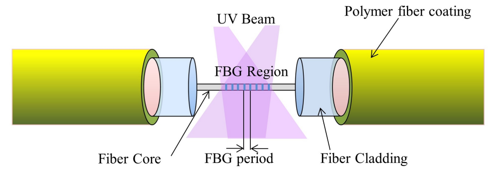


Fig. 1. Fiber Bragg grating fabrication method.

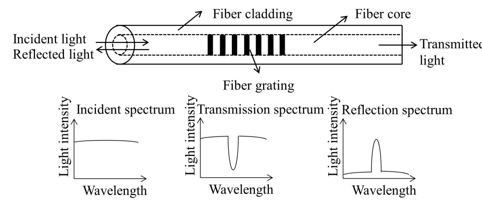


Fig. 2. The working principle of fiber Bragg grating [16].

optical fiber sensors have appeared in the field of medical rehabilitation. Such as fiber Bragg grating sensors, Fabry-Perot sensors, side polished fiber sensors, and micro-nano fiber sensors.

1) Fiber Bragg Grating (FBG) Sensors: Usually, gratings are periodically recorded on the fiber of the point to be measured, which can increase the sensitivity when measuring external physical quantities. FBG sensors are usually fabricated as shown in Fig.1. It is a fabrication method for recording in a standard single-mode fiber using a spatially-varying way with an intense ultraviolet laser. The working principle of the fiber Bragg grating is shown in Fig.2. When a beam of light with a broad spectrum is incident into the grating, the light of a certain central wavelength λ_B is reflected, and the remaining wavelength continues to transmit along the original direction. The reflected wavelength can be expressed as

$$\lambda_B = 2n_{eff}\Lambda. \tag{1}$$

Where n_{eff} is the effective refractive index of the fiber core and Λ is the grating period. Under the action of strain or temperature, n_{eff} and Λ will change, which will lead to the change of the reflection wavelength [14], [15]. It is expressed by the formula as

$$\Delta\lambda_B = [(1-P_e)\varepsilon + (\alpha+\zeta)\Delta T]\lambda_B. \tag{2}$$

Where $\Delta\lambda_B$ is the variation of the central wavelength of the grating, ε and ΔT are the variation of strain and temperature, and P_e, α and ζ are the photo-elastic constant, the thermal expansion coefficient and the thermo-optic coefficient, respectively. The corresponding relationship between the wavelength change and strain or temperature change is calibrated to calculate strain or temperature acting on the fiber Bragg grating. The wavelength multiplexing function of the fiber Bragg grating sensor can realize multi-parameter long-distance distributed measurement, which is mostly used for safety monitoring and fault diagnosis in the engineering field.

2) Fabry-Perot (F-P) Sensors: Reference [17] has summarized the fabrication methods of the standard F-P sensors (as shown in Fig.3), including the chemical etching method [18]–[20]: using chemical reactions to introduce bubbles into

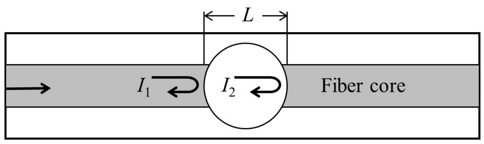


Fig. 3. F-P sensor.

optical fibers to form interference microcavities; laser processing method [21]–[25]: the microcavity is fabricated by femtosecond laser [26]; direct fusion method [27]–[31]: the single-mode optical fiber and the special optical fiber with a gap are directly fused to form a microcavity; polymer-assisted filling method [32]–[35]: in the process of direct fusion, a polymer material is added to the special optical fiber to form a closed air cavity. It has been successfully applied in large bridges, petroleum and coal mines, aerospace, and other fields, and has shown obvious advantages [17].

Another diaphragm-based sensor with a similar structure to the F-P sensor mainly uses a tubular device fixed at the tip of the optical fiber and a diaphragm to form a microcavity to sense external physical quantities. The mechanical properties of the diaphragm determine the sensitivity of the diaphragm-based sensor. The diaphragm is a highly flexible material, such as polymer material, graphene, and composite diaphragm.

The structure of the diaphragm-based optical fiber sensor [36], [37] is shown in Fig.4(a). The tip of the fiber is fixed in a ceramic ferrule with an outer diameter of 2.5 mm, and a ceramic sleeve with a side opening (about 3 mm in outer diameter) holds and assembles it. The facet of the fiber pigtail is polished at an angle of 8° to reduce the Fresnel effect. A 10-layer graphene diaphragm [36], or a composite diaphragm (aluminum sheet and polymer material) [37] is glued to the top of the sleeve with UV-curable adhesive, the diaphragm is pressed onto the sleeve and slightly wrapped around the hollow sidewalls of the cavity, then it is stretched and tension is applied to keep the diaphragm flat. When measuring the sound signal [36], the graphene diaphragm is covered with polymethyl methacrylate (PMMA) with a total thickness of about 5 nm; when measuring pulse signals [37], the optical fiber probe is combined with a sports wristband and fixed on the wrist to improve the convenience and comfort of pulse waveforms measurement, as shown in Fig.4(b).

When a sound wave or pulse signal is applied to the sensor tip, the diaphragm vibrates or deforms accordingly, resulting in a change in the length between the diaphragm and the end of the fiber, realizing the phase modulation of the reflected light. The relationship between diaphragm deformation and optical phase change is shown in (3).

$$\Delta\varphi = \frac{4\pi nh}{\lambda}. \tag{3}$$

where h is the deflection of the diaphragm, λ is the center wavelength of the incident light, n is the refractive index of fiber core, and $\Delta\varphi$ is the phase change. In reference [38], the deflection h of the diaphragm can be expressed, as in (4), where v is the Poisson's ratio of the diaphragm, E is the diaphragm's Young's modulus, and P is the pressure caused

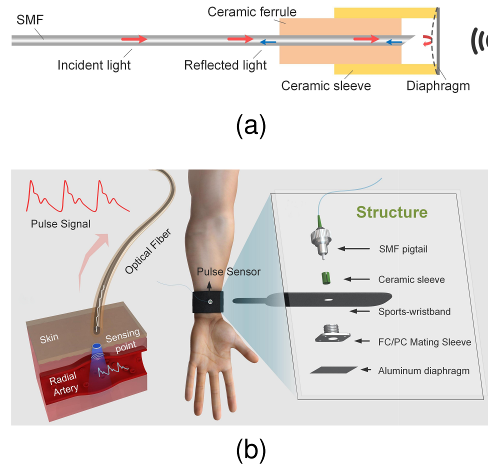


Fig. 4. Schematic diagram of the sensor tip [37]. (a) Detection mechanism of external parameter; (b) Diaphragm-based optical fiber sensor encapsulated method.

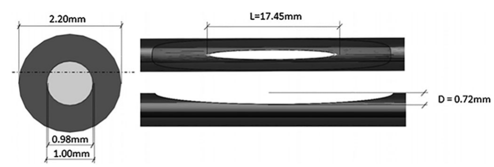


Fig. 5. Structure diagram of the polishing area [40].

by the pulse or sound signal, t and a are the thickness and radius of the diaphragm, respectively.

$$h = \frac{3(1-v^2)P}{16Et^3}a^4. \tag{4}$$

3) Side Polished Fiber (SPF) Sensor: By polishing the sides of the cladding and core of the polymer fiber, most of the cladding and a small part of the core are removed to increase the attenuation of light in the fiber and form a sensitive area. The structure diagram of the optical fiber sensitive area is shown in Fig.5, which shows the side view, top view, and front view of the sensitive area structure. During the optical fiber side polishing process, polishing lengths, depths, and curvature radii will have a certain influence on the sensitivity of the optical fiber sensor [39].

In reference [40], the side polished sensor was made by using polishing equipment and different abrasives under different grinding times and environments. In addition to using industrial equipment, it can also form sensitive areas through 3D printing technology. The polymer optical fiber is inserted into the 3D printed plexiglass mounting fixture, and the side of the core is polished by using diamond sheets of different particle sizes to form a D-shaped sensitive area with a certain depth and length [41].

In reference [42], a dynamic geometric analysis model considering the stress-optical effects and viscoelastic response of polymer fiber is proposed when analyzing the optical loss of side polished fiber under dynamic bending. In static measurement, the optical power ratio of side polished polymer fiber before and after bending is

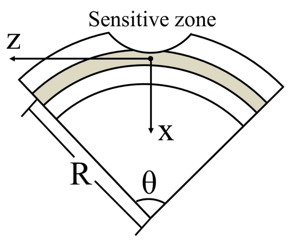


Fig. 6. Polymer optical fiber under bending.

$$\frac{P_{out}}{P_{in}} = \frac{(S_{CO} - S_0)\sin^2(\theta_b)}{S_{CO}\sin^2(\theta_c)}. \quad (5)$$

Where S_{CO} is the cross-sectional area of the fiber core, S_0 is the maximum cross-sectional area after material removal, θ_c is the total reflection angle, and θ_b is the corrected total reflection angle after bending the sensitive area, namely

$$\theta_b = \theta_c\sqrt{(1 - \frac{2a}{R\theta_c^2})}. \quad (6)$$

Where R is the radius of curvature of the fiber when it is bent, as show in Fig.6. During dynamic measurement, the refractive index change caused by stress-optical effects [43] is

$$\Delta n_x = \Delta n_y = \frac{n_c^3 q_{12}\sigma(t)}{2}, \Delta n_z = \frac{n_c^3 q_{11}\sigma(t)}{2}. \quad (7)$$

Where q_{ij} is the stress optical tensor, n_c is the refractive index of the fiber core, and $\sigma(t)$ is the stress tensor [44]. Because of the viscoelasticity of the material, it can be expressed as

$$\sigma(t) = \sigma_0 \exp(\frac{-t}{\tau}). \quad (8)$$

Where τ is the time constant and σ_0 is the initial stress. This method verifies the effectiveness of POF sensor for curvature measurement.

According to different physical quantities, it can be made into refractive index sensor, humidity sensor, temperature sensor, optical power sensor, strain sensor, sound pressure sensor, biochemical sensor, vector magnetic field sensor [45].

4) Micro-Nano Fiber (MNF) Sensors: the standard commercial single-mode optical fiber is about 125 μm, which is larger than that of the micro-nano fiber. Silica glass optical fibers have poor material flexibility and low measurement accuracy. The diameter of the micro-nano fiber is usually close to or smaller than the wavelength of the guided light (about 1 μm or less), which can perform excellent stability and reliability during long hours of work. In references [46], [47], a skin-like wearable optical sensor (SLWOS) was developed. To investigate the temporary response, SLWOS (80 μm thickness, 1.2 μm diameter) was used to measure mechanical vibration and detect acoustic waves and wrist pulse. The response time was about 10 μs below 20 kHz, which was more than three orders of magnitude faster than that of fast-response electronic skin sensor (10-30ms [48]–[50]. SLWOS (thickness 50 μm, diameter 780 nm) has a sensitivity of 1870 kPa^{-1} when detecting pressure below 0.2 Pa, which is much higher than that of high-performance electronic skin sensor (0.55-192 kPa^{-1} [51]–[53]). Fig.7 is a typical illustration of the taper stretching process of the micro-nano fiber sensor (MNFs) during flame heating. A hydrogen flame is used to heat the fiber, and under a certain tension, the fiber is stretched with a gradually decreasing diameter until the desired length or diameter of the fiber taper is achieved. Since the fabricated MNF is connected to the standard fiber through the tapered regions at both ends, it is often referred to as the "biconical" fiber taper or MNF [54].

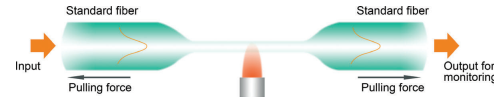


Fig. 7. Schematic diagram of flame-heated taper drawing of a MNF from a standard optical fiber [54].

Micro-nano fibers with different diameters have different percentages of the light field propagating in the form of evanescent waves around them [55]. This part of the evanescent waves can sense changes in some external environment parameters, so it can be made into a highly sensitive sensor. Evanescent wave is a kind of electromagnetic wave produced on the interface of two different media due to total reflection, and its amplitude decays exponentially with the increase of perpendicular depth of the interface. As shown in Fig.8. Evanescent wave field strength [56] can be described as

$$E = E_0 \exp(\frac{-z}{d_p}). \quad (9)$$

Where E_0 is the initial radiation intensity, z is the distance of evanescent field perpendicular to the interface, and d_p is the penetration depth of evanescent wave. Under non-adiabatic condition, the standard single-mode fiber is drawn into multi-mode micro-nano fiber by flame. When the diameter of micro-nano fiber is less than 12 μm, interference occurs between HE_{11} mode and HE_{12} mode at the waist of micro-nano fiber [57], and the interference intensity is expressed as

$$I = I_1 + I_2 + 2\sqrt{I_1 I_1}\cos(\delta). \quad (10)$$

I_1 and I_2 are the intensity of the two modes respectively, and δ is the phase difference caused by the optical path difference between the core and cladding modes [58], which is expressed as

$$\delta = \frac{2\pi L}{\lambda}(n_{eff}^{co} - n_{eff}^{cl}). \quad (11)$$

Where L is the length of the tapered waist of the MNF fiber, λ is the wavelength, and n_{eff}^{co} and n_{eff}^{cl} are the effective refractive indices of the core and cladding, respectively.

However, prefabricated MNFs exposed to air are highly sensitive to environmental disturbances (such as direct physical pressure) or pollution (such as dust adsorption), which can lead to unpredictable changes in optical signals. To make MNFs into wearable sensors, the light field must be properly managed. Reference [46] used a thin layer of polydimethylsiloxane (PDMS), a highly flexible and biocompatible polymer with a refractive index (n = 1.40) slightly lower than that of silica (n = 1.46), to encapsulate MNF and isolate evanescent

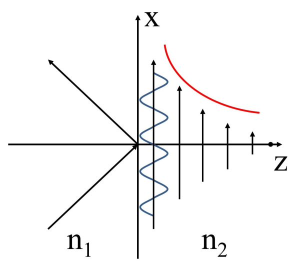


Fig. 8. The principle of evanescent wave propagation.

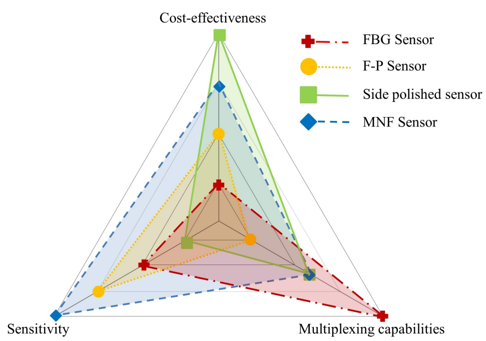


Fig. 9. Performance distribution chart of sensitivity-enhanced optical fiber sensors.

waves while maintaining its high mechanical flexibility and low optical loss. In the PDMS layer, MNF can be made into various shapes. Due to its high sensitivity, it is often made into a highly sensitive skin-like wearable optical sensor to detect high-frequency vibration, wrist pulse, and the human voice [46].

5) Performance Comparison of Sensitivity-enhanced Optical Fiber Sensors: We analyzed the materials, fabrication and application properties of the sensitivity-enhanced optical fiber sensors in Fig.9.

In the process of sensor fabrication, it is necessary to precisely control the period and length of the grating when recording the grating on the optical fiber. In industrial production, a femtosecond laser is often used for recording, and its production cost is relatively high. Compared with the standard F-P microcavity fabrication process, the diaphragm encapsulated process of the F-P sensor is more complicated. Moreover, because of the wavelength/phase interrogation principle of the FBG sensor and the F-P sensor, they need to use an expensive interrogator compared with the intensity interrogation sensor, which increases the signal processing cost. The fabrication of the MNF sensor requires precise micro-and nano-fiber diameters, which is more difficult than the fabrication of the sensitive area of the side-polished polymer fiber sensor. Most of the side-polished optical fiber sensors are made of polymer materials with a core diameter of 980 μm to make sensitive areas with different parameters. They can be produced by industrial methods (such as polishing machines) or by 3D printing technology. And they only need small photodetectors for signal interrogation. It is relatively convenient and the cost is relatively low.

In terms of sensitivity, the smaller the diameter of the fiber,

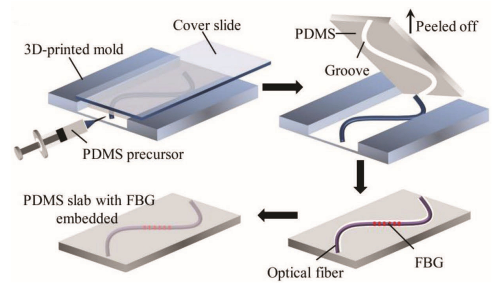


Fig. 10. Optical fiber sensors embedded in flexible substrates [12].

the stronger the sensitivity. The MNF sensor is the fiber sensor with the smallest diameter and has the highest sensitivity; when measuring the two basic physical parameters of strain and temperature, for the fiber Bragg grating sensor of pure fused silica, when the center wavelength is 1550 nm, the axial strain sensitivity of the sensor is about 1.22 $pm/\mu\varepsilon$, the temperature sensitivity is about 10.8 $pm/{}^\circ\mathrm{C}$ [59], [60]; while the strain sensitivity measurement results of Fabry-Perot sensors with various fabrication methods are all greater than 3 $pm/\mu\varepsilon$, and the temperature sensitivity is about 18.3 $pm/{}^\circ\mathrm{C}$ or even higher [61], Fabry-Perot sensors mostly use phase interrogation, so the sensing sensitivity of interferometric Fabry-Perot sensors is higher than that of wavelength-modulated fiber grating sensors.

In terms of multiplexing function, fiber Bragg grating can reflect waves of a certain wavelength according to the determined grating parameters and has strong wavelength multiplexing ability; while the F-P sensor is difficult to be used as a multiplex fiber sensor because of its interference modulation principle. The other two types of sensors are intensity-modulated, and their multiplexing ability is not outstanding, but spatial multiplexing can be achieved by array arrangement, or each sensor can achieve temporal multiplexing under a certain light source activation frequency.

B. Flexible Materials Fabricated or Encapsulated Optical Fiber Sensors

When monitoring the angle of human joints or torso and measuring the awareness of muscle extension and flexion, increasing the flexibility of the sensor can better fit the human joints, muscle tissue, and adapt to the dynamic range of these parameters. Such sensors mainly include flexible materials fabricated or encapsulated optical fiber sensors.

1) Flexible Materials Encapsulated Optical Fiber Sensors: In reference [12], a 3D printing mold was used to fabricate a flexible sensor, and the fabrication process is shown in Fig.10. First, the PDMS liquid was injected into the mold with the optical fiber groove and cured at room temperature, then the PDMS substrate was stripped from the mold, the fiber was buried in the groove of the substrate, and the PDMS liquid was injected to cover the bottom of the groove, then it was cured at high temperature. The encapsulated optical fiber sensor is easily bent, twisted, and stretched, exhibits mechanical properties similar to skin.

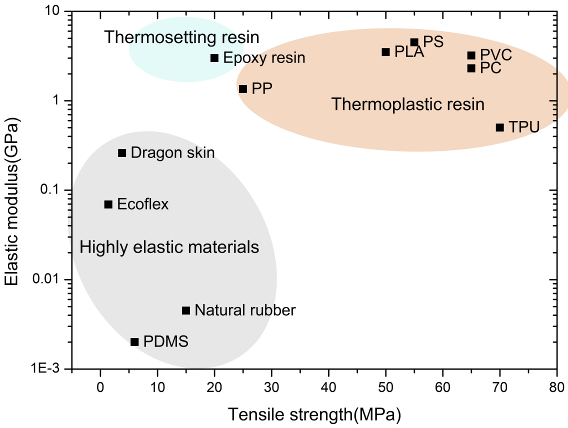


Fig. 11. Tensile strength and elastic modulus of different optical fiber encapsulation materials.

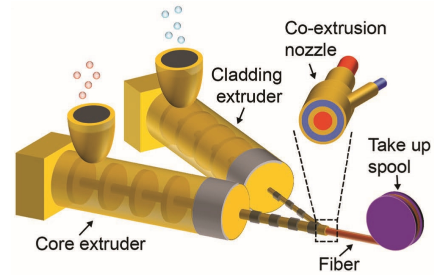


Fig. 12. Co-extrusion process [79].

Due to the shear or torsion of wearable sensors during wear and tear and movement, the sensors are prone to breakage. The packaging material should ensure that the sensor is intact and not damaged, and the packaging method should be simple. It is worth noting that if the elastic modulus of the encapsulation material does not match the skin properties, there will be some loss of strain transmission. Therefore, the performance analysis of packaging materials is very important. The elastic modulus and thermal expansion coefficients of diaphragm materials including glasses, ceramics, polymers, and metals have been reviewed in detail in the reference [62]. On this basis, we further analyze the properties of polymer materials, including thermosetting, thermoplastic resins, and highly elastic materials. When measuring pressure, thermosetting resins (epoxy resin [63], [64]) and thermoplastic resins (PC [11], PS [11], PLA [65], PP [66], TPU [66]) are often selected. For tensile strain measurement, some highly elastic materials such as PDMS (Sylgard 184 silicone elastomer) [67], [68], Ecoflex 00-50 [69], Ecoflex 00-30 [70]–[72], Dragon skin 20 [8], [73], [74] and natural rubber [75] are mainly used. And PVC [76]–[78] is fabricated as a sensing foil to monitor strain in some references. Fig.11 shows the elastic modulus (or 100% modulus) and tensile strength of different packaging materials. It provides a reference for selecting the appropriate packaging material to obtain the synergistic deformation with the human skin to accurately transmit the skin stress and strain, and to ensure the reasonable size and structure design of the sensor.

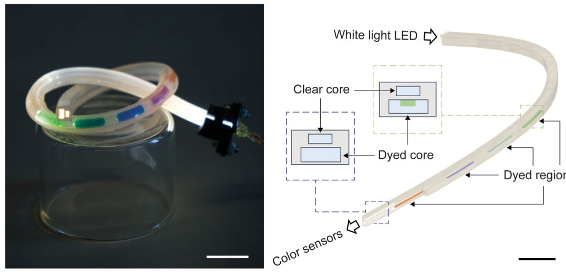
2) Flexible Materials Fabricated Optical Fiber Sensors: Reference [79] evaluated 10 commercially available thermoplastic elastomers. By analyzing the refractive index and elongation at break of the materials, the polystyrene-based polymer Star Clear 1044 (refractive index n = 1.52, elongation at break ε_{max} = 693%) was chosen as the core material, and the fluorinated polymer Daikin T-530 (n = 1.36, ε_{max} = 580%) was chosen as the cladding material. The reason for the choice of materials is that they have very different refractive indices and show similar mechanical properties in the same stress-strain experiments. To make the selected materials into optical fibers, they designed a co-extrusion process, as shown in Fig.12, which consisted of two single-screw extruders. Two selected polymer melts were fed into a custom-made concentric co-extrusion nozzle to form a core-cladding structure. Downstream of the nozzle, the fiber was cooled by the surrounding air and rolled up from a spool. The strain limit of the optical fiber made of these elastomer materials can reach 300%, the bending radius can be close to 0, and the indentation depth can reach 62.5% of the diameter of the optical fiber. The optical fiber sensor shows excellent mechanical properties during stretching, bending, and indentation.

A new type of optical fiber sensor was recently published in the journal Science [80]. According to the fabrication structure and principle of the sensor, the author named it a stretchable lightguide for multimodal sensing (SLIMS), as shown in Fig.13. Its core is made of polyurethane elastomer (Clear Flex 30, Smooth-on Inc.) with a high refractive index (n_{core} = 1.47); the cladding is made of silicone elastomer (Dragon Skin 20, Smooth-on Inc.) with a low refractive index ($n_{cladding}$ = 1.41). The fiber consists of two rectangular cross-section cores stacked on top of each other, separated by cladding material. One of the rectangular cores is transparent, and the other is doped with absorbing colored dyes (EP7701, Eager Polymers). The embedded dyes occupy only a fraction of the core cross-sectional height and are not in contact with the cladding material. The pattern in axial direction can be set as discrete color blocks or continuous gradient color bars, as shown in Fig.13(b), two red-green-blue (RGB) color sensor chips (TCS3472, ams AG) are coupled at the end of the fiber, It can be used not only as of the intensity output of dyed core and transparent core but also as a colorimetric detector. The use of polymer materials increases the flexibility of the sensor. When the sensor is affected by external physical quantities, the force point can be decoupled according to the color change of the fiber end face.

Optical fiber sensors made of flexible materials enable the measurement of large deformation parameters of the human body. Its small interrogator improves portability. The excellent performance of the material makes the sensor very valuable in the scenario of human body movement rehabilitation.

III. Human Biomechanical Parameters Measurement

Patients with stroke or muscle/joint injury, need to resume their daily life through basic training such as breathing training, upper limb joint movement, and walking exercises. Breathing training can not only effectively help patients im-



(a) (b)

Fig. 13. Stretchable lightguide for multimodal sensing [80].

prove their cardiorespiratory capacity, but also monitor abnormal conditions in rehabilitation training promptly; upper limb joint training can gradually improve patients' self-care ability, including the common movements of hand functions such as grasping, pinching, and gripping, and shoulder-elbow-wrist joint coordination exercise for dressing, drinking, and eating; walking is a complex movement completed by multiple joints and systems of the human body, by monitoring the hip-knee-ankle joint angle and plantar pressure, the patient's wrong gait can be corrected, normal gait guidance can be performed. According to different training requirements, optical fiber sensors can be embedded in flexible materials to make sensors that fit human skin, or embedded in exoskeleton devices to measure different physical quantities in single-point, array, and distributed ways. Using optical fiber multiplexing technology, a variety of information can be integrated and processed to achieve the purpose of intelligent analysis and control. Table I lists the applications of optical fiber sensing in measuring human biomechanical parameters.

A. Pulse, Heart Rate, or Respiratory Rate Monitoring

Pulse, heart rate, and respiratory rate are the basic parameters that characterize human physiological health. Under normal conditions, a person's pulse and heart rate are the same, about 60-100 beats per minute, and the breathing rate is about 12-20 beats per minute. In daily monitoring, the patient's abnormal pulse can be detected in time by obtaining the complete pulse waveform and analyzing the characteristic points. The sensing of pulse waveforms by optical fiber sensors has important reference significance for disease diagnosis and daily monitoring. Optical fiber sensors can be placed on different parts of the human body to obtain the pulse, heart rate, or respiration rate.

When monitoring the patient's physiological parameters, we pay more attention to the detection of small deformation and the comfort of the patient's wearing during the measurement process. The optical fiber sensor can be placed on the neck, wrist, and fingers for pulse detection. The neck-fit optical fiber sensor [81] is a plastic optical fiber sensor based on the modulation of reflected light intensity. The sensor includes a cylindrical holder to fix the POF in the center position at a distance of 3 mm from the end face of the sensor head, a circular rigid aluminium coated reflector adhesive (diameter is 5 mm) is taped to the skin of the subject's carotid artery, and the cylindrical support is attached to the reflector. The slight movement of the carotid artery will change the distance between the reflector and the fiber to modulate the reflected light, enabling non-invasive monitoring of the carotid artery pulse. Diaphragm-based sensors made by the same principle [36], [37], [86] are also commonly used for pulse monitoring at the wrist. To improve the sensitivity of small deformation sensing, the reference [67] drew the ordinary optical fiber into a micro-nano fiber and embedded it in the PDMS film of the hybrid plasmonic microfiber knot resonator (HPMKR).

There are two ways to monitor respiratory rate with optical fiber sensors: the optical fiber sensor combined with the elastic band is placed on the waist or abdomen [8], [9], [87], [88], and the optical fiber sensor combined with the humidity sensitive material is placed under the nose [7]. A sensor that integrates a smartphone with a plastic optical fiber [87], uses a flashlight as a light source, a camera as a photodetector, and uses a 3D printed connector for optical coupling. The sensing area is composed of two POF optical fibers separated by a certain distance. With the change of respiration, the displacement between the two fibers changes accordingly, and the coupled output power spectrum is used to characterize the breathing rate; flexible FBG sensor encapsulated by silicone rubber [8], singlemode-multimode-singlemode fiber (SMS) structural wearable optical fiber sensors [88], and polymer optical fiber sensors [9] are placed on the chest or abdomen to monitor respiratory rate or heart rate with the repetitive strain caused by human respiration or heartbeat. For ease of wearing, the sensing element of FBG is functionalized with a hygroscopic coating material [7] which is expanded and contracted by nasal airflow, and the strain of the coating material is transferred to the FBG to monitor the respiratory rate, the design does not require the use of a breathing mask, which improves wearing comfort.

B. Body Surface Temperature and Humidity Monitoring

The measurement of temperature and humidity by optical fiber sensors is mostly used in environmental monitoring in the fields of industry, agriculture, and construction, and relatively few applications in the measurement of body surface temperature and humidity. For patients with stroke or body injury, in their rehabilitation training, real-time monitoring of body surface temperature and humidity and synthesizing other physical information can more comprehensively characterize the patient's training state and degree of fatigue. The measurement of such parameters by optical fiber sensors has certain advantages in measurement safety and accuracy. In reference [89], aiming at the problem that the high complexity of the human body heat transfer process makes it difficult to accurately describe it mathematically, the BP neural network combined with the FBG temperature measurement system was proposed to model the temperature field of the human body surface. This temperature field was realized by measuring at a limited position on the human body surface.

C. Joint Angle Monitoring

Measuring the range of motion of limb joints such as finger-wrist-elbow, hip-knee-ankle joints is a basic step to evaluate

TABLE I

MEASUREMENT OF HUMAN PHYSIOLOGICAL AND MOTION INFORMATION BASED ON OPTICAL FIBER SENSING

Reference	Detection Position	Types of Optical Fiber Sensors	Performance
[81]	Carotid artery	POF sensor	The sensitivity coefficient is $727\mu V/\mu m$; displacement resolution is $0.1\mu m$.
[67]	Wrist	Hybrid plasmonic microfiber knot resonator	The strain response is below 1‰.
[9]	Chest or abdomen	Side polished POF sensor	It can simultaneously monitor heart rate and respiratory rate independent of body movement and monitoring location.
[7]	Hanging on the ear	Humidity sensitive material encapsulated FBG sensor	The mean absolute percentage error of respiratory rate is less than 2%.
[80]	Finger joint	The stretchable lightguide for multimodal sensing	It has a sensitivity to stretching that falls within 2-5 dBϵ^{-1}. It can simultaneously captures proprioceptive motion of three finger joints and external sensation of external pressing.
[82]	Elbow joint	Side polished POF sensor	The angular error is $6.33°$ for different bending speeds and $5.31°$ for different participants.
[70]	Shoulder, elbow, wrist, knee joints or abdomen	A skin-like and stretchable optical fiber	The strain limit can reach 100%; the hysteresis error is less than 4.07%.; the angle measurement range:0-120°.
[79]	Knee or finger joints	Flexible materials fabricated optical fiber sensor	The strain limit can reach 300%, the bending radius can be close to 0, and the indentation depth can reach 62.5% of the fiber diameter.
[74]	Back	Flexible materials encapsulated FBG sensor	The sensitivity coefficient is $0.2nm/m\varepsilon$.
[83]	Ankle joint	F-P sensor	The average sensitivity is $0.029 \pm 0.001mW$.
[84]	Plantar position	Polymer FBG sensor	The minimum sensitivity coefficient is $7.71pm/KPa$; the maximum sensitivity coefficient is $8.51pm/KPa$.
[85]	Plantar position (Carpet)	Side polished POF sensor	The average relative error is 2.9%.

the mobility of patients with muscle, bone, or nerve injury, and it is also one of the metrics to assess the range and degree of joint motor function damage.

The development of optical fiber sensors in joint angle monitoring can be summarized into the following four points. The optical fiber sensors have developed from the monitoring of a single joint angle [69], [79] to the monitoring of multiple joint angles [80]. The encapsulation or fabrication material for optical fiber sensors has been converted from silica glass [69] to more flexible materials [79], [82], [90], [91]. The interrogation method has a trade-off between wavelength/phase interrogation [90] and intensity interrogation [79], [80], [92]. The development of the sensor fusion system of optical fiber sensing combined with other types of sensors. Such as the fusion of optical fiber sensor and IMU [93], has higher measurement accuracy than a single sensor.

D. Muscle Extension and Flexion Monitoring

The optical fiber sensors can sense human motion intentions by detecting the patient's muscle extension and flexion or human-computer interaction force to control the exoskeleton. The flexion or extension intention of a pair of muscles in the arm (forearm) can be detected using the microbending loss principle of the optical fiber sensor [94]. It can be used to control the grasping of objects with the hand exoskeleton, which can be used by people with hand disabilities. Table.II lists the applications of optical fiber sensing in rehabilitation devices.

In the future development, optical fiber sensors can be combined with various sensing devices, such as EEG, EMG, or other sensing devices, to comprehensively judge the patient's movement intention, and assist exoskeleton equipment to achieve precise control.

E. Shape Monitoring

Shape sensing is to place an optical fiber sensor on the object that needs to measure the shape, the strain on the surface of the object is converted into the change of fiber curvature and torsion, and the shape of the object is calculated by plane or space reconstruction algorithm. It is mainly used in building structure monitoring and medical surgical instruments. At present, some researchers have also applied it to the shape sensing of limb movements in human rehabilitation training. They have realized the visualization of the patient's movement posture, which can assist doctors to judge the standardization of rehabilitation movements and customize personalized movement trajectories for patients. In reference [10], optical fiber sensors were fabricated into shape and angle sensors to be mounted on human wrist joints and fingers, as shown in Fig.14, they monitored the spatial motion curves of two degrees of freedom of wrist dorsiflexion-palmflexion and radial-ulnar deviation, realized a high-performance real-time motion capture system with an average error of 1.49 mm for the shape sensor at the distal tip and 0.21° error for the angle sensor. Ma [97] realized the measurement of the joint motion angle based on fiber grating string, combined with

TABLE II
APPLICATIONS OF OPTICAL FIBER SENSING IN REHABILITATION DEVICES

Reference	Detection Position	Types of Optical Fiber Sensors	Performance
[94]	Forearm flexors and extensors	Optical fiber microbend loss sensor	It can real-time judgment flexion and extension awareness of muscles to complete the grasp and release of objects.
[95], [96]	Shank flexors and extensors	Flexible encapsulated polymer FBG sensor	The relative error of human-computer interaction force is less than 4.5%.
[4]	Exoskeleton knee joint	Side polished POF Sensor	The root mean square error of exoskeleton knee joint angle measurement is 3.8°.

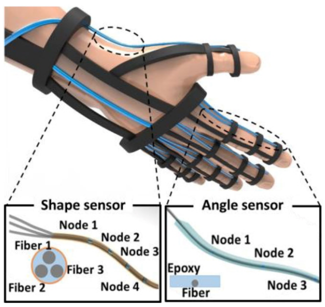


Fig. 14. Hand motion capture system [10].

arm length parameters, inverted the three-dimensional shape of limbs, and applied machine learning algorithm to complete posture classification to achieve the purpose of upper limb posture recognition. Reference [98] measured the joint angles of the lower limbs based on optical fiber sensing, which can accurately and visually capture the movement intention, spatial position and posture of the lower limbs of the human body, and identify various motion modes of the lower limbs.

F. Gait Monitoring

Neurodegenerative diseases such as spinal muscular atrophy, multiple sclerosis, cerebrovascular and cardiovascular diseases, and physical aging can lead to gait impairment [99]. Gait analysis can detect negative deviations from normal gait patterns and their causes to quantify the parameters involved in lower extremity movement and analyze the movement mechanisms that control the human body [100]. According to the gait cycle pattern, parameters such as anthropometry, spatiotemporal, kinematics, kinetics, and dynamic EMG can be monitored to assess the health status of patients.

Gait curves can be obtained by monitoring knee joint angle [101], ankle joint angle [83], [102], or plantar pressure [84], [85], [103]–[105], as shown in Table I. The ankle joint angle measurement [83] focuses on the reconfigurability of the sensor. By designing the corresponding adapter for the sensor, the sensor can be repeatedly installed and removed. Polymer FBG sensors [84] and silica FBG sensors [103] are commonly used in pressure insoles. The polymer optical fiber has a higher strain limit, fracture toughness, bending properties, impact resistance, and a higher sensitivity coefficient than the optical fiber sensor with the same structure made of silica. The interrogation principle of these two kinds of optical fibers is to interrogate the shift of the output light wavelength to obtain the plantar pressure. Reference [104] embedded a side-polished polymer optical fiber sensor array into the insole to measure the ground reaction force and gait events recognition. In addition, plantar shear force was also measured in reference [105], which is helpful in the analysis and diagnosis of plantar pathology. The ground reaction force and spatiotemporal gait parameters are measured by the intelligent pressure pad [85], which is low-cost and does not need to be worn and is suitable for indoor gait monitoring, but the walking range is limited. Because the signal interrogation method of the intelligent pressure pad is intensity interrogation, it is simpler than the former wavelength interrogation methods.

G. Distributed Physical Quantities Monitoring

In addition to the single-point monitoring summarized above, optical fiber sensors can also be used for quasi-distributed and distributed monitoring. Reference [106] proposed a sensor based on the change of light intensity of polymer optical fiber to realize quasi-distributed multiplexing technology in the way of time multiplexing, as shown in Fig.15, which represents the sensor multiplexing technology in three scenarios. The first scenario is a lower limb extension-flexion motion orthosis fixed on a seat, using polymer sensors to quantify the extension-flexion angle and human-computer interaction force during flexion-extension motion. The second scenario is a modular lower limb exoskeleton with gait assistance. In addition to the angle and human-computer interaction force, the treadmill is also equipped with a POF sensor to measure the ground reaction force. The third scenario is that people use breathing belts and insoles to monitor breathing rate and gait events while walking. Scenario 1 and 2 multiplex a single fiber, using five light sources as the input light sources for each sensor, which are activated sequentially in a predefined order and frequency, and only one LED is activated at a time, and the output response matrix is obtained using two photodetectors. To accurately interrogate each sensor's signal, compensation technology is also applied to eliminate the influence of other sensors. This quasi-distributed multiplexing technology is the unique advantage of optical fiber sensors. Multiplexing methods using single-point optical fiber sensors also include wavelength division multiplexing, carrier frequency division, spatial division, and coherent domain multiplexing [107], which is to achieve quasi-distributed multi-parameter measurement by expanding the number and types of single-point sensors on the sensing line.

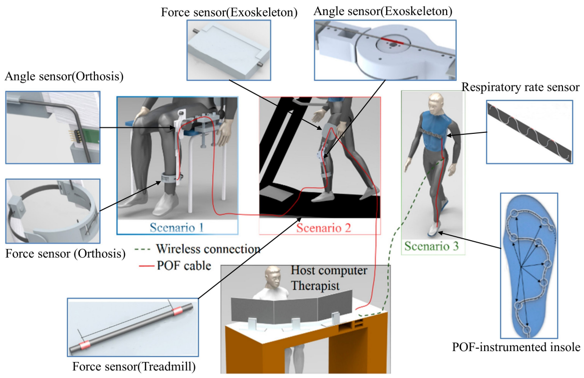


Fig. 15. The optical fiber sensor multiplexing technology [106].

Different from single-point optical fiber sensors, distributed optical fiber sensors detect external parameters through a variety of scattered light intensity, frequency, and phase information in the fiber. Its sensing unit is an optical fiber, and the measured physical quantity depends more on the interrogator and algorithms than the optical fiber itself. The interrogation technology of distributed optical fiber sensing includes optical time domain reflectometry (OTDR), optical frequency domain reflectometry (OFDR), and optical coherence domain reflectometry (OCDR). Among them, the distributed sensing system based on OFDR of back Rayleigh scattering has high spatial resolution and large dynamic range [108]–[110]. Its detection principle is to detect temperature, strain, and vibration in non-extensible structures by interrogating back Rayleigh scattering signals in optical fibers, and it is widely used in structural health monitoring, biomedical applications, border security, and pipeline intrusion monitoring [110]. Although distributed sensing is rarely used in the monitoring of biomechanical parameters, the parameters such as 3D shape [111], strain [109], temperature [109], pressure [112], and airflow [113], which are suitable for use in the field of medical rehabilitation. With the development of flexible optical fiber materials and the advancement of interrogation technology, the high sensitivity and high spatial resolution of the distributed optical fiber sensing system has a great possibility to be used to monitor the temperature distribution in the local area of human skin, the pressure and touch of the skin, the distribution of plantar pressure, and human body posture, which has greater advantages over quasi-distributed optical fiber sensing such as FBG arrays. Therefore, distributed optical fiber sensing has significant potential in the application of medical rehabilitation in the future.

H. Comparison of Application Scenarios of OFSs and Electronic Sensors

The wide application of optical fiber sensors in medical rehabilitation is a good solution for them to become wearable systems. We compare the performance of optical fiber sensors and electronic sensors in different scenarios. As shown in Table III. We can find out that the main advantages of the optical fiber sensor are: (1) They are not affected by the measuring environment; Optical fiber sensors are passive optical device, and their safety in measuring body surface humidity is higher than that of electronic sensor; In MRI environment, optical fiber sensors are less susceptible to interference compared with electronic sensors. (2) Multiple parameters can be measured independently or simultaneously; In the comprehensive evaluation of human physiological and motion parameters, it is necessary to wear many different types of electronic sensors, while distributed multi-parameter measurement can be realized by using optical fiber sensors in one optical fiber or multichannel optical fiber sensing with one interrogator. (3) Cost-effectiveness; when measuring plantar pressure based on intensity interrogator, the pressure insoles [4] and smart carpet [85] made with optical fiber sensors are cheaper than commercial electronic insole (such as Xsensor X4) and plantar pressure distribution measurement system (such as Zebris plantar pressure measuring plate). (4) Sensing performance advantages: A SLWOS was developed in reference [46]. Its response speed and pressure sensitivity are higher than electronic sensors.

The advantages of electronic sensors are reflected in (1) high integration, which is widely used in the field of medical rehabilitation; (2) high measurement precision and accuracy, and newly developed sensing devices must use the data of electronic sensors as ground truth.

IV. Discussion

From the above review and analysis, the current wearable optical fiber sensing system is mostly a monitoring system of human biomechanical parameters, but not further used as feedback information for human health assessment and robot on-demand assistance. How to integrate a wearable fiber optic sensing system into an exoskeleton assistive device is still in the proof-of-concept stage, as in [106], this is the first time optical fiber sensor is used as a multiplexed and integrated solution, which provides a research direction for the realization of compact, fully integrated and low-cost systems in the future and application in the clinical environment. However, the integration of the optical fiber sensing system is not enough, which is not comparable to the wide clinical application of electronic sensors, there are still many challenges in the structural design of the sensor, the selection of the measurement position, the interrogation of the signal, the multi-parameter measurement, and the production cost. These problems are discussed in detail and the directions for solving these problems are pointed out.

Wearable sensors must have high sensitivity and flexibility. Its structure needs to accurately sense the state changes of the points to be measured, fit the joints/muscles, and meet the motion range of various parts of the human body. The requirements of sensor wearability are achieved by flexible encapsulation or fabrication and reasonable arrangement of optical fiber sensors. Specifically, it is necessary to analyze the mechanical properties of flexible materials and verify the stability and repeatability of materials for long-term monitoring; flexible encapsulation/fabrication materials have a certain impact on the sensor's sensitivity, the use of humidity-sensitive materials to encapsulate FBG sensors [7], the use of diaphragm-embedded sensors made of graphene [36], composite materials [37], these materials improve the sensor's

TABLE III
COMPARISON OF APPLICATION SCENARIOS OF OFSS AND ELECTRONIC SENSORS

Application	Optical fiber sensing system	Electronic sensing system
Pulse/heart rate	SLWOS, fast response time (10 μm) below 20Hz [46], [47].	Commercial smart watches, health monitoring bracelets: high integration, accurate measurement.
Muscle deformation	SLWOS: high sensitivity to micro-deformation (Pressure sensitivity coefficient is 180 kPa^{-1} [46], [47]).	High-performance electronic skin electronic skin (Pressure sensitivity coefficient is 0.55-192kPa-1 [51]–[53]).
Respiratory rate	Humidity-sensitive material-encapsulated optical fiber sensor, which measures respiratory rate by monitoring changes in nasal airflow humidity, the percentage error of respiratory rate is $\leq$ 2.29 % and deviation $\leq$ 0.31 [7]. The sensor adopts the earhook type to improve comfort and acceptability.	Gas metabolism analyzer (COSMED), data accuracy $\pm$0.02 %, but it requires wearing a mask.
Body surface	Distributed FBG sensor, the highest prediction error of body surface temperature field is 0.33°C [89].	Commercial electronic thermometer, it is easy to use.
Joint angel/ Angular velocity	SLIMS, which can measure multi-joint angles in a distributed manner; simultaneously captures proprioceptive motion of three finger joints and external sensation of external pressing [80].	IMU: requires frequent calibration; Video capture systems: limited to laboratory environments.
Shape (posture)	Multi-core FBG sensor, 1.49 mm average error at the distal tip of the sensor (1.9% average error over the full length of the sensor) [10].	Video capture systems: It has high measurement accuracy, low delay, and it is very stable.
Gita parameters	SPF sensors, Optical fiber sensing insoles [4] and smart carpet [84], which are based on intensity demodulation, do not require expensive interrogator. So they are cost-effective.	Commercial electronic insoles (Xsensor X4) and plantar pressure measurement platforms (Zebris) are expensive.
Mult-parameters	It only needs one fiber or multiple fibers with one interrogator.	It requires different electronics.
In an exoskeleton-assisted rehabilitation system	Anti-electromagnetic interference.	They are susceptible to interference from actuators.

sensitivity, enabling it to monitor the human body's respiration, pulse, heart rate and other tiny deformation information; while the PDMS film is used to isolate the evanescent wave of the micro-nano fiber [46], reducing the sensitivity of the micro-nano fiber, making it more stable when measuring the human pulse signal. However, the flexible materials will cause the sensor to have a certain hysteresis problem [114], [115]. It is necessary to adopt hysteresis compensation technology to make the measurement parameters more accurate. When measuring human biomechanical parameters, the effect of the relative displacement and friction force between the sensor and the contact surface of the human body on the measurement accuracy should be considered. Kinesiology tape can be used to improve the problem of relative movement between the sensor and the skin during joint angle measurement; a correction algorithm (removing baseline drift algorithm) can also be used to solve the problem of waveform dirft when measuring human respiration rate with a breathing belt. In order to meet the motion range of each joint angle, the design of the incorporation of the sinuous-shaped FBG structure and the stretchable substrate [12], or the continuous bending of the fiber according to the principle of fiber bending loss [70], can greatly improve the strain limit of optical fiber sensing. However, when distributed optical fiber sensors are used for dynamic measurement of human body parameters, the influence of bending loss on scattered echoes should be avoided.

There are various sensor placement positions for gait monitoring. Gait curves can be displayed based on hip/knee/ankle angles, plantar pressure or muscle deformation data. Scholars selected the gastrocnemius, soleus and tibialis anterior muscles of the calf as the main measurement positions by studying the active relationship between the gait cycle and lower limb muscle groups [116]. Therefore, when using optical fiber sensing to monitor muscle deformation in the future, we can refer to this type of muscle group that contributes more in daily movement as the preferred measurement location.

Although the current FBG/F-P interrogator and optical reflectometry have complete functions, they are expensive, relatively bulky, unable to achieve the purpose of portability, and cannot be widely used. In future clinical applications, with the improvement of device structure and interrogation algorithm, the whole sensing system will be lighter, miniaturized, and easier to wear.

For wavelength-modulated optical fiber sensors, multi-parameter measurement can be achieved on one fiber, which is its unique advantage. Phase- and intensity-modulated optical fiber sensors can also be used to realize multi-parameter measurement functions arranged in time or space. The 2D sensing carpet based on distributed optical fiber sensing realizes high-resolution measurement of position and pressure, which has higher spatial resolution and measurement accuracy than quasi-distributed measurement [117]. In addition, Multi-dimensional modeling [80] and machine learning methods can be used to decouple multi-joint angle parameters. In view of the complexity and poor accuracy of gait division, machine learning methods are also commonly used in gait phase recognition and gait online prediction [118]. Multi-parameter measurement can realize more comprehensive analysis and evaluation of the patient's disease.

The complex structure design and the selection of phase/wavelength interrogation methods can highlight the performance of the sensor, but greatly increase the cost. Some wearable sensors, such as plantar pressure insoles, are consumables and need to be customized to the patient's needs. In the future, in order to reduce the cost reasonably, the design of the sensor can be simplified. For the measurement of the joint angle, the combination of the adapter and the sensor unit can be used [83] to realize the reuse of the sensor. There is a trade-off between performance and cost-effectiveness in monitoring of human biomechanical parameters.

V. Conclusion

For the needs of wearable sensing technology, combined with the advantages of optical fiber sensing, this paper provides a comprehensive review of the application of optical fiber sensing in medical rehabilitation, and summarizes the development of different optical fiber structures, flexible material encapsulation and fabrication methods, interrogation techniques, human body signal monitoring, and intelligent control of assistive rehabilitation equipment. By single-point, array, and distributed multi-sensor monitoring and fusion processing, it can provide disease prevention and diagnosis for patients with movement disorders, formulate personalized rehabilitation programs, and comprehensively evaluate patients' rehabilitation effects. In the future, the selection of optical fiber sensors needs to comprehensively consider the aspects of the sensor fabrication, material properties, interrogation and signal methods. The development of optical fiber sensors in medical rehabilitation mainly focuses on the further improvement of material properties, the miniaturization of the interrogator, multi-sensor fusion processing methods (such as machine learning methods), and real-time data processing.

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