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Review

Fiber Optic Intensity-Modulated Sensors: a Review in Biomechanics

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Abstract: Fiber optic sensors have a set of properties that make them very attractive in biomechanics. However, they remain unknown to many who work in the field. Some possible causes are scarce information, few research groups using them in a routine basis, and even fewer companies offering turnkey and affordable solutions. Nevertheless, as optical fibers revolutionize the way of carrying data in telecommunications, a similar trend is detectable in the world of sensing. The present review aims to describe the most relevant contributions of fiber sensing in biomechanics since their introduction, from 1960s to the present, focusing on intensity-based configurations. An effort has been made to identify key researchers, research and development (R&D) groups and main applications.

Keywords: Biomechanics, fiber optic sensors, intensity-modulated sensors

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1. Introduction

Biomechanics is the mechanics applied to living bodies with special emphasis given to the human body. It is a field of a confluence of several disciplines from engineering, medicine and sports, such as mechanics, anatomy, physiology, orthopaedics, rehabilitation, ergonomics, kinesiology, motor control and many others.

A major topic is movement analysis. It helps to assess the body's kinematics and dynamics, either to optimize an athlete's skill and his performance, either to assess gait patterns and postures of injured subjects. Some good examples of the optical technology applied to this topic are: (1) 3D motion capture systems using the advanced digital optical technology [1–4]; (2) pedobarographs or optical pressure platforms [5–8]; (3) fiber optic goniometers [9, 10].

Clinical biomechanics is also an important topic. Among a wide variety of applications, it includes the design of orthopaedic devices, such as prosthesis and implants. Thus, the issues of materials biocompatibility, their physical and mechanical properties, along with finite element analysis and mechanical tests to optimize the design and predict the devices performance, their durability and efficacy, are frequently reported. Consequently, *in-vitro*, *ex-vivo* and *in-vivo* studies are regular in biomechanics. More recently, interesting topics including tissue, cell and molecular biomechanics have been introduced. A good example of the optical technology applied to this topic is near infrared (NIR) spectroscopy (NIRS), allowing to assess the

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optical properties of cartilage and evaluate low-grade lesions [11–14]. The clinical biomechanics also poses interesting challenges in developing sensors for minimally invasive procedures, capable of not disturbing the natural biomechanics of body structures, mainly if *in-vivo* applications are pursued. That is why optical fibers sensors hold enormous potential for the use in the biomechanics. Due to the biocompatibility of the high-purity fused silica glass (SiO_2), an optical fiber has the potential to neither adversely affect the physiological environment, nor be adversely affected by it [15]. Other important attributes that will be discussed in this paper include small size, light weight, geometrical flexibility, chemical inertness, electric and thermal insulation, and immunity to electromagnetic interference [16–19].

An optical fiber guides light making possible to illuminate and capture images from the inside of the body. Indeed, the initial optical fiber based systems were proposed for endoscopic procedures, still before the 1960s [20]. It was, however, the possibility of using an optical fiber to carry information that revolutionized the world of communication. Furthermore, an optical fiber allows to relate a change in radiation properties (intensity, optical frequency, phase and polarization) with a change in a physical quantity (e.g., strain and pressure), and this possibility is also introducing substantial changes in the world of sensing.

The initial fiber optic sensors were proposed during the 1960s, based on intensity-modulated configurations. Since that time, a myriad of solutions have been presented covering many configurations and applications. However, the vast majority have been used for research and investigational purposes. With some exceptions, few have presented turnkey solutions, and fewer have reached commercialization. Table 1 is a list of companies offering fiber optic sensing solutions for the biomechanics and other related applications. Nevertheless, these companies and fiber optic sensors remain unknown to many engineers, biomechanists, clinicians and researchers. Most likely, this is related to the fact that their education and practice are focused on conventional sensors and non-optical technologies.

Table 1 Companies in the market offering fiber optic sensors suitable for biomechanics applications.

Company	Local, country	Website
5DT Inc.	Irvine, CA, USA	www.5dt.com
ADInstruments, Inc.	Colorado Springs, CO, USA	www.adinstruments.com
Arrow International, Inc (Teleflex Medical)	Research Triangle Park, NC, USA	www.arrowintl.com
BioTechPlex	Escondido, CA, USA	www.biotechplex.com
Camino Laboratories (Integra LifeSciences)	Plainsboro, NJ, USA	www.integralife.com
Delsys Inc.	Boston, MA, USA	www.delsys.com
Endosense, SA	Geneva, Switzerland	www.endosense.com
FISO Technologies	Québec, Canada	www.fiso.com
InnerSpace Medical, Inc.	Tustin, CA, USA	www.innerspacemedical.com
InvivoSense	Trondheim, Norway;	www.invivosense.co.uk
LumaSense Technologies	Santa Clara, CA, USA	www.lumasenseinc.com
Luna Innovations	Blacksburg, VA, USA	www.lunainnovations.com
MAQUET Getinge Group	Rastatt, Germany	http://ca.maquet.com
Measurand Inc.	New Brunswick, Canada	www.measurand.com
Neoptix Inc.	Québec, Canada	www.neoptix.com
Opsens	Québec, Canada	www.opsens.com
Radi Medical Systems (St. Jude Medical Systems AB)	Uppsala, Sweden	www.radi.se
RJC Enterprises, LLC	Bothell, WA, USA	www.rjcenterprises.net
Samba Sensors	Västra Frölunda, Sweden	www.sambasensors.com

The present review aims to identify the most relevant contributions in biomechanics oriented fiber optic sensing, pointing out applications, researchers and research and development (R&D) groups that have been working in the field and related areas.

2. Sensor classification

Fiber optic sensors can be classified accordingly to their working principles into some major categories. One of them relies on the modulation by the measurand of the light intensity, identified as intensity-modulated configurations, the first to be reported in the literature. Nowadays, they stand for a mature solution in many applications and are relatively simple to interrogate [19]. Sensing devices based on fiber Bragg gratings and Fabry-Pérot structures are also of great interests and have already been applied in the biomechanics. Compared to intensity-modulated schemes, they stand for higher sensitivity and resolution, but at the expense of relatively complex interrogation/detection techniques [21]. Our approach has been focused on intensity-modulated configurations, but all of them should be addressed if the full spectrum of biomechanics applications has to be known.

Most common configurations of intensity-modulated sensors applied in biomechanics are:

(1) An optical fiber with its tip placed in front of a movable reflecting membrane/mirror. The optical fiber guides the light of the source to the fiber tip, and under the influence of the measurand, the original membrane distance to the fiber tip changes, as well as the intensity of the reflected light that is coupled by the same fiber or another fiber parallel to the first one (Fig. 1). As it will be seen, first studies made use of similar configurations. However, instead of a single optical fiber, bundles of optical fibers have been used as waveguides.

(2) An optical fiber submitted to bending or curvature. These actions will result in light loss into the cladding and lead to a decrease in the light intensity (Fig. 2).

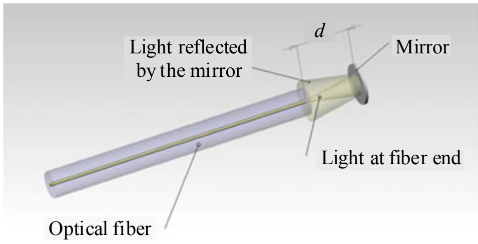

Fig. 1 An optical fiber placed in front of a movable reflecting membrane/mirror: the back-reflected intensity decreases when the distance, *d*, increases.

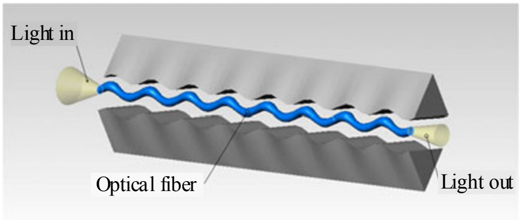

Fig. 2 Light losses due to microbending.

3. Earlier intensity-modulated configurations

The initial papers reporting fiber optic sensors were published in the early 1960s. They were based on intensity-modulated schemes and initially proposed for the intravascular and cardiac applications. In 1960, Michael Polanyi (American Optical Company, Southbridge, MA) and Robert Hehir (St. Vicent Hospital, Worcester, MA) presented an optical system for measurement of the *in-vivo* oxygen saturation and dye concentration in the blood [22] (Fig. 3).

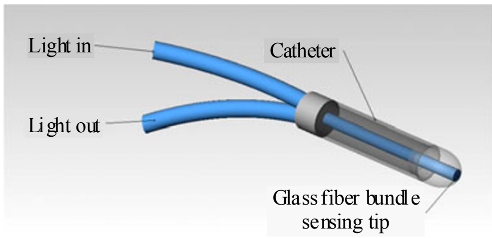

Fig. 3 Schematic drawing of the Polanyi sensor (adapted from[23]).

The innovative contribution of the system was its sensing probe, made of two glass fibers bundles located within a catheter (about 150 fibers with about 50 μm in diameter each) [22–24]. These two bundles have been used as waveguides, one to guide the filtered light from a tungsten lamp source to the tip of the catheter, the other to guide the back-scattered and diffusely reflected light, modified in its spectral distribution due to blood interaction, into a photocell. The pulses of the source were located at 805 μm and 660 μm, to measure the oxygen saturation, and at 900 μm and 805 μm, to measure the dye concentration [23]. The reflected light has been analyzed spectrophotometrically, on the basis of the linear relationship between the measurand and the ratio of the intensities of the two reflected wavelengths [25]. The following years have been particularly prolific using this or similar techniques in the laboratory and clinical environments [26–35].

It was also in the 1960s that fiber optic sensors became interesting for pressure measurement [36–39]. They intended to solve the drawbacks of standard fluid-filled catheters [40], such as hydrostatic artifacts caused by body movements and the necessity of flushing them to maintain accuracy [41]. The working principle of these new sensors also is based on the variation of the light intensity. To sense pressure, the light from the source is guided to a movable membrane which, under pressure, reflects the light back to a photodetector [36, 37]. Several US patents were presented at that time [42–44]. Among all contributions, the work of Lekholm and Lindström at the Research Laboratory of Electronics of the Chalmers University of Technology (Gothenburg, Sweden) deserves to be highlighted. Authors have presented a sensor for *in-vivo* blood pressure measurement [37, 39] (Fig. 4).

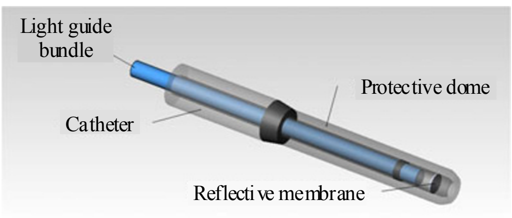

Fig. 4 Schematic drawing of the Lekholm and Lindströn sensor (adapted from [39]).

The above sensor was extensively described, covering the theoretical topics of fiber optics properties, membrane reflection, operation modes, number of fibers and their distribution, membrane mechanics, volume displacement, frequency dependence and limitations [39]. Error sources, sensitivity and miniaturization, failure and redundancy were also addressed [39].

One of the innovative features of the sensor was its miniaturization, evidencing sensor heads of only 0.85-mm (unshielded) and 1.5-m diameters. These heads consisted of air-filled chambers covered by a 6-μm pressure sensitive membrane of beryllium-copper. As in previous works, the guiding system was made of two independent optical fiber bundles, one to guide the light, from a gallium-arsenide light emitting diode (LED) source to the sensor head [later versions included a microminiature glow lamp powered by the direct current (DC)], the other to guide the reflected light into a photodetector. Near the reflective membrane, the optical fibers were randomly distributed allowing for higher miniaturization of the sensor head [39]. Another interesting feature of the sensor was its insensitivity to mechanical vibrations, shocks, and movements due to a light and stiff membrane. The cross sensitivity to temperature has been observed. Nevertheless, under temperature variations from 20 ℃ to 37 ℃, zero drift was reported after about 40 s [39]. The initial fabricated probes had a flat frequency response from static pressure to 200 Hz [37], increasing to 15 kHz in the following experiments [39]. After successful tests on one dog and one man [37], clinical tests have followed [39].

Similar intensity-modulated sensors with their membranes located at the tip of the sensor have since been reported for intravascular and intracardiac pressure measurement [45, 46]. Nevertheless, sensors with membranes located at the tip of the probe could lead to erroneous intravascular readings due to tip collisions with the blood vessels or the ventricular walls (the so-called wall or piston effect) and promote clot formation for long periods of monitoring [47, 48]. These drawbacks could be reduced by changing the location of the sensing membranes to the sides of the probe. Taylor *et al*. in 1972 [47] and Matsumoto *et al*. [48] in 1978, implemented this feature in fiber optic sensors, intended to monitor multiple physiologic changes, such as the cardiac output, oxygen saturation, dye clearance, intravascular pressure, and heart rhythm (Fig. 5). Nevertheless, tip and side-hole configurations have been adopted up to today. In fact, the most important achievement in the following years was miniaturization of sensor probes using microfabrication techniques [49–53].

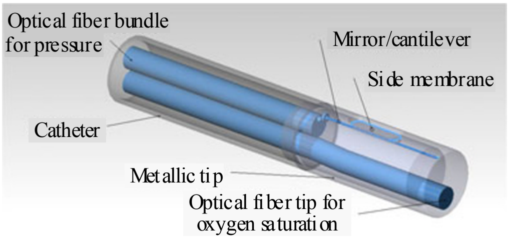

Fig. 5 Schematic drawing of the Matsumoto sensor, intended for pressure and oxygen saturation measurement, adapted from [48] (a side membrane has been used for pressure measurement and a tip configuration for oxygen saturation).

Besides intravascular pressure measurement, similar intensity-modulated configurations to the one proposed by Lekholm and Lindström [37, 39] have been explored to measure pressure in other sites of the human body. For example, in the 1970s, Epstein *et al*. [54] and Wald *et al*. [55, 56], both from the Department of Neurosurgery and Neurology of the New York University Medical Center, were the first to apply optical fibers to measure the intracranial pressure (ICP). Vidyasagar *et al*. [57, 58] adapted the technique for non-invasive purposes through the measurement of the anterior fontanel pressure in newborns. Authors stated the advantage of electric insulation provided by optical fibers to eliminate the risk of electric shocks. The system they have used was probably the first to be commercially available (Ladd Intracranial pressure monitoring device, Model 1700, Ladd Research, Williston, VT). However, while a significant correlation between the anterior fontanel pressure and ICP was reported, the same was not observed by others [59]. It seems the extradurally technique has the disadvantage of signal damping and a tendency to read higher than the true ICP [60].

4. Earlier commercial solutions

The configuration proposed by Lekholm and Lindström [37, 39] was also the basis for the development of Camino pressure sensors, probably the most widespread dual-beam referencing intensity-modulated based sensors (Camino Laboratories, San Diego, CA, USA; acquired by Integra LifeSciences; Plainsboro, NJ, USA) [61]. In 1996, Keck reported the company had been producing around 60 000 devices/year [62]. This transducer-tipped catheter consisted of a tip enclosed in a saline-filled sheath with side holes (Fig. 6). A pressure sensitive diaphragm varied its distance to the optical fibers changing the intensity of the reflected light.

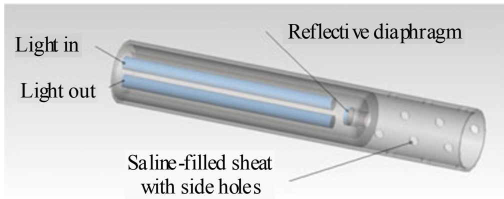

Fig. 6 Schematic drawing of earlier Camino sensors (adapted from [63]).

Following the above original contributions, Camino sensors became popular in the 1980s, and since that time they have been extensively used for pressure measurement in different sites of the body, as in the brain, muscles and joints.

In the late 1980s, Crenshaw *et al*. [63], from the Division of Orthopaedics and Rehabilitation of the University of California (San Diego, CA, USA) and the NASA-Ames Research Center (Moffett Field, California, USA), were the first to apply Camino sensors (model 110-D) to measure intramuscular pressure (IMP), either in animals or in human volunteers. These sensors proved to be insensitive to hydrostatic artifacts caused by body movements and capable of long-term measurement (≈2.5 h) without flushing them to maintain accuracy [63]. Conversely, long-term measurement was also associated with patient discomfort, probably due to the size and rigidity of the polyethylene sheath enclosing the sensor. Even so, these IMP sensors were used in many biomechanics applications, such as during isometric and concentric exercises [64]; to demonstrate that IMP varied with the muscle depth [65]; to study compartment syndrome following prolonged pelvic surgery [66]; and to analyze muscles contribution during gait [67].

Pedowitz *et al*. [68], also from the Division of Orthopaedics and Rehabilitation of the University of California, applied Camino sensors to measure intraarticular pressure (IAP), namely during continuous passive motion of the knee joint, a common post-surgery therapeutic procedure. In the following years, IAP was also monitored in cadaveric glenohumeral joints to study its relation with the range of movement of the shoulder joint [69]; during typing tasks to measure cubital tunnel pressures [70]; and in patients suffering from cubital tunnel syndrome [71, 72].

It was, however, for ICP measurement that Camino sensors became popular, namely the model 110-4B. They were considered to be accurate and reliable for ICP monitoring, evidencing high-quality readings under laboratory and clinical conditions, a good correlation with strain gauge sensors and fluid-filled systems, insensitivity to hydrostatic artifacts and no flushing or infusion requirements [73–78]. On the other hand, they also underwent extensive scrutiny leading to identification of several drawbacks and questioning their routine use, particularly in clinical practice. Reported drawbacks included sensor failure (e.g., breakage, cable kinking, probe dislocation, abnormal readings), contamination, infection, hemorrhage, drift, and magnetic resonance imaging (MRI) incompatibility due to the presence of ferromagnetic components [74, 76–88].

Alternative sensors were proposed, particularly using Fabry-Pérot configurations [41, 89, 90], but their description is away from the scope of the present review.

Meanwhile, a good example of novel applications of light intensity-modulated sensors supported by reflective membranes is the radiofrequency (RF) ablation catheter with force feedback, presented by Polygerinos *et al*. [91, 92], from the Department of Mechanical Engineering of King's College of London. Three plastic optical fibers were aligned inside a plastic catheter in a circular pattern to provide a three axes force sensing system (Fig. 7). The sensor was tested in an artificial blood artery showing a working range of 0 to 1.1 N, a resolution of 0.04 N and good dynamic response.

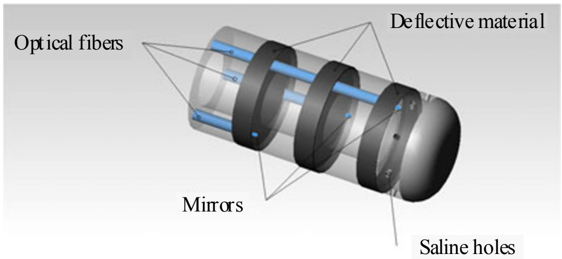

Fig. 7 Schematic drawing of the Polygerinos sensor (adapted from [91, 92]).

5. Intensity-modulated sensors based on bending

Intensity-modulated schemes based on macro- or micro-bending were proposed for biomechanical applications. As for almost intensity-based sensors, they were easy to fabricate and require simple interrogations techniques if fine precision was not required [93–95]. One of the first applications in biomechanics was in dentistry. In 1995, Kopola *et al.* [96], from the University of Oulu (Finland), proposed a device to measure human biting forces consisting of a mouthpiece, made of two stainless steel plates, and a microbending fiber optic sensor placed between them. The sensor was able to measure forces ranging from 0 to 1000 N with a resolution of 10 N [96].

Also in the 1990s, another group from Finland, led by Paavo Komi, at the Biology of Physical Activity Department of the University of Jyväskylä, in collaboration with researchers from the Laboratoire de Physiologie, GIP Exercise (Lyon, France), made important contributions in the study of tendons and ligaments biomechanics. They explored fiber optic sensors as an attempt to reduce the errors introduced by large conventional buckle transducers and minimize the subject's complaints [97]. In their first study, a needle was used to guide a 500-μm-diameter optical fiber into the rabbit common calcaneal tendon [97]. The fiber had a polymethol metachrylate core and a fluorinated polymer cladding. After removal of the needle, the tensile loads applied to the tendon were able to bend the optical fiber leading to changes in light intensity. The fiber was illuminated by an infrared LED, with the central wavelength at 820 nm, and the detector was an integrated circuit photodiode. To assess tendon forces, the system was calibrated using static equilibrium conditions. Hysteresis was negligible, and despite a slight time delay for the optical sensor response, a good agreement with a reference strain gauge transducer was obtained [97]. *In-vivo* studies followed their first *ex-vivo* experiment. The majority of them resulted from cooperation between the University of Jyväskylä and other institutions, such as the Institute for Biomechanics of the German Sport University of Cologne (Cologne, Germany) [98–100]. These studies reported the Achilles tendon force contribution during locomotion [101, 102]; individual muscle contributions to the Achilles tendon force [98]; leg muscles contributions to perform standardized jumps [103]; the muscle behavior during jump skills [99]; and the interaction between lower leg muscles and the Achilles tendon in walking [100].

Compared to buckle transducers, a clear advantage of these sensors is their minimally invasive impact since there are smaller and only require an anesthetic cream, applied to the skin surrounding the tendon, instead of anlar anesthesia [102]. However, the validity of previous studies has been questioned. Contradicting the original findings by Komi *et al*. [97], a nonlinear relationship was observed between the sensor output and the tendon force, requiring the use of third order polynomials for adequate fitting [104]. Hysteresis [104, 105], cable migration [104, 106], loading rate [105, 106], tendon creep [107], calibration procedures [108] and skin movement artifacts [106] were also pointed as possible sources of error in force prediction. To diminish these sources of error is a challenge because soft tissues are complex structures with nonlinear, visco or poroelastic properties requiring the most accurate sensors and techniques to obtain precise measurements.

The macrobending losses of an optical fiber have also been explored to monitor respiratory and cardiac functions, namely through the fiber optic respiratory plethysmography (FORP) technique. This non-invasive technique was firstly described by Augousti *et al*. in 1993 [109], at the School of Life Sciences of the Kingston University (United Kingdom) and was based on a notional geometrical model of the human respiratory system that consisted of two stacked connected cylinders, with the top cylinder representing the thorax and the lower abdomen [110]. It was presented as an alternative to the respiratory inductive plethysmograph technique, considered to be expensive and susceptible to electromagnetic interference [109, 111]. Authors also presented a improved version based on a novel figure-of-eight loop configuration, contributing for the increased linearity of response, less mechanical resistance and hysteresis [112]. The FORP technique was also explored by others [113–115]. Such an example was the FORP system presented by Davis *et al*. [113], from the Centre for Imaging and Advanced Optics of the School of Biophysical Sciences and Electrical Engineering (Swinburne University of Technology, Melbourne, Australia). The system was improved to monitor children high-frequency, low amplitude chest wall movements, claiming for no risk of electric noise and shock [114].

The possibility to apply optical fibers into textiles and create smart wearable clothes to monitor vital functions and motion was an important application in biomechanics.

Initial contributions were focused on gloves to assess the hand/fingers motion and interact with virtual environments. A large variety of sensors have been employed, including strain gauges, bend sensors, fiber optics, pneumatics, Hall effect sensors, among others [116–119]. Actually, initial gloves prototypes, such as the Sayre glove, developed in 1977 by Thomas de Fanti and Daniel Sandin (University of Illinois,Chicago, IL, USA), were based on light attenuation caused by bend of a flexible tube (not an optical fiber), with a light source at one end and a photocell at the other [116]. However, it was commercialization of DataGlove^{TM} that triggered research about these devices and spread their popularity worldwide [119]. DataGlove^{TM} was developed by Zimmerman *et al*. [120] at VPL Research (VPL Research, Inc.; acquired by SunMicrosystems which was acquired by Oracle Corporation, Redwood Shores, CA, USA), the first company to sell virtual reality gloves and the pioneer in 3D computer graphics. It consisted of a hand to machine the interface device providing real-time motion of the hand. A neoprene glove incorporating several sensors and technologies was used. The optical part of it consisted of patented optical goniometers intended to measure the fingers joints motion [121]. These devices were made of a flexible black rubber tube with a reflective inner wall coated with aluminum spray (a possible embodiment). An optical source and a photosensitive detector (it could be an optical fiber), were placed at the ends of the tube allowed for measurement of light attenuation concomitant with the bend of the tube. Wise *et al*. [122], from the Rehabilitation Research and Development Center of VA Medical Center (Palo Alto, CA, USA), evaluated DataGloveTM performance for clinical application. Inclusion of abduction/adduction measurement, as well as the wrist motion and full characterization of the thumb movement, has been recommended to be used as an effective clinical tool [122].

Fifth Dimension Technologies (5DT, Irvine, CA) developed another glove using optical fibers to sense the fingers motion. The sensor was modulated by the intensity to sense the fingers angular motion with a 5- (one joint per finger) or 14-sensor arrangement (two joints per finger) [123, 124]. It has been used as the primary manual input device for virtual reality technologies [123] and to improve hand function in adolescents with cerebral palsy, by means of the in-home gaming technology [124, 125]. It was, however, an expensive device (the current lower price for one glove is US$995) [126]. The price increased in versions incorporating further sensors or intended for specific applications, such as for the MRI environment. In fact, other affordable sensors were presented, such as the non-optical bend sensor (Flexpoint, South Draper, UT) developed by Simone *et al*. [117] for 24-hour daily life monitoring of finger motion. The sensor total cost was less than US$40 [117].

Miniature fiber optic goniometers seemed also interesting to study the association between highly repetitive movement and musculoskeletal disorder. The impact of typing was a possible application, which was studied by Nelson *et al*. [9] (General Motors, Orion Assembly Center, MI, USA), using opto-electric finger goniometers, developed in the Biodynamics Laboratory of the Ohio State University (Columbus, OH, USA). A similar study, using commercial sensors (Shape Sensors, Measurand Inc., New Brunswick, Canada), was performed by Jindrich *et al*. [10] at the Department of Environmental Health of the Harvard School of

Public Health (Boston, MA, USA).

The current research has been focused on the process of embedding fiber optic sensors into textile. In 2006, the OFSETH European project (Optical Fiber Sensors Embedded into technical Textile for Healthcare – OFSETH), a €3.5 million project made its contribution in the field [127, 128]. Several configurations, such as intensity-modulated, fiber Bragg gratings and optical time domain reflectometry (OTDR), were explored for healthcare applications. Project results were published for respiratory monitoring in the MRI environment [129–132] and pulse oximetry using near infrared spectrometry [133].

To conclude our approach to intensity-modulated sensors, a reference to pressure mapping devices seems mandatory. Several companies such as Tekscan Inc. (South Boston, MA, USA) and Novel GmbH (Munich, Germany) already offered powerful accurate electronic based systems at the relatively low cost for many biomechanical applications. Thus, optical fiber based systems should be capable of competing with these standard technologies. Meanwhile, some of their limitations were described. Tekscan sensors were based on conductive elastomers, which might exhibit nonlinear response, hysteresis, and gradual voltage drift [134]. The novel sensors used capacitive-based transducers, which could be affected by electrical interference and suffered from low spatial resolution, drift, and high sensitivity to temperature [134]. Moreover, with both technologies only normal loads and pressures could be measured. Until now, few alternative contributions have been presented in this field. The work of Wang *et al*. [135], from the Departments of Mechanical Engineering and Orthopaedics and Sports Medicine of the University of Washington (Seattle, WA, USA), is of particular interest because it represents the first contribution in the field using a bend loss technique. A 2×2 array of multimode fibers, embedded into the high-compliance material and forming four orthogonal intersection points, was developed to form the basic sensing sheet. Under compressive loading, light attenuation caused by physical deformation of the fibers at the intersection points allowed to calculate the (x, y) coordinates of the pressure point and the corresponding normal stress. To obtain shear stress, two layers of the basic sensing sheet, placed between gel/polymeric shoe insole pads, were used. In this way, the relative difference between the corresponding pressure points allowed calculation of the amount of shear. Repeatable results were obtained under bench mechanical loading tests. The minimum detectable vertical and shear forces were 0.4 N and 2.2 N (at 60º pitch angle), respectively. To address some limitations of the previous configuration (e.g., low spatial resolution, consistent and accurate manufacturing of the sensor, cost and noise), a batch process to fabricate Poly (dimethylsiloxane) (PDMS) as the optical medium, and a neural network technique to provide an accurate description of the force distribution were proposed [134, 136, 137]. After successful bench tests, the same group recently presented a full-scale foot pressure/shear sensor, capable of measuring normal forces ranging from 19.09 kPa to 1000 kPa [138].

6. Final remarks

Intensity modulated fiber optic sensors applied for biomechanics have been reviewed. Usually, they fall into one of two categories: a reflective membrane/mirror that changes its distance to the fiber tip; or an optical fiber that bends accordingly to the action of the measurand. While the first configuration has been used to sense pressure in many sites of the human body, the second seems to offer a wide range of applications from tendons force measurement to respiratory monitoring and goniometric applications.

Since the middle 1960s, the fiber optic technology has progressed at an astonishing rate, triggered by the increasingly demands of large capacity communication networks, turning out the optical fiber systems to be nowadays the backbone of the information society. This movement benefits the development of other applications of the optical fiber, most notably in the sensing domain where the intrinsic characteristics of the fiber remarkably match the ideal requirements of a sensor system. Indeed, besides the fiber being the sensing element and the communication (telemetry) channel, it brings the optical field (optical power) to the measurement region, eliminating the need of additional wiring for power delivery to the sensor, a need in many sensing approaches as is the case of electrical sensing in most of its applications. Therefore, the fast development of this sensing technology and its utilization in a diversity of areas are not surprising, as is the case of biomechanics. Here, what has been achieved so far, reviewed in this paper when considering measurand induced optical power modulation, is a clear indication of how valuable the fiber sensing approach is when addressing this field and provides an insight into what can be achieved in the future.

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