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Fiber-Optic Sensors for High-Precision Temperature and Strain Measurements

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Description

Due to their inherent properties compact size, quick response time, low cost, and insensitivity to electromagnetic fields optical fiber sensors have been a promising sensing device for decades. Based on the development of optical fiber fabrication techniques, such as fiber Bragg gratings, long-period gratings, Raman scattering, and Brillouin scattering, numerous configurations have been successfully reported. Typically, sophisticated devices are required for the fabrication of these configurations, which comes at a relatively high cost. A simpler structure based on multimode interference was the first structure to address these factors. The multimode interference effect, which is present in multimode waveguides, has been thoroughly examined. A so-called single-mode–multimode–single-mode fiber sensor, which consists of a brief segment of multimode fiber sandwiched between two single-mode fibers, is the fundamental optical fiber structure used to construct a multimode interference device [1].

Single-mode-multimode-single-mode fiber structure has gotten critical consideration somewhat recently due to its interesting unearthly elements, high responsiveness, simple manufacture, and likely low expenses. It also demonstrates excellent compatibility with other photonic structures and devices. To improve sensor performance, such as multi-parameter measurement or higher sensitivity, it can, for instance, connect to other fiber structures with ease and great adaptability. As a result, it can be used as a variety of sensors in a variety of situations, including biomedical sensors, temperature sensors, curvature sensors, humidity sensors, strain sensors, and position sensors.

Image resolution and imaging depth are two of the most crucial parameters for evaluating imaging performance. The frequency or wavelength of the sound waves used in ultrasound imaging directly influences the resolution. For normal clinical ultrasound frameworks, sound wave frequencies are in the ten megahertz system and yield spatial goals as fine as 150 µm. The advantage of ultrasound imaging is that sound waves of this frequency are easily transmitted into the majority of biological tissues. As a result, it is possible to obtain images of structures within the body that are several tens of centimeters deep. The sound recurrence is a significant boundary in ultrasound imaging since enhancing picture goal for a given application while compromising picture entrance depths is conceivable. High recurrence ultrasound has been created and explored broadly in research facility applications as well as a few clinical applications. Frequencies of and higher have resulted in resolutions [2].

However, biological tissues significantly reduce the attenuation of high-frequency ultrasound, and this attenuation increases roughly in proportion to the frequency. As a result, high frequency ultrasound imaging can only go as deep as a few millimeters. It is also important to keep in mind that the ability to focus sound waves determines the transverse resolution of ultrasound. Since sound is generally more difficult to focus than light, transverse resolutions for ultrasound are lower than for. The resolution of current imaging technologies ranges. OCT's inherent high resolution makes it possible to image aspects of the cellular and architectural morphology of tissues. The majority of biological tissues significantly scatter light, which is the primary drawback of optical imaging. Except for the eye, optical scattering restricts image penetration depths in most tissues [3].

Last but not least, it is helpful to point out that the contrast mechanisms of OCT, ultrasound, and microscopy differ. A mismatch in the acoustic impedance of ultrasound scattering between various tissues results in ultrasound images. This produces distinctions in the power of reflected or backscattered sound waves. OCT imaging employs light and is sensitive to variations in optical scattering indexes between tissues. Finally, differences in optical reflection or transmission through thin sections produce images in microscopy. Stains can be used to selectively enhance contrast between various structures in histopathology. As a result, studies are required to establish a foundation for interpreting OCT images in terms of clinically relevant pathology because the appearance of OCT images is generally distinct from that of either ultrasound or histopathology.

Electronics cannot directly measure the echo time delay of light because of its extremely high velocity, as is the case with ultrasound. Sound travels at about the same speed through water as light travels at about 3 x 108 m/sec. The formula where is the echo delay, z is the distance that the echo travels, and v is the velocity of the sound wave or light wave can be used to calculate distance or spatial information from the time delay of reflected echoes. A resolution on the scale of approximately corresponds to the measurement of distances or dimensions, which is typical for ultrasound. The reverberation time delays related with light are very fast. For instance, the typical OCT measurement of a structure at a resolution of 10 microns corresponds to a time resolution of approximately. On this time scale, electronic direct detection is impossible. Correlation or interferometry techniques must be used instead. Low-coherence interferometry is one way to measure the echo time delay of light. Fiber optics and optoelectronic devices were the first applications of low-coherence interferometry. Low-coherence interferometry was first used in biomedicine in ophthalmology to precisely measure axial eye length and corneal thickness [4].

Backscattered light's echo time delay and intensity are measured using low-coherence interferometry by comparing it to light that has traveled a known reference path length and time delay. A Michelson-type interferometer is used for the measurements. Light from a source is directed onto a beam splitter. One of the beams hits the sample that needs to be imaged, and the second beam follows a reference path that has varying lengths and timestamps. A photodetector located at the interferometer's output detects the interference between the sample's backscattered light and the reference arm's reflected light. As the relative path lengths change, interference fringes will be observed if the light source is coherent. However, if short pulse or low-coherence light is used, interference between the sample's reflected light and the reference path can only occur if the two paths are the same length within the light's coherence length. By detecting and demodulating the interference output of the interferometer while scanning the reference path length, one can measure the echo time delay as well as the intensity of backscattered light from sites within the sample.

One class of utilizations where OCT could be particularly strong is where customary excisional biopsy is risky or unthinkable. OCT, for instance, is able to provide high-resolution images of pathology that cannot be obtained using any other method. In ophthalmology, for instance, retina biopsy is not an option. Imaging of the morphology of atherosclerotic plaques in the coronary arteries is another scenario in which biopsy is not an option. The majority of myocardial infarctions, according to research, are the result of cholesterol-laden coronary artery plaque rupture, followed by thrombosis and vessel occlusion. Plaques with a fibrous cap that is structurally weak pose the greatest risk of rupture. The microstructural characteristics of these plaque morphologies cannot be determined, and conventional radiologic methods make it difficult to identify

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them. Due to the high proportion of occlusions that result in sudden death, it is important to identify patients at risk for myocardial infarction and high-risk unstable plaques. OCT may be useful for diagnostic intravascular imaging, helping to guide interventional procedures like atherectomy and stratifying risk [5].

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Conflict of Interest

None.

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