Chirped fiber Bragg grating sensor for pressure and position sensing

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Abstract. We present a chirped fiber Bragg grating sensor that should be suitable for esophageal motility studies. The device uses the time-dependent group delay response of a chirped fiber Bragg grating to measure the peristaltic pressure wave that propagates down the esophagus with the transport of a bolus to the stomach. In contrast to existing transducers that only measure at discrete points, the output of this device is a continuous function of length along the esophagus. This paper presents *ex-vivo* experimental results. There is a linear relation between the wavelength location of the maximum phase perturbation and the position along the sensor where the perturbation occurred. The maximum phase change itself is directly proportional to the magnitude of the applied load at a specific position.
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Subject terms: chirped Bragg grating; pressure sensor; position sensor; esophagus; motility studies.

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1 Introduction

Disorders of the esophagus, a 25- to 30-cm-long hollow muscular tube that transports food from the mouth to the stomach, are very common.[1] Gastro-esophageal reflux is one of the most prevalent symptoms. Although less prevalent, esophageal motility disorders such as diffuse spasm of the esophagus and achalasia may cause unpleasant symptoms such as noncardiac chest pain and dysphagia. They can only be diagnosed accurately and categorized by esophageal function studies. In these cases esophageal manometry and pH monitoring are indicated. Esophageal manometry measures the function of the body muscle of the esophagus and the associated sphincters, ring-like muscular valves at the top and bottom ends of the esophagus, by recording pressure profiles along the length of the esophagus.[2] A strong contraction of the previously relaxed upper sphincter, which follows the entry of a bolus into the esophagus, travels at a velocity of between 3 and 4 cm/s towards and involving the lower sphincter. The pressure generated by this wave is between 5.3 and 13.3 kPa with fundamental frequencies ranging from dc to about 4 Hz. Up to eight harmonics need to be recorded for a variety of pressure waves. The maximum rate of change of the pressure wave lies between 40 kPa/s for the mid-esophagus and about 500 kPa/s for the pharynx.[3] Currently, manometry gives the most accurate measurements of the strength of the muscular contractions as well as the timing and coordination of the amplitudes and durations of contractions along the esophagus. It also gives accurate readings of the normal ranges for the resting body, percentage relaxation and the timing of the relaxation of the two sphincters, as well as the position of the lower sphincter.[4]

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Although primitive manometric studies were already possible during the late 1800s, it was not until the late 1960s that accurate measurements were possible with improved pressure recording techniques and the introduction of a low-compliance perfusion system.[5] After further development and better understanding of the esophageal physiology and disease, manometry and pH monitoring became the tools for stationary and ambulatory clinical evaluation of patients exhibiting symptoms that can be attributed to motility disorders. Presently systems of water-perfused catheters or solid-state transducers are commonly used. Solid-state transducers are more convenient to use in ambulatory studies. Schneider et al.[6] introduced a fiber optic Fabry-Perot sensor that records local asymmetric pressure with constant sensitivity over the sensor surface area at discrete points along the upper gastrointestinal tract and that does not require water as a transmitter medium.

Fiber Bragg grating sensors have many desirable properties such as high sensitivity, complete absence of electrical shock hazard, and immunity to electromagnetic interference.[7,8] We investigated experimentally a single slender catheter containing a lightwave-based pressure sensor that was proposed by us previously,[9] for application in esophageal manometry. The transducer comprises a chirped fiber Bragg grating with a polymer coating to convert hydrostatic pressure to longitudinal strain. The output is a continuous function of length along the esophagus. The proposed manometer has the potential to be a lightweight instrument for ambulatory use as well.

2 Principle of Operation

We use a continuous wave modulation technique to measure the group delay characteristic of a chirped fiber Bragg grating.[10] A tunable laser that is intensity modulated by a

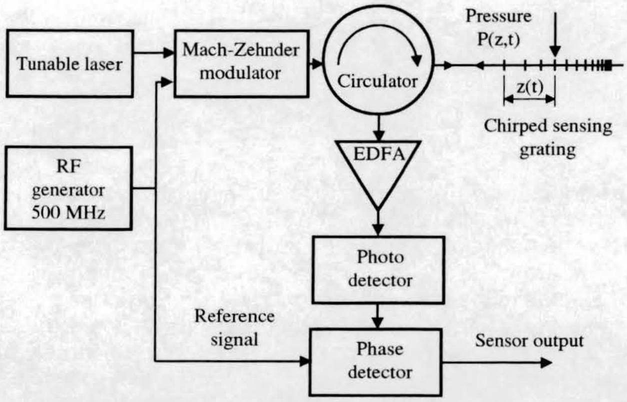

Fig. 1 Chirped fiber Bragg grating pressure sensor.

Mach-Zehnder modulator illuminates the grating. Upon reflection from the sensing grating, a circulator routes the light to the photodetector, followed by a phase detector. Localized pressure that is applied to the grating causes a perturbation to its phase delay characteristic. The phase detector produces a time-dependent signal representing the phase characteristic of the perturbed grating. Details in the phase characteristic allow us to determine the position and the magnitude of the applied pressure wave propagating along the grating. Figure 1 shows the experimental setup.

The intensity of the light at the output of the Mach-Zehnder modulator can be presented as:

$$I=I_0+I_1\sin(\Omega_0 t), \tag{1}$$

where I_0 is the mean output intensity of the laser, I_1 is the amplitude of the intensity modulation, and Ω_0 is the frequency of modulation in rad/s. The signal at the output of the photodetector is:

$$V_{PH}=K\sin\left(\Omega_0 t+2\pi\frac{2L_1(t)}{\Lambda}+\phi_f+\phi_e\right), \tag{2}$$

where K represents the conversion coefficient of the photodetector and amplifier, L_1 is the fiber length from the circulator to the point in the chirped Bragg grating from where the light is reflected, ϕ_f represents all fixed delays associated with propagation of light in the fiber, and ϕ_e represents all electrical delays. The wavelength Λ of the modulation signal in the fiber is given by:

$$\Lambda=\frac{2\pi c}{\Omega_0 n}, \tag{3}$$

where c is the velocity of light in vacuum, and n is the effective refractive index of the fiber core. For a perfect linearly chirped grating with no external perturbation, and with a linear sweep of the laser wavelength, L_1 is a linear function of time:

$$L_1(t)=\gamma t+L_0, \tag{4}$$

where γ is the slope and L_0 is a constant. If the grating is perturbed, γ becomes time-dependent. Thus, by measuring the phase difference between the reference (modulation) signal and the signal at the output of the photodetector for a complete wavelength scan, we can derive the position within the chirped grating where it was perturbed. The corresponding magnitude of the perturbation is obtained from the value of the local change in the phase slope γ.

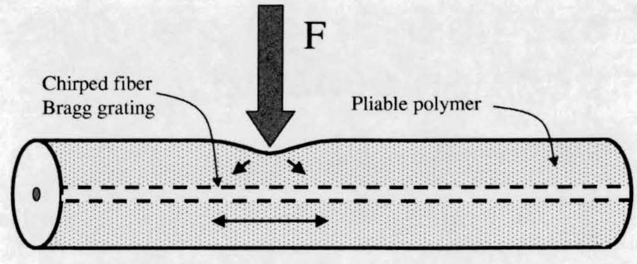

Fig. 2 Schematic representation of polymer-coated chirped fiber Bragg grating.

The pressure sensitivity of the resonance wavelength of the fiber Bragg grating is given by[11]:

$$\frac{\Delta\lambda_B}{\Delta P}=\lambda_B\left[-\frac{(1-2\nu)}{E}+\frac{n^2}{2E}(1-2\nu)(2\rho_{12}+\rho_{11})\right], \tag{5}$$

where E is the fiber Young's modulus, ν is its Poisson's ratio, and ρ_{12} and ρ_{11} are components of the strain-optic tensor. Xu et al. found $\Delta\lambda_B/\Delta P$ experimentally to be 3 pm/MPa for λ_B nominally equal to 1533 nm.[11] This sensitivity is much too low for the intended application. However, by coating the optical fiber with an appropriate polymer, transverse pressure is converted to axial strain, thereby increasing the pressure sensitivity of the Bragg grating significantly. This is depicted schematically in Fig. 2.

3 Experiment

To evaluate the proposed sensor, we manufactured a chirped fiber Bragg grating by exposing photosensitive single mode fiber with 248-nm pulsed UV light through a linearly chirped phase mask. The laser fluence was tapered off towards the ends of the 120-mm-long phase mask to reduce phase delay ripple. This yielded a useful grating length of 100 mm. To convert lateral pressure into axial strain, the fiber containing the grating was encapsulated with a polymer jacket. A special applicator was developed to force the polymer under pressure through a die while simultaneously feeding the optical fiber through the center of the die. The encapsulated chirped grating was incorporated in the system shown schematically in Fig. 1. We measured the phase shift versus wavelength characteristic for the grating by scanning the wavelength of the tunable laser and by recording the corresponding phase changes. There is a tradeoff between system sensitivity and system linearity because of the range limitation of the phase detector. To ensure that the phase excursion did not exceed 360 deg, we used a modulation frequency of 500 MHz. The slope of the phase characteristic was 2.82 rad/nm over a 1.1-nm bandwidth. Because the length of the grating is known, the wavelength-to-distance conversion factor of the experimental sensor should therefore be equal to 0.011 nm/mm. The time response of our phase detector limited the scan rate to 5 Hz. Although this is too low for dynamic measurements

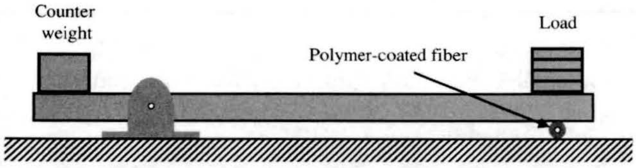

Fig. 3 Schematic representation of the load applicator.

in the esophagus where one would require at least 40 Hz, it is adequate for the static measurements during evaluation of the measurement concept.

To determine its wavelength versus position characteristic, we loaded the transducer sequentially at 10-mm intervals along its length with the load applicator shown in Fig. 3. The load was kept constant by using a fixed weight of 55.2 g at each point. Figure 4 shows the measured phase change of the perturbed grating with respect to the unperturbed grating, and Fig. 5 depicts the position of the peak phase change as a function of the loading position on the grating. The slope of the solid line representing a linear least-squares fit to the data is 0.0116 nm/mm.

In the next experiment, various loads (weights) were applied to the grating at one position by use of the load applicator. The corresponding phase delays were measured. Figure 6 depicts the measured phase change as a function of wavelength. As can be seen in Fig. 7, the peak phase change is sensitive to the load. The solid line represents the least-squares fit of a second-order polynomial to the data. The error bars represent the standard deviation obtained from the data in Fig. 4.

4 Discussion and Conclusions

We demonstrated experimentally the feasibility of using an encapsulated chirped fiber Bragg grating as a position-dependent pressure sensor that is suitable for use as an esophageal pressure sensor. A special loading device allowed us to apply a transverse force at various positions along the length of the transducer. The experimental device

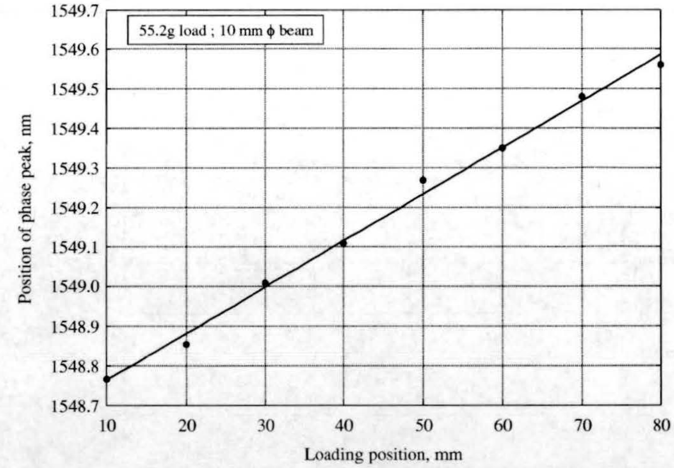

Fig. 5 Wavelength of peak phase change as a function of load position.

comprised of a 100-mm-long chirped fiber Bragg grating with a bandwidth of 1.1 nm. As is to be expected, the wavelength of the phase disturbance peak is a linear function of the position of the point of application of the pressure. The calculated slope from the linear least squares fit to the data is equal to 0.0116 nm/mm. It agrees well with the expected value of 0.011 nm/mm obtained from the measured phase characteristic of the grating. The results show that the measured phase change is approximately a quadratic function of the load. This is probably fortuitous, because the load applicator was not designed to apply a uniform lateral pressure to the transducer. We expect the relationship between the unilateral load and the strain in the fiber to be highly nonlinear for the cylindrical load applicator. Nevertheless, the standard deviation in the measurements of 0.02 rad is small enough (6.2% of full scale) to warrant further investigation and improvement of the transducer and the load applicator. There also seems to be a slight load-dependence of the position of the peak phase excursion. From Fig. 6 the peak phase change shifts by

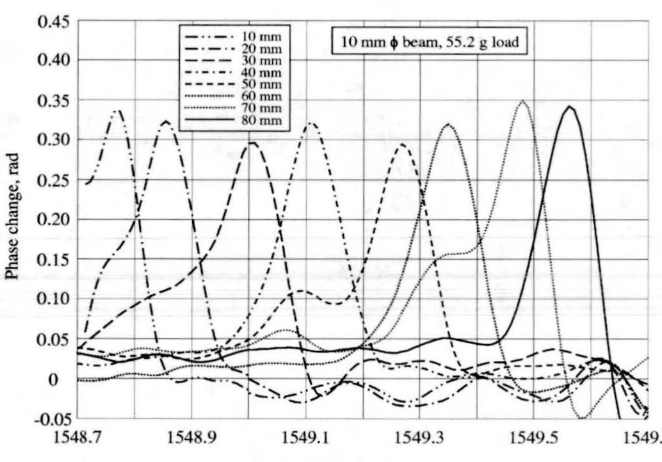

Fig. 4 Phase change versus wavelength with the loading position as parameter.

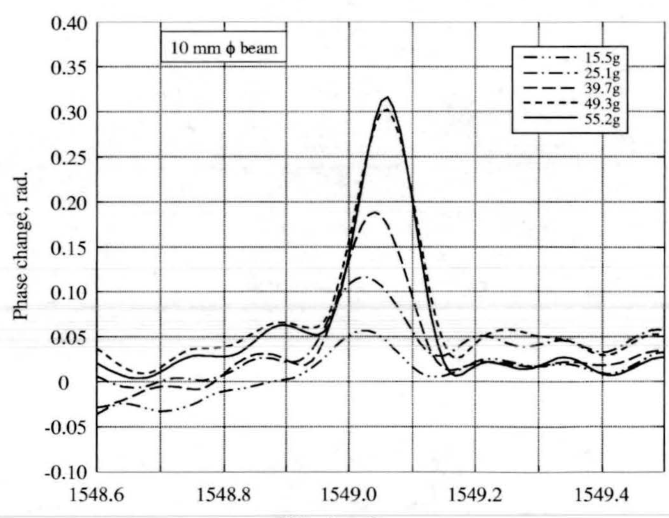

Fig. 6 Effect of strain magnitude on phase disturbance.

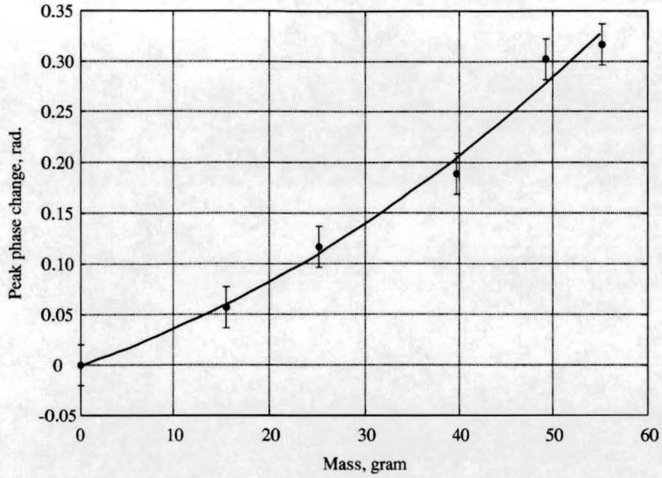

Fig. 7 Peak phase change as a function of load.

0.03 nm for the load increasing from 15.5 g to 55.2 g, implying a position uncertainty of ±1.5 mm due to this effect.

This work summarized proof-of-concept research for the development of a fiber optic esophageal pressure sensor. It encodes the position of lateral pressure applied at a point along the length of the transducer as a peak in a phase change in the wavelength domain, with the magnitude of the phase change proportional to the pressure. The next phase of the project involves the development of a 25-cm-long probe that will be suitable for *in-vivo* experiments.

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