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To cite this article: N Raicevic *et al* 2016 *J. Phys.: Conf. Ser.* **682** 012008

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Monitoring of respiratory volumes by an long period grating sensor of bending

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Abstract. Here, we present a method of respiratory volumes monitoring using a single fiber-grating sensor of bending. Measurements are conducted using simple monochromatic interrogation scheme that relies on a photodiode measurement of the power transmitted through a long period grating (LPG) sensor at fixed wavelength. Good sensor accuracy in measurements of tidal and minute respiratory volumes for different types of breathing is achieved.

1. Introduction

Long-period fibre gratings have been used as bend-rejection filters, gain-equalizers, and sensors [1, 2]. Their sensing properties stem from sensitivity of the fibre cladding modes to changes in environmental parameters such as temperature and the refractive index of the surrounding medium, and on forces applied to the fibre (pressure, strain, torque, bending). The sensing applications of LPGs are in biomedicine [3, 4, 5, 6, 7], medicine [8, 9, 10], environmental studies [11], radiation dosimetry [12], etc. LPGs can be sensitive to a number of parameters at the same time, which is known as cross-sensitivity and is detrimental in all applications except in multi-parameter sensing schemes [13].

Here we investigate application of LPG sensors of bending in volumetric measurement of air inhaled and exhaled during breathing. The idea to use the sensor of bending is motivated by the discovery of Konno and Mead [14] that the volume of the inhaled air is correlated to the thoracic and abdominal movements caused by breathing. Further incentives for the use of LPGs for this purpose are the non-invasiveness of the measurement method and immunity to electrical noises. As the end application we target the non-invasive ventilation (NIV) during which a mechanical ventilator provides respiratory support to a patient. The current clinical standard relies on air-flow sensors interfaced with the patient though an oronasal mask or a mouthpiece. Such an interface promotes air leaks that introduce errors in the measurement of respiratory volumes, making NIV success dependent on the capability of the ventilator to deal with air leaks or on clinician's observation of the patient's chest wall movement [15, 16]. Here we review our investigations of the LPG bending measurement scheme as one possible solution fully independent of the mask.

Content of this paper is as follows: in Section 2 we describe the working principle of LPG sensor and support this by the results of numerical modelling. We further present the full experimental characterization of the grating and describe a simple monochromatic scheme that was used for

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respiratory monitoring. In Section 3 we review the results of volumetric measurements of breathing performed on a set of 15 healthy volunteers. Conclusions are given in Section 4.

2. Long period grating sensors of bending

2.1. Principles of operation

The LPG is one-dimensional periodic perturbation of the fibre refractive index profile or geometry along the fibre axis. Period of LPG varies from several tens to several hundreds of micrometers which, for NIR light, can induce out coupling of fundamental core mode and formation of resonant co-propagating cladding modes [1]. The formed cladding modes represent loss of light from the core, and can be recognized as attenuation bands (AB) in the transmission spectrum. Attenuation bands occur at discrete, resonant wavelengths (λ) that satisfy phase-matching condition between core (n_{co}^{eff}) and v -th cladding mode (n_{clad}^{eff}):

$$\lambda_v = (n_{co}^{eff} - n_{clad}^{eff})\Lambda \quad (1)$$

Here Λ represents the grating period. In order to calculate the mode distribution of core and cladding modes fulfilling Eq. (1), finite element method [17] has been performed due to its flexibility in meshing different geometries and shapes. Theoretically, mode analysis was done for LPGs inscribed in the core of standard SMF-28 with the following parameters: $n_{co}=1.44922$, $n_{cl}=1.444024$, $r_{co}=4.1\mu\text{m}$ (core radius), and $r_{cl}=62.5\mu\text{m}$ (cladding radius). The refractive index modulation representing the grating is set to $\Delta n_{co}=5.75\cdot 10^{-4}$. Intensity distributions of core mode and four lowest cladding modes formed in single mode fibre are presented in figure 1a).

The transmission spectrum of LPG, with ABs marked, is given in figure 1b). Beside resonant wavelength, AB is characterized with magnitude and width. Magnitude of each AB is influenced by overlap of core and cladding mode as well as by LPG length. Width of AB is related to fibre dispersion and the grating dispersion.

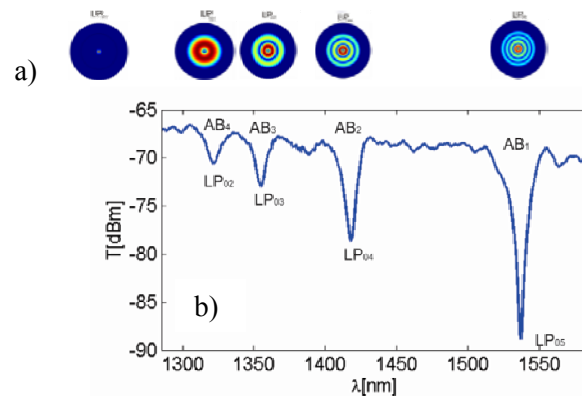


Figure 2. a) Intensity distribution of core mode (LP₀₁) and cladding modes LP₀₂, LP₀₃, LP₀₄, LP₀₅ and LP₀₆; b) Transmission spectrum detected using Optical Spectral Analyzer and broad band optical source. Four ABs are occurring.

To model the bending of the grating, conformal transformation has been used. With this procedure it is possible to model curved fibre with equivalent straight one whose refractive index is modified [18] by the expression:

$$n(x) = n_{core} \left(1 - \frac{x}{R} \right) \quad (3)$$

where n_{core} is the core refractive index, R is the curvature radius and x is the distance from fibre center. Strictly speaking, the dependence on x is exponential, but since $x/R \ll 1$, it can be approximated by the linear term. The change in refractive index profile comes from the strain-optic effect and results in cladding mode redistribution. In figure 2 we presented intensity distribution of modes LP_{05} for curvatures $K = 0 \text{ m}^{-1}$, $K = 0.5 \text{ m}^{-1}$ and $K = 1 \text{ m}^{-1}$ obtained for a single mode fibre with LPG of period $\Lambda=340\mu\text{m}$ at appropriate wavelengths. Mode redistribution results in changes in the resonant wavelength, magnitude and shape of AB, which all can be detected by the spectrum analyzer. In the case of LP_{05} mode, bending induces reduction in the core-cladding mode overlap and hence a reduction in the magnitude of the attenuation band, while the asymmetrisation of the grating refractive index profile changes its shape.

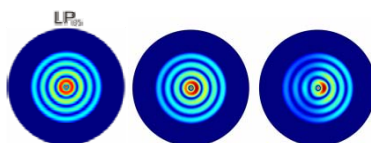


Figure 2. Intensity distribution of mode LP_{05} before and after LPG bending.

2.2. Sensor characterization

Grating response to bending is analyzed utilizing gauge specially designed for this purpose, figure 3a). Measurement of the transmission spectra are conducted using and broadband optical source (Spectral Products ASB-W-005) and Optical Spectral Analyzer (OSA) (YOKOGAWA AQ6370C). Response to bending of an LPG sensor is presented in Fig 3 b). It can be seen that transmission spectrum of this sensor exhibits a red shift with increase of the curvature. The working range in which the sensor has linear response is marked by gray in figure 4 a).

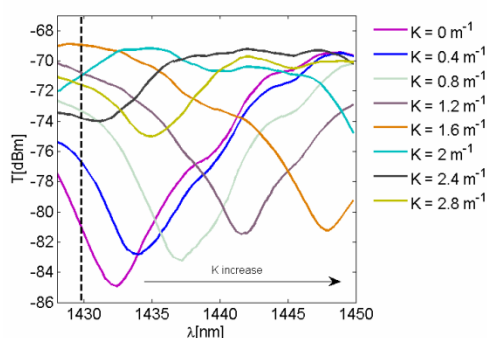
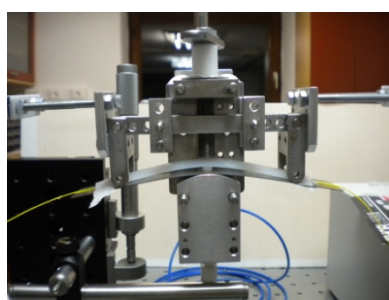


Figure 3. (a) Gauge for characterization of the fibre sensors of bending. (b) Spectral response of the LPG to bending.

Changes in the transmitted power at a fixed wavelength (dashed line in figure 3b)) are also presented as a function of the grating curvature, figure 4 b) (K-T plot). It can be seen from figure 3 b) that the slope and the width of the linear grating response depend on the choice of the laser wavelength.

The above results indicate that in curvature recognition, both the resonant wavelength and transmitted power can be used as measurement observables. As the high cost of the OSA limits its widespread use in medical applications, we have investigated a simple monochromatic interrogation

scheme that relies on a photodiode measurement of the power transmitted through an LPG sensor at fixed wavelength. For this purpose a distributed feedback laser with a central wavelength 1470.73 nm is used, with narrow tunable range of 5nm, which is controlled via temperature or current.

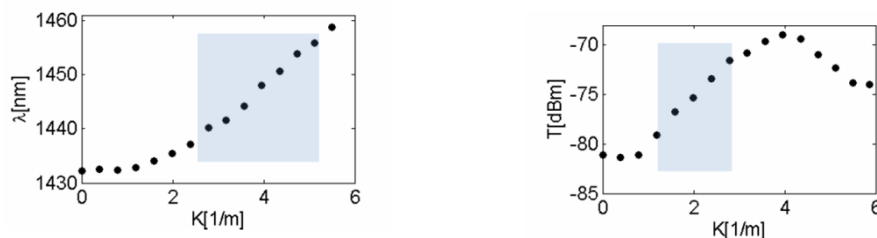


Figure 4. (a) K- λ dependence; (b) K-T dependence; Shaded area represents linear response of LPG to bending.

3. Application in respiratory monitoring

We have applied LPGs to continuous monitoring of the respiratory volumes based on the measurement of chest wall movements. The proposed solution uses LPG curvature sensor attached to the patient's thorax and a simple and cost-effective monochromatic interrogation scheme described above [19, 20].

The measurement protocol consists of two steps, calibration and test measurements. As a reference instrument we used a spirometer (SpiroTube, Thor Medical, Budapest). After eliminating the drift by zeroing the end of each breath, the calibration function was determined by different order polynomial fitting of a sensor to a simultaneously recorded spirometer signal. Finally, the linear scaling was chosen as optimal regarding robustness and accuracy. The calibration function was determined for each subject and used for calculation of the respiratory volumes in all test measurements performed on the same subject.

A number of measurements on 15 healthy volunteers, aged 32.6 ± 8.0 years, were used as a statistical set for the data analysis. During all measurements the subjects were in supine position equivalent to the position of the patients during mechanical ventilation. LPG sensor was attached to a surface of the thorax in an area with stiff underlying tissues and fixed by an elastic bandage that goes around subject's torso and over the sensor. Since LPGs can sense the heart beats, sensor was positioned at the right side of the thorax.

4. Results and discussion

A typical set of the calibration and test results is shown in figure 5. Scatter plots a) and b) show linear correlation between the change of the local torso curvature and change in lung volume measured by spirometer. The calibration curve is represented by an inner solid line. The outer lines designate the Δ region that contains 68% of all measured values, and corresponds to standard deviation. For the calibration measurements, subjects were asked to breathe naturally with typical tidal volumes of 400-800 ml, and for the tests to perform one minute of each natural, shallow (approximately half the tidal volume of natural breathing) and a combination of deep, natural and shallow breathing (hereinafter called mix type of breathing). The calibration uncertainty Δ_{calib} of the whole group of $(6.8 \pm 2.2)\%$ and test uncertainty Δ_{test} of $(7.9 \pm 2.4)\%$, $(11.4 \pm 2.9)\%$ and $(10.2 \pm 3.8)\%$ for natural, shallow and mix breathing, respectively, validates the correlation fit.

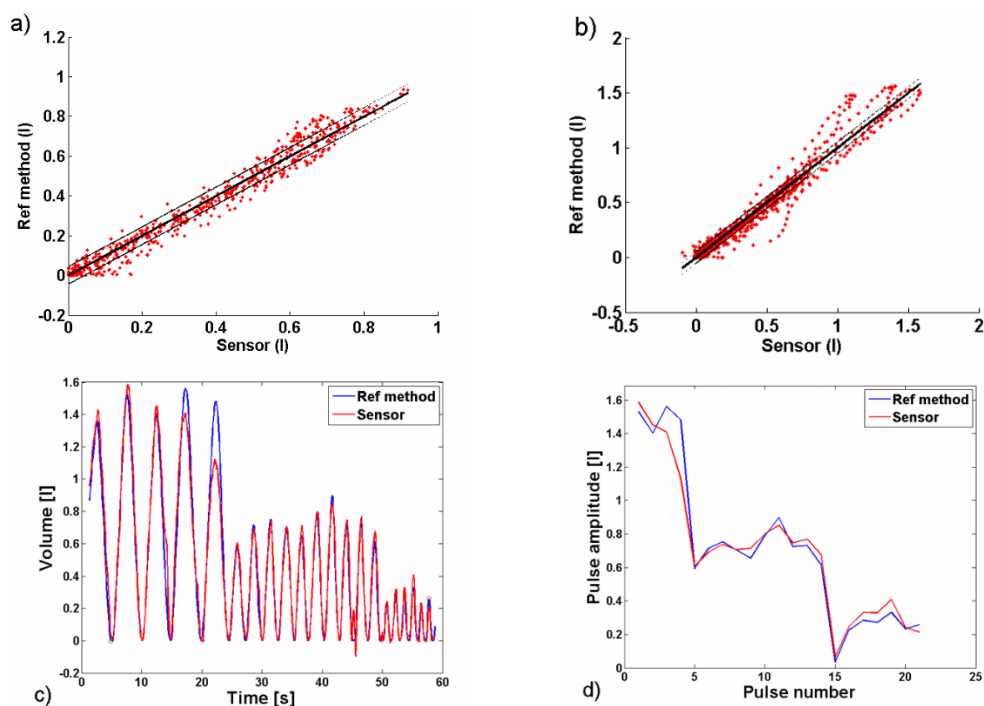


Figure 5. a), b) Scatter plot (dots) and the calibration function (inner line) and Δ region (outer lines) in calibration and test step, respectively; c) calibrated sensor output; d) tidal volume extracted from c).

Tidal volumes shown in figure 5 d) were obtained as a difference between maximum (end-inspiratory) and minimum (end-expiratory) of the same breathing cycle. Sum of the tidal volumes over one minute yields the minute volume and represents a typical average measure used by doctors. We calculated the measurement error of the LPG sensor with respect to the spirometer in the following way: the mean tidal volume error was defined as the mean relative error of tidal volumes over a number of respiratory cycles, and the minute volume error as the relative error of the minute volumes. These errors were used as the performance metric [19]. Results obtained for the whole group show the tidal volume error of $(10.8 \pm 3.8)\%$ with the minimum error of 0.4% and the maximum error of 16.6%, and the minute volume error of $(8.7 \pm 4.6)\%$ with the minimum error of 0.2% and the maximum error of 18.2%, all calculated for normal breathing.

When compared to other measurement methods based on the discovery of Konno and Mead, the proposed method of respiratory volume monitoring using a single LPG curvature sensor is simpler, easier to implement and less susceptible to signal distortion. Additionally, it does not require subject cooperation for calibration, does not suffer from large baseline drift, and exhibits good calibration and/or measurement accuracy for clinical use [19]. Its main disadvantage is susceptibility to the body movements not originating from respiration, which can be largely corrected by the advanced signal postprocessing and better sensor fixation to the body.

4. Conclusion

We have presented the application of an LPG sensor of bending in monitoring of respiratory volumes. The method is based on the observation of a linear correlation between the changes in the local torso curvature and the lung volume. A good agreement of tidal and minute volumes measured simultaneously by the LPG sensor and spirometer, proves the sensor accuracy and consistency in time. The measurements were performed by a single LPG sensor, analyzed here in detail, and a monochromatic measurement scheme that requires only one narrowband laser diode and a photodiode.

Therefore, besides the mandatory satisfactory accuracy, the proposed scheme is characterized by simplicity and a potential low-cost, which indicates its suitability for application in non-invasive mechanical ventilation.

Acknowledgment

The authors acknowledge financial support from the Ministry of Education, Science and Technological Development of Serbia (Project No. III45010) and thank Lj. Hadžievski, B. Bojovic, A. Danicic, M. Vukcevic, T. Allsop and D. J. Webb for collaboration and their contribution to this work.

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