A miniature optical breathing sensor

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Abstract: We demonstrate a novel miniature optical breathing sensor based on an Agarose infiltrated photonic crystal fiber interferometer. The sensor detects the variation in relative humidity that occurs between inhaled and exhaled breath. The sensor interrogation system can determine the breathing pattern in real time and can also predict the breathing rate and the breathing status during respiration. The sensor is suitable for monitoring patients during a magnetic resonance imaging scan where use of sedatives and anesthetics necessitates breathing monitoring; electronic sensors are not suitable in such an environment and a visual observation of the patient's respiratory efforts is often difficult.

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1. Introduction

Breathing is a human vital sign; it is the process that moves varying volumes of air into and out of the lungs. This is required to provide an adequate oxygen (O$_2$) supply to meet the energy production requirements of the body and maintain a suitable acid-base status by removing carbon dioxide (CO$_2$) from the body. The act of normal breathing has a relatively constant rate. The rate is noted by observing the frequency of breathing. The number of breaths per minute is called the breath rate.

Breathing monitoring is one of the most important elements of assessing physiological state. It can provide valuable information related to cardiac, neurological and pulmonary conditions [1]. Breath rate monitoring is important during certain imaging and surgical procedures where the patient needs to be sedated or anesthetized [2,3] because respiratory failure is difficult to predict and in just a few minutes life-threatening conditions can arise due to its failure. On the other hand, some illnesses such as apnea, tachypnea, hyperpnea, hypopnea and Cheyne-Stokes respiration syndromes can be diagnosed by detecting alterations in breathing patterns such as delayed breathing, voluntary breath holding, shortness of breath, hyperventilation, or abnormal respiratory rate etc. [4,5]. Continuous monitoring of respiratory activity is also of great importance in the case of infants susceptible to sudden infant death syndrome.

Breathing can be monitored utilizing nose exhaled air (using a humidity sensor, temperature sensor, capnometer or spirometer) or by monitoring the movement across the thoracic cavity/chest from the rise and fall of the abdomen, or with a plethysmograph. The most popular commercially available sensors are electronic sensors [6]. Electronic breathing sensors are not suitable when patients are, for example, in a magnetic resonance imaging (MRI) system, or during any oncological treatment that requires the administration of radiation or high electric/magnetic fields. MRI is a powerful, non-invasive way of obtaining detailed internal images of the human body which probes the inside of the body with strong magnetic fields. Sensors including metallic parts or electrical conductive wire are inappropriate during MRI, since they disturb the imaging process and also can cause burns on the patient’s skin [7–9]. Respiratory rate and apnea may be monitored during MRI procedures using an end-tidal carbon dioxide monitor or a capnometer. These devices measure the level of carbon dioxide during the end of the respiratory cycle (i.e., end-tidal carbon dioxide). The interface between the patient for the end-tidal carbon dioxide monitor and capnometer is a nasal or oro-nasal cannula that is made from plastic. This interface prevents potential adverse interactions between the monitor and the patient during an MRI procedure but some delay in measurement exists because the air has to reach the sensor from the subjects nose through the cannula [3,7]. In addition, the patients cannot be monitored during the transport in or out of the MRI room since such MRI compatible monitoring systems are not easily transportable and often exclusively used for MRI examination due to their cost.

Fiber optic sensors are advantageous because of their electromagnetic immunity and because they can be interrogated remotely as demonstrated in many application areas. Breath

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monitoring using optical fiber sensors can be undertaken by detecting the contraction and expansion of the patient’s chest and abdomen that occur during breathing [10–15]. This can be done by means of highly-sensitive strain, bending or pressure sensors set in a strap, belt or patch attached to the patient’s body [10–15]. However this technique is only capable of recording body movements associated with respiratory effort. Therefore, such optical fiber based respiratory rate monitoring techniques do not detect apneic episodes related to upper airway obstruction (i.e., absent airflow despite respiratory effort) and may not provide sufficient sensitivity for assessing patients during MRI procedures. The cross sensitivity to extraneous body movements will also affect the performance of these kind of sensors making them less suitable for accurate breath monitoring. Breathing can also be monitored using a fiber optic air flow sensor placed close to the patient’s nose or mouth [16] but it will show high cross sensitivity to vibrations associated with body movements.

Another approach to breath rate monitoring is using fiber-based humidity or temperature sensors placed close to the patient’s nose or mouth since the air exhaled has higher humidity and is warmer than the inhaled air [4,17–21]. Air in the lungs is essentially saturated with water at body temperature of 37°C [22]. Most of that water is supplied by evaporation from the membranes lining the nose and as a result, these surfaces are cooled a bit. When one exhales, the breath passes over these cooled surfaces and loses some of its moisture, thus conserving at least some of the body water. However, the air exhaled is still saturated (100% relative humidity) or very close to it.

Recently we have developed a miniature optical humidity sensor based on an Agarose infiltrated photonic crystal fiber interferometer (AI-PCFI) [23]. Compared to the existing optical-fiber-based humidity sensors, the sensor proposed in [23] has the advantages of a very compact size, high resolution, fast response time, ease of fabrication. The sensor head is also low cost and thus offers the potential to be disposable. Disposable sensors are required while monitoring breathing from the nose or mouth because exhaled breath condensate can contaminate the sensor head making the sensor unsuitable to reuse on another patient. The end-type sensor head configuration offers the advantages of being able to operate in environments which demand a compact probe-type sensor and also reduced system complexity as only one interconnecting fiber is needed. These advantages underpin the motivation to investigate disposable sensors for breathing monitoring in a clinical situation using AI-PCFI. In this paper we demonstrate a breathing pattern and breathing rate sensor developed using an AI-PCFI. The sensor registers a change in the received optical power as a function of time as the air is exhaled making it suitable for monitoring breathing patterns. It has been demonstrated that by appropriate signal processing one can determine the breathing rate and the breathing status during expiration. The demonstrated sensor is also suitable for monitoring a decrease in respiratory rate, hypoxemia, or airway obstruction associated with the use of sedatives and anesthetics in the MRI environment.

2. Experimental demonstration and discussion

The complete experimental sensor system is composed of a light source - tunable laser (Anritsu, Tunics plus CL/WB), a fiber optic coupler/circulator (FOC), the AI-PCFI (relative humidity sensor), an optical detector (PX Instrument Technology, PX2000-306) and a PC with a breath analysis application program, as shown in Fig. 1. The AI-PCFI is composed of a small length of Photonic crystal fiber (PCF) fusion spliced to the end of a standard single mode fiber (SMF). The PCF in the sensor head has a microhole collapsed region near the splicing point and the free end of the PCF is infiltrated with Agarose.

Fusion splicing of the PCF to the SMF is undertaken using the electric arc discharge of a conventional splicer. During the splicing process, the voids of the PCF collapse through surface tension within a microscopic region close to the splice point [23–26]. In our experiment, PCF type LMA10 from NKT Photonics, designed for endless single-mode operation, was used. It has four layers of air holes arranged in a hexagonal pattern around a solid silica core. The light guidance mechanism in such a fiber is by means of modified total internal reflection. The dimensions of the LMA-10 PCF simplify alignment and splicing with
the SMF with a standard splicing machine and minimize the loss due to mode-field diameter mismatch compared to other PCFs. After fusion splicing, the PCF is cleaved using a standard fiber cleaving machine so that the end of the PCF exposed the local environment behaves as a reflecting surface. To improve the humidity sensitivity compared to the photonic crystal fiber interferometer sensor proposed in [24] in this case the microholes of the PCF are infiltrated with Agarose by immersing the tip of the PCF in a hot Agarose solution. The solution is prepared by dissolving 1 wt% Agarose (Sigma-Aldrich, A9045) in distilled water [23,27]. The size of the fabricated AI-PCFI is 1 mm in length and 125 μm in diameter; the length includes both the 250 μm hole collapsed regions and the approximately 100 μm long Agarose-infiltrated region. Such a small sensor length gives a large fringe spacing (calculated fringe spacing for a 1 mm length photonic crystal fiber interferometer is 330 nm [23]). An advantage of choosing a compact length is that the spectrum remains very stable when the sensor is subjected to vibrations, most particularly from airflow in this case. Good stability against vibrations and air flow is required for reliable breathing monitoring. The effective refractive index (RI) of the cladding mode of the device depends on the RI of the Agarose material infiltrated into the microholes of the PCF. The RI of the Agarose changes with the ambient relative humidity (RH), which in turn changes the modal propagation constant of the cladding mode. As a result, a phase change is induced between the interfering core and cladding modes, which in turn causes the shift of the interference pattern. A detailed calibrated RH response of the AI-PCFI is given in [23]. So by monitoring the humidity-induced changes in the interference fringes in the reflection spectrum of the device due to the breath exhale, the breathing rate and pattern can be measured.
In our experiment the optical power change obtained due to the spectral shift induced by changes in humidity during the breathing cycle is monitored. For this purpose a tunable laser output at a wavelength of 1550 nm is fed to the AI-PCFI via the FOC and the reflected optical power from the AI-PCFI is measured using a detector. Accurate measurement of pulmonary ventilation or breathing often requires the use of devices such as masks or mouthpieces coupled to the airway opening. For this purpose the device was mounted in an inexpensive, disposable plastic oxygen mask which in turn was attached to a volunteer’s nose and mouth and secured with the elasticized headband of the mask (See Fig. 1.). The sensor was set in such a way that it was kept approximately 5 cm from the patient’s nose to avoid condensation on the device. We estimated the distance between the tip of the nose and the tip of the sensor using a conventional measuring scale. In our study this distance varies depending on the patient’s nose size because the sensor is fixed on the oxygen mask (Fig. 1). However in practice this distance can be maintained by fixing the sensor on to a moving platform inside the oxygen mask or by suitably fixing the sensor inside the mask with a prior knowledge of the patient nose size. The normal baseline RH signal depends on the room humidity and also on the distance of the sensor from the patient’s nose. The time-dependent RH signal when the patient is breathing normally is equal to the ambient RH (< 80% RH) during inhalation and ~100% RH during exhalation. We have not investigated the detail packaging of the AI-PCFI in this study. But it is important to point out that before using the sensor in a clinical environment the sensor would have to be packaged inside a suitable plastic tube to prevent the transmitted light reaching the volunteer’s eyes, face or skin.
The optical power received by the detector is acquired in real time (with a maximum delay of the order of ms) using the breath analysis application program based on LabVIEW platform. Figure 2. shows a screen shot of the user interface of the breath analysis application program, for the case of a regular breathing pattern. The real time breathing response of the sensor system is displayed by the application program user interface. During breath expiration due to an increase of the ambient humidity in the vicinity of the sensor, the reflected power received by the detector decreases [23]. In the real time breathing response trace there are valleys which represent exhalation and peaks which represent inhalation. The real time breathing response of the sensor system is displayed by the application program user interface. During breath expiration due to an increase of the ambient humidity in the vicinity of the sensor, the reflected power received by the detector decreases [23]. In the real time breathing response trace there are valleys which represent exhalation and peaks which represent inhalation. The real time breathing response trace there are valleys which represent exhalation and peaks which represent inhalation. The lower plot in Fig. 2. is the breathing state indicator calculated from the breathing pattern using a preset threshold. It has two states, a high level represents the state after air is exhaled and a low indicates the state after air is inhaled. Each time when the power decreases below the preset threshold value, the breath count (BC) is incremented once and it is displayed in the user interface of the program. Since breath rate is the number of breaths per minute, after every 60 seconds the counter is reset. Average Breath Rate (BR) is calculated as BR = (BC*60)/ET where ET is the elapsed time since the last reset. For user convenience the elapsed time, breath rate and breath status are also displayed in the user interface/front panel of the application. Breathing status is described as follows ‘NORMAL’ when BR is between 10 and 20; ‘LOW’ when BR < 10; ‘HIGH’ when BR > 20.

Figure 3 shows a screen shot of the user interface for an irregular breathing pattern, where the subject ceases breathing for a few seconds, in order to demonstrate that our sensor system is capable of monitoring breathing abnormalities in real time. The valleys shown in the upper...
plots of Figs. 2 and 3 are due to the air breathed out by the volunteer and the area inside each of these valleys can be correlated with the amount of air breathed out. Thus by applying an appropriate signal analysis and calibration, the breathing air volume could be estimated from our sensor response. The value of breathing volume could be used as an indicator of potential respiratory dysfunction and subject’s pulmonary health status.

Since the dimensions of the sensor head are very small, the breathing response is unaffected by the turbulence and vibrations that result from air flow during breathing and other mechanical effects that may occur due to the movement of a patient. Given that the interrogation system measures the reflected intensity of light a possible source of error could arise from random variations in the received power resulting from variations in the bend radius at any bends in the fiber connecting the sensor to the interrogation system. To prevent such a failure the fiber cable connected to the sensor would need to be packaged to avoid small bend radii and significant changes in bending radius. Other fluctuation sources are the variations in the ambient RH, the distance between patient’s nose and the sensor and the fluctuations in the RH of the exhaled air. The source power fluctuation, wavelength drift and the detector noise might also contribute to the measurement errors in the system demonstrated. However selecting a threshold power equivalent to a higher RH than the ambient RH should overcome the effect of these fluctuation sources on monitoring the breath rate. Finally it should be point out that failure of the sensor may occur if a person coughs or contaminates the device with fluids (saliva, sputum, etc.). However, these issues can be avoided or overcome by embedding the device in a nasal clip, or by adequate packaging.

Because respiratory depression and upper airway obstruction are frequent complications associated with the use of sedatives and anesthetics, monitoring the respiratory rate, hypoxemia, and the detection of airway obstruction are important during the administration of these drugs. This is particularly important in the MRI environment because visual observation of the patient's respiratory efforts is often difficult. Also since the anesthetist cannot accompany the patient, it is essential that the patient is monitored remotely from the neighboring control room. The magnetic fields can interfere with electrical equipment, meaning that conventional electronic sensors cannot be used during the MRI scan. The sensor demonstrated in this paper is a potential solution for these problems. Finally while a tunable laser source was used in the demonstration described here, in order to reduce the cost of the system for real-world applications the tunable laser can be readily replaced with a low cost fixed wavelength laser diode. A high power change of >5 dB will be obtained using our sensor during breathing even in event of the source intensity and/or center wavelength drift. In addition a shift in the dynamic signal power level of the sensor can be addressed easily in the LabVIEW program by suitably shifting the threshold, so that the source power fluctuation and the wavelength drift of a low cost source will not degrade the accuracy of the breath rate monitoring.