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A New Method for Respiratory-Volume Monitoring Based on Long-Period Fibre Gratings *

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Abstract— **Respiratory–volume monitoring is an indispensable part of mechanical ventilation. Here we present a new method of the respiratory-volume measurement based on a single fibre-optical long-period sensor of bending and the correlation between torso curvature and lung volume. Unlike the commonly used air-flow based measurement methods the proposed sensor is drift-free and immune to air-leaks. In the paper, we explain the working principle of sensors, a two-step calibration-test measurement procedure and present results that establish a linear correlation between the change in the local thorax curvature and the change of the lung volume. We also discuss the advantages and limitations of these sensors with respect to the current standards.**

I. INTRODUCTION

In medicine, mechanical ventilation is a method to mechanically assist or replace spontaneous breathing. The ventilators are equipped with sensors designed to detect the start of patient's inspiratory effort in response to which the ventilator will pump air into the patient's respiratory system [1]. Continuous measurement of the tidal volume Vt (the volume of air moved into or out of the lungs) is necessary to ensure adequate and timely mechanical ventilation.

The ventilators can be used either invasively, by using tracheal intubation (a tube is inserted through the nose or mouth and advanced into the trachea), or non-invasively, by using an oronasal mask or a mouthpiece. In the non-invasive ventilation, tidal volume is monitored by measuring air flow and is hence very often affected by oronasal mask air leaks and thus unreliable. Most efforts invested in the development of direct monitoring methods independent of air leaks have been based the discovery of Konno and Mead [2] that the thoracic and abdominal movements caused by breathing can give a good measure of the volume of the inhaled air. It has been used to develop several methods and devices: Respiratory Inductance Plethysmography [3], Fibre-optic Respiratory Plethysmography [4], Optoelectronic Plethysmography [5], Respiratory Movement Measuring Instrument [6] and Plethysmography based on LPG sensors [7]. Each of these techniques has some drawbacks that impose limits to its usage in ambulatory respiration monitoring, the most detrimental of them being a large baseline drift, difficult or 'poor' accuracy of calibration, and the measurement accuracy insufficient for clinical use [8,9,10,11].

In this paper we describe a new method and apparatus for continuous monitoring of the respiratory tidal and minute volumes based on measuring of thoracic movements. The basis of the method is the existence of a good correlation between a change in the lung volume and a change in a local curvature of humane torso. In this paper we present a prove of this dependence, which we reach by using a novel fibre-optic long-period grating (LPG) curvature measurement scheme and the standard volume measurements by spirometer and pneumotachograph. The proposed single-sensor scheme is non-invasive, simple, low-cost and easy to implement.

The paper is structured as follows. Section II describes the LPG sensor, interrogation scheme and the measurement protocol. Section III shows the results of measurements on a set of healthy volunteers. These results provide a proof of the correlation between torso curvature and lung volume and of the feasibility of the proposed sensing scheme. Advantages and limitations of the new method and the scheme are discussed in Section IV. Conclusions and directions of future work are given in Section V.

II. METHOD

A. *Long-period grating sensors and interrogation scheme*

The LPG is a device that consists of a periodic change in the refractive index or the fibre geometry along the fibre length with the typical period of several hundred micrometers. It couples the light from the core mode to the resonant co-propagating cladding modes of the fiber thereby producing attenuation bands at discrete wavelengths in the transmission spectrum. The resonant wavelength and the spectral profile of an attenuation band are sensitive to the force applied to the fibre (strain, bending, load) and local environmental conditions (temperature, humidity, referactive index of the surrounding material) [12, 13]. Bending induces strain and stress across the fiber and consequently a change in its refractive index via the strain-optic effect. The corresponding changes in the propagation of the core and, in particular, cladding modes are observed as changes in

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position, shape and amplitude of the resonant bands in transmission spectrum.

The full grating characterisation can be obtained by using an optical spectrum analyzer, but the cost of such a scheme is a practical obstacle to its common clinical use. Here we use a simple monochromatic interrogation scheme that relies on the measurement of power transmitted through an LPG, see Fig.1. Such a scheme requires tuning of the grating into the resonance with the laser, which is achieved by a judicious positioning of the sensor on a patient's torso. The sensor was fixed by an elastic bandage that goes around the patient's torso and over the sensor. In this manner it is easier to manipulate with the sensor in order to find a good working point than by using a tape. The final version of the interrogator will automatically tune the laser wavelength to the optimal working point within the grating resonance, which will allow for application of self-adhesive sensor patches.

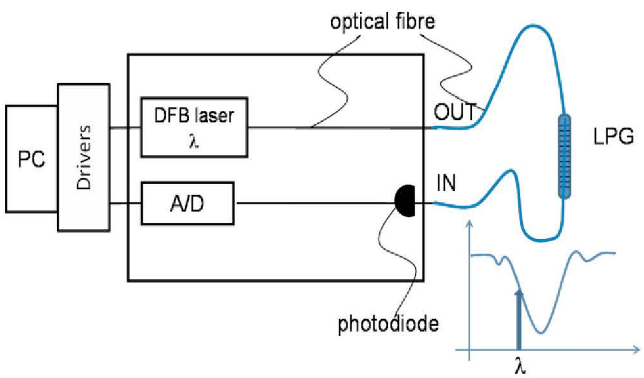

Figure 1. Sensing scheme. The interrogation module consists of a fibre-coupled narrowband temperature- and current-stabilized laser diode and a photodiode for the measurement of power transmitted through the sensor. The inset shows a typical LPG transmission spectrum.

B. *Measurement protocol*

The measurement protocol comprises two steps, calibration and test measurements [14].

The calibration function for the conversion of an LPG-sensor curvature signal to volume is obtained by performing simultaneous measurements by an LPG sensor and a standard reference over a short interval of breathing. The reference measurements were performed by a spirometer (SpiroTube, Thor Medical, Budapest) and a pneumotachograph (C2, Hamilton Medical, Boanduz, Switzerland). The measurement interval must be at least one breathing cycle (breath in and out) and should cover the whole relevant breathing range. All measurements reported here lasted 1 min each. The used reference instruments measure the air velocity and calculate the volume by integration of the air flow during inspiration and expiration. As the speed of sound depends on the temperature, humidity and pressure of the flowing air, which may be different during inspiration and expiration, a baseline drift may appear in the reference respiratory volume signal. This drift was eliminated by subtracting the difference between the baseline drifts of the sensor signal and the reference signal from the reference signal. The drifts were calculated by polynomial fitting. In this way, the calibration eliminates the excess baseline drift in the reference volume measurement without distorting the natural change in end-expiratory volumes. Upon the elimination of the drift, the calibration function is found by a polynomial fit (the 1st order was sufficient) of a sensor to the drift free reference signal. The calibration function is then used in calculation of the respiratory volumes from all test measurements performed under the same conditions.

During all measurements the volunteers were in supine position equivalent to the position of patients during mechanical ventilation. Several sensor positions were tested and the zone of ribs 6-8 on the right side of the rib cage was chosen as optimal, Fig. 2. The distance of this zone from the heart also ensures that the impact of the heart beat on the sensor signal is reduced. A number of healthy voluneteers were measured, 5 with the spirometer and 5 with the pneumotachograph as a reference.

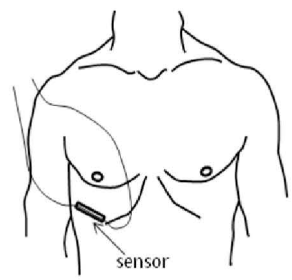

Figure 2. Sensor position on thorax.

III. RESULTS

A typical set of calibration and test results is shown in Fig. 3. The scatter plot in (a) shows that the correlation function between the sensor signal and the volume measured by the spirometer is linear. The corresponding calibration function was obtained by a linear regression from the scatter plot. This and a series of similar test results prove our hypotesis that the torso curvature and the lung volume are correlated. The calibration uncertainty Δ is defined as a radius of the region around the calibration curve that contains 68% of all measured values. The results of a dynamic test measurement in Fig. 3(b) show a good agreement between the sensor and the spirometer with an error of less then 10% for a variety of tidal volumes. Fig. 3(c) shows the tidal volume (the volume of the breathed-in air) in each respiratory cycle. Integration of the whole signal over one minute yields the minute volume.

Tables 1 and 2 give the minute volumes measured by LPG sensors. In the measurements represented by Table 1 the calibration was done using the spirometer and in the measurements represented by Table 2 using the pneumotachograph as a reference. The measurements were performed on a set of volunteers with the normal body-mass index (BMI in the range 18.5 – 25). Sensors had up to 8% error with respect to the spirometer and up to 4% error with respect to the pneumotachograph. The relative error of the calibration calculated as the RMS error per pulse volume of the test signal does not exceed 10%. The good agreement of the sensors with both standards proves their accuracy and consistency in time, as well as validity of the measurement procedure.

Another important parameter of a sensor is its sensitivity defined as the smallest signal that the sensor can detect. The sensitivity of the current LPG interrogation scheme is 0.3 ml. Accuracy of the method, however, depends on the positioning of the sensor and the contamination by a mechanical signal originating from the heart beat. Indeed, the dynamic range of the current interrogation scheme of 0 - 200 Hz allows for a simultaneous detection of the cardiac and respiratory signals. We have found that by positioning of the sensor to the right side of the rib cage the heart beat mechanical signal can be significantly reduced or eliminated. The accuracy of such a tidal volume measurement is better than ±100 mL with respect to the spirometer and better than ±50 mL with respect to the pneumotachograph.

TABLE I. MINUTE VOLUME MEASUREMENT WITH A SPIROMETER AS A REFERENCE

No	Sex	BMI	Minute Volume			Δ [l]
			sensor	spiro meter	error [%]	
1	M	24.0	12.47	12.15	2.7	0.079
2	F	21.8	10.67	10.51	1.5	0.079
3	M	23.4	22.31	21.56	3.5	0.077
4	M	23.6	16.02	15.79	1.4	0.085
5	F	23.0	11.09	10.32	7.5	0.092

TABLE II. MINUTE VOLUME MEASUREMENT WITH A PNEUMOTACHOGRAPH AS A REFERENCE

No	Sex	BMI	Minute Volume			Δ [l]
			sensor	Pneumo tach.	error [%]	
1	F	22.1	4.80	4.79	0.26	0.013
2	M	24.0	4.98	5.05	1.29	0.032
3	M	23.1	6.59	6.81	3.12	0.060
4	M	22.4	8.35	8.63	3.28	0.110
5	F	23.0	7.83	7.83	0	0.054

IV. DISCUSSION

Due to the outcoupling of light from the core to the radiation cladding modes, long-period gratings are sensitive to changes in environmental parameters. Here, the grating cross-sensitivity to the surrounding refractive index has been avoided by using a progressive three-layered fibre whose outer cladding serves as an isolator [15]. The effects of rapid temperature fluctuations have been reduced by encapsulating the fibre into a low-temperature curing silicone rubber [16], which is also convenient for application on the surface of the human body. The biggest problem caused by the grating sensitivity to the parameters other than bending is its sensitivity to the forces instigated on the fibre by the upper-body movements that a patient makes voluntarily or not. Possible strategies for elimination and alleviation of this problem are fabrication of sensors as patches that can be attached to the body and/or automatic correction of the sensor working point upon a movement of the patient.

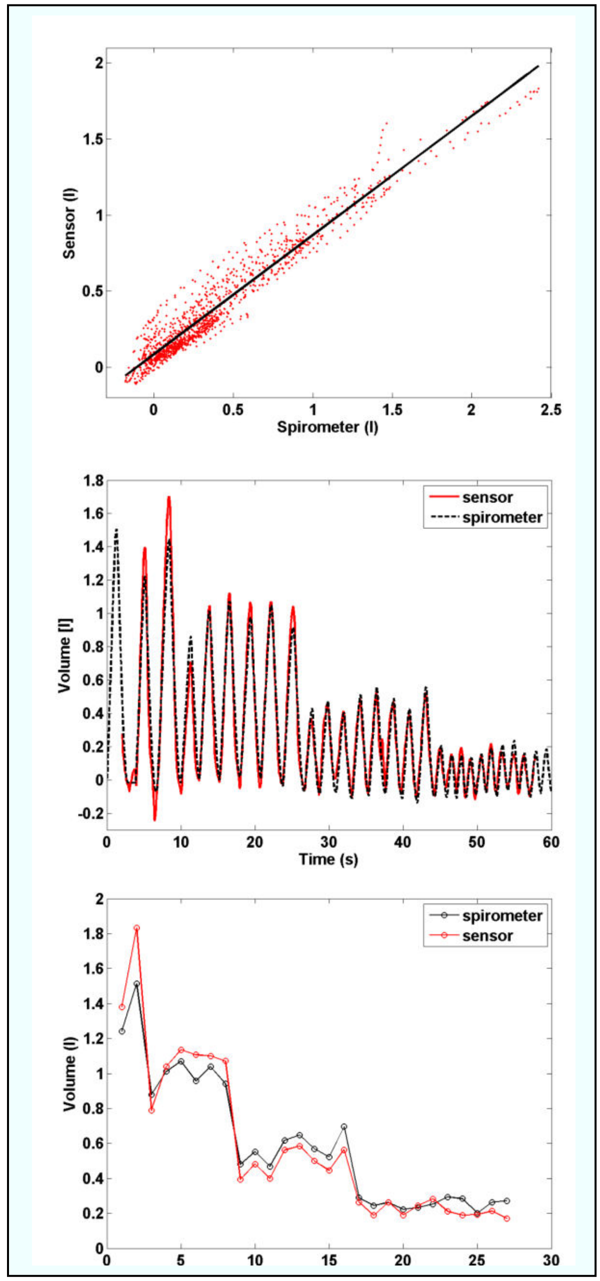

Figure 3. (a) Scatter plot (dots) and the calibration function (line), (b) Calibrated sensor output, (c) Tidal volume extracted from (b).

Fibre gratings are linear sensors and as such can be arranged to construct 2D sensor networks. This level of complexity is necessary for measurements of absolute volumes [16] or for shape reconstruction [17]. While the grating network is also capable of measuring tidal volumes, the single-sensor scheme is simpler, easier to implement and less susceptible to signal distortion due to the movements of the soft abdominal tissues.

By direct monitoring of local changes in the torso curvature in mechanically ventilated patients, LPG sensors eliminate the flaws of air-flow measurements that are associated with leaks from oronasal masks [18].

Moreover, flow-based measurements of respiratory volumes suffer from baseline drifts caused by changes in environmental parameters, mainly temperature [18]. Although the state-of-the-art pneumotachometers compensate these dependences in different ways, even the most sophisticated devices have tidal volume errors in the range of ±5%, see e.g. D-lite flow and airway pressure sensor (GE Healthcare, Helsinki, Finland) [19]. The curvature measurements performed by LPGs are not significantly affected by fast changes in environmental parameters. The observed small baseline drift in the LPG curvature signal is caused only by a natural change in end-expiratory volume.

If the heart beat mechanical signal contaminates the useful signal, it can be removed by one of the signal-separation techniques, such as those applied in separation of electrocardiogram from other electrical physiological signals [20, 21]. On the other hand, the simultaneous detection of the respiratory and cardio signals opens the door to the development of sophisticated and versatile monitoring schemes.

V. Conclusion

We have presented a novel sensing scheme for the real-time monitoring of lung volume and the measurement of tidal and minute volumes. The scheme is based on a long-period grating fibre-optic sensor of bending and the assumption of the correlation between the local torso curvature and the lung volume. By applying a two-step calibration-test measurement procedure to a series of healthy volunteers, we have found that this correlation is linear in the normal breathing range. The demonstrated technique is not affected by air-leaks and drifts that are limiting factors in the currently used air-flow measurement methods. As it does not require an oronasal mask, it is also more convenient for a patient. The major source of errors in the proposed scheme is its sensitivity to the body movements not associated with breathing. Its elimination, along with the development of a tuneable-laser interrogation scheme, is a priority of the future research.

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