FIBER OPTIC SENSOR FOR THE MEASUREMENT OF RESPIRATORY CHEST CIRCUMFERENCE CHANGES

A. Babchenko, B. Khanokh, Y. Shomer, and M. Nitzan
Jerusalem College of Technology, Department of Applied Physics/Electro-Optics, Jerusalem 91160, Israel

(Paper JBO-211 received July 20, 1998; revised manuscript received Nov. 1, 1998; accepted for publication Nov. 2, 1998.)

ABSTRACT

A fiber optic sensor for the measurement of the respiratory depth has been developed. The sensor is composed of a bent optic fiber which is connected to an elastic section of a chest belt so that its radius of curvature changes during respiration due to respiratory chest circumference changes (RCCC). The measurement of light transmission through the bent fiber provides information on its changes in curvature since a higher fraction of light escapes through the core-cladding surface of a fiber bent to a lower radius of curvature. The sensor can quantitatively measure the RCCC, although in relative terms, and it is sensitive enough to detect changes of the chest circumference due to the heart beat. Measurements of the RCCC were simultaneously performed with photoplethysmography (PPG)—the measurement by light absorption of the cardiac induced blood volume changes in the tissue—and a significant correlation was found between the RCCC and some parameters of the PPG signal. The fiber optic respiratory depth sensor enables a quantitative assessment of the respiratory induced changes in the cardiovascular parameters.
[S1083-3668(99)00402-5]

Keywords fiber optic sensor; respiratory depth measurement; photoplethysmography.

1 INTRODUCTION

Optical fibers are used extensively in medicine both for diagnosis and for treatment. Diagnostic fiber optic sensors are mainly based upon sending light through a fiber to the measurement site, and measuring the light reflected or scattered from the tissue or from a transducer at the end of the fiber. The reflected light is conveyed to the optical detector through either the same fiber or a different one. The sensors that measure the light reflected from the tissue itself include invasive or noninvasive oximeters, laser Doppler flowmeters, and sensors for fluorescence and temperature measurements.[1–4] The other sensors use an appropriate transducer attached to the measurement site at the tip of the fiber for measurement of temperature, pressure, *p*H, or concentrations of other analytes.[1,5–7] In some sensors the whole surface of a portion of the fiber serves as the measurement site by using the evanescent light through the unclad fiber for exiting fluorescence in molecules attached to the fiber surface.[8] In the Bragg grating fiber sensor a periodic modulation of the fiber core refractive index was used for measurement of the blood temperature in the magnetic resonance imaging (MRI) environment.[9] In the sensors discussed above the optic fibers are only utilized for transmission of light to the measurement site and back to the photodetector in order to achieve remote sensing.

In another category of fiber optic sensors the fiber itself is used for measurement of the required parameter. Interferometric measurement of the phase change due to tension induced elongation of the optic fiber was used in nonmedical applications as a strain sensor.[10] This method of phase change measurement provides assessment of displacements of the order of magnitude of light less than a wavelength and is not suitable for medical applications. Another fiber optic displacement sensor was introduced by Stenow and Öberg[11] that utilizes the leakage of light from a fiber under multiple periodic longitudinal microbending. At an optimal wavelength of the periodic mechanical perturbation, the coupling between core modes is enhanced, thereby increasing the radiative losses from the fiber. This sensor enabled measurement of the limb circumference change during occlusion plethysmography. A theoretical calculation of the bend loss in a fiber was performed for periodic microbending in a multimode fiber[11] and for a single bend in a single mode fiber (or a few modes fiber)[12] by analyzing the mode coupling at the bends. The loss of a single bend in a large core multimode optical fiber was calculated by analyzing the propagation of the eva-

Address all correspondence to Prof. Meir Nitzan, Fax: 972-2-6751200.

1083-3668/99/$10.00 © 1999 SPIE

nescent waves of the higher-order modes in the cladding,[13] or by ray tracing.[14] In the current study a fiber optic sensor based on measurement of a single bend loss was used for assessment of respiratory chest circumference changes (RCCC). The measurement of the changes in the light transmitted through the fiber provides a means of assessing the changes in the fiber radius of curvature.

Proper functioning of the chest cavity is vital for adequate breathing, and several methods have been developed for measuring and monitoring the respiration rate and the inspiration depth.[15,16] The breathing pattern can be monitored either by directly measuring the airflow or air volume displacement during breathing or by measuring body dimensional changes between expiration and inspiration.[15,16] Direct measurements of airflow for mere *detection* of inspiration or expiration, such as air temperature sensors, are very simple and easy to use, but their application is limited to the detection of apnea (cessation of breathing) or to measurement of the respiratory *rate*. Quantitative direct measurement of the tidal volume (the volume of inspired air during the respiratory cycle) requires a nonleaking connection to the patient's airways (a mask over the mouth and nose or an endotrachial canula) and is therefore not appropriate for most clinical uses and physiological studies.[16,17] Another type of method for evaluation of the breathing pattern is based on measurement of parameters that depend on body dimension changes, such as transthoracic impedance plethysmography or respiratory induction plethysmography, which, respectively, measure the electric resistance of the chest or the self-inductance of insulated coils which are wrapped around the patient's chest (or abdomen). These techniques can provide quantitative information about *relative* changes in tidal volume and even measure it in absolute terms after suitable calibration.[15–17] These methods are used for clinical tests of pulmonary function, and are also applied in different physiological studies,[18–20] such as the investigation of respiratory sinus arrhythmia, i.e., the increase of heart rate during inspiration, which occurs due to activity of the autonomic nervous system.

2 MATERIALS AND METHODS

2.1 RCCC SENSOR

The sensor for the measurement of the respiratory chest circumference changes was composed of a quartz-polymer optical fiber of 400 μm core diameter bent into the shape of an ellipse (Figure 1). The bent fiber is connected to a chest belt. The principle of operation of the sensor can be qualitatively described by the decrease of the angle of incidence of the light rays in the bent region relative to that of the straight region (Figure 2), resulting in radiation loss of some of the rays which are totally reflected in the straight fiber. A light source and a photodetector were attached by means of connectors to the two ends of the fiber for measurement of the light transmission through it. As a light source, an infrared light emitting diode (LED) was used, emitting 845 nm wavelength light at an angle of 5°, which was modulated at a frequency of 3 kHz in order to eliminate background noise. The output light intensity was detected by a P–I–N silicon photodiode

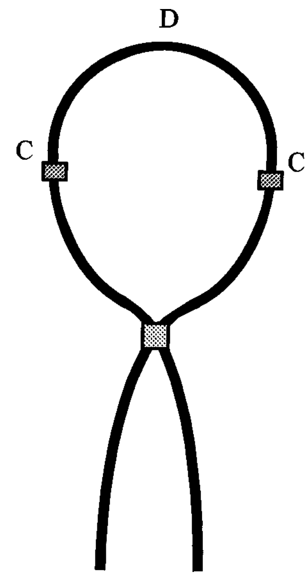

Fig. 1 Bent optical fiber of the RCCC sensor. C indicates the points of connection to the nonelastic part of the belt. D is the region between the two C regions.

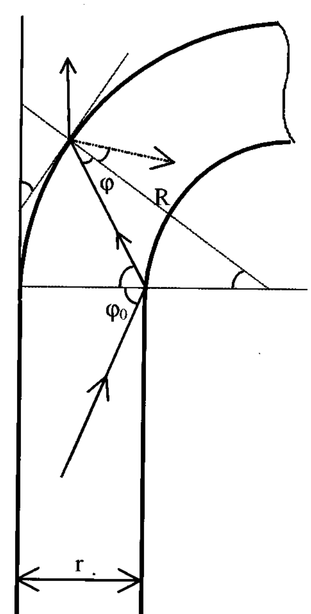

Fig. 2 Reflection and refraction of light in the bent optical fiber. φ_0 and φ are the angles of incidence in the straight and bent regions, respectively, of the optical fiber. The ray incidenting the core surface in the straight region is totally reflected, while the ray incidenting the surface at an angle smaller than the critical angle is partially transmitted.

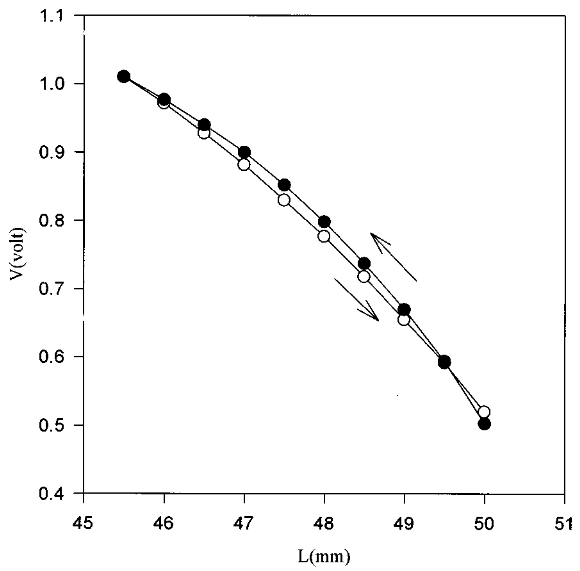

Fig. 3 Output voltage V of the fiber optic sensor as a function of the distance L between the two C points of Figure 1, measured during an increase and decrease of L. The chest circumference changes for normal respiration of a male adult are 2–3 mm.

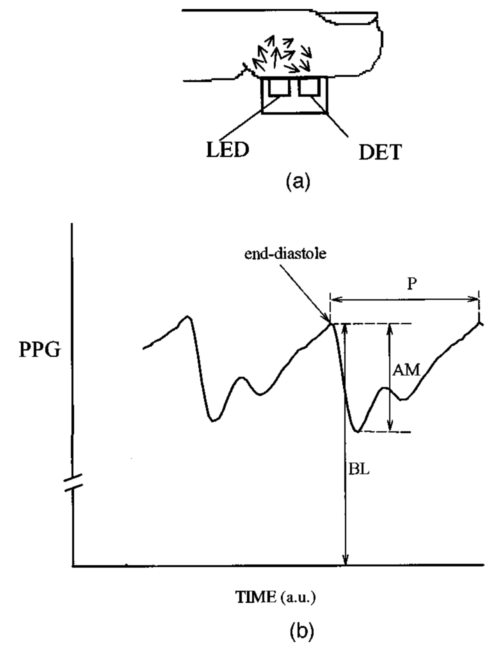

Fig. 4 (a) PPG probe, attached to the fingertip. LED—light source, DET—photodetector. (b) PPG signal as a function of time. The signal decreases during systole and increases during diastole. BL—the baseline of the PPG pulse; AM—its amplitude; P—its period.

with a spectral response of 0.56 A/W at 845 μm. The photodetector's output was demodulated and amplified, and the amplified signal was digitized and sampled (500 samples per second) by an analog-to-digital (A/D) converter.

The chest belt was divided into two sections: a short elastic part, which can be lengthened during inspiration, and a flexible nonelastic longer part whose length does not change due to the respiration process. The fiber, bent approximately to a circle of about 40 mm in diameter, was connected at two points (C in Figure 1) to the edges of the flexible nonelastic section, parallel to the elastic section, so that the radius of curvature in the two C points decreased during inspiration, resulting in greater light loss in the bends. Although the radius of curvature in the D region (the region between the two C points) increased, the light transmission through the total fiber decreased during inspiration.

The linearity of the chest circumference sensor was checked by stretching the elastic part of the belt with a translation stage and measuring the sensor output light intensity as a function of the distance between the two C points (see Figure 1). The results are depicted in Figure 3, showing that the deviation of the sensor output from linearity is smaller than 10% in the range of 2–3 mm, which is the typical increase in chest circumference, during normal inspiration. The maximal difference between the sensor output during increase and decrease of the sensor length is equivalent to 0.25 mm, which is about 10% of the typical RCCC.

2.2 PHOTOPLETHYSMOGRAPHY MEASUREMENT

In order to show the change of cardiovascular parameters between inspirium and expirium, the photoplethysmography[21,22] (PPG) was measured simultaneously with the RCCC measurement. PPG evaluates the increase in tissue blood volume during heart contraction (systole) by measuring the resultant increase in light absorption during systole. The PPG probe included an infrared LED (880 nm) as the light source which emits modulated light into the tissue, and a P–I–N photodetector located 13 mm from it which measured the light scattered from the tissue under the skin. The probe was attached to the right-hand index fingertip of the subject, and the hand was held at heart level. The photodetector output was demodulated, amplified, digitized, and sampled by the same A/D converter used for the breathing depth measurement. Figure 4 shows the PPG probe and the PPG signal, which depicts oscillations in the tissue blood volume due to oscillatory heart activity. The PPG signal was automatically analyzed for each pulse to determine the baseline (BL) of the pulse, its amplitude (AM) and its period (P), which is equal to the heart period. Each of these three parameters oscillates in the respiratory frequency[23,24] due to the influence of inspiration on the cardiovascular system. Fluctuations of the PPG signal parameters at frequencies lower than that of the respiration were filtered out by smoothing the original curve of the parameter (by means of moving an average of five adjacent points) and subtracting it from the original curve. The high frequency fluctuations in the BL, AM, and P were qualitatively compared to the output of the chest circumference changes sensor.

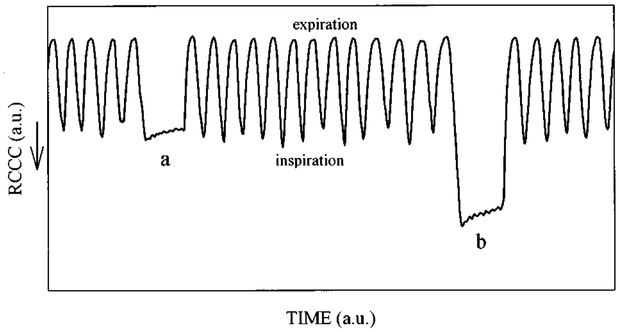

Fig. 5 Respiratory chest circumference changes (CCC) as a function of time. The arrow indicates the direction of the chest circumference increase. (a), (b) Periods of respiratory arrest after inspiration. Note the oscillations in the CCC signal during the respiratory arrest periods.

Measurements of the respiratory induced fluctuations of the PPG signal parameters were performed on eight healthy adult male subjects aged 21–56 years. The examinations were performed in the sitting position, after a 10 min rest. Three to four examinations were performed on each subject; each included a 1 min simultaneous measurement of chest circumference and PPG signal fluctuations during normal breathing.

3 RESULTS

Figure 5 shows the chest circumference measurement during normal breathing and during short periods of breathing arrest for one of the subjects. After a normal breathing period, the respiration was arrested for about 7 s [Figure 5(a)], and after another period of normal breathing, the subject stopped breathing after a deep breath for about 7 s [Figure 5(b)]. Then the subject continued to breathe normally. Note the small oscillations during the respiration arrest periods. Also note that for different expirations the sensor output level is almost constant, while changes in the sensor output at the different inspirations are significant.

Figure 6 is an enlarged presentation of the oscillations in chest circumference obtained in another study during the respiration arrest period measured simultaneously with PPG measurements. Since the PPG signal measures the changes in tissue blood volume due to heart activity, the periodic changes in chest circumference which correlate well with the PPG signal are also due to the heart pulse. It can be seen that the systolic rapid increase of the circumference sensor signal leads the systolic rapid increase of the PPG pulse, as can be expected, since the pressure pulse propagates in finite blood velocity and it arrives at the finger with a time delay relative to that of a heart contraction.

Figure 7 shows a simultaneous recording of the PPG and chest circumference during normal breathing for one of the subjects. After the examination, the PPG signal was analyzed and the minimum and the maximum of each pulse were digitally determined in order to obtain the baseline, the amplitude, and the period for each pulse. The low frequency fluctuations of these PPG parameters were filtered out as described above, and the high frequency oscillations of these three PPG parameters as a function of time were then drawn. The result is shown in Figure 8 (circles) together with the curve of the respiratory chest circumference changes. The P, AM, and BL curves are positively correlated with the RCCC curve. Similar results were obtained in most of the other examinations, but the value of the correlation coefficient changed significantly in different examinations of the same subject. In three examinations the correlation coefficient of the BL with the RCCC was inverse.

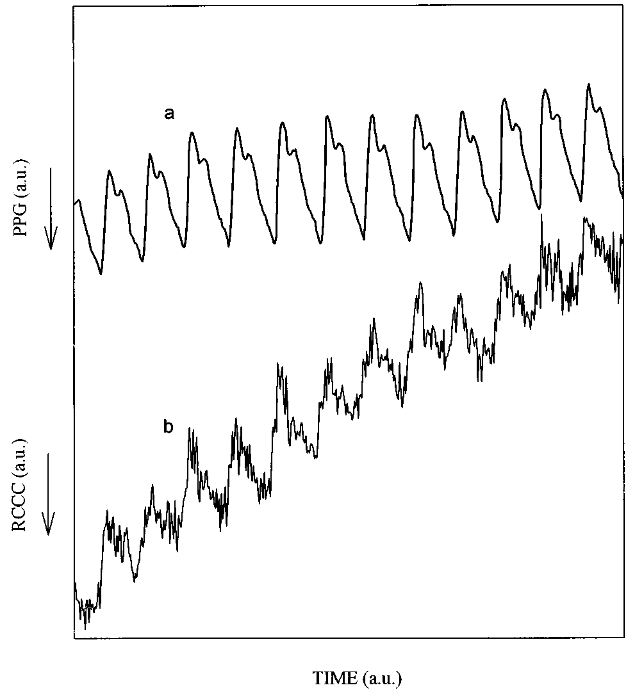

Fig. 6 Simultaneous measurement of PPG on the fingertip (a) and chest circumference changes (b) during respiratory arrest as a function of time. The PPG signal is inverted.

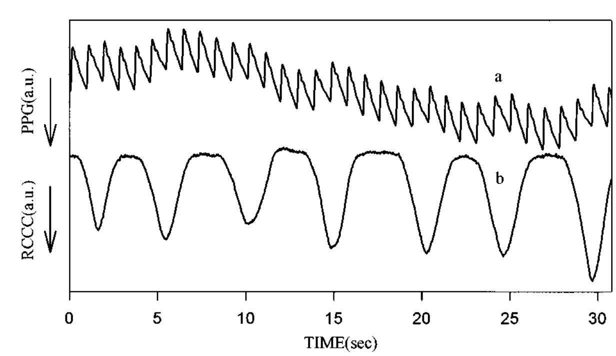

Fig. 7 Simultaneous measurement of the PPG signal on the fingertip (a) and respiratory chest circumference changes (b) for normal breathing as a function of time.

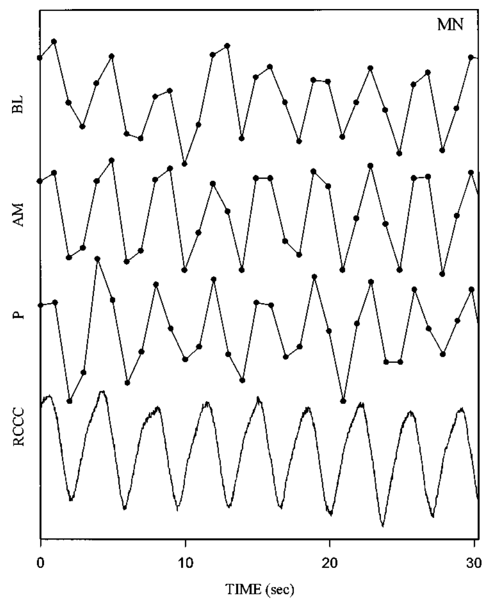

Fig. 8 Values of parameters BL, AM, and P of the PPG pulses and synchronous measurement of the respiratory chest circumference changes.

4 DISCUSSION

In the current study an intensity-modulated fiber optic sensor for the measurement of the respiratory chest circumference changes was constructed and examined. The sensitivity of the sensor is high, as can be deduced from its ability to detect the chest circumference changes due to the heart beat during respiratory arrest (Figure 5). The sensor is almost linear in the circumference changes range of normal breathing and can therefore provide quantitative assessment of breathing depth in relative terms. In order to assess breathing depth in absolute terms, two belts, around the chest and around the abdomen, are required in addition to suitable calibration. Since the sensor is based on a multimode fiber and a LED of 5° radiant angle as a light source, the launching conditions do not significantly affect the sensor sensitivity. The dependence of the bend loss on the temperature[25] is not expected to interfere with the RCCC sensor, since the temperature of the sensor does not change during the examination. Furthermore, the temperature dependence reduces significantly for radii of curvature above 10 mm,[25] while the fiber of the RCCC sensor has a radius of curvature of about 20 mm. The minor influence of the possible changes in temperature on the sensor output can be deduced from the almost constant level of the latter in different expirations (see Figures 5 and 7).

Preliminary simultaneous examinations of the respiratory chest circumference changes and the high frequency fluctuations in three PPG signal parameters, BL, AM, and P, revealed positive correlation of the RCCC signal with the P and AM for all the examinations and with the BL for most of them. The heart period decreases (heart rate increases) during inspiration, an effect known as respiratory sinus arrhythmia[18–20] which originates from the influence of the respiration on the heart rate through direct coupling between the respiratory and the autonomic centers and through a mechanical influence on venous pressures and flows. The significant correlation between the BL or AM and the RCCC is in agreement with other studies in which the respiratory induced variability of the PPG signal was used for the monitoring of the respiratory rate[26] and the respiratory volume.[27] The BL is inversely related to the total fingertip blood volume, and the AM is directly related to the compliance to pressure of the arterial system in the fingertip.[24] The inspiratory decrease of the BL and AM seem to be related to the higher arterial blood pressure during inspiration,[28,29] since higher arterial blood pressure is correlated with higher arterial blood volume and a higher tone of the arterial walls. The increase in tissue blood volume results in an increase in the total light absorption by the tissue, and the increase in the tone of the blood vessel wall results in a lower systolic increase in the blood volume in the blood vessels. Consequently, both the BL and AM generally decrease during inspiration. Since the vascular system is complex, there also are other mechanisms by means of which the respiration can affect the blood vessels, hence the negative correlation that was found in some examinations between the BL and the RCCC. A more comprehensive study of this issue is being performed in our laboratory.

5 CONCLUSION

In the current study, a fiber optic device was developed and used for measurement of chest circumference changes during breathing. The device is sensitive, simple, and easy to use and can therefore be used for monitoring respiration rate and depth.

The correlation found between the respiratory chest circumference changes and the high frequency variability of the PPG parameters proves that the sensor enables the evaluation of the respiratory induced oscillations in the heart rate (respiratory sinus arrhythmia) and other cardiovascular parameters. The respiratory changes in the heart rate and other parameters are due to the regulatory mechanism of the autonomic nervous system, and a method for a quantitative evaluation of this effect is important both for physiological cardiovascular studies and for clinical neuropathic examinations.

Acknowledgments

This research was supported by the E. W. Joseph Fund and by the Israeli Ministry of Science. One of the authors (B.K.) was supported by the Israeli Ministry of Immigrant Absorption.

REFERENCES

1. J. I. Peterson, ''Optical sensors,'' in *Encyclopedia of Medical Devices and Instrumentation*, J. G. Webster, Ed., pp. 2121–2134, Wiley, New York (1988).
2. H. Cai, S. Larsson, and P. A. Oberg, ''Single fiber laser-Doppler flowmetry—Dependence on wavelength and tip optics,'' *J. Biomed. Opt.* **3**, 334–339 (1998).
3. V. G. Artjushenko, A. A. Lerman, A. P. Kryukov, E. F. Kuzin, V. N. Ionov, N. A. Afanasyeva, V. S. Letohov, V. V. Sokolov, G. A. Frank, S. Romano, and W. Neuberger, ''MIR-fiber spectroscopy for minimal invasive diagnostics,'' *Proc. SPIE* **2631**, 92–97 (1995).
4. A. Katzir, H. F. Bowman, Y. Asfour, A. Zur, and C. R. Valeri, ''Infrared fibers for radiometer thermometry in hypothermia and hyperthermia treatment,'' *IEEE Trans. Biomed. Eng.* **36**, 634–636 (1989).
5. L. H. Lindstrom, ''Miniaturized pressure transducer intended for intravascular use,'' *IEEE Trans. Biomed. Eng.* **17**, 207–215 (1970).
6. J. L. Gehrich, D. W. Lubbers, N. Opitz, D. R. Hansmann, W. W. Miller, J. K. Tusa, and M. Yafuso, ''Optical fluorescence and its application to an intravascular blood gas monitoring system,'' *IEEE Trans. Biomed. Eng.* **33**, 117–131 (1986).
7. S. C. Liao, Z. Xu, J. A. Izatt, and J. R. Alcala, ''Real-time frequency domain temperature and oxygen sensor with a single optical fiber,'' *IEEE Trans. Biomed. Eng.* **44**, 1114–1121 (1997).
8. J. D. Andrade, R. A. Vanwagenen, D. E. Gregonis, K. Newby, and J. N. Lin, ''Remote fiber-optic biosensors based on evanescent-excited fluoro-immunoassay: Concept and progress,'' *IEEE Trans. Electron Devices* **32**, 1175–1179 (1985).
9. Y. J. Rao, D. J. Webb, D. A. Jackson, L. Zhang, and I. Bennion, ''Optical in-fiber Bragg grating sensor systems for medical applications,'' *J. Biomed. Opt.* **3**, 38–44 (1998).
10. H. W. Haslach and J. S. Sirkis, ''Surface-mounted optical fiber strain sensor design,'' *Appl. Opt.* **30**, 4069–4080 (1991).
11. E. N. D. Stenow and P. A. Oberg, ''Venous occlusion plethysmography using a fiber-optic sensor,'' *IEEE Trans. Biomed. Eng.* **40**, 284–289 (1993).
12. H. F. Taylor, ''Bending effects in optical fibers,'' *J. Lightwave Technol.* **LT-2**, 617–628 (1984).
13. A. A. P. Boechat, D. Su, D. R. Hall, and J. D. C. Jones, ''Bend loss in large core multimode optical fiber beam deliver systems,'' *Appl. Opt.* **30**, 321–327 (1991).
14. I. S. Melnik, I. V. Kravchenko, N. A. Denisov, S. M. Dets, and T. V. Rusina, ''Transmission of straight and curved multimode optical fibers,'' *Proc. SPIE* **2631**, 226–233 (1995).
15. M. A. Sackner and B. P. Krieger, ''Noninvasive respiratory monitoring,'' in *Heart-lung Interactions in Health and Disease*, S. M. Scharf and S. S. Cassidy, Eds., pp. 663–695, Marcel Dekker, New York (1989).
16. M. R. Neuman, ''Neonatal monitoring,'' in *Encyclopedia of Medical Devices and Instrumentation*, J. G. Webster, Ed., pp. 2015–2034, Wiley, New York (1988).
17. M. T. Antonio-Santiago and B. C. Clutario, ''Pulmonary function testing,'' in *Pulmonary Physiology: Fetus, Newborn, Child and Adolescent*, E. M. Scarpelli, Ed., Chap. 20, Lea and Febiger, Philadelphia (1990).
18. K. Yana, J. P. Saul, R. D. Barger, M. H. Perrot, and R. J. Cohen, ''A time domain approach for the fluctuation analysis of heart rate related to instantaneous lung volume,'' *IEEE Trans. Biomed. Eng.* **40**, 74–81 (1993).
19. M. Eriksen and K. Lossius, ''A causal relationship between fluctuations in thermoregulatory skin perfusion and respiratory movements in man,'' *J. Auton. Nerv. Syst.* **53**, 223–229 (1995).
20. P. Z. Zhang, W. N. Tapp, S. S. Reisman, and B. H. Natelson, ''Respiration response curve analysis of heart rate variability,'' *IEEE Trans. Biomed. Eng.* **44**, 321–325 (1997).
21. A. Fronek, *Noninvasive Diagnostics in Vascular Disease*, pp. 22–27, McGraw–Hill, New York (1989).
22. A. V. J. Challoner and C. A. Ramsay, ''A photoelectric plethysmograph for the measurement of cutaneous blood flow,'' *Phys. Med. Biol.* **19**, 317–328 (1974).
23. M. Nitzan, H. de Boer, S. Turivnenko, A. Babchenko, and D. Sapoznikov, ''Power spectrum analysis of the spontaneous fluctuations in the photoplethysmographic signal,'' *J. Basic Clin. Physiol. Pharmacol.* **5**, 269–276 (1994).
24. M. Nitzan, S. Turivnenko, A. Milston, A. Babchenko, and Y. Mahler, ''Low frequency variability in the blood volume and in the blood volume pulse measured by photoplethysmography,'' *J. Biomed. Opt.* **1**, 223–229 (1996).
25. R. F. Liu, X.-C. Xi, W.-M. Li, and D.-C. Chao, ''A fiber optic temperature sensor based on bend losses,'' *Proc. SPIE* **1572**, 180–184 (1991).
26. L. G. Lindberg, H. Ugnell, and P. A. Oberg, ''Monitoring of respiratory and heart rates using a fiber-optic sensor,'' *Med. Biol. Eng. Comput.* **30**, 533–537 (1992).
27. A. Johansson and P. A. Oberg, ''Estimation of respiratory volumes from the photoplethysmographic signal,'' *Med. Biol. Eng. Comput.* **35**, 270 (1997) (abstract).
28. J. P. Saul, R. D. Berger, P. Albrecht, S. P. Stein, M. H. Chen, and R. J. Cohen, ''Transfer function analysis of the circulation: Unique insights into cardiovascular regulation,'' *Am. J. Physiol.* **261**, H1231–H1245 (1991).
29. J. B. Madwed, P. Albrecht, R. G. Mark, and R. J. Cohen, ''Low-frequency oscillations in arterial pressure and heart rate: A simple computer model,'' *Am. J. Physiol.* **236**, H1573–H1579 (1989).