Blood pressure manometer using a twin Bragg grating Fabry-Perot interferometer

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ABSTRACT

We propose the use of optical fiber Bragg gratings in a non-invasive blood pressure waveform monitor. Bragg gratings can be written in a Fabry-Perot interferometric configuration to yield a method of strain measurement that has both a high resolution and a wide unambiguous range. This fiber Bragg grating Fabry-Perot interferometer (FBGI) can be used as a sensor to detect strain resulting from blood pressure applied to the walls of an artery situated near the patient's skin. Strain measurements taken on the skin surface, typically over the radial artery at the wrist, are encoded as phase shifts of the FBGI signal. These phase shifts may be obtained by the analytic representation of the interferometer signal in the wavelength domain or by Fourier analysis in the frequency domain. For the proof of concept a realistic physical model was constructed to simulate pressure conditions at the actual sensor location. The operation of the device is demonstrated by measurements of pressure-pulse waveforms obtained in real-time. This sensor was also successfully tested on human patients, and these results are also presented. Since it yields continuous readings of blood pressure non-invasively, further application of the optical manometer may yield an alternative to conventional sphygmomanometry.

Keywords: optical fiber sensors, fiber Bragg gratings, interferometry, sphygmomanometry

1. INTRODUCTION

In-fiber Bragg grating sensors have been successfully implemented to detect physical changes, especially strain and temperature, by observing the shift in Bragg wavelength induced by these variables[1]. When combining the characteristics of Bragg gratings and fiber optic interferometric sensors, the result yields a method of static strain measurement that has both a high resolution (in the nanostrain level[2]) and a wide unambiguous range[3]. The Bragg gratings act as in-fiber partial reflectors to form a low finesse interferometer. An advantage of using Bragg gratings in this configuration is that the sensor output is a shift in phase, and no longer a very small wavelength shift. Figure 1 depicts such a twin-grating all-fiber Fabry-Perot interferometric sensor, formed by writing two identical Bragg gratings (with the same central wavelength) a small distance apart.

One of the few truly scientific measurements that are done during clinical assessment is the procedure of taking a patient's blood pressure, and medical personnel spend a large amount of time performing it[4]. This vital sign is extremely important since it can help to detect serious negative health effects such as heart attack, stroke and kidney failure. Inaccuracies in blood pressure measurement could cause incorrect diagnosis, which leads to errors in treatment. Decisions that are based on these readings could be so dramatic as to influence the quality of the remainder of a patient's life[4], for example the treatment of hypertension (high blood pressure) could involve the prescription of medication, the adaptation of a patient's diet and an exercise routine not normally followed by that person.

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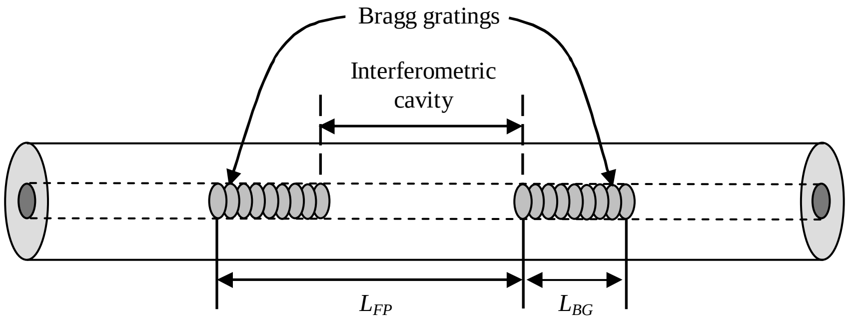

Fig. 1. Twin-grating Fabry-Perot interferometer

Blood pressure measurement is traditionally done using a mercury sphygmomanometer, which was first introduced by Riva-Rocci (1896) and later modified by Korotkoff (1905)[4]. Although it is still the standard in the field, its replacement is imminent due to a number of factors. These include the following: the toxicity of mercury, possible inaccuracies due to observer error and the fact that the variability of blood pressure tends to be ignored when using this method for single readings. The latter factor is perhaps the most prevalent, as it demands the use of a profile of blood pressure measurements taken over a period of time (usually 24 hours), which the mercury sphygmomanometer does not provide directly unless such measurements are repeated regularly. Improvements on this techniq ue have been attempted in the form of automated devices (ambulatory blood pressure measurement) but the fact remains that most of these devices have been shown to be inaccurate and have thus not been accepted into modern clinical practice[4].

Traditional sphygmomanometry makes use of occlusion of an artery with an inflatable cuff, as do more modern techniques such as oscillometry. However, inaccuracies in blood pressure readings are common when incorrect cuff sizes are used. On the other hand, devices that do not use full arterial occlusion, such as tonometry, often provide readings that are difficult to reproduce. A new blood pressure manometer needs to be at least as accurate as the present apparatus, and should overcome the potential errors associated with the current measurement techniques.

2. THEORY

A reflection spectrum of the optical fiber Bragg grating Fabry-Perot interferometer (FBGI) mentioned above comprises two component parts: the reflection spectrum of the Bragg gratings that forms the envelope of the response, and the cosinusoidal modulation due to interference between the beams reflected from the gratings[2]. Perturbations in strain ($\Delta\varepsilon$) applied along the fiber axis induce a change in both of these components without changing the envelope of the spectral response.

The shift ($\Delta\lambda_B$) in Bragg wavelength (λ_B) is given by[5]:

$$\Delta\lambda_B = \lambda_B \left(1-\rho_\alpha\right)\Delta\varepsilon \tag{1}$$

where ρ_α is the photo-elastic coefficient of the fiber. The cosinusoidal modulation will shift with the same amount if the measurand is applied uniformly across the length of the FBGI configuration. A measurand-altered reflection spectrum is described by[2]:

$$R'(\lambda) = R\left(\lambda-\Delta\lambda_B\right) = 2R_{BG}\left(\lambda-\Delta\lambda_B\right)\left[1+\cos\left(\frac{4\pi n_{eff} L_{FP}}{\lambda-\Delta\lambda_B}\right)\right] \tag{2}$$

$R_{BG}(\lambda)$ is the reflection spectrum of a single Bragg grating and L_{FP} is the distance between the two gratings, which should be longer than a few multiples of L_{BG}, the length of each of the gratings.

A demodulation algorithm for this sensor is based on the shifting property of the Fourier transform. The unstrained reflectance spectrum is taken as reference and, for each consequent reflectance spectrum obtained when axial strain is applied to the fiber, the exact phase difference can be calculated. The resolution of the FBGI can be increased by increasing its length L_{FP}. It yields absolute measurements because the shift in the Bragg wavelength is also found from the position of the envelope function.

3. PROPOSED SENSOR IMPLEMENTATION

Operation of this sensor is based on the assumption that it is possible to measure blood pressure externally by monitoring the force exerted by the blood on the wall of a patient's artery located near the skin. If a sensor placed on the surface of the skin can accurately measure such small strain artifacts, its measurements can be directly translated into pressure readings. The best physical position for measuring blood pressure non-invasively using this technique is at the wrist, because of the proximity of the radial artery to the surface of the skin.

Due to the high resolution of strain detection that can be obtained from a twin-grating FBGI, this structure was chosen as the sensing element required to create the proposed optical fiber blood pressure manometer. Such a sensor is ideally suited to measurements in the clinical environment, because it is an electrically passive device that cannot present any shock hazard to patients and medical personnel.

Figure 2 illustrates the proposed application of the optical fiber blood pressure manometer in a human patient. A non-inflatable cuff is used to secure the fiber containing the FBGI sensor on the surface of the skin at a point directly over the radial artery at the wrist.

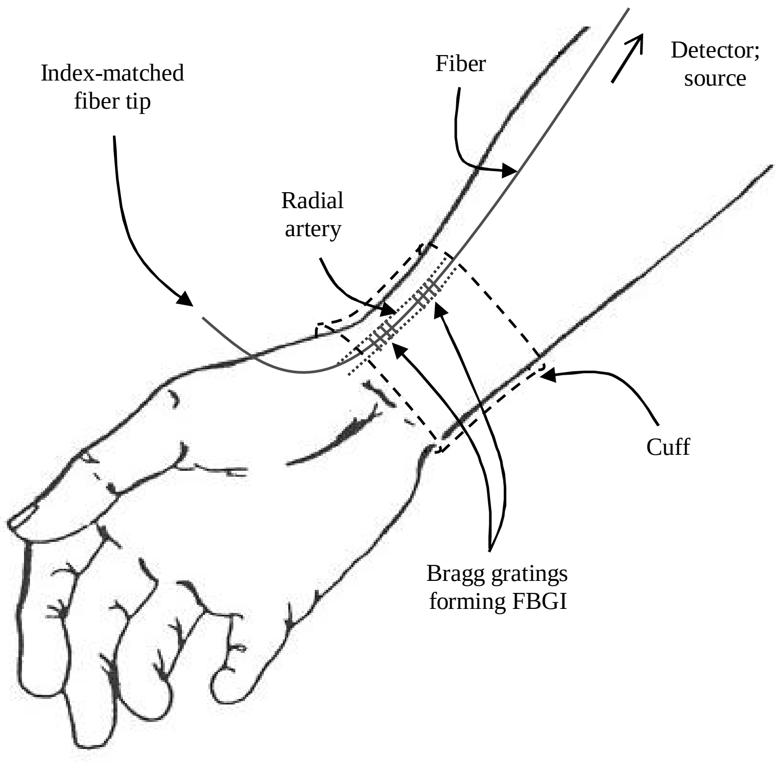

Fig. 2. Diagram illustrating proposed FBGI placement on skin surface

4. EXPERIMENTAL WORK

After writing two Bragg gratings in the desired interferometer configuration in photosensitive single-mode fiber, the sensor was characterized for its sensitivity to strain and temperature. To create the gratings, we used UV-illumination of the fiber core by a 248 nm KrF excimer laser beam through a phase mask with a period of 1.017μm.

For the proof of concept a realistic physical model (Fig. 3) was constructed to simulate pressure conditions at the actual sensor location. This was done using a stepper motor-controlled peristaltic pump to generate a pressure pulse waveform. For calibration purposes a piezo-resistive sensor measured the water pressure in the tubing. To obtain realistic strain readings, enough fluctuation in pressure was required so that an upper and lower pressure level, representing the systolic and diastolic blood pressures respectively, could be recorded. The in-flow sensor showed a gauge pressure of 139.56 mmHg at the peak pressure point, and the lowest pressure in the cycle was 9.69 mmHg below atmospheric pressure. Considering that the pressure pulse in a human artery has a systolic pressure of 120 mmHg, on average, and a diastolic pressure of approximately 80 mmHg[4], the model's pressure range was sufficient for experimentation purposes.

The twin-grating interferometric sensor was mounted externally on the wall of the pliable 1 mm thick silicone tubing used in the model. The physical properties of the tubing also served to simulate the damping of the pressure pulse by the tissue and skin that cover the radial artery.

A novel method was employed to interrogate the twin-grating optical manometer. Instead of measuring static strain values by scanning across the reflectance spectrum at each new level of 'blood' pressure, the pulse waveform was obtained in real-time once the peristaltic pump had been switched on, allowing for dynamic strain measurements to be taken. This was done by fixing the wavelength of a tunable narrowband laser diode source at a specific wavelength (typically at the point of maximum gradient of the unstrained reflectance spectrum) and then measuring the voltage output from a photodiode as a function of time.

Several values of peak pressure were generated during the continuous measurement of the pulse waveform, and this was done by merely constricting a section of the model's tubing to a certain degree with a drip-closing mechanism to maintain that pressure in the tubing.

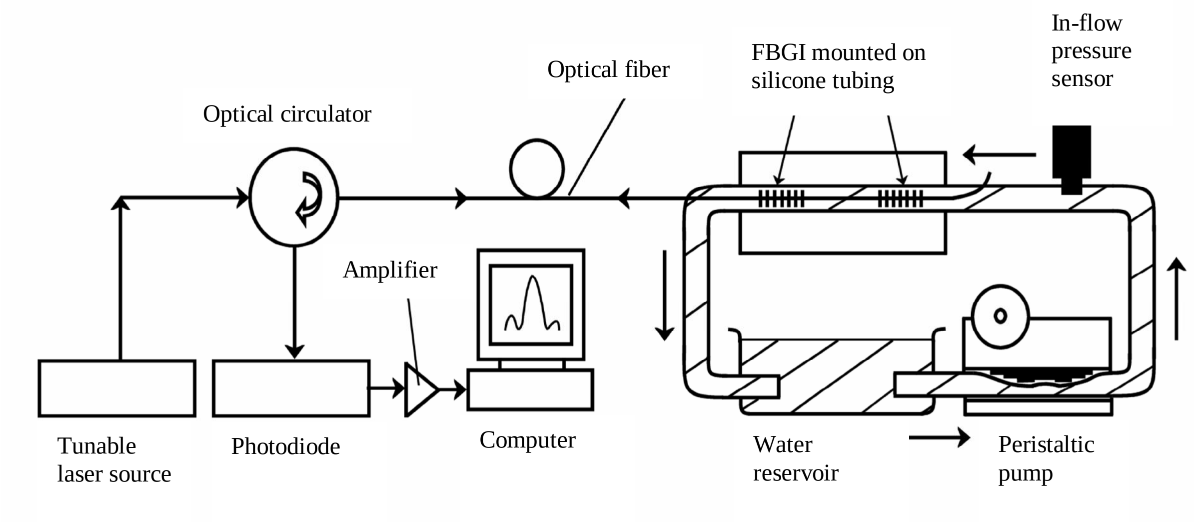

Fig. 3. Complete experimental setup for proof of concept of the optical blood pressure manometer

5. RESULTS

Strain characterization of the manufactured twin-grating FBGI was performed by applying successive axial strain values across the length of the sensor: one end of the fiber containing the gratings was mounted on a movable stage that could be controlled by stepper motors, whereas the other end of the fiber was mounted in a fixed position. Each corresponding reflectance spectrum was compared to that of the unstrained fiber that served as reference, to determine the relationship between strain applied and phase change obtained. Figure 4 displays the approximately linear relationship between phase difference and applied strain that was obtained practically.

Various experiments were conducted to investigate and evaluate the properties of the proposed optical blood pressure manometer. The phase change in the interferometer spectrum due to the small strain variations caused corresponding voltage fluctuations in the photodiode output. As the strain amplitude is small, the output voltage is directly proportional to the strain waveform. Figures 5 and 6 show recordings of real-time waveforms obtained by the twin grating interferometric sensor mounted on the model's tubing. We confirmed with the in-flow piezo-resistive pressure sensor that the peak pressure readings in these two cases were 93 mmHg and 139 mmHg, respectively. The measured pressure was increased to a maximum value of 139 mmHg by totally constricting the tubing – Fig. 6 shows the detail in the pulse waveform that was detected with the twin-grating interferometer. The second spike occurring after each main pressure peak is attributed to backlash caused by reflection of the pressure pulse from the constricted end of the tubing.

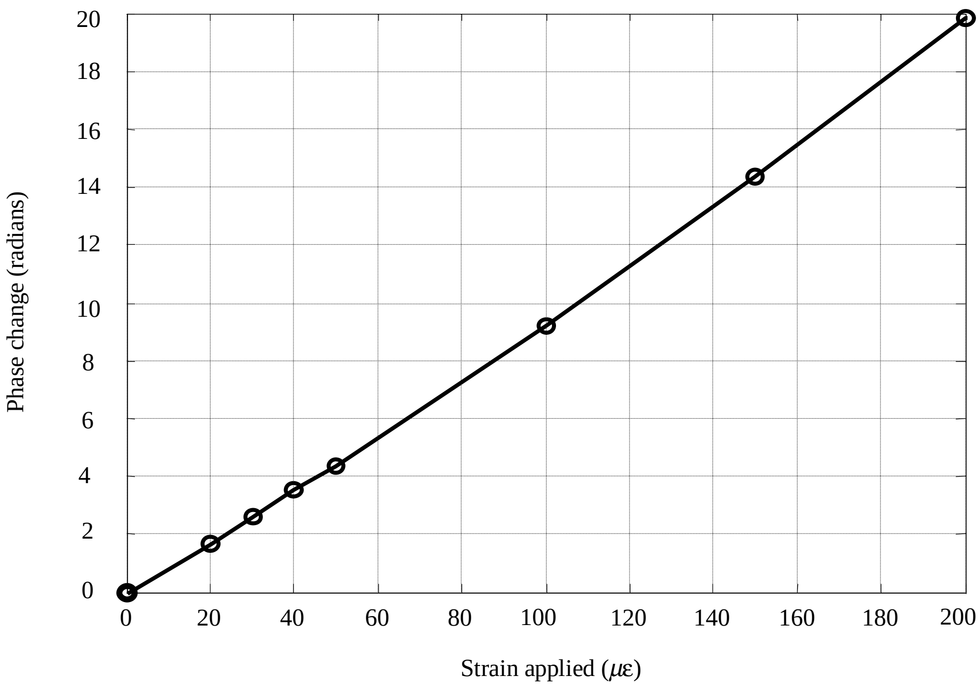

Fig. 4. Plot of phase difference versus applied strain for the twin-grating interferometer

In subsequent experimentation, we also mounted the described twin grating interferometric sensor on the wrist of a human test subject by adhering the section of fiber containing the Bragg gratings to the skin over the person's radial artery. Successful blood pressure measurements were obtained and these readings are displayed in Fig. 7. A conventional sphygmomanometer reading performed on the 23-year old patient (immediately after the fiber-based measurements were taken) showed that his systolic blood pressure was in the region of 110 mmHg, while his diastolic blood pressure was about 80 mmHg. The results in Fig. 7 therefore demonstrate that the optical manometer can indeed be implemented in the proposed configuration to record actual blood pressure profiles in real-time. An initial, accurate reading from a mercury sphygmomanometer may be needed to calibrate the output of the proposed sensor (each time a new patient's blood pressure is measured, for example). Clinical practice still relies upon the continuous measurement of blood pressure by invasive means (i.e. inserting a catheter into the artery) [4]. The optical manometer has been shown to provide truly continuous blood pressure waveforms, which may be an improvement on external techniques of continuous blood pressure measurement that have recently been developed (but have not yet been accepted for use in clinical practice).

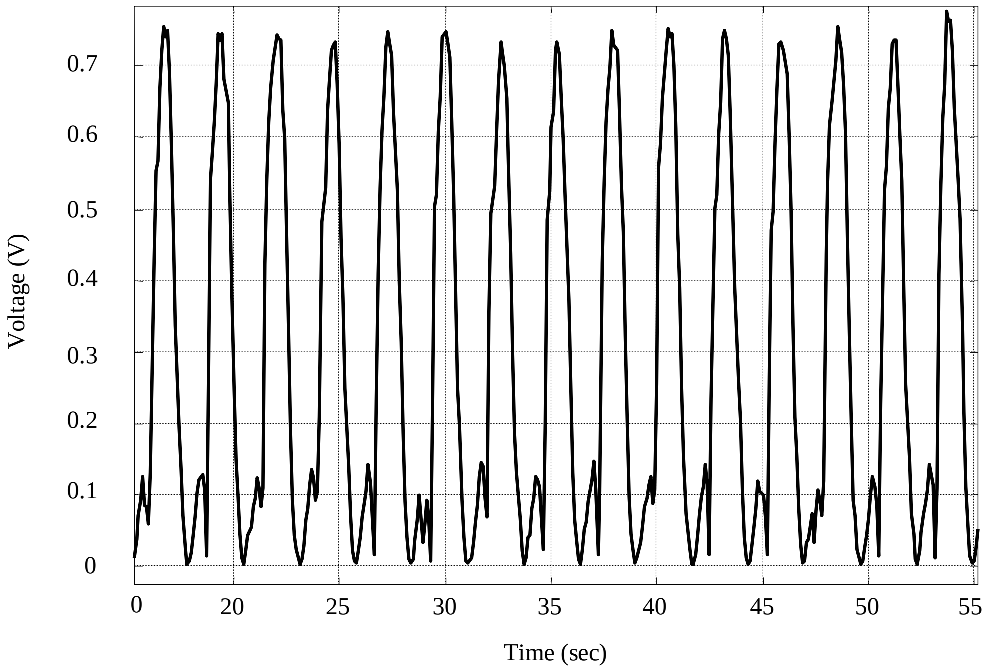

Fig. 5. Detail of pressure waveform for 93 mmHg 'systolic'

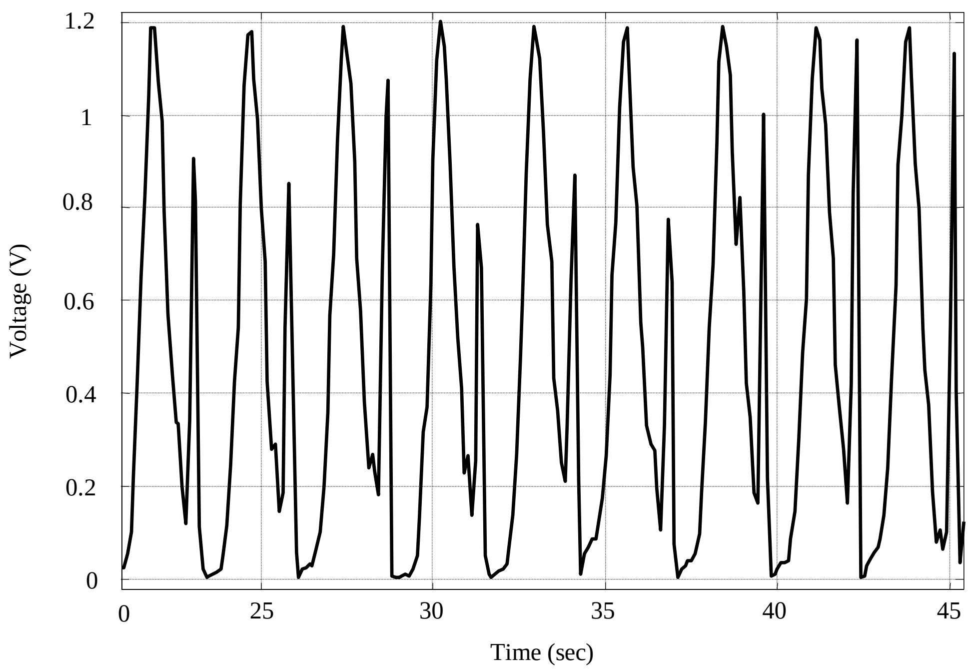

Fig. 6. Detail of pressure waveform for 139 mmHg 'systolic'

The blood pressure profile in Fig. 7 was evaluated by comparing it to that obtained with the *SphygmoCor®* Vx Pulse Wave Velocity System (a commercial device that provides continuous blood pressure waveforms based on tonometry and ECG recordings), as displayed in Fig. 8. These results were obtained from a 24-year old patient (with a measured systolic blood pressure of 114 mmHg and a diastolic blood pressure of 86 mmHg) and can therefore be used to draw a fair comparison with those in Fig. 7. It appears that a greater amount of detail is apparent in the readings taken with the optical fiber blood pressure manometer.

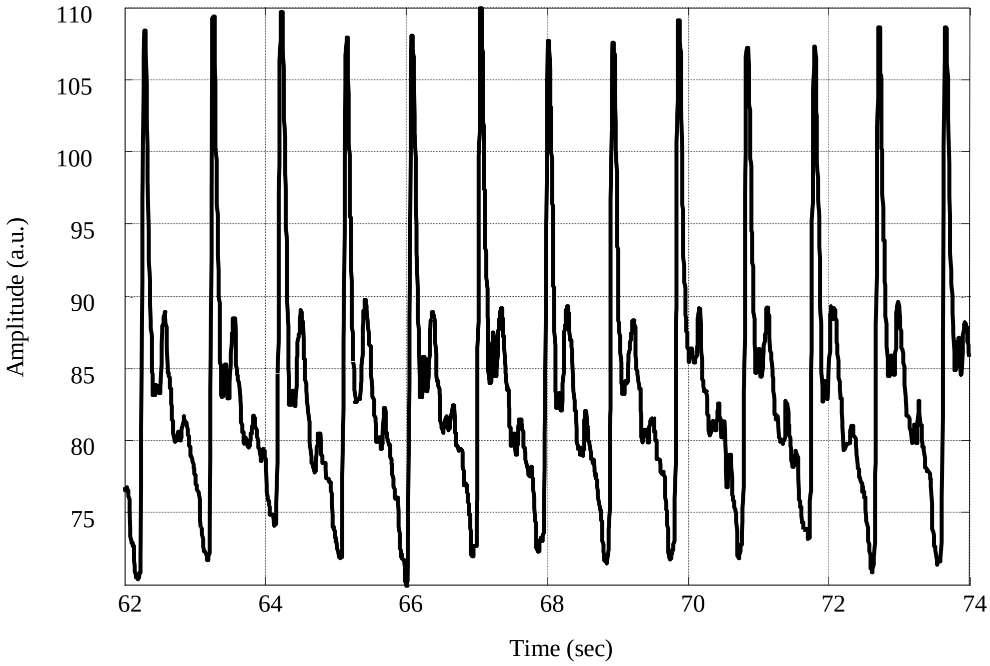

Fig. 7. Blood pressure waveform recorded in real-time from the FBGI sensor mounted on patient's skin (patient's pulse rate: 56 beats per minute)

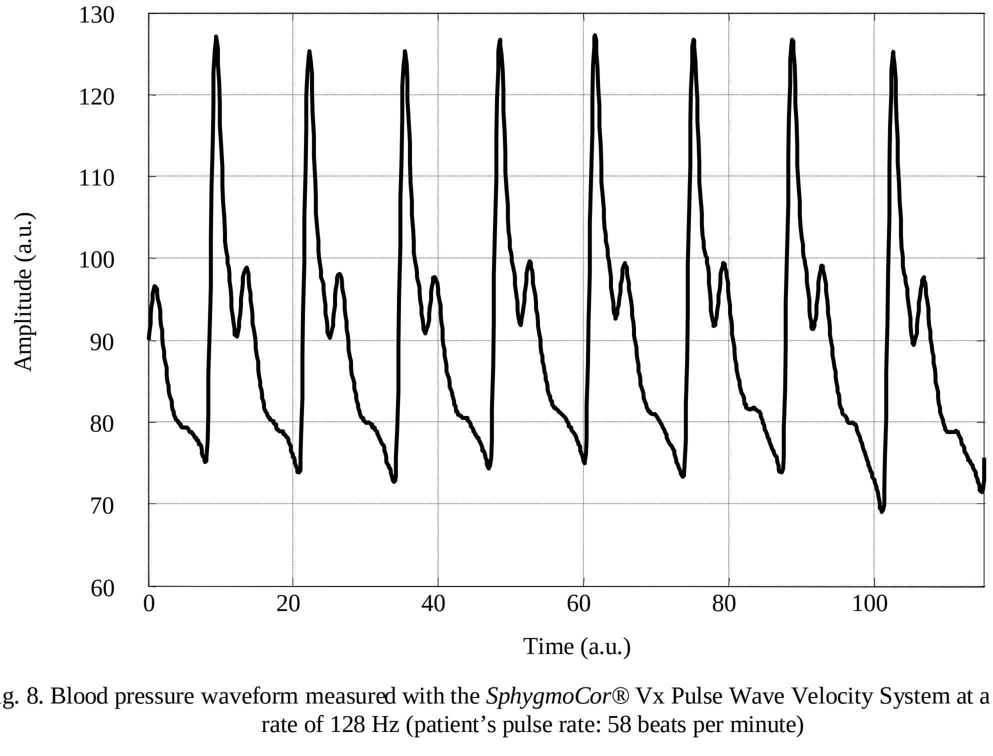

Fig. 8. Blood pressure waveform measured with the *SphygmoCor®* Vx Pulse Wave Velocity System at a sampling rate of 128 Hz (patient's pulse rate: 58 beats per minute)

6. CONCLUSION

We have demonstrated an optical blood pressure manometer that not only measures accurate systolic and diastolic blood pressures once it has been calibrated correctly, but also provides a continuous pressure waveform. This is highly beneficial to blood pressure measurement technology because the sensor described here takes external measurements, and yet it shows the potential to provide data that is currently only acceptable in clinical practice when obtained from invasive blood pressure measurements.

Successful temperature compensation techniques for use in the proposed optical blood pressure manometer are currently under investigation. For example, temperature sensitivity of the twin-grating sensor can be reduced sufficiently by adding a temperature feedback loop to control the wavelength of the narrowband tunable laser. If fluctuations in temperature do occur, this wavelength will be shifted to compensate for these changes and thereby maintain an output voltage level due to strain only.

Further research towards a commercial optical blood pressure manometer includes more rigorous trials on human patients to determine the clinical accuracy of this sensor. Due to the continuous nature of measurements performed with the proposed sensor, successful implementation may lead to this optical manometer being considered as an alternative to conventional sphygmomanometry in certain situations.

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