Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Wearable, Cuff-less PPG-Based Blood Pressure Monitor with Novel Height Sensor

Phillip A. Shaltis*, Andrew Reisner, and H. Harry Asada

Abstract— **A truly wearable non-invasive blood pressure (NIBP) sensor -- light-weight, compact, unobstrusive, and essentially unnoticeable to the patient -- could revolutionize healthcare delivered beyond the traditional walls of medical facilities, offering new ways to care for patients in their everyday surroundings. This paper presents results from our work towards the development of a self-contained, wearable blood pressure sensor. A PPG-based approach to blood pressure monitoring is presented. The design enables significant miniaturization of traditional oscillometric devices without the need for occlusive circumferential pressures. It will be shown how natural raising and lowering of the arm replaces the need for bulky actuators. Additionally, a dual-accelerometer height sensor that is tetherless is proposed and supported by experimental results.**

I. Introduction

Monitoring beat-to-beat arterial blood pressure (BP) with a sensor virtually imperceptible to the wearer, for continuous periods of weeks, months, or years, could prove revolutionary in the diagnosis and treatment of chronic hypertension and heart failure, as well as a monitoring tool for convalescing individuals, and individuals in hazardous duty (such as firefighters or soldiers). Indeed, intensive 24-hour BP monitoring, using a conventional oscillometric cuff has been shown to offer superior prognostic data and end-points for treatment, for chronic hypertensive subjects [1]. Yet no existing BP sensor modality offers all of the features of an ideal wearable sensor: non-invasive, light-weight, low-powered, unobstrusive, motion tolerant, and trivial to place.

Most of the 24-hour ambulatory BP monitoring studies used cuff-based devices which are based on the oscillometric method. The oscillometric method uses the relationship of the external pressure and the magnitude of the arterial volume pulsations to estimate arterial pressure [2]. For oscillometry, it is therefore essential that an air-filled, compliant bladder be the source of the external pressure, so that the arteries' pulsations are not mechanically constrained. Traditional oscillometry is far from ideal for long-term monitoring: it is motion intolerant, requiring the user keeps the monitored arm immobile whenever the cuff inflates for a measurement. Second, the circumferential squeezing of the arm is often uncomfortable.

In this study, we describe the development and preliminary validation of a miniaturized, noninvasive BP sensor which utilizes photoplethysmography (PPG). PPG is an optic signal related to the volumetric pulsations of blood in tissue [3], which in turn is related to arterial pressure pulsations. Features of the PPG signal have been correlated with important hemodynamic phenomena, including arterial blood pressure, however, because there are several technical hurdles which complicate this relationship, previous attempts at utilizing this modality have yielded only modest results [4]. Our work combines the salient features of oscillometric sensors and tonometric sensors into a simple, compact device which can be worn at the fingerbase. Validation studies conducted within our lab have demonstrated the potential of this method for reliable measurement of the patient's mean arterial pressure (MAP).

To implement the above PPG-based MAP measurement, a few technical issues must be addressed and a practical design concept must be created. First, the MAP measurement requires an effective method for providing relative arterial-based volumetric changes. Not only must the sensor be compact and power efficient, but also it has to be attached to the skin stably and comfortably without requiring a large pressure. We propose a PPG-based sensor design that guarantees stable, reliable measurements. Second, this method necessitates a height sensor for measuring the hydrostatic pressure offset of the PPG relative to the heart. Such a height sensor must also be wearable, i.e. compact, lightweight, and of low power consumption. A novel height sensor using MEMS accelerometers will be presented which meets these criteria. The application of these design solutions is manifested in a miniaturized, wearable blood pressure sensor.

II. Methods

A. PPG Calibration Using Modified Oscillometry

We have developed a PPG-based blood pressure sensor. It has been suggested that the PPG-based, volumetric changes are related to the patient's ABP through a nonlinear vascular compliance relationship [5]. Let P_{tm} be the transmural pressure, that is, the pressure difference across a vascular wall:

Manuscript received April 3, 2006. This work was supported in part by the Sharp Corporation.

P. A. Shaltis* is with the Department of Mechanical Engineering at the Massachusetts Institute of Technology, Cambridge, MA, 02139 USA (email: pshaltis@mit.edu).

A. Reisner* is with the Massachusetts General Hospital Boston, MA, 02114 USA (email: areisner@partners.org).

H. H. Asada is with the Department of Mechanical Engineering at the Massachusetts Institute of Technology, Cambridge, MA, 02139 USA (email: asada@mit.edu).

1-4244-0033-3/06/$20.00 ©2006 IEEE.

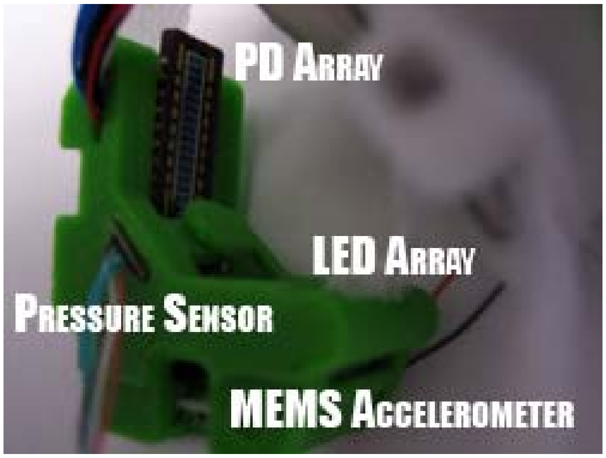

Fig. 1. Picture of the complete ring sensor monitoring unit.

$$P_{tm} = P_{MAP} - \rho \cdot g \cdot h - P_{cuff} \quad (1)$$

with P_{MAP} as the mean arterial blood pressure, $\rho \cdot g \cdot h$ as the hydrostatic pressure offset occurring when the pressure measurement site is not located at the same height as the heart, and P_{cuff} as the cuff pressure applied to the outside of the blood vessel from an external pressure source. For traditional oscillometric sensors the cuff pressure is modified to locate the zero transmural pressure point where the vascular wall becomes most compliant, i.e. the slope of the volume-transmural pressure curve is the steepest (hydrostatic offset pressures are neglected by requesting the measurement occur at the same height as the patient's heart). The largest amplitude pulsation transmitted through the cuff therefore indicates when the known cuff pressure is equal to the internal MAP.

Analogously, the volumetric change of the blood vessel can be detected with PPG (PD Array: Hamamatsu S8558, LED Array: Elekon ELM-3003). At the zero transmural pressure point, the amplitude of the PPG signal becomes a maximum. For long-term monitoring considerations, the cuff pressure must be kept below 75 mmHg to maintain normal blood flow. Therefore, we propose two solutions to accommodate this requirement as follows: instead of circumferentially applied pressure, our device applies a pressure to the tissue directly overlying an artery at pressures significantly lower than the above threshold (measured by an Entran EPL-D02-10P/Z1); additionally, instead of modifying P_{tm} by applying a range of large cuff pressures, the transmural pressure is modified by requesting that the subject simply raise and lower their arm, effectively altering the hydrostatic pressure and consequently the transmural pressure.

Based on the above formula, we proposed the following ABP measurement method:

1. While the cuff pressure is set to a proper, the arm with the PPG sensor is raised so that the reference pressure ($P_r = \rho \cdot g \cdot h + P_{cuff}$) varies in the vicinity of the mean arterial pressure (MAP).
2. The amplitude of the PPG signal is examined to find the zero transmural pressure point, where:

$$P_{MAP} \cong P_r = \rho \cdot g \cdot h + P_{cuff} \quad (2)$$

More specifically, the protocol utilized for our PPG calibration is as follows: The instrumented wearer's hand is raised to a comfortable height above the heart; typically this is between 40 and 60 cm. After an initial rest period of two minutes, the arm height is decreased from the highest position above to the heart to the lowest position below the heart in decrements of 10 cm. Data are collected at each arm position for a period of 20 seconds before offline post-processing [6].

B. Tetherless Height Sensor Using Accelerometers

Clearly, any unsupervised blood pressure monitor, including the proposed calibration method, requires reliable measurement of the subject's arm height (relative to the heart). Most existing options include either fluid-filled tubing or video motion tracking, which can be cumbersome or impractical. A tetherless, less obtrusive solution would clearly be preferable for most patients. To fulfill this need we have implemented a dual MEMS accelerometer approach to height measurement. The influence of the angle of gravity on the respective outputs of the two sensors is the key to the height sensor measurement. One accelerometer is attached to the upper arm with the axis of the accelerometer aligned to the longitudinal direction of the upper arm, ACC_1 in the figure. The other accelerometer is contained within the PPG sensor at the finger base, where either the x-axis or y-axis of the accelerometer is aligned with the longitudinal direction of the forearm, ACC_2 in the figure.

Let ℓ_1 and ℓ_2 be the lengths of the upper arm and the forearm, respectively, and ℓ_0 be the height of the shoulder joint relative to the heart (Figure 2). From the figure the height of the heart relative to the PPG sensor is given by either (3) or (4) depending on the accelerometer axis utilized.

$$h = \ell_1 \cdot \cos\theta_1 + \ell_2 \cdot \cos\theta_2 - \ell_0 \quad (3)$$

$$h = \ell_1 \cdot \sin\theta_1 + \ell_2 \cdot \sin\theta_2 - \ell_0 \quad (4)$$

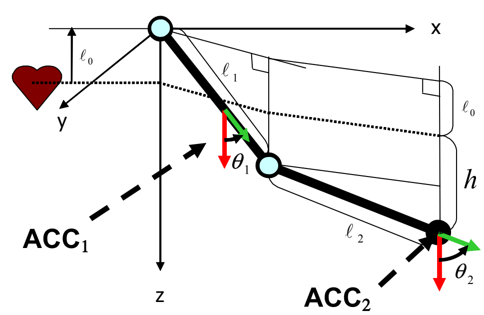

Fig. 2. Definition of height measurement terms in three dimensions.

We note that arm lengths may be different from person to person depending on body height as well as on the location of the sensor. However, given the patient's body height, ℓ_1 and ℓ_2 can be approximated to average lengths based on standard body proportions. It should be noted that when the patient is lying, the third term in the above equation, i.e. ℓ_0, must be deleted since the shoulder would be at the same hydrostatic height as the heart.

C. Post-Processing and MAP Measurement Algorithm

PPG amplitude measurements were conditioned as follows: First, the sampled raw PPG data were passed through a 2nd order high-pass digital Butterworth filter (Sampling Rate: 200 Hz, F_c = 0.7 Hz) to remove the non-pulsatile DC component of the waveform. The filtered waveforms were next given to a peak detection algorithm, such that the beat-to-beat global maximum and minimum values of each PPG waveform could be separated.

Once each of the beat-to-beat amplitudes were separated, spurious data points were removed by first averaging the amplitudes over five consecutive beats and by then further passing the averaged data points through a moving window 3-element median filter ($\bar{y}_{AC\_PPG}$). The subject's mean arterial blood pressure can then be determined by combining the pressure data from the recorded height ($\rho \cdot g \cdot h$), the cuff pressure data (P_{cuff}) and the conditioned PPG amplitude data ($\bar{y}_{AC\_PPG}$). The measured cuff pressure and the hydrostatic pressure measurements are simultaneously averaged over the duration of each detected beat and are then summed together to provide a single reference pressure measurement corresponding to each conditioned amplitude measurement.

By noting the reference pressure which produces a maximum PPG amplitude, the underlying mean arterial pressure can be determined as shown in equations (5) and (6).

$$h_{zero-Ptm} = \arg \underset{h}{Max} \left[\bar{y}_{AC\_PPG} \right] \tag{5}$$

$$P_{MAP} \cong \rho \cdot g \cdot h_{zero-Ptm} + P_{cuff} \tag{6}$$

Since the hydrostatic challenge yields a set of discrete reference pressures and PPG amplitudes it is necessary to fit the data to a continuous non-linear model to better estimate the true maximum amplitude. Although there are many non-linear functions capable of representing the expected bell-shaped compliance curve, we have selected a standard Gaussian model for our analysis.

$$\hat{\bar{y}}_{AC\_PPG} = \theta_1 e \left[-\left(\frac{P_r - \theta_2}{\theta_3} \right)^2 \right] \tag{7}$$

where $\hat{\bar{y}}_{AC\_PPG}$ is the estimated AC amplitude of the conditioned PPG, P_r is a continuous set of reference pressures, which also contains the reference pressure values used to generate the model, and θ_i are the fitting coefficents for the model. The fitting method utilized for the data consisted of an offline non-linear least squares approach utilizing a traditional Levenberg-Marquardt algorithm [7]. Data greater than 50% of the maximum measured amplitude were utilized for the non-linear fit. Each of the results was compared to the MAP provided by a simultaneous Finapres (Ohmeda 2300, Finapres BP Monitor) measurement.

III. Results

A. PPG-based MAP Measurements

Following the experimental procedure outlined previously, preliminary calibration data were collected for 4 healthy volunteers (2 male, 2 female, average age: 29 ± 10 years). All data were collected in accordance with an experimental protocol approved by the Massachusetts Institute of Technology's Committee on the Use of Humans as Experimental Subjects (COUHES Approval No. 0403000233) and following Federal regulations for the protection of human subjects established by 45 CFR 46.

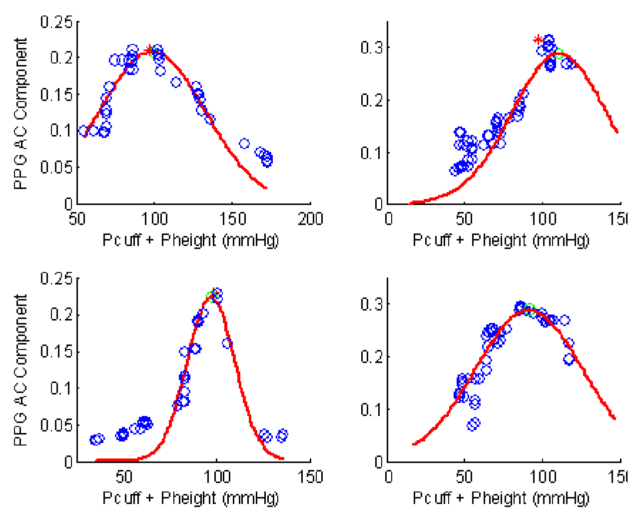

Fig. 3. Comparison of the compliance curves with Guassian fit models generated by the hydrostatic challenge.

The performance of procedure for estimating the patients' mean arterial pressures was compared using a standard Bland-Altman comparison. The Bland-Altman statistics (limits of agreement represent the 95% confidence interval) are provided in Table 1.

Table 1. Bland-Altman 95% limits of agreement and the percent error between the Finapres and the proposed MAP estimation methods.

	Bland-Altman
Finapres – Raw Data	[6.91, -9.04]
Finapres – Gaussian Fit	[10.57, -17.79]

B. Tetherless Accelerometer-based Height Sensor

The accelerometer-based method for arm height estimation was found to have an excellent agreement with actual arm heights, as measured manually with a ruler.

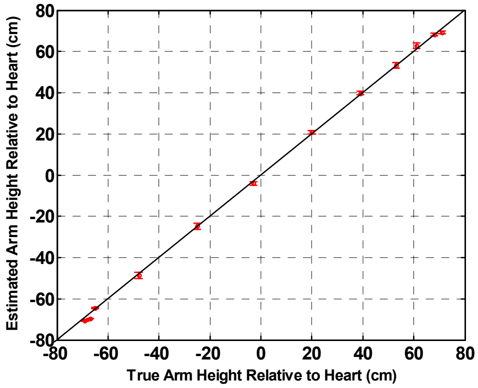

Fig. 4. Comparison of the estimated and actual measured arm heights relative to the patient's heart. (error bars denote 95% confidence).

IV. Discussion

A. Hydrostatic Oscillometry

We have proposed a hydrostatic oscillometric approach towards wearable BP monitoring. As was previously noted, traditional oscillometry requires the use of a high pressure occlusive cuff for non-invasive blood pressure (NIBP) measurements. Actively utilizing occlusive pressures has two main draw backs for wearable devices; large applied pressures can be both uncomfortable and may cause damage to the local tissue if sustained for long periods of time. Also, for non-invasive sensors where path length changes significantly affect sensor performance (such as for PPG sensors), large changes in cuff pressure are unacceptable and can lead to disconnects between the measured PPG signal and the underlying MAP. Our proposed solution ensures that the locally applied pressure may be kept at a lower, constant level without large power consuming actuators. We believe this will permit normal blood flow and will be less intrusive to the wearer. In fact, the natural arm motions of the wearer may be used to continuously re-calibrate the sensor. Additionally, the hydrostatic challenge does not disturb the PPG light path, since it affects the reference pressure from inside of the blood vessel. Yet, it creates the same effect as the external cuff pressure.

There are tradeoffs to this approach, though. The calibration is limited by the arm length of the subject and consequently may require at least temporarily higher cuff pressures for certain subjects to properly identify the calibration curve. It should also be noted that there are limitations to the rate at which a hydrostatic calibration may be performed. Previous studies conducted in our lab indicate that a rest period of at least two minutes prior to calibration reduces the effects of rapid hydrostatic changes [6]. Additionally, since the measurements are provided by a peripheral PPG sensor we must acknowledge that there may be particular monitoring environments where the single measurements may become unreliable, such as during changes in temperature or changes in mental state. Further investigations are clearly required to demonstrate both the potentials and limitations for this approach under less-controlled monitoring conditions.

B. Accelerometer-based Height Sensor

The current embodiment of the height sensor consists of two dual-axis MEMS type accelerometers. The dual axis sensors have demonstrated robust performance for arm motions occurring within a plane. However, the calibration breaks down for changes in arm/hand orientation. To overcome this limitation, it seems natural that adding data from a third axis will further extend the utility of the height sensor for any arm/hand orientation.

A second consideration associated with the height sensor is that the performance of the height sensor is limited to low acceleration monitoring conditions. Large accelerations, accelerations several times larger than gravity, would easily overwhelm the measured gravity influence and consequently limit the accuracy of any corresponding height change measurements. For this reason, we recommend that calibrations be performed following the prescribed time interval guidelines.

V. Conclusion

This paper presents significant progress towards the development of a self-contained, unobtrusive, miniaturizable blood pressure sensor. A modified, hydrostatic-based oscillometric method has been proposed. The method enables significant device miniaturization without the need of a high pressure, actuated cuff. To accommodate the need for reliable height measurements, we have demonstrated the utility of a tetherless, MEMS accelerometer height sensor. The sensor is compact, unobtrusive, and require less than 5V of power for operation.

References

[1] Stewart, R.B. and G.J. Caranasos, Medication compliance in the elderly. Med Clin North Am., 73(6): pp. 1551-63, 1989.

[2] Ando, J., et al, "Pressure-volume relationships of finger arteries in healthy subjects and patients with coronary atherosclerosis measured non-invasively by photoelectric plethysmoraphy," Jpn Circ J, 55(6): pp 567-575, 1991.

[3] Millasseau, S.C., et al, "Noninvasive assessment of the digital volume pulse. Comparison with the peripheral pressure pulse," Hypertension, 36(6): pp. 952-956, 2000.

[4] Awad, A.A., et al., "How does the plethysmogram derived from the pulse oximeter relate to arterial blood pressure in coronary artery bypass graft patients?," Anesth Analg, 93(6): pp. 1466-1471, 2001.

[5] Langewouters, G.J., Zwart, A., Busse, R., and Wesseling, K.H., "Pressure-diameter relationships of sements of human finger arteries," Clin. Phys. Physiol. Meas., 7(1): pp. 43-55, 1986.

[6] Shaltis, P., Reisner, A., Asada, H., "Calibration of the Photoplethysmogram to Arterial Blood Pressure: Capabilities and Limitations for Continuous Pressure Monitoring," 2005 27th Annual International Conference of the IEEE/EMBS, Shanghai, China, Sept. 1-4, 2005.

[7] Marquardt, D.W., "An Algorithm for the Least-Squares Estimation of Nonlinear Parameters." SIAM Journal of Applied Mathematics, 11(2): 431-441, June 1963.C. J. Kaufman, Rocky Mountain Research Lab., Boulder, CO, private communication, May 1995.