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Development of a Palpable Carotid Pulse Pressure Sensor Using Electromagnetic Induction

Harutoyo Hirano, Tomohiro Fukuchi,
Graduate school of Engineering
Hiroshima University
Higashi-Hiroshima, Hiroshima, Japan
{harutoyo, fukuchi}@bsys.hiroshima-u.ac.jp

Yuichi Kurita,
Faculty of Engineering
Hiroshima University
Higashi-Hiroshima, Hiroshima, Japan
kurita@bsys.hiroshima-u.ac.jp

Akihiko Kandori, Yuko Sano,
Central Research Laboratory
Hitachi Ltd.
Kokubunji, Tokyo, Japan
{akihiko.kandori.vc, yuko.sano.hd}
@hitachi.com

Ryuji Nakamura, Noboru Saeki,
Masashi Kawamoto, Masao Yoshizumi,
Institute of Biomedical and Health Sciences
Hiroshima University
Minami-ku, Hiroshima, Hiroshima, Japan
{r-nacamura, nsaeki, anekawa}@hiroshima-u.ac.jp
yoshizum-tky@umin.ac.jp

and Toshio Tsuji
Faculty of Engineering
Hiroshima University
Higashi-Hiroshima, Hiroshima, Japan
tsuji@bsys.hiroshima-u.ac.jp

***Abstract* - This paper proposes a novel non-invasive palpable sensor for measuring carotid pulse pressure. The unit consists of a sensing plastic chip, a pair of coil printed circuit boards, a pair of springs attached between the circuit board and the plastic chip. The distance between the boards is monitored from the displacement of the springs, and the information is converted into a voltage signal based on electromagnetic induction. In this study, the optimal forces externally applied to the proposed sensor were first examined to allow accurate measurement of carotid pulse wave amplitude variations, and it was found that the force applied when the measured maximum amplitudes of the sensor were obtained yielded the best performance. Next, carotid pulse waves were measured using the sensor with these optimal forces, and the results were compared with carotid pulse pressure values measured using a commercial pulse wave transducer. The coefficients of correlation between the two were 0.9 or more. It was therefore concluded that the proposed sensor enables non-invasive measurement of carotid pulse waves.**

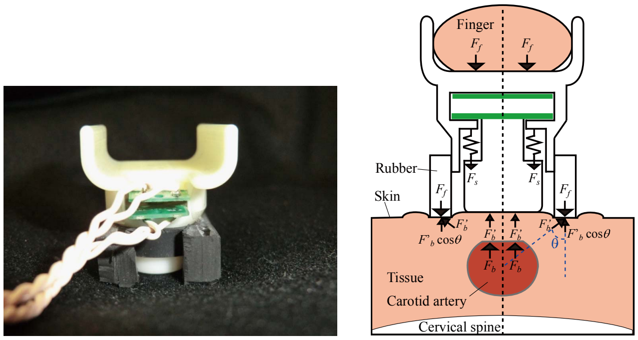
Index Terms - Carotid pulse pressure, Palpation, Non-invasive measurement, Magnetic sensor, Electromagnetic induction

I. Introduction

The carotid artery is an elastic vessel between the cephalon and the cervical part for channeling blood to the brain. Its pulse pressure waves reflect the function and morphology of the ventriculus sinister, the aortal valve and the aorta [1]. Thus, pulse waves measured from the carotid artery provide doctors with important information for the diagnosis of cardiac function. The carotid artery is also the nearest palpable artery to the heart, and its location directly below the skin allows qualitative checking of vital signs by medical staff at emergency scenes and clarification of the amount of blood flowing to the patient's brain [1]. Accordingly, the development of a system that enables immediate determination of pulse pressure from the carotid artery is expected to allow the collection of quantitative data to support more efficient treatment at a hospital.

Several non-invasive methods of measuring carotid arterial pressure have been proposed [2] – [5]. Ishii *et al*. [2] used an electrostatic transducer to measure changes in air pressure based on the pulse pressure of the carotid artery. The results showed a strong correlation between waves measured using the device and those measured with an invasive sensor [2]. However, the device can only detect the changes in air pressure, not in blood pressure, and is also affected by body motions. In other studies, measurement sensors based on the tonometry approach (e.g., an array-type sensor incorporating many micro-pressure sensors [3] and a pen-type transducer [4], [5]) were developed. However, it takes time to set up carotid pressure measurement using these methods, so that these sensors cannot be effective at emergency scenes. Similarly, it is difficult to attach commercial pen-type transducers (e.g., SPT-301, Millar Instruments) in the normal direction of the vascular diameter and to measure carotid pulse pressure stably and accurately. Moreover, in order to use such sensors for measurement during palpation, they need to transmit the pulse pressure of the patient to the measurer. However, no sensor that can be palpated by medical staff during pulse pressure measurement from the carotid artery has yet been developed.

This paper describes the development of a carotid artery pulse pressure sensor that can be palpated during measurement. Specifically, we propose a sensor based on the following criteria: (I) the measurer must be able to palate the carotid pulse pressure using the sensor; (II) the sensor must be small and easy to attach/detach; (III) the position of the sensor during pulse pressure measurement must be stable; and (IV) the sensor must be able to measure continuous pulse pressure and have a high signal-to-noise ratio.



(a) (b)

Fig. 1 The proposed sensor: (a) the sensor unit; (b) diagram showing its structure and the equilibrium of forces between biological tissues and the sensor

II. Method

Figure 1 (a) shows the structure of the proposed sensor. The system unit consists of a DC power source, an inverter circuit, a converter circuit, a pair of coil printed circuit boards and a sensing plastic chip. A pair of high-friction rubber pieces is attached to the sensor. The sensor has dimensions of 24 mm (W) × 30 mm (D) × 27 mm (H) and weighs 10.6 g.

The method of measuring carotid pulse pressure with the proposed sensor is based on the tonometry method [3] – [7]. An arterial vessel is pressed until its surface is flattened for attachment of the sensing plastic chip. This technique is unaffected by the stiffness and tension of the arterial vessel, but accurate measurement of arterial pressure requires the presence of stiff matter (e.g., bone) behind the artery. In light of the cervical spine's location relatively near the carotid artery, arterial pressure can be measured using the tonometry method [3] – [5].

Figure 1 (b) shows the equilibrium of forces between biological tissues and the proposed sensor in the measurement of carotid pulse pressure. Here, F_b is the force in the normal direction based on intravascular pressure, F'_b is the force applied to the contact surface between the high-friction rubber piece and the skin caused by the intravascular pressure, F_f is the normal force applied from the high-friction rubber piece toward the skin surface, θ is the angle between F'_b and F_f. The forces applied to the proposed sensor are assumed to be bilaterally symmetric. According to Fig. 1, when noise components such as body motions are input into the proposed sensor, both the sensing plastic chip and the high-friction rubber pieces are displaced. Accordingly, the intravascular pressure can be measured without any effect of these components. It should be noted that the intravascular pressure component F'_b is transmitted to the measurer's finger through the proposed sensor because F'_b is applied to the high-friction rubber piece. The measurer can thus palpate the carotid pulse pressure using the sensor.

The force F_S acting on the spring in the proposed sensor is equal to the force obtained by subtracting the normal force applied to the high-friction rubber piece ($F'_b \cos\theta$) from the normal force applied to the sensing plastic chip (F'_b). Thus, F_s is expressed as

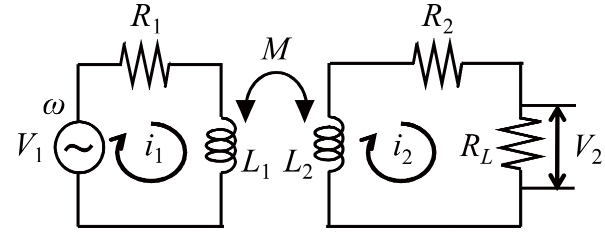


Fig. 2 Equivalent circuit of the proposed sensor

$$F'_b(t) - F'_b(t)\cos\theta = F_s \quad . \tag{1}$$

F'_b is therefore expressed as

$$F'_b(t) = \alpha k \Delta x \quad , \tag{2}$$

where k is the spring constant of the proposed sensor, Δx is the displacement from the spring's natural length, and $\alpha = 1 / (1-\cos\theta)$.

Figure 2 shows the equivalent circuit of the proposed sensor. A direct current supplied by a DC power source or a battery is changed to an alternating current of 20 kHz using an inverter circuit. The AC power is then transmitted using electromagnetic induction from the primary coil to the secondary coil, and converted back to DC power using a rectifier and a smoothing circuit. The DC voltage is then amplified using a post amplifier and passed to the phase shift detection circuit before being regulated using a 10-Hz first-order low-pass filter. The circuit equation for Fig. 2 is therefore expressed as

$$\begin{pmatrix} R_1 + j\omega L_1 & -j\omega M \\ -j\omega M & j\omega L_2 + R_2 + R_L \end{pmatrix} \begin{pmatrix} i_1 \\ i_2 \end{pmatrix} = \begin{pmatrix} V_1 \\ 0 \end{pmatrix} , \tag{3}$$

where j is an imaginary unit, ω is the angular frequency, R_1 and R_2 are the combined resistances of the primary and secondary circuits, L_1 and L_2 are the self-inductances of the coils, M is the mutual inductance between the coils, V_1 is the source voltage, and i_1 and i_2 are the currents of the circuits. In this study, the same flattened types were used for the primary and secondary coils, the combined resistance R_2 was negligible compared to the output resistance R_L, and the mutual inductance M depended on the distance between the coils ($d - \Delta x$) according to Neumann's formula [8]. The output voltage $|V_2|$ is therefore expressed as

$$|V_2| \approx |i_2| R_L = \left| \frac{j\omega V_1}{R_1 + j\omega L_1} M(\Delta x) \right| . \tag{4}$$

Additionally, the pressure P applied to the sensing plastic chip is expressed as

$$P = \frac{k\Delta x}{S} \quad , \tag{5}$$

where S is the area of the contact surface of the sensing plastic chip. The output voltage $|V_2|$ is therefore expressed by the following equation using (4) and (5):

$$|V_2| = \left| \frac{1}{\alpha + j\beta} M(P) \right|, \tag{6}$$

where

$$\alpha = \frac{L_1}{V_1},$$

$$\beta = -\frac{R_1}{\omega V_1},$$

$$M(P) = \frac{\mu}{4\pi} \sum_{n=1}^{N} \sum_{n=1}^{N} \int_0^{2\pi} \int_0^{2\pi} \frac{\left(a + \phi(n-1)\right)^2 \cos(\theta_1 - \theta_2)}{\sqrt{2\left(a + \phi(n-1)\right)^2 \cos(\theta_1 - \theta_2) + \left(d - \frac{S}{k} P\right)^2}} d\theta_1 d\theta_2 .$$

Consequently, the output voltage $|V_2|$ corresponding to the pressure P applied to the proposed sensor is obtained.

III. Experiment

Non-invasive measurement of carotid pulse pressure using the proposed sensor was performed. First, to explore the optimal force F_f applied to the sensor, carotid pulse pressure waves recorded using the proposed sensor were compared with radial artery pulse waves simultaneously recorded using the conventional arterial tonometry method. Waves from the left carotid artery were measured using the proposed sensor, and blood pressure and electrocardiogram signals were also measured from the left radial artery and the chest using a physiological monitor (BP-608, Omron Colin). All signals were simultaneously measured at 1,000 Hz using an analog-digital card (CSI-360116, Interface). The five healthy male subjects (ages ± S.D.: 22.2 ± 0.4) used for the study were seated in a resting state, and signals were measured over a period of one minute. F_f was applied to the proposed sensor by the measurer's finger toward the vascular radius. Optimal force F_f was decided from the amplitude of the carotid pulse pressure waves measured using the proposed sensor and from the coefficient of correlation between 30 waves measured using the proposed sensor and the physiological monitor.

Next, to verify the measurement accuracy of carotid pulse pressure waves measured by the proposed sensor with the optimal force F_f, waves recorded with the proposed sensor were compared with others recorded using a commercial pulse wave transducer (TK-701T, Nihon Kohden). The carotid pulse pressure waves were first measured with the proposed sensor, and then again using the pulse wave transducer from the same part of the subject's left carotid artery. All signals were measured at 1,000 Hz using an analog-digital card via a dialysate connector (AK-605H, Nihon Kohden). Both sets of measured waves were then passed through a 10-Hz second-order infinite impulse response (IIR) low-pass filter and a 0.3-Hz first-order IIR high-pass filter. The filtered waves were then resampled using the electrocardiogram R-wave timing, and the coefficient of correlation between both sets of waves was calculated. The signal-to-noise ratio was also calculated for the waves measured using the proposed sensor. Informed consent was obtained from all study subjects before the experiments were performed based on the Declaration of Helsinki.

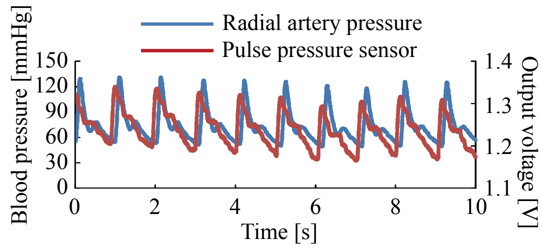


Fig. 3 Example of waves measured using the proposed sensor and those measured using a tonometry sphygmomanometer

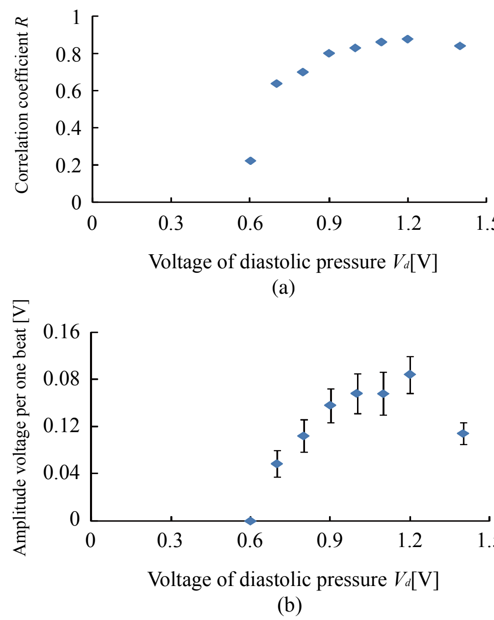


Fig. 4 Variation of parameters with changes in diastolic voltage V_d: (a) coefficients of correlation between the pulse waves measured using the proposed sensor and those measured using a tonometry sphygmomanometer; (b) amplitude voltage per beat measured using the proposed sensor

IV. Results

Figure 3 shows examples of waves measured using the proposed sensor and the pulse pressure waves of the radial artery. Both are from Subject A, and the waves measured using the proposed sensor were recorded when the force F_f was applied to the sensor so that the output voltage corresponding to diastolic pressure V_d was 1.2 V. Figure 4 (a) shows the calculated coefficients of correlation between the pulse waves measured using the proposed sensor and those measured using a tonometry sphygmomanometer, and Fig. 4 (b) shows the results of amplitude voltage per beat measured using the proposed sensor with R-wave timing. Figure 4 (a) indicates correlation coefficients of 0.8 or more ($p < 0.01$) when voltage V_d was changed from 0.9 to 1.4 V, and Fig. 4 (b) indicates that the amplitudes of voltage $|V_2|$ depended on force F_f applied to the proposed sensor. In addition, the maximum correlation coefficient in Fig. 4 (a) corresponds to the maximum voltage

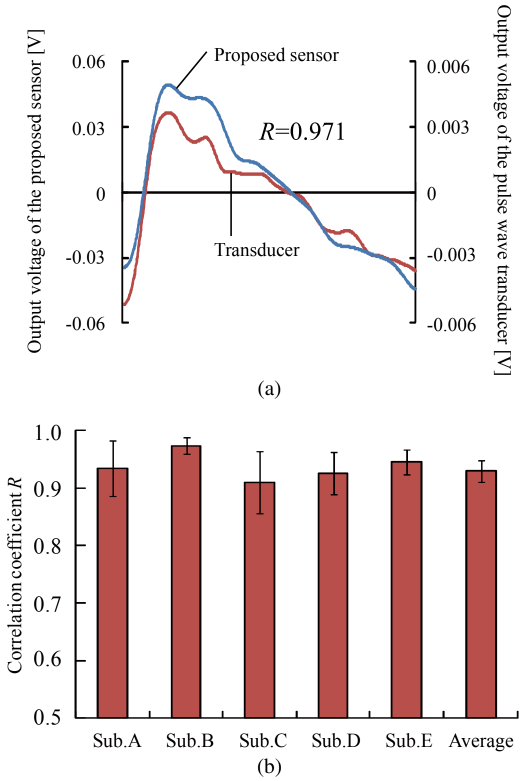


Fig. 5 Measurement results using the proposed sensor and a commercial pulse wave transducer: (a) comparison of output voltages for the proposed sensor and the pulse wave transducer; (b) coefficient of correlation between waves measured using the proposed sensor and those measured using the pulse wave transducer

amplitude V_d in Fig. 4 (b). This tendency was confirmed for all subjects. Note that the measurer was able to palpate the subjects' carotid arteries while measuring waves using the proposed sensor. For this, optimal force F_f was decided as the point at which maximum amplitude of output voltage $|V_2|$ for the proposed sensor was obtained during wave measurement using the sensor.

Figure 5 (a) shows examples of the measured waves for one beat using the proposed sensor with the optimal force F_f and using the commercial transducer, both of which were extracted from Subject B. From the figure, it can be seen that a significant similarity between the two was observed, and the determination coefficient R^2 was 0.94 ($p < 0.01$). Figure 5 (b) shows the coefficients of correlation between waves measured using the proposed sensor and those measured using the commercial transducer for all subjects. The average correlation coefficient was 0.920 ± 0.024, and all values were 0.9 or more. The signal-to-noise ratio of the waves measured using the proposed sensor was 12.71 ± 1.14 dB.

V. Discussion

Figures 3 and 4 show that the maximum coefficient of correlation between the left carotid artery pressure waves measured using the proposed sensor and the pulse pressure waves of the radial artery was obtained when the output voltage corresponding to diastolic pressure V_d of output voltage V_2 was 1.2 V. The coefficient of correlation between the two waves and the amplitude of output voltage $|V_2|$ decreased in the range of $V_d > 1.2$ V because F_f applied to the proposed sensor may flatten the vascular wall of the carotid artery excessively, causing disorder in the waves measured. Conversely, the coefficient of correlation between the two sets of waves and the amplitude of output voltage $|V_2|$ decreased in the range of $V_d < 1.2$ V because the surface of the vascular vessel may not be flattened by the force F_f acting on the proposed sensor which was too small. It was therefore concluded that the optimal force F_f was obtained when the maximum amplitude of output voltage $|V_2|$ for the proposed sensor appeared during measurement.

The proposed sensor achieved a suitable signal-to-noise ratio to allow measurement of pulse pressure waves from the carotid artery because its high-friction rubber pieces prevented sliding on the skin. It can therefore be concluded that the waves measured using the sensor were free of the effects of body motion. Figure 5 shows a high coefficient of correlation between waves measured using the proposed sensor and those measured using the commercial transducer for all subjects. It was therefore confirmed that the sensor enabled accurate non-invasive measurement of carotid pulse waves regardless of the subject.

VI. Conclusions

This paper outlines the development of a sensor for non-invasive measurement of carotid pulse pressure based on the following requirements: (I) the measurer must be able to palate the carotid pulse pressure using the sensor; (II) the sensor must be small and easy to attach/detach; (III) the position of the sensor during pulse pressure measurement must be stable; and (IV) the sensor must be able to measure continuous pulse pressure and have a high signal-to-noise ratio. It was found that the sensor can be used to measure pulse waves from the carotid artery. In future work, a method of sensor calibration for blood pressure measurement will be considered, and a first-aid measurement system using the proposed sensor will be explored.

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