An Improved Artery Pressure Sensor Design That Minimally Disturbs Blood Pressure and Flow

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Abstract

This paper reports the design of an in-vivo pressure sensor that can be implanted around the artery to measure blood pressure. As traditional arterial pressure sensors are completely wrapped around the artery, they reduce blood pressure and mass flow rate inside the artery. Two designs were proposed here to overcome these negative effects. The designs consist of two identically sized piezoelectric sensors placed opposite each other around the artery and a substrate whose endpoints are tied by medical suture to fix the sensors. The piezoelectric sensor allows to detect blood pressure as an electrical signal during the arterial pressure wave. The proposed designs were analyzed by the finite element simulations to evaluate the improvements in blood pressure and mass flow rate compared to previously reported sensor designs. In this respect, it has been proven that the blood pressure and mass flow rate values are improved from 99.88% to 99.97% and from 85.34% to 94.37%, respectively.

1. Introduction

Blood flow in the body is a critical parameter for human health status. Blood flow state can be evaluated by pressure and mass flow rate values. If the blood flowing from the heart to the organs is not proper due to the surgical operations such as organ transplantation, organ failure may occur. For these reasons, continuous measurements of artery pressure after the surgical operations provides the opportunity for medical intervention. In addition, it is important that the blood flow is not restricted during the pressure measurement, so that the measurement does not have a negative effect.

It has been reported that in vivo sensors are located to the artery in two different ways to measure blood pressure in last researches [1]. When the sensor is located to inside of artery, artery pressure is measured directly. However, there is a risk of blood clotting and some complications occurring. Therefore, most of works in the literature have focused on sensors placed outside the artery, which indirectly measure pressure through the expansion and contraction of the wall.

Sensor design which consists of inductor and capacitor for real-time wireless monitoring of blood pressure, is presented [2]. The sensor packed with flexible material is placed around the artery. It was proved by in-vitro experiments that blood pressure could be detected when there was a resonance frequency shift in the sensor. A blood pressure sensor is developed by utilizing fringe effect in capacitor was reported in [3]. The sensor package covered with biodegradable material was wrapped around the artery and it was demonstrated through in-vitro and in-vivo experiments that the fringe capacitance was affected during blood pressure change. The change in capacitance was obtained as a shift in the resonant frequency with the inductor on the sensor. The capacitance change caused a shift in the resonant frequency with the inductor on the sensor. The piezoresistive sensor, which is widely used for the detection of blood pressure by placing it into the artery, has been reported in a research where it is applied to outside of artery [4]. When the artery expands and contracts, a change in resistance occurs due to the use of graphene in the sensor. In the literature, there are many researches using sensor types with different detection mechanisms such as surface acoustic waves [5], optics [6], impedance plethysmography [7] to detect blood pressure around the artery.

Piezoelectric materials, which can convert the mechanical effect into an electrical signal, are used for direct blood pressure detection from the artery. Potkay et al. designed the biocompatible piezoelectric sensor by encapsulating the polyvinylidene fluoride (PVDF) material with medical grade silicon [8]. The sensor was wrapped around a latex tube that mimics an artery. The device measured maximum output voltage of 1.2 V and generated power of 16 nW for pressure detection by in-vitro experiments. Another similar design in which PVDF material is preferred, has been reported by Cheng et al. [9]. Both in vivo and in vitro tests of the sensor which is coated with polyimide (PI) material, have been demonstrated. They reported that the blood pressure sensitivity of the device was 14 mV/mmHg and that it produced a maximum instantaneous power of 40 nW.

It has been observed that most in-vivo pressure sensors placed outside the vessel, which have been developed so far, completely surround the artery and therefore limit radial expansion. This limitation reduces the mass flow rate of blood and damages the organ which is connected with respective to artery. In addition to this, it changes the actual value of the pressure to be measured. To overcome these limitations, segmentation of piezoelectric sensing elements was proposed in [10]. The artery wall and blood were simulated separately to find the rate of improvement in flow. However, since the artery wall is a flexible material, it is more accurate to analyze blood and artery wall together by considering the more complex physics involving simultaneous fluid-solid interaction. Here, different designs are presented for applying the piezoelectric sensor around the artery to overcome the limitations mentioned above. The fluid-solid interaction was analyzed utilizing the finite element method and the improvement in the mass flow rate and pressure of the blood was proven for each design.

2. Design

Design consists of sensor material and substrate. Polyvinylidene fluoride (PVDF), which has biocompatibility, was preferred for the piezoelectric sensor material. It was wrapped around the artery as two identical separate pieces. A single piece spans one-quarter in the first design and one-eighth in the second design of the artery wall. Polydimethylsiloxane (PDMS) was chosen as the substrate material to fix the PVDF. Thanks to the high flexibility property of PDMS, it does not cause the wall to be very restricted when the artery is expanded. The two ends of the PDMS are fixed to the artery by tying them with medical suture. Schematics of these designs are shown in Fig. 1.

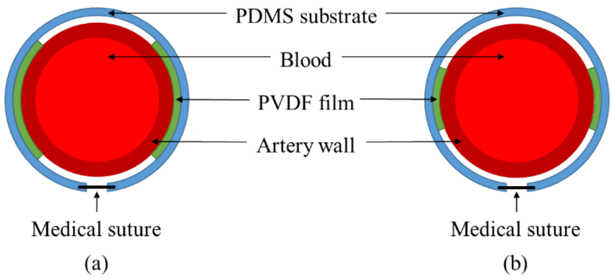

Fig. 1. Schematics of the sensor designs, (a) Design 1, (b) Design 2

Due to the gap between the PDMS substrate and the artery wall in the designs, sensor allow the vessel wall to expand during the blood pulse wave. The blood pressure to be measured will be more accurate because the expansion is not restricted. Also, the change in blood flow will be kept to a minimum and the negative impact of the sensor on the body will be reduced.

3. Finite Element Simulations

The Lagrangian method is insufficient for the solution of large deformations of the fluid. Also, inaccurate results can be obtained when the Euler formulation is applied to solids [11]. The arbitrary Lagrangian-Euler (ALE) method is used for fluid-structure interaction (FSI) problems so that the fluid domain mesh follows the deformation of the structural domains. In this respect, ALE method is used to analyze the problem of blood pressure and flexible arterial wall together in order to obtain accurate results.

The blood flow and artery deformation was simulated using the Fluid-Structure Interaction multiphysics module of the COMSOL finite element analysis software. The following of simulation, blood pressure and mass flow rate in the middle of the artery were calculated. To demonstrate the advantages of the two proposed designs, three sets of models were simulated as specified below.

- Artery with sensorless.
- Artery with sensor that completely surrounds the artery wall.
- Artery with the designed sensor.

It is not possible to compile the entire arterial system in simulation. In this respect, only 30 mm part of an artery was selected. The three sets of models simulated for Design 1 are shown in Fig. 2.

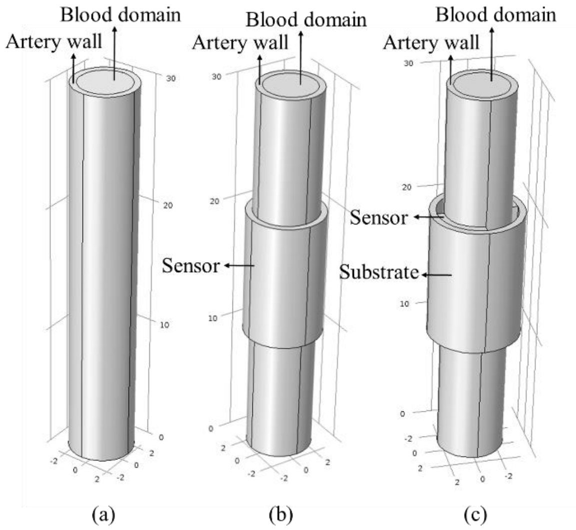

Fig. 2. Geometries of sets of models for Design 1, (a) artery with sensorless, (b) artery with sensor that completely surrounds the artery wall, (c) artery with the Design 1

Due to the blood pressure is in wave form, time dependent analysis was performed. The minimum and maximum values of the inlet pressure waveform were chosen as 80 mmHg (10.665 kPa) and 120 mmHg (16 kPa), which are the blood pressure values of a healthy person, respectively. The heart rate was selected as 80 beats per minute and the first three pulses were evaluated. Fig. 3 shows the waveform of the inlet blood pressure with 80 beats per minute heart rate used in the simulation.

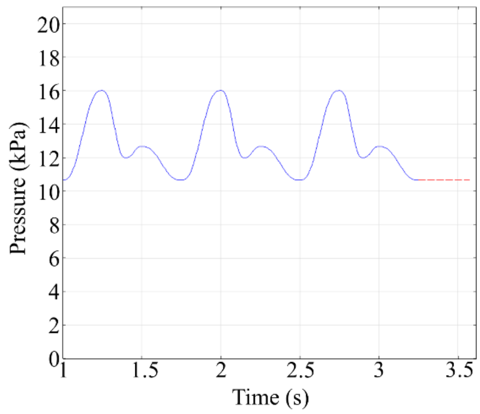

Fig. 3. The waveform of the inlet blood pressure used in the simulation

The outlet pressure was chosen to be 99.5% of the inlet pressure by considering the friction-induced pressure drop in the artery. The parameters and values used in the simulations were selected by considering values in [1]-[10] and listed on Table 1. It was assumed that the Young's Modulus values of the artery, sensor, and substrate were the same.

Table 1. Design parameters used in simulation

Parameter	Value	Parameter	Value
Artery Young's modulus	500 kPa	Model length	30 mm
Artery Poisson's ratio	0.45	Sensor length	10 mm
Artery density	1120 kg/m^3	Substrate length	10 mm
Blood density	1060 kg/m^3	Artery inner radius	2 mm
Blood viscosity	0.0035 Pa.s	Artery wall thickness	0.5 mm
Sensor and substrate Poisson's ratio	0.45	Sensor thickness	0.5 mm
Sensor and substrate density	1120 kg/m^3	Substrate thickness	0.5 mm

In alignment with the previous related studies reported in the literature [12-14], blood flow was modeled to be Newtonian, laminar and incompressible. Thus, the viscosity of the blood will be equal to its initial value during the simulation. Since the velocity of blood adjacent to the wall in arteries changes at the same rate as the wall, a no-slip condition was assumed. When the inlet pressure is 10.665 kPa and 16 kPa, blood pressure profile inside the artery for model sets and proposed designs is shown in Fig. 4 and Fig. 5, respectively.

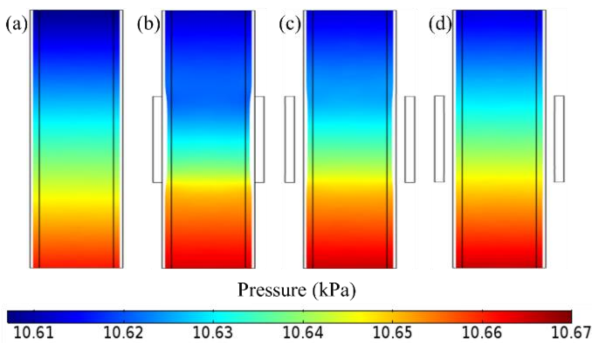

Fig. 4. When the inlet pressure is 10.665 kPa, blood pressure profile inside the artery, (a) sensorless, (b) circular sensor, (c) Design 1 sensor, (d) Design 2 sensor

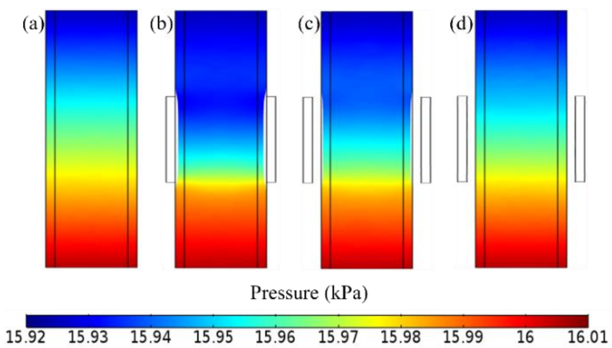

Fig. 5. When the inlet pressure is 16 kPa, blood pressure profile inside the artery, (a) sensorless, (b) circular sensor, (c) Design 1 sensor, (d) Design 2 sensor

The mass flow rate and blood pressure were calculated by arithmetic mean considering the values at the times of the input waveform at its maximum and minimum. Calculated results that are listed in Table 2, show the actual blood pressure drops to around 99.88% for the circular sensor, compared to the case with no sensor in the artery. It was obtained that the ratios for the Design 1 sensor and the Design 2 sensor were 99.92% and 99.97%, respectively, so there is a small improvement. A greater effect was shown in the mass flow rate than in the pressure. Although 85.34% of the mass flow rate in the sensorless artery was achieved in the circular sensor design, it yielded 88.95% and 94.37% for the design 1 sensor and design 2 sensor, respectively. Since the proper functioning of the organs is directly related to the amount of blood coming from the relevant artery, this improved ratio is critical.

Table 2. Calculated results

Values	No sensor	Circular sensor	Design 1 sensor	Design 2 sensor
Pressure (kPa)	12.6343	12.6198	12.6252	12.6310
Mass flow rate (kg/s)	0.00498	0.00425	0.00443	0.00470

The simulation results are analyzed for the cases where the Young's Modulus of the artery, the substrate and the sensor material were assumed to be equal, and the improvement rates were revealed. The Young's Modulus values of the sensor materials reported in the literature are generally higher than that of arterial tissue. Simulation applications were performed to investigate the effect of the sensor material on blood pressure and mass flow rate. The values of blood pressure and mass flow rate obtained from the simulation for different values of the sensor Young's Modulus are shown in Fig. 6 and Fig. 7, respectively. Note that although PVDF has a Young's Modulus value higher than 1 GPa, the Young's Modulus of the substrate and the sensor were chosen to be equal in the simulations as this would allow for more closely observing the effect of the existence of sensor on pressure and mass flow rates. Higher sensor Young's Modulus results can easily be extrapolated from the obtained data. Although there is a minor change in blood pressure, a major change in mass flow rate of blood was observed. In addition, it was noticed that the change in both parameters decreased at high Young's Modulus values.

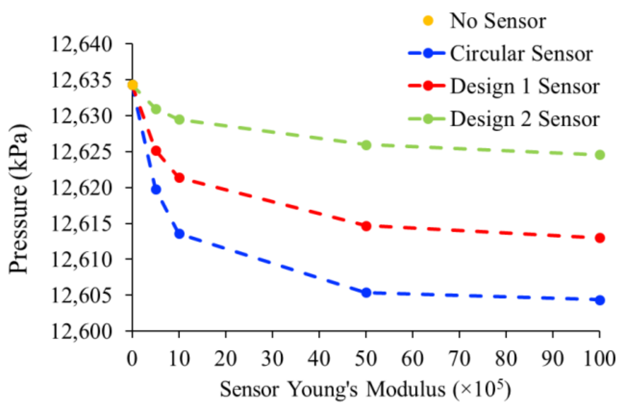

Fig. 6. The change of blood pressure with respect to different Young's Modulus value of the sensor

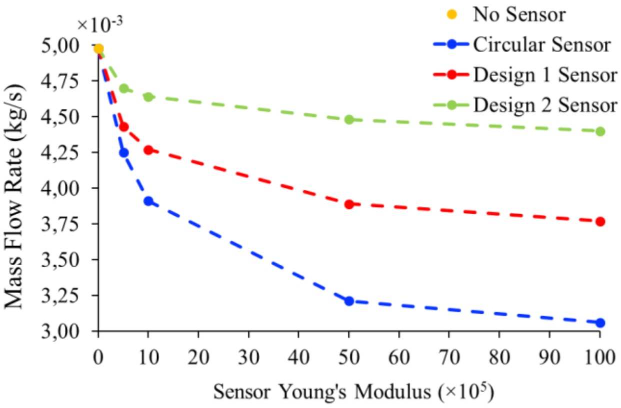

Fig. 7. The change of mass flow rate with respect to different Young's Modulus value of the sensor

4. Discussion

The simulation results reported in this work clearly demonstrate that the sensor, which partially surrounds the artery, restricts blood flow less than the ones that completely surrounds the artery. The purpose of Design 1 and Design 2 is to apply to the piezoelectric sensor module for detecting blood pressure around the artery. The advantage of these sensor designs reveals with increasing sensor Young's Modulus values. Simulation applications are in progress to determine how much signal the sensor will give at any given size. After the device size is decided with respect to the simulation results, it will be manufactured using microfabrication technologies and test it.

5. Conclusions

This paper reports implantable blood pressure sensor designs that can detect the pressure inside the artery. The traditional sensors that surround the artery restrict blood flow and therefore harm organ health. Different designs were recommended to overcome this problem. Simulations performed with the finite element method proved that the proposed designs cause small changes in arterial mass flow rate and blood pressure. This causes blood pressure measurement to be more accurate and healthy after organ transplantation.

6. Acknowledgment

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