Self-Powered Implantable Electromagnetic Device for Cardiovascular System Monitoring Through Arterial Wall Deformation

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Abstract. In this paper, we present the potential of a device, originally designed for energy harvesting, to form a self-powered medical implant that monitors critical parameters of the cardiovascular system. The original design consists of a coil that deforms with an artery inside magnetic field applied by two permanent magnets. We fabricated the device, and developed appropriate experimental setup that simulates blood flow and arterial wall pulsation with adjustable frequency and pressure. The voltage and power of the moving coil, as well as the pressure inside the tube simulating the pulsating artery were measured at different frequencies. In-vitro experiments and theoretical analysis showed that the voltage induced across the coil's terminals can provide information on blood pressure, heart rate, arterial wall deformation and velocity.

Keywords: Self-powered · Implantable devices · Cardiovascular system · Arterial wall deformation · Blood pressure monitoring · Energy harvesting

1 Introduction

Blood pressure monitoring is of utmost importance for the prognosis and treatment of various diseases. High Blood Pressure (HBP) has no specific symptoms, but it can cause serious damage or even death. Heart attack, aneurism, atherosclerosis, stroke, kidney damage and vision loss can be some of the consequences when HBP is left untreated [1].

Cuff-style monitors are the most common non-invasive commercial devices for blood pressure monitoring, but they usually provide intermittent measurements and their accuracy is lower than that of invasive methods [2, 3]. A cuffless, non-invasive approach to estimate blood pressure is through Pulse Transit Time (PTT), which is defined as the time of propagation of a pulse to travel in artery. However, the correlation between PTT and blood pressure can be affected by viscoactive and other drugs, thus reducing the reliability of the method [4]. Arterial catheterization is an invasive method that offers

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continuous and accurate measurements, but in some cases it can cause complications such as bleeding, hematoma, nerve damage, arterial thrombosis and infection [5].

Medical implantable devices have become quite popular during the last decades, since they permit continuous and precise measurements of critical physiological parameters. CardioMEM's Champion implantable device is a commercially available pressure monitoring system. An implantable blood pressure monitor that uses MEMS capacitive sensors has been fabricated [6], while an implantable wireless device that provides pressure measurements through a Surface Acoustic Wave (SAW) resonator has also been proposed [7]. Moreover, an implantable sensor that measures blood pressure through PTT by using an accelerometer has also been developed [8].

A major issue associated with implantable devices arises from the limited lifetime of their battery, which imposes the need for the patient to undergo surgery for their replacement. To address this problem, energy harvesting devices have been developed, which exploit energy sources either in the environment or in the human body, in order to recharge or power medical implants. Among these energy sources, the cardiovascular system seems very promising, since the motion of the heart and the arteries is a continuous and periodical source of mechanical energy, which can be converted to electrical energy.

Using an energy harvesting device as a sensor would be quite beneficial to the power efficiency of a medical implantable device. A device that exploits the pulsation of aorta to produce electrical energy through a piezoelectric thin film and functions as blood pressure sensor has been recently constructed [3]. Moreover, a mechanism that harvests energy from arterial wall deformation through a coil that moves in magnetic field has been developed [9]. The purpose of our work is to demonstrate that an energy harvester based on the design proposed in [9], can be used to monitor blood pressure and other cardiovascular parameters.

2 Methods

2.1 Sensor Description

We consider an energy harvesting device which consists of two ring magnets placed in parallel through metallic holders and a flexible coil [9]. Both permanent magnets are magnetized along the artery axis. The coil is placed between the magnets and it is constructed with side loops as depicted in Fig. 1(a). The purpose of the side loops is to make the coil flexible, in order to deform freely without restraining arterial wall movement. The artery comes through the holes of the magnets and the main loop of the coil (Fig. 1(b)). The coil moves inside the magnetic field applied by the magnets, and as a result, an alternating voltage proportional to its velocity is induced.

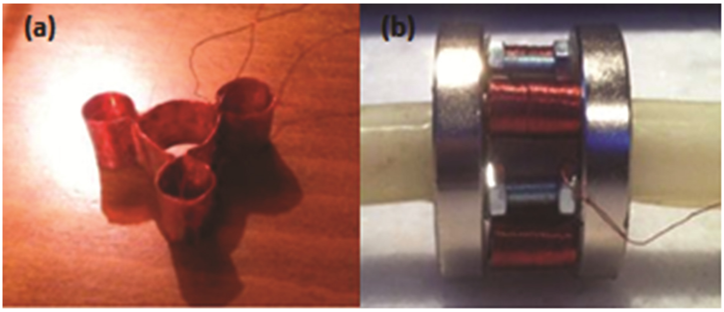

Fig. 1. **(a)** Flexible coil with side loops which is wound around the artery. **(b)** Energy harvesting device [9] used in our experiments.

2.2 Theoretical Analysis

A theoretical analysis was developed to support the scenario, according to which the energy harvesting device of Fig. 1 can operate as a pressure sensor. Figure 2 illustrates the ring magnets, the artery, and the coordinates system. The z-axis is parallel to the artery and the arterial wall deforms along the radial r-direction. Moreover, l and s denote the length of the magnets and the spacing between them, respectively, while R_{in}, R_{out} are the inner and outer radius of the ring magnets, R_0 is the artery's undeformed inner radius and d is the arterial wall's thickness.

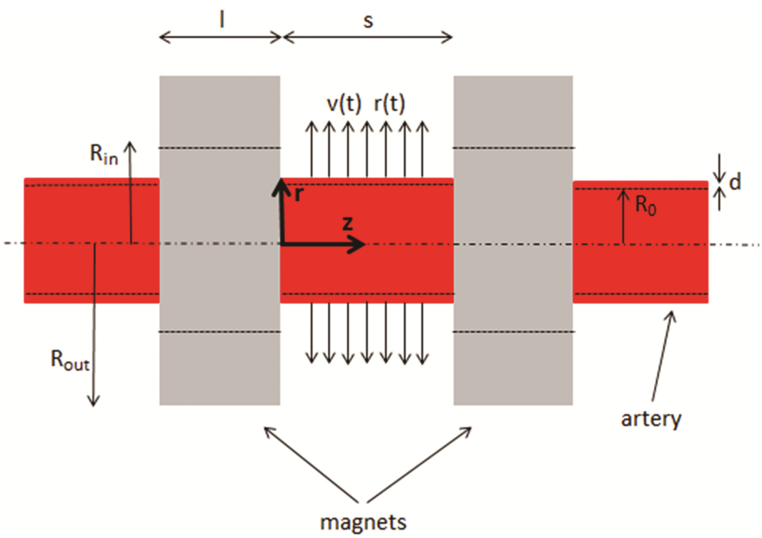

Fig. 2. Ring magnets, artery and coordinates system

The coil moves perpendicularly to z-axis, so only the z-component of the magnetic field contributes to the induced voltage. Assuming a cylindrical coil without side loops, then, the radial position r(t) and velocity v(t) of the coil are identical with those of the artery. The induced voltage V(t) can be determined by the following integral [9]:

$$V(t) = -\int_{c} v(t) \times B_{z2}(r(t), z)dl \tag{1}$$

where c is the curve defined by the coil and B_z is the z-component of the magnetic field. The latter (B_z) is determined in [9] for given dimensions (R_{in}, R_{out}, l) and remanent field (B_r) of the magnets.

The total resistance (R_c) of the coil depends on the resistivity (ρ_w), length (L_w) and surface (A_w) of the wire. In our hypothetical scenario of a cylindrical coil that deforms radially together with the artery, the length of the wire is a function of time and depends on its radial position r(t) [9]. Thus, the total resistance of the coil is given by:

$$R_c = \rho_w \frac{L_w(t)}{A_w} = \rho_w \frac{\pi N r(t)}{\pi d_w^2/4} \tag{2}$$

where N is the number of windings that form the coil and d_w is the diameter of the wire. The power transfer of the device is maximum when the coil is connected in parallel with a resistance equal to its own resistance and can be expressed as [9]:

$$P_{max}(t) = \frac{(V(t)/2)^2}{R_c(t)} \tag{3}$$

In order to demonstrate the function of the device as a pressure sensor, a relationship between the pressure inside the artery and the induced voltage needs to be determined. In our analysis, the artery is treated as an elastic, cylindrical tube, in which a periodic pressure pulse is applied. A detailed mechanical analysis is carried out and the following relationship between the artery's inner radius R(t) and blood pressure P(t) is obtained:

$$R(t) = \frac{R_0}{1 - \frac{\left(P(t) - P_e\right)}{Ed} R_0(1 - m^2)} \tag{4}$$

where R_0 is the undeformed inner radius of the artery, E is the Young's modulus of the material that forms the arterial wall, m is the Poisson's ratio, d is the wall's thickness and P_e is the environmental pressure which is considered as the pressure reference point. It is noted that the radial position of the coil r(t) corresponds to the outer radius of the artery (r(t) = R(t) + d).

2.3 Fabrication

Sensor

We used Neodymium ring magnets with inner and outer radius of R_{in} = 7.5 mm and R_{out} = 20.5 mm, respectively, and length l = 10 mm. Both magnets are magnetized along their length and their remanent magnetic field is B_r = 1.2 T. Non- magnetic, stainless steel screws were used as holders to place the magnets in parallel and at constant spacing. The coil was constructed using enamelled copper wire with diameter $d_w = 0.2$ mm. For the above given parameters, an analytical model of the energy harvesting device was developed using Matlab. From our simulations, we determined that the power generation is maximum when the spacing between the magnets is s = 2 cm, and selected this value for our in-vitro experiment.

Experimental Setup

In order to test the energy harvesting device, an experimental setup was constructed to simulate the arterial wall pulsation. An elastic tube made of platinum cured silicon rubber with $R_0 = 6$ mm undeformed inner radius, d = 1 mm wall thickness and E = 1.97 MPa Young's modulus was used to simulate the artery. An open loop was formed by connecting the elastic tube to a rigid tube made of Plexiglas, as depicted in Fig. 3(a). A catheter was then inserted to the latter through a thin hole on its wall, and the loop was filled with water. An Edwards Truwave pressure disposable transducer integrated in a sterile pressure monitoring kit was used to receive pressure measurements inside the open loop through the catheter. The elastic tube was placed inside the energy harvesting device. Both ends of the open loop, the connection between the elastic and rigid tube, as well as

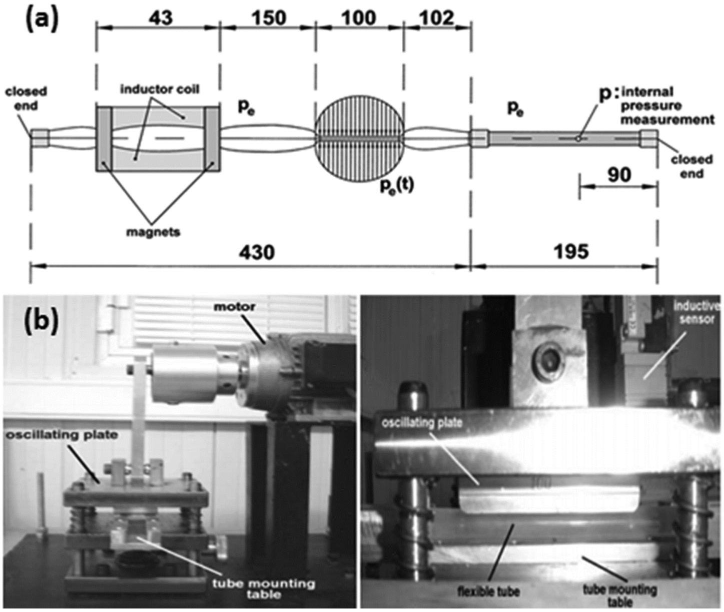

Fig. 3. (a) Representation of the open loop and the experimental set up. The elastic tube comes through the energy harvesting device and is connected to a rigid tube, where a catheter is inserted. A pressure pulse is applied through partial compression and decompression of an elastic tube's segment. (b) Motor-based compression mechanism. The elastic tube is inserted in the tube mounting table and is compressed and decompressed through an oscillating plate with adjustable compression rate. The motor drives the plate with controlled frequency.

the hole where the catheter was placed, were air tightly sealed. Moreover, all the air inside the loop was removed through a valve connected to its one end.

Blood flow and arterial wall deformation were reproduced by applying a periodic pressure pulse inside the tube, through a motor-based compression device, as shown in Fig. 3(b). A 10 mm long segment of the elastic tube was placed in the tube mounting table of the compression device and was periodically partially compressed and decompressed by a reciprocating flat plate. The latter moves due to the motor's rotation and the compression-decompression frequency can be controlled through the velocity of the motor. Special care is taken so that the latter is parallel to the tube axis and as a consequence the compression is applied uniformly along the tube. The device also contains a mechanism so that both the flexible tube mounting table and the reciprocating flat plate can be moved in small steps up and down, adjusting the compression of the elastic tube. The amplitude of the pressure pulse was adjusted in order to achieve a physiological arterial wall deformation, which is typically 10% of the artery diameter [9].

The time waveform of the pressure measured by the Edwards Truwave pressure disposable transducer was obtained by HBM QuantumX MX440A Analog to Digital Converter (ADC). An instrumentation amplifier was used to amplify the voltage induced in the coil and its output was sampled by a microcontroller Arduino Uno. The output power was measured by connecting a resistor of 8.2 Ω to the coil's terminals, which is approximately equal to the coil's resistance.

3 Results

Table 1 shows the peak open circuit voltage of the coil, as well as the mean output power at four different frequencies corresponding to a different heart rate (sleep, rest, tachycardia and exercise). Figure 4 demonstrates the internal pressure P (Fig. 4(a)), the elastic tube's inner radius R (Fig. 4(b)) and wall velocity dR/dt (Fig. 4(c)), as well as the coil's denoised open circuit voltage V_{oc} (Fig. 4(d)) over time, at a rate of 119 bpm (tachycardia). The waveforms of the tube's inner radius and wall velocity were derived from our pressure measurements, by using Eq. (4) and by differentiating the radius.

Table 1. Peak open circuit voltage and mean output power for frequencies corresponding to typical heart rates of a person at the states of sleep, rest, tachycardia and exercise

Heart rate (bpm)	V_{peak} (mV)	P_m (nW)
52.7	2.3	19.1
82.2	2.4	28.1
119	2.5	52.2
139.1	2.7	68.5

From Fig. 4(c) and (d), we can see that the waveforms of tube's wall velocity and the output voltage have similar shape, as expected, given that the induced voltage is proportional to the velocity of the coil, which deforms along with the elastic tube. Given that the velocity of the arterial wall v(t) is the first time derivative of the radius R(t) (v(t) = dR(t)/dt), blood pressure can be derived from the output voltage of our sensor

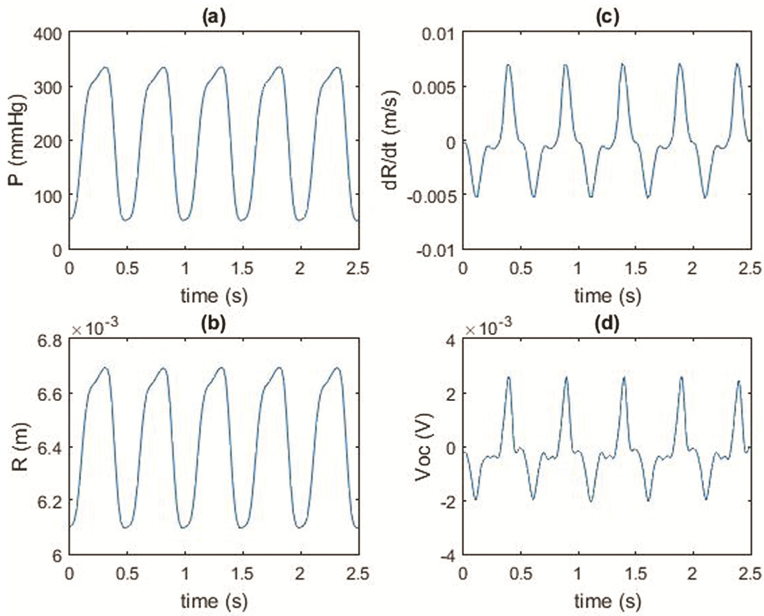

Fig. 4. Time waveforms of (a) internal pressure, (b) elastic tube's radius, (c) elastic tube's wall velocity, (d) open circuit voltage of the deforming coil

and vice versa through Eqs. (1) and (4). We also note that all the quantities presented in Fig. 4 have the same frequency, which in our experiments corresponds to the heart rate. Moreover, the produced power and voltage increases with increasing frequency of the pressure pulse. This can be attributed to the fact that the tube deforms faster at higher rates.

4 Conclusion

In vitro experiments and theoretical analysis indicate that the output voltage of the coil can be used to estimate cardiovascular parameters such as blood pressure, heart rate, arterial wall deformation and velocity. The next steps of our research would be to further study the voltage variation as a function of the desired bio-parameters by running the experiment for different values of pressure and construct a prototype with smaller dimensions for in vivo experimentation. Its biocompatibility is also a matter that needs to be investigated. Furthermore, an appropriate packaging would be necessary to prevent possible interaction with other implantable electromagnetic devices, or electromagnetic interferences from external sources. Though there are several issues that need to be addressed prior to its in vivo application, the presented energy harvesting method has the potential to form an implantable medical device that offers continuous monitoring of the cardiovascular system and simultaneously produces electrical energy to power itself.

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