A MODEL OF THE ARTERIAL WALL INTERACTION WITH A BLOOD PRESSURE SENSOR TRANSDUCER FIXED IN PROXIMITY

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***Abstract*— MEMS technology is an option for the development of a pressure sensor which allows the monitoring of several bio-signals in humans. In this work, a comparison is made between the typical elasticity and viscosity presented in several arteries in the human body and those present in MEMS silicon microstructures based on membranes in proximity with an arterial wall. The main purpose is to identify which types of microstructures are mechanically compatible with human arteries. The ultimate goal is to integrate a blood pressure sensor which can be implanted in proximity with an artery. The expected benefits for this type of sensor are mainly the reduction in problems associated with the use of bulk devices through the day and during several days. Such a sensor could provide precise blood pressure readings in a continuous or periodic form, i.e. information that is especially important for some critical cases of hypertension patients. The modeling work involved in this paper, accounts for the analysis of micro displacements present in the membrane of a MEMs silicone microstructure placed directly on a human arterial wall, at different heart rates. The modeling includes the effects of elasticity and viscosity of the silicone structure on the pressure measurement. Additionally, the sensitivity of the membrane to detect slight variations in the blood pressure is presented.**

***Keywords*— MEMS, Bio-MEMS, Blood Pressure Sensors.**

I. INTRODUCTION

Several models exist that suggest different forms of interpretation based on mathematical models of the elastic behavior of the arterial wall, the majority of these models are based on the viscoelasticity theory. Some of the models that have been proposed to solve the strain-stress relation of the arterial wall are: elastic parameters only (Rand, 1968) and viscoelasticity (Armentano *et al.*, 1995; Salvucci, 2007). Furthermore, a typical arterial wall consists of three layers: an innermost layer, "the intima", mainly composed of endothelial cells, a middle layer, "the media", composed of elongated smooth cells, elastin and collagen, and an outer layer "the adventitia", comprising a varied number of elastic sheets, bundles of collagen fibrils, and a network of elastic fibrils (Valdez-Jasso, 2009).

The different layers of the arterial wall can be modeled as springs, with an elastic of Young modulus E, or dashpots, with a viscosity modulus η. The arterial wall exhibits the characteristics of stress relaxation, creep and hysteresis (Fung, 1993). The basic mechanical models that are often used to discuss the viscoelastic behavior of materials are: the Maxwell model, the Voigt model and the Kelvin model (also called the standard lineal solid), all of which are a combination of linear springs and dashpots. The combined structure of the arterial wall in this paper has been modeled using a modified Maxwell model (Armentano *et al.*, 1995).

There is a clear tendency to the use of implanted bio-medical devices. An implanted blood pressure sensor could be the best option for the continuous or periodic monitoring of this parameter in some critical cases of hypertension in humans. We propose in this work an implantable MEMS pressure sensor in proximity with one of the human arteries. Several options exist, and different studies have been carried out in animals such as rats and sheep. These include carotid, femoral and radial arteries to mention a few examples. We believe that a sensor placed in the outside and touching an appropriate artery, such as the radial artery, could be accessible for implantation with minimal surgery

The sensor design is based on a silicon piezo-resistive membrane with a CMOS circuit amplifier and an RF section to transmit the signal via wireless. The expected size of a MEMS-CMOS pressure sensor is similar to a grain of rice, for instance: 1.4mm x 1.4mm x 7mm. The aim is to design a wireless, battery less, bio-compatible device in proximity with a human artery. Similar designs already exist (Potkay and Brooks, 2008), for example, to monitor the blood pressure in small laboratory animals (Cong *et al.*, 2012), a clip size wireless monitor of endovascular-repaired abdominal aortic aneurysms (MacKenzie *et al.*, 2005) and an implantable microsystem for tonometric blood pressure measurement with a size of 3x6.5x10 mm (Ziaie and Najafi, 2001). Most of these sensors are based on the tonometric principle.

In this paper an extension of the modified Maxwell model to include the silicon rubber layer of the bio-compatible packaging and the silicon membrane of the MEMS pressure sensor, is proposed. Our model links the viscoelastic parameters of the arterial wall to the

viscoelastic parameters of the sensor to give an integral view of the arterial wall–sensor system.

II. METHODS

A. Model and Designs

Figure 1 shows the mechanical models used to discuss the viscoelastic behavior of materials. The Maxwell body is a dashpot arranged in series with a spring. The Voigt body is a dashpot in parallel with the spring. The Kelvin body is a combination of the Maxwell and the Voigt body. In these models μ is the spring constant, η is the coefficient of viscosity, u is the total displacement, u_1 is the displacement of the spring, u'_1 is the displacement of the dashpot, and F is the total force acting on the body.

If we solve u(t) when F(t) is a unit-step function, the results are called creep function. The question is: which is the right model to choose for modeling an artery. One approach is to compare the experimental curves of relaxation, creep, hysteresis, and frequency response with those of the theoretical models. One model that has been proposed that fits experimental data for arteries is the shown if Fig. 2 and it is called the Modified Maxwell Model (Armentano *et al.*, 1995). In this case, the spring constant is replaced by the Elastic or Young Modulus (E of the arterial wall is given by the elastin fibres, and the non-linear behavior is given as the collagen) in mmHg, and the Viscosity Coefficient by the Viscosity Modulus (η) in mmHg·s. The Parallel Elastic Component (PEC) corresponds to the elastic component composed by elastin and collagen fibres. As reported in previous works, the elastin fibers are constantly submitted to all levels of deformation, whereas the collagen fibers are gradually "recruited" as the strain degree increases. This is indicated by the hook shapes in the diagram. There is also a contractile element (CE) in series with a pure elastic spring. The mass of the arterial wall corresponds to the M block.

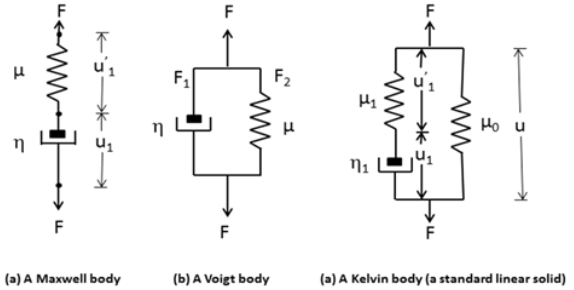

Figure 1. Basic Viscoelastic Models.

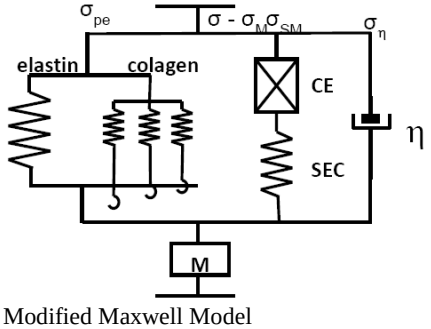

Figure 2. Modified Maxwell Model

Figure 3 shows the proposed design of blood pressure sensor. It consists of a silicon chip 4 mm long, 400 μm thick (the thickness of the silicon wafer) and 1.5 mm wide plus 500 μm of silicone rubber surrounding the chip. The design of the chip includes an area for energy capture via electromagnetic induction, similar to an RFID chip in order to make battery less operation possible, the piezo-resistive silicon membrane, and the CMOS section with the electronics necessary to amplify the weak signal obtained from the transducer. The amplified signal is used to modulate a 200 kHz carrier signal delivered to the square spiral antenna to send the blood pressure information signal across the layers of tissue to an external device located for example in the wrist of the patient. The external device can have display, battery, a microcontroller, and a memory to store the measurements for one or more days. The sensor device is planned to be touching the external layer of the target artery. Also, it is very important that some pressure be exerted to the artery to flatten it to some extent in order to be able to make the measurement using the tonometric principle (Ziaie and Najafi, 2001).

As we can see for the shadowed area beneath the silicone membrane of the chip, the section that is going to be modeled by the chip includes the three layers of the arterial wall plus approximately 900 μm of silicone rubber (500 μm of the biocompatible packaging and the inner 400 μm that serve as mechanical coupling linking to the silicon membrane of the sensor) and the 10 μm piezo-resistive silicon membrane.

The model we propose for the analysis is shown in Fig. 4. We have simplified some parts of the Modified Maxwell Model (Rand, 1968). In this way we could obtain preliminary, interesting results. The simplifications are the following: i) the non-linear response of the addition of successive collagen fibers is not taken into account, ii) only one elastic element due to collagen was considered, iii) the contractile element is not to be considered as we are working on the assumption that no drugs are administrated for excitation of the smooth muscle of the artery, iv) the effect of the mass (M) of the arterial wall is neglected due to the low frequency of the signal. On the other hand, we have added an elastic element and a viscous element due the presence of a silicon rubber layer and also an equivalent elastic element due to the silicon membrane. The viscous component of the silicon membrane is assumed to be zero.

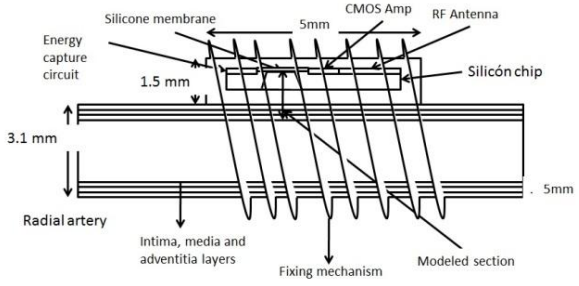

Figure 3. - Design of the MEMS-CMOS pressure sensor indicating the section that is going to be modeled using viscoelasticity analysis, for the intima layer to the silicone membrane.

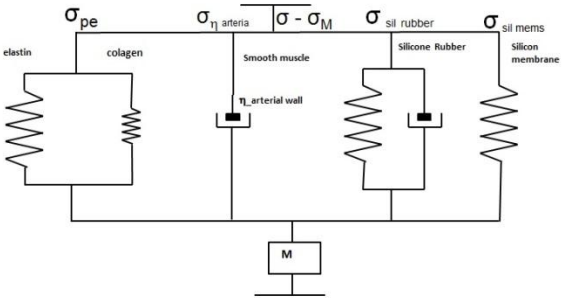

Figure 4. Extended model to cover the silicone rubber layer of the sensor and the silicon membrane of the piezoresistive transducer

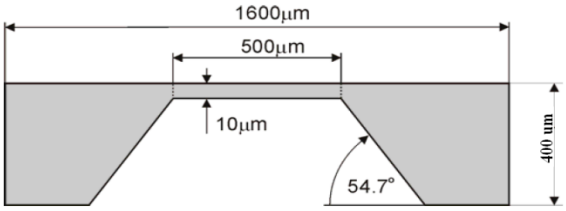

Figure 5. Geometry and dimensions of the piezoresistive pressure sensor based on a silicon membrane.

The constitutive equation of the extended model of Fig. 4 is then:

$$\sigma = E_e\varepsilon + E_c\varepsilon + \eta_a\frac{d\varepsilon}{dt} + E_{sr}\varepsilon + \eta_{sr}\frac{d\varepsilon}{dt} + E_{equiv-sm}\varepsilon \quad (1)$$

where σ is the total stress corresponding to the blood pressure input, E_e is the elastin elastic modulus, ε is the deformation, E_c is the collagen elastic modulus, ηa is the arterial wall viscous modulus, E_{sr} is the silicone rubber elastic modulus, η_{sr} is the silicone rubber viscous modulus, and $E_{equiv-sm}$ is the silicon membrane equivalent elastic modulus, respectively.

The membrane design is shown in Fig. 5. Figure 6 shows the input signal that is introduced in our model. This input signal has typical characteristics of a blood pressure signal in human artery walls, where the systolic stage (first peak) and the diastolic stage (second peak) of a heart beat can be observed, with minimum and maximum values of blood pressure ranging around 80-120 mmHg and a heart rate of 60 beats/min. It is worth to mention that this numerical signal is calculated via the sum of different sinusoidal signals. In order to consider estimation errors of this numerical input compared to the blood pressure in humans, we consider deviations of this numerical input as it can be observed in Fig. 6, where curves in dash lines represent the possible estimation errors of the numerical signal. However, these differences can also be considered as deviations of the blood pressure, typically present in different individuals. The effects of these differences in the input signal (stress) and the effect of the heart rate (from 60 to 120 beats per min) on the micro-displacements of the silicone membrane (strain) are analyzed in order to obtain the sensitivity of the proposed device.

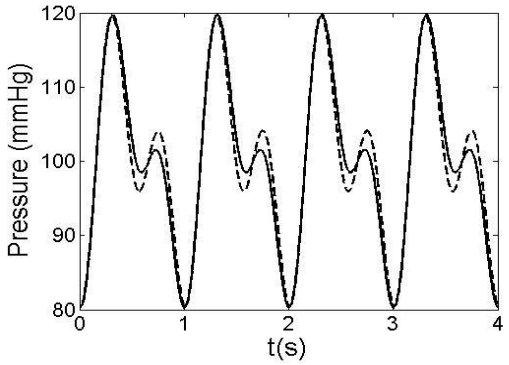

Figure 6. Artery pressure signal with a heart rate of 60 beats/min.

B. Data analysis

To evaluate the strain function described in Eq. 1, we have to modify it as follows:

$$\frac{1}{\frac{d\varepsilon(t)}{dt}} = \frac{1}{\frac{1}{E_e}\frac{d\sigma(t)}{dt}} + \frac{1}{\frac{\sigma(t)}{\eta_a}} + \frac{1}{\frac{1}{E_{sr}}\frac{d\sigma(t)}{dt}} + \frac{1}{\frac{\sigma(t)}{\eta_{sr}}} + \frac{1}{\frac{1}{E_{equiv-sm}}\frac{d\sigma(t)}{dt}} \quad (2)$$

where $\varepsilon(t)$ corresponds to the total deformation caused in the membrane or in the artery wall depending on whether the elastic and viscosity terms of the silicone rubber and the membrane are considered or neglected in Eq. 2. To solve it, we consider the values of E_e=10 Mdyn/cm2, η_a=0.1seg-Mdyn/cm2, E_e=1x106 N/m2, η_a=100 seg-N/m^2, and $E_{equiv-sm}$=1x10^6 N/m^2 as reported previously (Salvucci *et al.*, 2007; Shin-Etsu, 2012). Also, it is worth to mention that in a similar way to the models presented in Figs. 2 and 3, our model does not consider the width of the different layers present in the artery wall and the layers of the MEMS device, due to the fact that the elasticity and viscosity terms are considered constants along these layers. We then can proceed to calculate the strain on both, the arterial wall and the membrane of the MEMS device by introducing an input pressure signal.

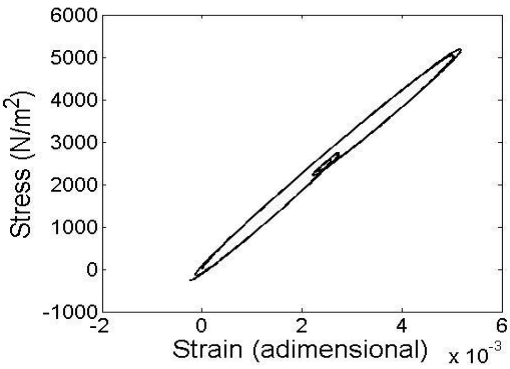

Figure7. Stress vs. Strain for an arterial wall without a MEMS device.

Figure 7 shows the result of the strain present on the arterial wall due to the blood pressure signal shown in Fig. 6. This stress is related to the strain via elasticity and viscosity of the arterial wall as it was indicated

in Eqs. 1 and 2. In this case, no effects of introducing a MEMS device are considered. As a consequence, the shape of the hysteresis response between the stress and strain obtained in Fig. 7 is similar to previously obtained results found in literature for human arterial walls (Salvucci *et al*., 2001; Schmalholz and Podladchikov., 2001; Zhang, 2005; Craiem, *et al.*, 2008). Furthermore, if we consider the mechanical effects of introducing our MEMS device, we can observe that the stress in the arterial wall remains without changes. Nevertheless, the strain values in the membrane compared to the artery wall values are reduced from 1x10-3 to $1x10^{-4}$. Initially this is due to the fact that the introduction of a quasi-rigid silicone material in the arterial wall decreases the elasticity of the analyzed system and as a consequence, the deformation present in the membrane is reduced although the hysteresis effect is conserved.

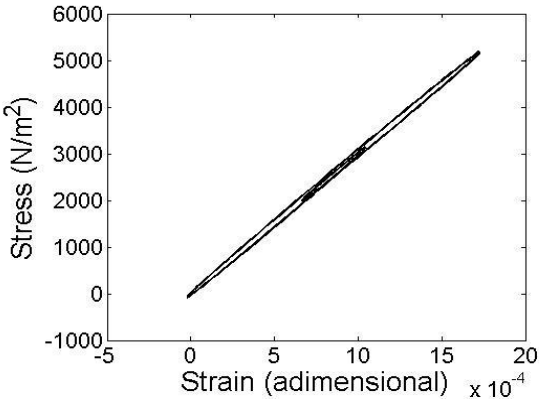

Figure8. Stress vs. Strain for the membrane of the MEMS device.

Figures 9 A) and 9 B) show with more clarity the reduction of the strain (deformation) presented in the membrane compared with the strain of the artery wall without the MEMS device. In both cases, the strain has the same shape and only differs in amplitude, although the effect of introducing deviations in the input signal is less notorious when the MEMS device is introduced (Fig. 9B). Additionally, we calculate the strain of the artery wall and the membrane of the MEMS device for different values of heart rates as it can be seen in Fig. 10, on which we could observe that the strain measurements of the membrane in the MEMS device can reach deformations of 0.0017µm, which is calculated if we consider a strain value of 0.0017 in a membrane of 10µm width, respectively. This value is sufficient to be measured with a piezoelectric component in circuit system. In fact, from Fig. 10 we can obtain the deformation sensitivity of the membrane of 0.0007 µm in a heart rate range of 60 beats/min, respectively.

III. CONCLUSION

We propose a biomedical microsystem for tonometric blood pressure based on deformations of a membrane in a MEMS device placed directly on an artery wall. We present the model to describe our blood pressure design which consists of a modified Maxwell Model and we numerically solve it for different pressure input signals, i.e., different heart rates and deviations in the shape of the input signal. We obtain that the introduction of our proposed system can reach deformation sensitivities of the membrane of around 0.0007 µm in a heart rate range of 60 beats/min, which is sufficient to be measured with a piezoelectric component in a circuit system. Also we found out that the material introduced in the artery wall, i.e., the silicone rubber, can reduce the noise presented in deviations on the shape of the input pressure signal. Our results are reproducible and contribute with new information for the improvement of novel blood pressure designs based on the tonometric principle.

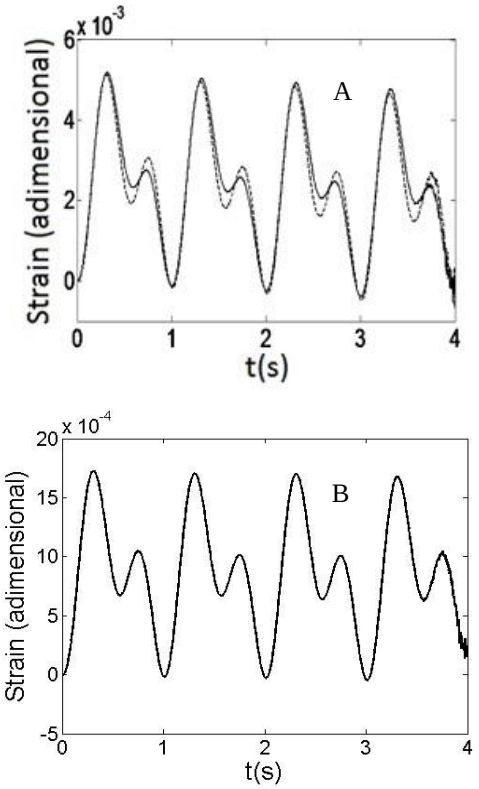

Figure 9. Strain values for two different input signals at: A) the arterial wall, and B) the membrane of the MEMS device. Solid and dashed lines correspond to the input signals shown in Fig. 6.

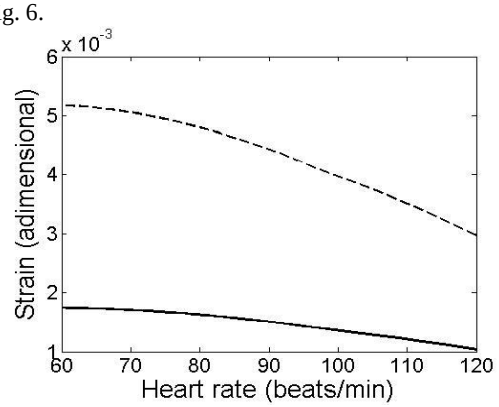

Figure 10. Strain values for the membrane (solid curve) and the arterial wall (dashed curve) for different values of heart rates.

ACKNOWLEDGEMENTS

M. F. Franco-Ayala is grateful to CONACYT, SIP-IPN and CIITEC-IPN for his MSc and PhD Studentship. D. E. Ceballos-Herrera thanks UANL for financial support. F. Martinez-Piñon, G. G. Perez-Sanchez and J. A. Alvarez-Chavez would like to thank CIITEC-IPN, SIP-IPN and CONACYT for funding their projects.

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Received: April 5, 2013
Accepted: August 26, 2013
Recommended by Subject Editor: Marcelo Martins Seckler