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A BIOIMPLANTABLE TELEMETERED BLOOD PRESSURE MEASUREMENT SYSTEM FOR TRANSGENIC MICE

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ABSTRACT

A system including an implantable integrated circuit (IC) for measuring the blood pressure of transgenic mice is implemented in 0.18 μm CMOS technology. The system specifications and circuit design challenges are summarized and discussed in this paper. The implant part of the system is to operate with a single miniature battery for more than two month. The measured data is transmitted wirelessly to a base station. The power consumption of the implantable system is 38 μW.

1. INTRODUCTION

In biomedical studies conducted on transgenic mice several biological parameters such as blood pressure, blood volume, body temperature, etc, are monitored for a relatively long period of time (several months) [1, 2]. In order to collect the data continuously during this period, the measuring device should be carried by the mouse for the whole period of study. Therefore, the measuring device should be as small as possible compared to the size and weight of the mouse, and should have the least impact on its normal life. The state of the art devices can be implanted inside the body of the mouse. The power consumption of the implantable part is a major concern since it directly affects the life time of the implant. In order to allow the mouse to have full normal activity, the measured data should be transmitted wirelessly to the outside of the body using an RF transmitter. The transmitted signal is then detected by a receiver in the base station and the biological parameters are extracted.

In this paper we describe a system implemented to measure the blood pressure of transgenic mice. In section 2, the system specifications and design challenges are described. MEMS based sensors often have non-ideal behavior. In section 3 the specifications of the sensor for blood pressure measurement are outlined, and a circuit technique to compensate sensor behavior is presented. Section 4 is dedicated to a system level technique to reduce the power consumption. The integrated circuit of the implant is discussed in section 5 and simulation results are presented. Finally, section 6 concludes the paper.

2. SYSTEM SPECIFICATIONS

The system has two parts: the implant and the base station. The implant uses a miniature MEMS strain gauge sensor which will be placed in the left ventricle of the heart to measure the blood pressure. A miniature battery supplies the energy for the implant. Using this battery the implant should be able to operate for more than two months. Therefore, the low power consumption of the IC is crucial. The role of the IC is to measure the blood pressure and transmit it to the base station wirelessly. The implant is capable of sending 2000 Samples/S which is approximately 200 times the heart beat rate of a mouse [1]. At the base station, the transmitted signal is detected and processed to extract the blood pressure. The data will then go to a data logger or PC for recording or further processing. The base station is implemented using off-the-shelf components since there is no limitation on its power consumption, size, and weight. However, the implant should be custom designed owing to stringent design constraints. Therefore, in this paper, we will focus our discussion on the implant part of the system.

The weight and volume of the implant should be kept less than 3.4 grams and 1.76 cm^3, respectively. This is due to the typical size of a transgenic mouse. The weight of 3.4 grams includes the sensor, the battery, the IC, and the antenna, with the battery being the heaviest part. Considering the commercial miniature batteries, the maximum allowable implant current should be less than 100 μA. Low power circuit and system level techniques are implemented to achieve power consumption target. The implant should be capable of working over a temperature range from 32 ^{o}C to 42 ^{o}C.

Figure 1 shows the block diagram of the implant. As can be seen in this figure, the sensed signal is passed to the signal conditioner. Subsequently, the signal is amplified, filtered, and transmitted. In this approach, the data is transmitted by analog means. This analog approach consumes less power compared to the digital counterpart where the signal must be digitized before transmission.

3. SENSOR SPECIFICATIONS AND COMPENSATION

The pressure sensor used in this application is a MEMS strain gauge. It has two resistors on a tiny steel plate. The values of the two sensor resistors change differentially (in opposite directions) as a result of a pressure exerted on the steel plate.

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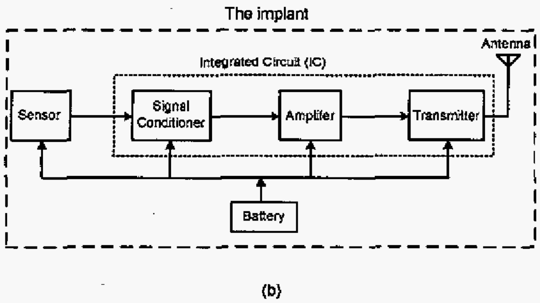

Fig. 1. Block diagram of the implant.

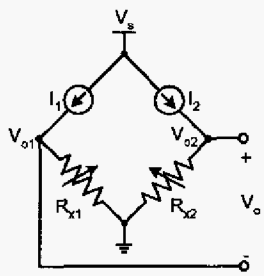

Fig. 2. Wheatstone bridge with current sources.

3.1 Sensor Specifications

The basic specifications of the strain gauge are as following:

(i) Nominal sensor resistance: 10,000 Ohm
(ii) Gauge factor (GF): 70-80
(iii) Total resistor tolerance: +/- 10-15 %
(iv) Maximum resistor mismatch of a sensor: 2.4%
(v) Temp. coefficient of resistance: +900 ppm/ ^{O}C
(vi) Temp. coefficient of gauge factor: -1440 ppm/ ^{O}C
(vii) Breakdown voltage: 20V

As can be seen, this sensor suffers from some non-idealities: (a) The resistance mismatch between the two resistors on a given sensor, (b) the temperature coefficient of the resistors (TCR), and (c) the temperature coefficient of the gauge factor (TCGF). Since the resistance value, and gauge factor are both temperature dependent, the compensation becomes a difficult task.

3.2 Thermal Compensation

The signal conditioner measures changes in resistor values and produce a corresponding differential voltage or current. A Wheatstone bridge configuration, shown in Figure 2, is employed for measurements. Two non-measuring resistances are replaced by two current sources, I_1, and I_2. This arrangement has two main advantages: (a) Current sources can be implemented using MOS transistors efficiently. (b) A Current source provides flexibility to deal with the non-idealities of the sensor.

Two sensor resistors, R_{x1} and R_{x2}, shown in Figure 2 can be represented by the following equations.

$$R_{x1} = R_{01}(1+GF.x) \qquad (1)$$

$$R_{x2} = R_{02}(1-GF.x) \qquad (2)$$

In equations (1) and (2), x represents the strain applied to the sensor and GF represents the gauge factor of the strain gauge sensor. R_{01} and R_{02} are the values of R_{x1} and R_{x2} at zero strain ($x=0$). Two different values are assigned for R_{01} and R_{02} to represent the mismatch between them. In the ideal case where there is no mismatch $R_{01}=R_{02}$. Equations (1) and (2) do not exhibit the thermal impact. Hence, these equations are modified as follows.

$$R_{x1} = R_{01}(1+TCR.\Delta T)[1+GF_0(1+TCGF.\Delta T).x] \qquad (3)$$

$$R_{x2} = R_{02}(1+TCR.\Delta T)[1-GF_0(1+TCGF.\Delta T).x] \qquad (4)$$

In above equations, TCR and $TCGF$ represent the temperature coefficient (TC) of resistor and gauge factor, respectively. To simplify the notation, in the following discussion we consider $k_1=TCR$ and $k_2=TCGF$. The currents of the two current sources in Figure 2 are also temperature dependent. Hence, we represent the current sources by following equations.

$$I_1 = I_{01}(1+TCI.\Delta T) = I_{01}(1+k_3.\Delta T) \qquad (5)$$

$$I_2 = I_{02}(1+TCI.\Delta T) = I_{02}(1+k_3.\Delta T) \qquad (6)$$

where k_3 = TCI, is the temperature coefficient of the current. Voltages V_{01} and V_{02} can be found utilizing equations 3-6.

$$V_{O1} = R_{x1}.I_1 = R_{01}.I_{01}.(1+k_1.\Delta T).(1+k_3.\Delta T).[1+GF_0(1+k_2.\Delta T).x] \qquad (7)$$

$$V_{O2} = R_{x2}.I_2 = R_{02}.I_{02}.(1+k_1.\Delta T).(1+k_3.\Delta T).[1-GF_0(1+k_2.\Delta T).x] \qquad (8)$$

The final output voltage V_O is equal to V_{O2}-V_{O1}. Manipulating the above equations and ignoring the second order effects (e.g. the terms containing $k_1.k_2$, etc) we obtain the following simplified equation for V_O.

$$V_O = V_{O2} - V_{O1} = (1+k_1\Delta T)(R_{02}I_{02} - R_{01}I_{01}) - xGF_0[1+(k_1+k_2)\Delta T](R_{02}I_{02} + R_{01}I_{01}) + k_3\Delta T(R_{02}I_{02} - R_{01}I_{01}) - xGF_0k_3\Delta T(R_{02}I_{02} + R_{01}I_{01}) \qquad (9)$$

The first term on the right side of equation (9) is due to the mismatch between the two sensor resistors. In order to compensate the mismatch, we adjust two current sources such that $R_{02}I_{02} - R_{01}I_{01} = 0$. In this case the output voltage is simplified to

$$V_O = -xGF_0[1+(k_1+k_2+k_3)\Delta T](R_{02}I_{02} + R_{01}I_{01}) \qquad (10)$$

According to the equation (10), in order to compensate the impact of the temperature on the output voltage, one should adjust the temperature coefficient of the current sources such that:

$$k_3 = -(k_1+k_2) \qquad (11)$$

This shows that using current sources with a specific TC instead of resistors in the Wheatstone bridge can help to eliminate the impact of temperature on the sensor. In CMOS technologies, a current source can be designed with a specific TC [3, 4]. Therefore, we can compensate the thermal impact on the sensor using appropriate current source.

4. REDUCING IC POWER CONSUMPTION

As mentioned earlier, the implant part of the system has a very stringent power consumption requirement. We have added a sleep mode to reduce the power consumption.

Figure 3 shows the block diagram of the implant with average power consumption per block. The voltage regulator provides a stable voltage of 1.0 V over a battery voltage range of 1.55 to 1.2 V. The timing control block controls the sleep/wake up timing of sensing block and the transmitter. The voltage regulator and the timing block consume a total average current of approximately 26 μA. The sensing block consumes a total average current of 70 μA. The transmitter takes an average current of 32 μA.

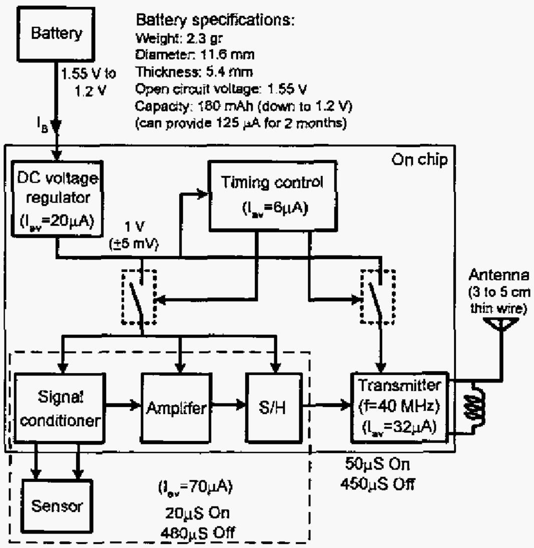

Fig. 3. The block diagram of the implant.

Considering that the IC operates continuously and ignoring the timing block, the total current drawn from the battery is approximately 122 μA. With this current, the battery can last for about two months. In order to increase the life time of the battery each of the blocks shown in Figure 3 are turned on for a short period of time and then turned off for the rest of the period to save power. The timing diagram of the IC is shown in Figure 4. In every period T, the sensing block is *on* for a period of T_1 (20 μs) and the transmitter is *on* for a period of T_2 (50 μs). For the rest of the period these blocks are *off*. An inactive time of 10 μs between T_1 and T_2 reduces the transmitter interference with the sensing block.

Using the timing scheme shown in Figure 4, the average current drawn by the IC over time period T is reduced to 32 μA. Hence, the battery life can be extended to more than seven months. However, the design of the base station becomes more sophisticated. This is not a major concern since the design requirements of the base station is not as stringent as that of the implant.

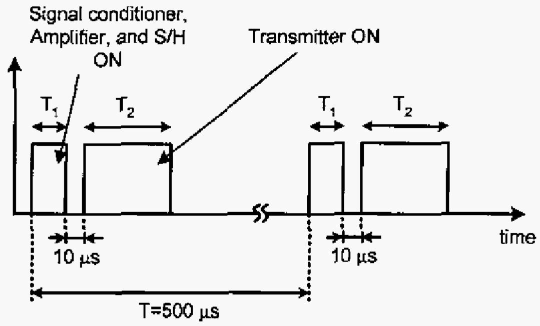

Fig. 4. The Sleep/wake up timing diagram of the IC.

5. THE DESIGNED IC

The individual blocks in Figure 3 are simulated and laid out for fabrication using the CMOS 0.18 μm technology. The main part of the signal conditioner is two current sources. The current sources are designed to have a temperature coefficient TC fulfilling the requirement dictated by equation 11. In our design, the resistors of the sensor have a TC of +900 ppm/^{o}C and the TC of the gauge factor of the sensor is -1440 ppm/^{o}C. Hence, current sources should have a TC of 640 ppm/^{o}C. Such an arrangement will have signal thermally compensated at the input of the amplifier. However, we should also consider the TC of the amplifier. In our design we have trimmed the TC of the current sources to cancel out the impact of temperature on the sensor as well as the amplifier. Therefore, current sources are designed with a TC of approximately +1400 ppm/^{o}C. In order to generate a current with the desired TC, we have used the configuration shown in Figure 5 where two current sources I_1 and I_2 with two different TCs drive a load. Form Figure 5 one can write:

$$I_L = I_1 + I_2$$

$$\frac{\partial I_L}{\partial T} = \frac{\partial I_1}{\partial T} + \frac{\partial I_2}{\partial T}$$

$$\frac{1}{I_L}\frac{\partial I_L}{\partial T} = \frac{I_1}{I_L}\frac{\partial I_1}{I_1 \partial T} + \frac{I_2}{I_L}\frac{\partial I_2}{I_2 \partial T}$$

$$\Rightarrow TCI_L = \frac{I_1}{I_L}TCI_1 + \frac{I_2}{I_L}TCI_2 \qquad (12)$$

According to equation 12 the TC of the load current is determined by the TC of the two current sources. In our circuit we have designed the current source I_1 with a positive TC and I_2 with a negative TC and combined them to get the required TC of +1400 ppm/^{o}C.

Figure 6 shows the current of one of the current sources as well as the output voltage of the amplifier with respect to temperature from 32^{o}C to 42^{o}C obtained from circuit simulations. The TC of the current of the current source is approximately +1436 ppm/^{O}C and the TC of the output voltage of the amplifier is only +45 ppm/^{O}C. Moreover, one of the current sources is designed to be digitally adjustable such that the mismatch between the two sensor resistors can be cancelled out. The amplifier of the IC is a differential pair amplifier with a gain of about 2. This gain is determined based on the full swing voltage at the output of the signal conditioner and the gain of the transmitter.

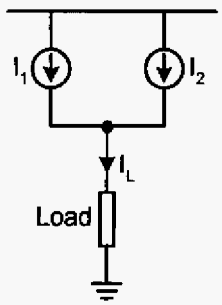

Fig. 5. Two current sources driving a load.

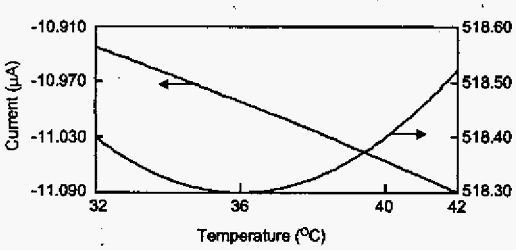

Fig. 6. The current source and amplifier outputs vs. temperature.

The transmitter is an LC tank VCO which is also used as an FM modulator. The inductor of this VCO is off the chip, however capacitances are realized using on chip MIM structures. Using off-the-chip inductor has some advantages. First, we can get inductors with a relatively high quality factor. Second, it provides means for adjusting the free running oscillation frequency of the VCO. The implant needs an antenna of about 3 to 5 cm. The length of the antenna is determined based on the experimental results. It is worth noting that in this application the base station is placed at a distance of 30 to 100 cm.

Figure 7 shows the final layout of the chip including several test structures. The post layout power consumption is simulated to be 38 μW.

6. CONCLUSION

In this paper we described a bio-implantable telemetered system for measuring blood pressure of transgenic mice. We showed how we can eliminate the non-idealities of the sensor using two current sources with specific TCs in a Wheatstone bridge configuration. The designed IC is laid out and submitted for fabrication in 0.18 μm CMOS technology. A sleep/wake up mode is applied to some of the blocks in the IC to reduce the power consumption. The implant consumes an average power of 38 μW. The simulation results show that the blood pressure can be measured with a resolution of 8-bits.

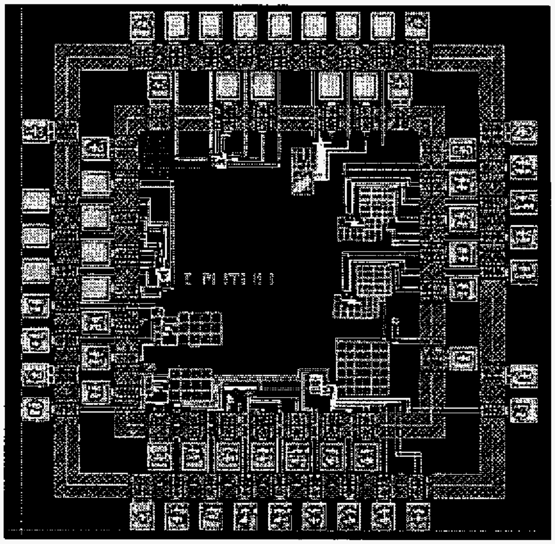

Fig. 7. The chip layout.

7. REFERENCES

[1] Demitrios Georgakopoulos, "Assessment of Cardiac Function in the Mouse by Pressure-Volume Relations with Applications to a Mouse Model of Familial Hypertrophic Cardiomyopathy", *JPhD dissertation*, Johns Hopkins University, July 2000.

[2] D. Georgakopoulos, "Estimation of Parallel Conductance by Dual Frequency Conductance Catheter in Mice," *American J. Physiology*, Heart Circulation Physiol, 279: H443-H450, 2000.

[3] Jiwei Chen; Bingxue Shi, "1 V CMOS current reference with 50 ppm/^{O}C temperature coefficient," *Electronics Letters*, Volume: 39, Issue: 2, 23 Jan. 2003, Pages: 209 – 210.

[4] Sansen, W.M.; Op't Eynde, F.; Steyaert, M., "A CMOS temperature-compensated current reference," *J. Solid-State Circuits*, Vol. 23 (3), June 1988, Page(s): 821-824.