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RF Interface for Intelligent Arterial Pressure Sensor Based in RFID Technology

Marcus Tadeu Pinheiro Silva[1], Flavio Henrique Vasconcelos[2] and Guilherme Augusto Silva Pereira[2]

[1] *Federal Center of Technological Education of Minas Gerais, Belo Horizonte, MG, Brazil. (pinheiro@ deii.cefetmg.br)*
[2] *Electrical Engineering Department, Federal University of Minas Gerais, Belo Horizonte, MG, Brazil. (fvasc@cpdee.ufmg.br)(gpereira@cpdee.ufmg.br)*

Abstract- This paper presents an intelligent wireless sensors device developed to measure blood pressure and heart rate in small laboratory animals, employing an invasive (in vivo implant) method. This sensor relies on the RFID (Radio-Frequency Identification) technology to transmit and receive data. A mathematical measurement model is derived in order to obtain the measurand (blood pressure and heart rate) best estimate and the uncertainty.

I. Introduction

In the last decades wireless technologies have experienced an enormous progress [1], in particular for use in battery powered sensor devices. However, if mobility brings benefits, the small batteries create the need for energy management and for specially designed low-power devices. If these rules are not strictly followed these benefits may become irrelevant due to replacement costs. This is even more stringent where it is not possible to have direct access to the sensors. It is especially true for wireless sensor implants used to retrieve information from animals. As a rule, power management requires some local processing, which is neither common nor a requisite for traditional sensors. Moreover, it is important to use standardized communication technology for cost reasons and for interoperability among vendors.

RFID (Radio-Frequency Identification) is an identification technique relying on RF signals to transmit information that can contribute for energy consumption reduction in the sensor subsystems. In the last few years this identification technology has spread to many fields[2].

In this paper it is proposed an intelligent wireless sensor device for use in blood pressure and heart rate measurements in small laboratory animals employing an invasive (in vivo implant) method that makes use of RFID technology to transmit and to receive data. It is also developed a mathematical measurement model [3][4] aiming to derive the blood pressure and heart-rate best estimates and the corresponding uncertainties.

II. RFID Systems

A typical RFID system is shown in Figure 1. In case the transponder doesn't have an explicit power supply, like a battery, it is called passive. Otherwise it is an active transponder. In the former case energy needed for operation is obtained from the RF signal generated by the interrogator. In active transponders, the battery is rarely used as a power supply to the RF interface subsystem. It can be used only to power memory and processor subsystems, excluding the RF interface. This approach is appealing for portable embedded systems with wireless communication, since the RF interface consumes most part of the energy in similar systems such as Bluetooth, ZigBee or IEEE802.11. In a portable embedded system the use of the RFID passive communication interface concept makes energy spending very low, and when compared to other systems where the RF stage is active, either autonomy will be improved or only a small battery would be necessary.

The main disadvantage of RFID transponders with passive interface is its very short range communication compared to systems with active interface. Furthermore, in order to maximize the communication distance, generally RFID systems has a data rate lower than systems with active RF interface.

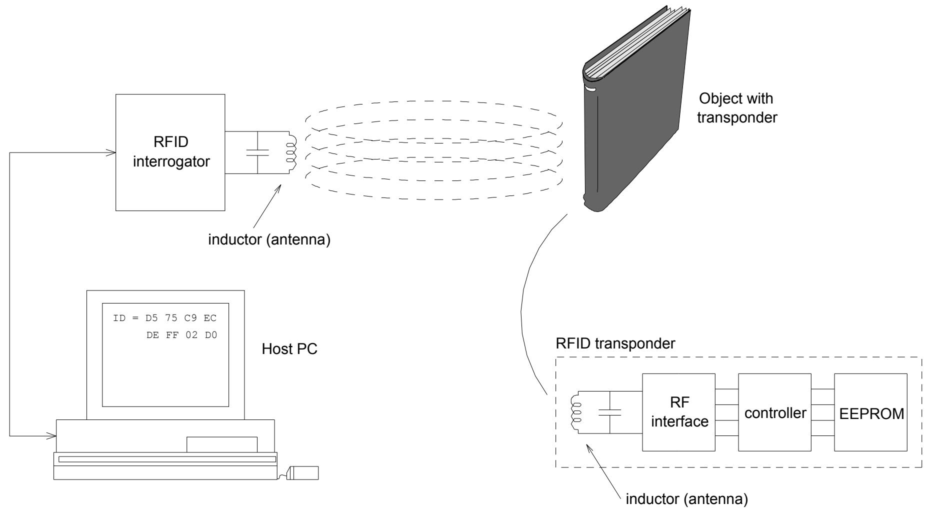

Figure 1. Typical RFID system.

The passive RF stage can be implemented based in different physical principles, such as electromagnetic wave radiation with load modulation or sequential surface acoustic wave. Figure 2 presents the most common approach in applications like smart card, animal identification, passport and public transport pass. Called inductive coupling with load modulation[5], this principle is applied in the intelligent sensor architecture under development. The upper part of this figure represents the transponder and the lower the interrogator. Whether the pointed resistor is inserted or not in the transponder LC circuit the load imposed to interrogator LC circuit is changed. In this way the information can be sent in the downlink direction without the transponder RF section drawn energy from any eventual battery present in the device.

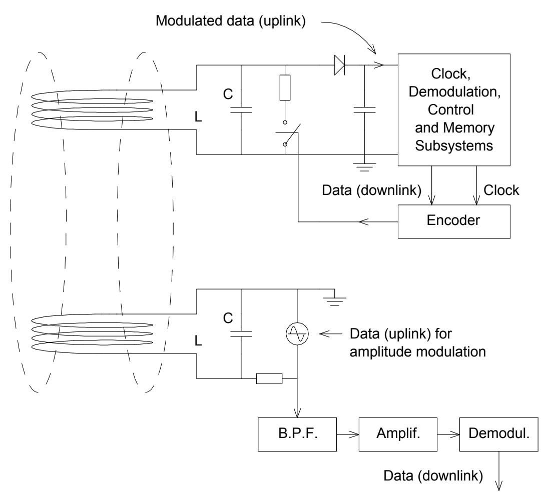

Figure 2. Inductive coupling with load modulation working principle.

III. Arterial pressure implant for small laboratory animals

Biological parameter measurement in animal model is a fundamental procedure in drug development protocols for human beings and animals [6]. Some of the most common measures have to be done in alert (non-asleep) animals and preferentially in those able to move freely. Otherwise the results can be impaired due to excessive stress originated by the measurement process. Cardiovascular parameters, as arterial pressure and heart rate, represent an example where in general the stress in the measurement process must be minimized. Indeed, there is an already commercial implant developed to do these measurements and transmitting the results by radio signals. These products have an active RF interface also in the animal side (implant). Since the power consumption in this type of interface is relatively high, the battery lifetime is limited to a few months. Moreover, the battery discharge with time affect measurement uncertainty as long this long time drift influences the system reference voltage. Most commercial systems are analogue in the animal side and there is no possibility of intelligent interaction between the implant and the remote monitoring subsystem. As an example, one can mention the case that to turn off the implant with a timer in the monitoring software the user has to move a magnet close the animal. Whenever he or she forgets to turn off the implant, the experiment can be ruined, since the battery could become flat in a few days.

Our interest is to get a biotelemetry system where the implant can be used in small rodents for extended time periods. In the proposed design, shown in figure-3, the RF implant stage has a commercial RFID transponder and includes a microcontroller. With such architecture the implant can be used much longer and present measurement uncertainty components invariables with time. Although it has been designed with focus in the arterial pressure implant it can be adapted for other applications with small effort. The pressure sensor element is a piezo-resistive commercial type, which is specifically designed to take biological pressure measures invasively. This sensor is to be attached to a vascular system by aorta cannulation in its abdominal segment. The sensor output represents the arterial pressure waveform, but it has a reduced voltage span which is not suitable for direct use with analogue to digital converters (ADC). Therefore, the signal generated in the sensor is applied to signal conditioner composed mainly by an Instrumentation Amplifier (IA). Usually, the ADC is a block embedded in the microcontroller. Through I2C serial communication the microcontroller receives commands and sends data to the RFID transponder. This transponder, which is compliant with the ISO 14443B standard [7], provides the passive RF interface, which operates in the ISM band of 13.56 MHz with inductive coupling with load modulation. All parts in the implant are low-power and work with reduced voltage (2.2V-2.5V).

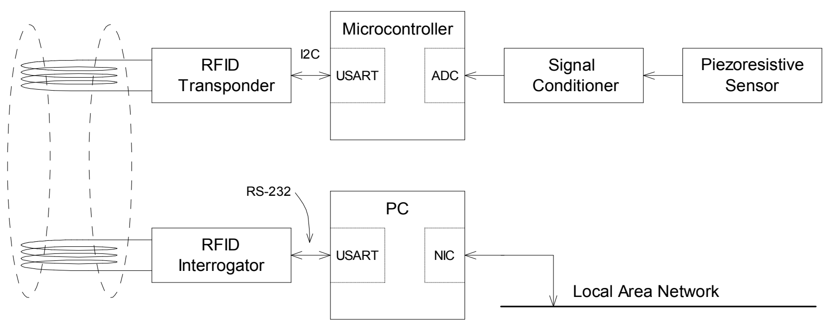

Figure 3. The arterial pressure measurement system.

The interrogator RFID in association with the transponder provides a RF interface which allows data and commands transmission between the pressure sensor and the host computer without energy consumption from the implant's battery.

IV. Measurement model for the implant

Figure 4 shows a block diagram of the arterial pressure sensor subsystem included in the biotelemetry system under development. The mechanical interface with the vascular system is a 0.4 mm diameter biocompatible tube (cannula) filled with a low compliance saline solution. One end of the cannula is introduced in the artery and the other is attached to a face of the sensor device. In the range of 40 to 200 mmHg, valid for arterial pressure, this interface can be assumed to behave linearly with unit gain. Moreover, the pressure wave in small mammals can reach a maximum rate of up to 13 Hz (800 bpm for mice) and the dynamic response of the interface (cannula) can transfer the pressure cycle to the sensor with negligible distortion. Therefore, the effect of block 1 components, shown in Figure 4 are insignificant and do not need to be considered in the mathematical measurement model.

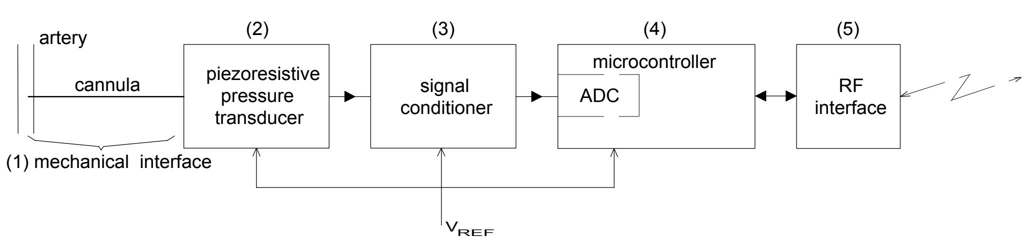

Figure 4. Implant's conceptual model.

Arterial pressure is measured with sensor devices which employ the so called load-cell [8]. The applied pressure causes a fractional deformation of a sensitive element. The mathematical relationship is given by Hook's law for elasticity. Using appropriate materials, the deformation changes the resistance of a circuit element. Making use of this knowledge a transducer can be used to convert pressure into a voltage. This is normally carried out with a Wheatstone bridge circuit configuration were at least one of the arms changes its resistance with applied pressure. For biomedical applications the transducer has four piezoresistive elements, one in each bridge arm. The piezoresistors are built up on the silicon membrane of the sensor [9]. The membrane deformation with applied pressure is transferred to each piezoresistor changing its resistance. Ideally, in this deformation one pair of resistors has a resistance variation equal and opposite in signal to the other pair. Considering that under no strain the initial pressure is zero, the pressure to Wheatstone bridge output voltage relationship can be obtained as follow.

From Hook's law, $P \propto \frac{\delta}{k}$,

where δ is the fractional resistance ratio and k is a constant in the range of 50 to 150 for silicon material. Using special circuit techniques for balancing the Wheastone Bridge, there is a correlation between the fractional resistance variation and the bridge output voltage, given by[10]:

$\delta = \frac{V_3}{A_3 V_{REF}}$.

Substituting this result in the expression above, will lead to

$$P \propto \frac{V_3}{k A_3 V_{REF}}. \qquad (1)$$

V_3 is the block 3 output voltage, V_{REF} is a stable reference voltage applied to the transducer and A_3 is the amplifier gain, if it is present at the bridge's output (block 3). The advantage here is that the pressure to voltage relationship tends to be linear [10]. Based on diagram of Figure 4, the basic mathematical model is given by:

$$P = \frac{V_4 + V_{\Delta 4}}{A_2 A_3 V_{REF}}. \qquad (2)$$

Equation (2) shows the pressure dependence on the voltage V_4, which is the digital voltage represented by a N-bits number from the Analog-Digital Converter (ADC). V_4 is related to V_2, pressure sensor output voltage, by gain factors A_3 (IA gain) and A_4 (ADC gain). A_4 ideally is equal to 1, so, $V_4 = A_3 V_2$. A_2 is a constant proportional to k, that in practical terms is the sensitivity parameter obtained from transducer's datasheet. Finally, $V_{\Delta 4}$ is a correction factor, unknown exactly, that is somehow related to the ADC resolution.

The basic model described in (2) disregards some other contributing factors. Analysing the datasheets for the devices in block 2 (MPXM2300[11]), in block 3 (AD8553[12]) and in block 4 (AD converter embedded in the MSP430[13] microcontroller) several other parameters were considered relevant to be included in the model. Equation (3) reflects a more realistic model, including other important contribution to the uncertainty of the result. It is important to emphasize that in (3) the desired applied pressure P is the output of the measuring system related to the input quantities.

$$P = \frac{V_4 + V_{\Delta 2} + V_{\Delta 3} + V_{\Delta 4}}{A_2 A_3 V_{REF}}. \qquad (3)$$

where $V_{\Delta 2}$, $V_{\Delta 3}$, $V_{\Delta 4}$, are correction factors, unknown exactly, due to respectively to block 2 (sensor), block 3 (signal conditioner) and block 4 (ADC) contributions that includes effects such as bias, offset and non linearity. It should be noted that all these input quantities are random variables, each one described by a pdf that have to be previously known. All terms in the right hand side of (3) are actually the expected value of each relevant input quantity. The measurand uncertainty can be calculated according to the Guide approach [3].

The measurement model above does not show any temperature dependence because the implant is supposed to work in a temperature controlled environment, such as the mammals body. Additionally, battery drift is almost negligible here, what causes V_{REF} be considered a constant. Consequently, equation (3) can be considered linear, making uncertainty determination easier.

V. Results

In the first phase of the development, the focus was on the interrogator. Therefore, an equivalent system to that shown in Figure 4 was designed and implemented. For the tests, a ready to use smart card based in the SRIX4K transponder [14] was chosen, which behaves like a tag fixed in an object. In these experiments the communication range was 37 mm.

In the second phase of the development, the primary goal was to establish the computational platform for the implant. Thus, blocks 1, 2 and 3 (Figure 4) were not implemented. The communication between the microcontroller in the implant and the host computer was sucessfully accomplished and the results are shown in Table 1. It is important to emphasize that during the experiments the energy drawn from RF field generated by the interrogator was sufficient to power on the implant's microcontroller. The clock frequency of this microcontroller is 2.9 MHz. This reduced version of the implant was able to capture signals applied to the ADC by an external waveform generator and send the digital signal to the application program running in the host computer. The results were obtained with the interrogator connected to a 7 cm length square planar inductor (antenna) able to deliver a RF power output of less than 100 mW.

Table 1. Results for communication with the microcontroller in the implant.

Range (mm)	36
Physical layer data rate (Kbps)	106
Application layer downlink data rate (Kbps)	6.2
Application layer uplink data rate (Kbps)	4.2

The weakness of the RFID technology is its short range. The data on table-1 show that the communication range obtained was not long enough to cover every point inside the animal cage (W=30cm x L=45 cm). In principle, there are two parameters that can be adjusted to improve these results. The first one is the antenna dimension, which can be increased. The second is the RF power generated by the interrogator, which also can be increased. The maximum range stablished in ISO 14443B systems is around 10 cm. This value is based in two points: i) the international regulation about power limit for radio emissions in the ISM band of 13.56 MHz (maximum 42 dBμA/m@10 m); and ii) the dimension of the square planar inductor (antenna) generaly used in smart card interrogators (around of 8 cm). Published results in [15] shows an extended range of 25 cm through the use of an interrogator with output power of 1.6 W connected to a circular planar inductor, 35 cm diameter. Obviously, these extended range results does not comply with the limit of radio emissions already mentioned. But, this is not a restriction if the system is operated in a specific environment where it is possible to provide screening in order to avoid generation of unacceptable interferences. In the proposed system the desired maximum distance interrogator-implant is about 15 cm. There is a practical rule for inductively coupled RFID systems which estimates that the maximum distance is approximately equal to the dimension of the square planar inductor [5]. Thus, in this case the square planar inductor should have dimension of 15 cm. Besides, the interrogator RF power would be raised; the present version has an output less than 100 mW.

The approach discussed above is not enough to solve the coverage problem. So, it is necessary to build a system with several communications cells in the cage, similarly to what is done in cell-phone systems, as illustrated in Figure 5. There is an increase in the system costs due to additional interrogators, but this is acceptable in view that this makes possible to the monitored animal to move inside the cage. Such monitoring allows to retrive data about locomotion activity, which is of interest to many biomedical experiments with animals.

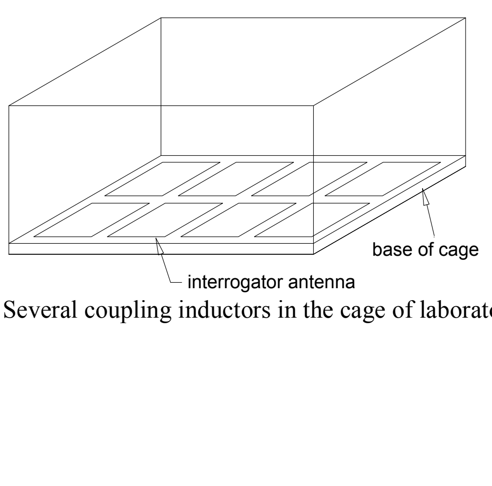

Figure 5. Several coupling inductors in the cage of laboratory animal.

VI. Conclusion

This work presented the application of the concept of passive RF interface, typical in RFID systems, in order to provide wireless communication for intelligent sensor devices. This has been done using an example of an invasive measurement system for cardiovascular parameters in small laboratory animals. The system uses off-of-shelf RFID devices which provides the RF interface for the sensor side (implant) of the system. The first results show the viability of the concept proposed.

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