Published in final edited form as:

Proc IEEE Radio Wirel Symp. 2011 ; 2011: 11–14. doi:10.1109/BIOWIRELESS.2011.5724342.

The Responsivity of a Miniaturized Passive Implantable Wireless Pressure Sensor

Hao Jiang[#], Di Lan[#], Ken Goldman⁺, Mozziyar Etemadi^{*}, Hamid Shahnasser[#], and Shuvo Roy^{*}

[#]School of Engineering, San Francisco State University, San Francisco, CA 94132

⁺H-Cubed, Inc., Olmsted Falls, OH 44138

^{*}Dept. of Bioengineering and Therapeutic Sciences, University of California, San Francisco, CA 94158

Abstract

A miniature batteryless implantable wireless pressure sensor that can be used deep inside the body is desired by the medical community. MEMS technology makes it possible to achieve high responsivity that directly determines the operating distance between a miniature implanted sensor and the external RF probe, while providing the read-out. In this paper, for the first time, an analytical expression of the system responsivity versus the sensor design is derived using an equivalent circuit model. Also, the integration of micro-coil inductors and pressure sensitive capacitors on a single silicon chip using MEMS fabrication techniques is demonstrated. Further, the derived analytical design theory is validated by the measured responsivity of these sensors.

I. Introduction

A batteryless implantable wireless pressure sensor is highly anticipated by the medical and space life sciences research community [1]. We proposed the MEMS pressure sensor in [2] and demonstrated pressure in spine [3], however, the overall system responsivity, which solely determines the operating distance between the implanted sensor and the external RF probe, was dramatically deteriorated when the size of the sensor was shrunk from 16 mm to 6 mm[4]. Further miniaturization of the sensor is needed to monitor the intraocular pressure in the eye[5]. The responsivity of the miniaturized sensor must be large enough so that it can be implanted deep inside the body.

The micro-coil inductor using the MEMS (Micro-Electro-Mechanical-System) technology is an attractive candidate to miniaturize the sensor inductor, which dominates the sensor size. The flexibility of MEMS technology brings in more design freedoms. However, the lack of design theory correlating the system responsivity to the sensor design makes it tedious to optimize responsivity for a given sensor size. In this paper, an analytical expression of the system responsivity versus the sensor design is derived based on an equivalent circuit model for the first time. Also, the resonant frequencies of miniature pressure sensors that integrate micro-coils with pressure sensitive capacitors on a single MEMS substrate are tested. Finally, the measured responsivity characteristics of these sensors is validated using the derived design theory.

II. The Theory

A sensor that consists of a pressure sensitive capacitor and a micro-coil inductor is shown in Fig. 1(b). The capacitance of the capacitor correlates to the external force due to the change of the distance between the two parallel plates. When the capacitor is paired with an inductor, a LC resonator is formed, as shown in Fig. 1(c). Its resonant frequency can be detected wirelessly via inductive coupling between the resonator and an external radiofrequency (RF) reader. Thus, the mechanical pressure on the capacitor can be estimated from the resonant frequency because the inductance of the sensor doesn't change with the mechanical loading. The RF probe has a separated orthogonal transmitter and receiver, as shown in Fig. 1(a). In this system, the transmitter and the receiver sweep frequency simultaneously. At the resonant frequency, the induced current in the sensor loop reaches its maximum, as does the current in the receiver. As long as the received signal is stronger than the background noise, the resonant frequency of the sensor can be detected in the receiver. Because the high-power transmitting signal is spatially filtered out by the orthogonal antenna, the receiver sensitivity can be maximized without concerns about its saturation[4].

A. The Equivalent Circuit Model

The LC based passive sensor system can be separated into 3 subcircuits, the transmitter loop, the sensor loop and the receiver loop. In the sensor loop, the sensor is modeled as an inductor L_S , a resistor R_S and a capacitor C_T in series, as shown in Fig. 1(c). The R_S counts for the conduction loss associated with the inductor, and the C_T is the sum of the inductor's parasitic capacitance and the capacitance of the pressure sensitive capacitor. In the transmitter loop, a RF power source is connected to a coil directly because of the broadband application, as shown in Fig. 1(e). The v_T is the source voltage of the RF power source, L_T represents the transmitter antenna, R_T is sum of the RF source resistance and the parasitic resistance of the transmitter antenna. The broadband receiver is simplified as an inductor L_R and a resistor R_R as shown in Fig. 1(d). The R_R is the sum of the receiver's load resistance and the receiver antenna's parasitic resistance. The i_T , i_S and i_R are the currents in the transmitter, the sensor and the receiver, respectively. The mutual inductance, between the transmitter and the sensor, and between the sensor and the receiver, is noted as M_{TS} and M_{SR} , respectively. Because of the orthogonal design of the transmitter and receiver antenna, the mutual inductance between transmitter and the receiver is ignored in this analysis [4].

B. The Derivation of R

Based on the above equivalent circuit model, the responsivity R , which equals to $\frac{v_R}{v_T}$ of the sensor system can be derived. The voltage and current in the transmitter, receiver and sensor loop can be expressed as the follows:

$$v_T = (R_T + sL_T)i_T - sM_{TS}i_S \quad (1)$$

$$R_R i_R + sL_R i_R + sM_{SR} i_S = 0 \quad (2)$$

$$\left(R_s + \frac{1}{sC_T}\right) i_s = sM_{TS}i_T - sL_S i_S - sM_{SR}i_R \quad (3)$$

To simplify the calculations, we assume $Z_T = R_T + sL_T$, $Z_R = R_R + sL_R$, $Z_S = R_s + \frac{1}{sC_{tot}} + sL_S$ and $M_{TS} \approx M_{SR}$. By solving the Eq. from (2) to (3), we have

$$\frac{v_T}{i_R} = Z_{TR} = Z_T + Z_R + Z_T Z_R \frac{Z_S}{\omega^2 M_{TS} M_{SR}}. \quad (4)$$

Z_{TR} is the transimpedance. Because the transmitter and receiver's antenna are well designed transmission lines, the inductance of the antenna is very small[4]. When the operating frequency is less than 100 MHz, the imaginary part of the Z_T and Z_R are negligible. Thus, the magnitude of the Z_T and Z_R is 50 and 25 (due to the power combiner [4]) Ω , respectively. Also, due to the large distance between the sensor and the antenna, $|\omega M_{TS}|$ and $|\omega M_{SR}|$ are much less than 1. Therefore, we have

$$|Z_T| \approx 50\Omega \ll \left| \frac{Z_T}{\omega M_{TS}} \right| \quad (5)$$

$$|Z_R| \approx 25\Omega \ll \left| \frac{Z_R}{\omega M_{SR}} \right| \quad (6)$$

The transimpedance in Eq.(4) can be approximated to:

$$Z_{TR} = \frac{v_T}{i_R} = \frac{Z_T}{\omega M_{TS}} \cdot \frac{Z_R}{\omega M_{SR}} \cdot Z_S \quad (7)$$

The induced voltage in the receiver v_R can be expressed as by $i_R Z_R$. Thus, the system responsivity is:

$$R = \frac{v_R}{v_T} = \left| \frac{\omega^2 M_{TS} M_{SR}}{Z_T Z_S} \right| \quad (8)$$

When the frequency reaches the resonant frequency ω_0 , the responsivity reaches the maximum as

$$R(peak) = \left| \frac{\omega_0^2 M_{TS} M_{SR}}{Z_T R_s} \right|. \quad (9)$$

C. The M_{TS}

Based on the definition of the mutual inductance [6], the induced voltage in the sensor loop caused by the transmitter can be expressed as:

$$v_s = M_{TS} \frac{di_T}{dt} = s M_{TS} \cdot i_T \quad (10)$$

The mutual inductance between the transmitter and the sensor M_{TS} is determined by the geometrical design of both the transmitter antenna and the inductor in the sensor, and their relative geometrical arrangement, which includes both the distance and the orientation. M_{TS} can be difficult to calculate without solving the 3D Maxwell's equations. Since we only focus on the design of the sensor itself, it is possible to establish a simplified expression for M_{TS} . We assume that the magnetic flux density B_T generated by the transmitter current i_T is uniform across the entire inductor coil of the sensor, because the out-dimension of the inductor coil d_{OUT} (4 and 8 mm in our case) is much smaller than one-eighth of the

wavelength ($\sim \frac{1}{8} \frac{3 \times 10^8 [\text{m/sec}]}{100 [\text{MHz}]} = 37.5 \text{ cm}$). From Faraday's law, we have

$$v_s = \frac{d(\oint B_T \cdot dS)}{dt} = s B_T \cdot S_{ind} = s(k_{TS} i_T) \cdot S_{ind} \quad (11)$$

Here, S is the area enclosed by the inductor coil. The k_{TS} is a constant to describe that the magnetic flux density and B_T is proportional to the transmitter current i_T . The S_{ind} is the total inner core area of an inductor. It can be estimated as the following if we simplify a spiral inductor to a set of concentric circular rings as described in [7].

$$S_{ind} = \sum_{n=1}^n \pi r_n^2 = \sum_{n=1}^n \pi \left(\frac{d_{out}}{2} - (n\omega + (n-1)s) + \frac{\omega}{2} \right)^2 \quad (12)$$

Here, the r_n is the inner radius of each ring, d_{out} is the outer dimension, ω is the metal width, s is the spacing between the metals, and n is the total number of turns of the inductor. Thus, the mutual inductance M_{TS} between the transmitter and the sensor can be expressed from the Eq. (10) and (11)

$$M_{TS} = k_{TS} \cdot S_{ind} \quad (13)$$

Eq. (13) shows that from the sensor design point of view, the responsivity of the sensor system is proportional to the total inner core area S_{ind} at a given distance, the relative orientation and the medium between the transmitter and the sensor.

D. M_{SR}

In the receiver loop, the induced voltage in the receiver v_R by the current in the sensor i_S can be expressed as:

$$v_R = s M_{SR} \cdot i_S \quad (14)$$

The v_R can also be derived from the Faraday's law:

$$v_R = s \psi_R \quad (15)$$

Here, ψ_R is the total magnetic flux intercepted by the receiver antenna. The ψ_R is also proportional to the magnetic flux generated by the sensor ψ_S . The definition of the inductance suggests that the ψ_S equals the $L_S i_S$. Therefore, the Eq. (15) can be expressed as:

$$v_R = s k_{SR} L_S i_S \quad (16)$$

Equating the Eq. (14) and (15), we have

$$M_{SR} = k_{SR} L_S. \quad (17)$$

From the sensor design point of view, the mutual inductance between the sensor and the receiver M_{SR} is proportional to its inductance.

E. The Summary of R

The Eq. (13) and (17) show the relationship between the sensor design and the mutual inductance. After we input Eq. (13) and (17) into Eq. (9), we have:

$$R(peak) = \left| \frac{\omega_0^2 (k_{TS} S_{ind}) (k_{SR} L_S)}{Z_T R_S} \right| \quad (18)$$

Since we only focus on the sensor design, we lump a number of terms in the Eq. (18) into a constant k . Therefore, the impact of the sensor design on the responsivity R can be described as:

$$R(peak) = k \left| \frac{\omega_0^2 A_{ind} L_S}{R_S} \right| \quad (19)$$

In a series RLC circuit as shown in Fig. 1(c), we have

$$\omega_0 = \frac{1}{\sqrt{L_S C_T}}; Q = \frac{1}{R_S} \sqrt{\frac{L_S}{C_T}}. \quad (20)$$

Thus, the responsivity R can be rewritten as:

$$R(peak) = k \omega_0 Q S_{ind} \quad (21)$$

Thus, the responsivity of this passive wireless sensor system is expressed by the fundamentals of a LC resonator, Q , ω_0 and S_{ind} . The issue of how to maximize the Q and ω_0 for a given size has been widely reported in many RFIC papers, *e.g.*, [8]. More specifically, to achieve high responsivity, we need to consider the following with respect to the sensor design:

- A high operating frequency favors the responsivity, however, signals at high frequency are more likely to be absorbed by the conductive bio-tissues[9]. Also, the Q could be degraded at high operating frequency due to the increase of the inductor resistance caused by the skin effect.

- A inductor with a large number of turns will help the responsivity due to large S_{ind} . However, more turns could result in higher inductor resistance and lower Q .
- Increasing the sensor's Q is always helpful for a given size and operating frequency.

III. Experiments

The sensor, which includes a pressure sensitive capacitor and an on-chip micro-coil inductor, is fabricated on a high resistivity Si wafer. The inductor coil is made of 20 μm thick copper that is deposited by the electroplating. The circular spiral inductor design is adopted here to achieve maximum Q for a given d_{out} and S_{ind} . The cross-section of the inductor is depicted in Fig. 2. A 20 μm wide and 0.5 μm thick copper line, which is sandwiched by 2 μm thick SiO_2 , is used as the underpass to connect the inductor's inner port to the pressure sensitive capacitor. The inductor region is passivated by 30 μm thick hardened polyimide on the top. Meanwhile, the Si underneath the inductor is completely etched away to minimize the inductor's substrate capacitance and increase the Q . There are 4 sensors fabricated and tested. The details of the sensor are listed in the Table I.

A network analyzer (HP8752A) is used to measure the sensor's responsivity. The frequency response of sensor is plotted in Fig. 3 after the transmission loss between the antenna and the input/output port of the network analyzer is calibrated out. The peak of each frequency response denotes the peak responsivity R_{peak} . The Q of the sensor is extracted from the ratio of its resonant frequency and its 3dB bandwidth.

To verify the analytical model of the responsivity in Eq.(21), the extracted $\frac{R_{peak}}{\omega_0 \cdot A_{ind}}$ versus Q is plotted in Fig. 4. The Fig.4 indicates that the sensor's responsivity at a given resonant frequency and the total inner core area is proportional to the quality factor Q of the LC resonator. This observation is consistent with the analytical model derived above. Therefore, improving the Q of the sensor inductor is the key to achieve high responsivity of this passive wireless sensor system.

IV. Conclusions

An analytical model of the responsivity in a passive LC based sensor system is established in this paper for the first time. A miniature passive pressure sensor with the micro-coil inductor fabricated by MEMS technology is also demonstrated. Both theoretically modeling and physical experiments suggest that the responsivity of an implantable LC resonator based passive wireless sensor can be improved by increasing the Q of the sensor for a given sensor size.

Acknowledgments

Dr. Rishi Kant and Neel Shad of University of California at San Francisco made numerous helpful suggestions. This work has been partially supported by NIH #1R43NS052939-1 and UNCFSP's NASA Science and Technology Institute grants. The authors would like to thank our corporate partners for allowing us to use their equipments.

References

1. Simons R, Hall D, Miranda F. Spiral chip implantable radiator and printed loop external receptor for RF telemetry in bio-sensor systems. IEEE Radio and Wireless Conference. Sep.2004 :203–206.
2. Roy S, Ferrara LA, Fleischman AJ, Benzel EC. Micro-electromechanical systems and neurosurgery: A new era in a new millennium. Neurosurgery. 2001; 49(4):779–798.
3. Ferrara L, Fleischman A, Togawa D, Bauer T, Benzel E, Roy S. An in-vivo biocompatibility assessment of MEMS materials for spinal fusion monitoring. Biomedical Microdevices. Dec; 2003 5(4):297–302.
4. Talman J, Fleischman A, Roy S. Orthogonal-coil RF probe for implantable passive sensors. IEEE Trans Biomed Eng. Mar; 2006 53(3):538–546. [PubMed: 16532781]
5. Collins CC. Miniature passive pressure transducer for implanting in the eye. IEEE Trans Biomed Eng. Apr; 1967 14(2):74–83. [PubMed: 6078978]
6. Irwin, JD. Basic Engineering Circuit Analysis. 7. John Wiley and Sons Inc; 2002.
7. Jiang H, Lan D, Shahnasser H, Roy S. Sensitivity analysis of an implantable LC based passive sensor. The 3rd Intl Conf on BioMedical Engineering and Informatics (BMEI'10). Oct.2010 :1586–1590.
8. Yue CP, Wong SS. On-chip spiral inductors with patterned ground shields for Si-based RF IC's. IEEE JSSC. May; 1998 33(5):743–752.
9. Christ A, Samaras T, Klingebrock A, Kuster N. Characterization of the electromagnetic near-field absorption in layered biological tissue in the frequency range from 30 mhz to 6000 mhz. Physics in Medicine and Biology. Sep; 2006 51(19):4951–4966. [PubMed: 16985280]

Fig. 1.

(a) The picture of the RF reader with an orthogonal antenna; (b) The microscopic top view of a fabricated sensor; (c) The equivalent circuit model of the sensor; (d) The equivalent circuit model of the receiver; (e) The equivalent circuit model of the transmitter.

Fig. 2.
The cross-section of the sensor inductor

Fig. 3.

The responsivity of the sensor when the distance between the sensor and the antenna is 1-cm.

Fig. 4.
The peak responsivity divided by its resonant frequency ω_0 and total inner core inductor area S_{ind} versus its Q

TABLE I

The fabricated sensors

Sensor (Number)	d_{out} mm	s μ m	n	C_S pF
#1	4	20	35	1.75
#2	4	20	35	1.70
#3	4	10	46	1.35
#4	8	20	98	0.14