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Electromagnetic fields: from dosimetry to human health

Investigation on an RFID planar coil for a wireless communicative aortic stent

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ABSTRACT

This paper presents an investigation on a potential biomedical RFID planar coil operating at 13.56 MHz for measuring the blood pressure level, mounted inside an abdominal aortic aneurysm. The results show that a wireless communication is achieved in a perimeter of 60 cm, while respecting the ISO 15693 standard for vicinity contactless cards. Measurement and electromagnetic simulation results show that the inductive coupling is deteriorating if the planar coil is in direct contact with water without correctly encapsulating this coil. An electric model is given for our proposed RFID planar coil immersed in water.

1. Introduction

Nowadays, the use of the radiofrequency identification (RFID) in wireless applications, such as in telemetry and telemedicine, is increasing considerably in the biomedical and health domains. Although the first wireless pressure sensors adapted to telemetry have been developed since at least the 1950s [1,2], their actual elaboration did not begin until the 2000s thanks to the successful combination between complementary metal oxide semi-conductors (CMOS) and technologies based upon micro-electromechanical systems (MEMS) [3–8] or to the combining of laminated sheets of copper-clad liquid crystal polymer (LCP) with expanded polytetrafluorethylene (PFTE) [9]. The realizations that seem more mature are those related to the sensors that are implemented to measure body temperature or aortic/ocular pressure [10,11]. For measuring the blood pressure level inside an aorta by wireless-link technology, two RFID tags, designed to work within the specific radiofrequency bands of the industrial, science and medical (ISM) [12], seems to be privileged. In the first band, one uses antennas working in the ultrahigh-frequency (UHF) band around 2.45 GHz, since these frequencies allow a compatibility with other communication standards, such as Bluetooth and ZigBee. In the second band, one uses planar coils working in the high-frequency band around 13.56 MHz in order to be able to communicate with other communication standard means through magnetic coupling. Thus, for this second band, the electromagnetic wave absorption by the body organs is minimized while maintaining a read distance of at least 10 cm to 1 m between the transponder (tag RFID with the chip) and the reader (external magnetic loop). In all cases, it is necessary to respect the rules given by the International Organization for Standardization (ISO), such as ISO 14443 or ISO 15693. RFID tags are classified, according to the presence of an energy supplier, as passive, semi-active or active. Also, it is evident that in order to ensure a good autonomy, it is essential to use lightweight battery.

In this paper, we present a cheap passive planar coil RFID that we have developed during our participation in the French project “Endoprothèse communicante” (ENDOCOM) [13] supported by the French National Research Agency (ANR). This

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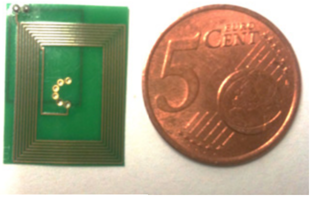
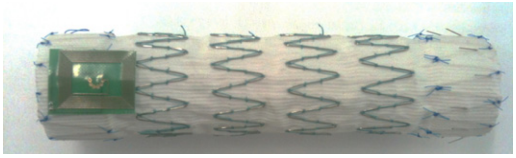


(a)



(b)

Fig. 1. (a) Planar coil compared at 5 cent euros coin, (b) planar coil on the aortic stent.

project aims to develop a communicating endoprosthesis to survey the blood pressure level evolution inside an abdominal aortic aneurysm (AAA) by means of non-invasive monitoring. Contrary to the device proposed in [14], in which an aortic stent is considered as a possible dipole antenna working at 2.4 GHz, the system proposed here is a wireless sensor network operating at 13.56 MHz and respecting the ISO 15693 standards for vicinity contactless cards. In the first part of our work, an electrical circuit model for the RFID planar coil is proposed through the use of an extraction method based on impedance measurement. Thus, the intrinsic capacitance parameter C_i of the required pressure sensor operating at 13.56 MHz is then determined. The value of the intrinsic resistance R_i of the required pressure sensor is chosen as a function of the minimal value of the Q factor that respects the restrictions imposed by ISO 15693 standards. The second part of our paper presents water measurement results when the RFID planar coil is immersed in water, with and without a protecting box capsule.

2. Design and measurement of the planar coil

An AAA is a pathological condition whereby a weakened artery wall may dilate up to dangerous proportions. One of the most common treatments consists in the insertion of an aortic stent graft, which is a medical device introduced into the artery lumen in order to diminish the pressure on the aneurysm wall and so avoiding its rupture. The main problem that appears when one uses the typical Credit Card RFID tag operating at 13.56 MHz around the aortic stent (which usually has a diameter of 2.5 cm) is that when the tag is bent, the equivalent electrical circuit is modified according to spatial conformation. Thus, the performances, namely in terms of resonance frequency as well as of quality factor, will be deteriorated. To avoid this problem, we proposed the RFID planar coil presented in Figs. 1(a) and 1(b), for which the curvature radius is negligible with respect to the diameter of the aortic stent. It is composed of windings having eight turns ($N = 8$) with dimensions of 1.5 cm × 1.85 cm. The conductors have a rectangular 35-μm-thick cross-section and are 100 μm wide. The distance between each neighboring conductor is 100 μm. The employed printed circuit board (PCB) is the FR4 Epoxy ($\varepsilon_r = 4.4$) with a nominal thickness of 100 μm. It is emphasized that the metallic wires trellises surrounding the stent do not influence the resonance frequency.

Fig. 2 shows the measured real impedance in free space and around the stent obtained using a vector network analyzer (VNA). The resonance frequency is the same with a resistance value of 20.6 kΩ. Typically, a pressure sensor (whose size is usually close to the millimeter) as that presented in [9,15,16] must be integrated at the middle of the RFID. The data exchange between the transponder and a reader located outside the human body becomes then possible for a sensor bias voltage V_i close to 2 V. As mentioned previously, pressure sensors can be designed either in combining CMOS and MEMS technologies or in combining LCP with PFTE technologies. In the two designs, the intrinsic electrical parameters C_i and R_i of the input impedance of an activated chip pressure sensor are of some tens of pF and kΩ, respectively. It is well known that the system can be viewed as a parallel RLC equivalent circuit, for which the total admittance $Y_T(\omega)$, the resonance frequency f_0 and the quality factor $Q_T(\omega)$ can be expressed as:

$$Y_T(\omega) = 1/R_T + \left(1 - L_p C_T \omega^2\right)/L_p \omega j \tag{1}$$

$$f_0 = 1/(2\pi R_T C_T) \tag{2}$$

$$Q_T = R_T/(\omega L_p) \tag{3}$$

with:

$$R_T = R_p // R_i$$

$$C_T = C_p + C_i$$

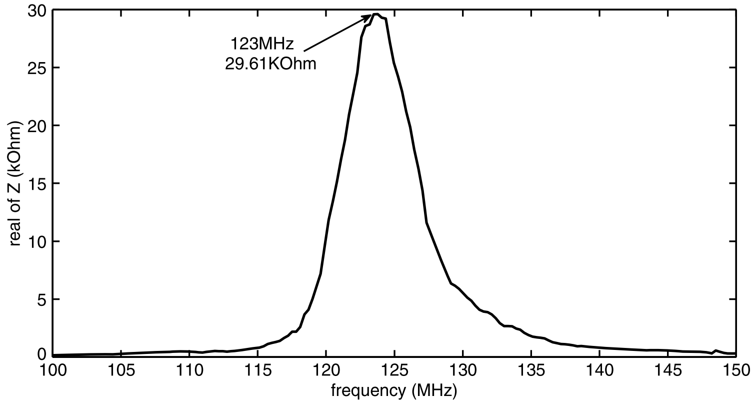


Fig. 2. Real impedance measurement of the planar coil.

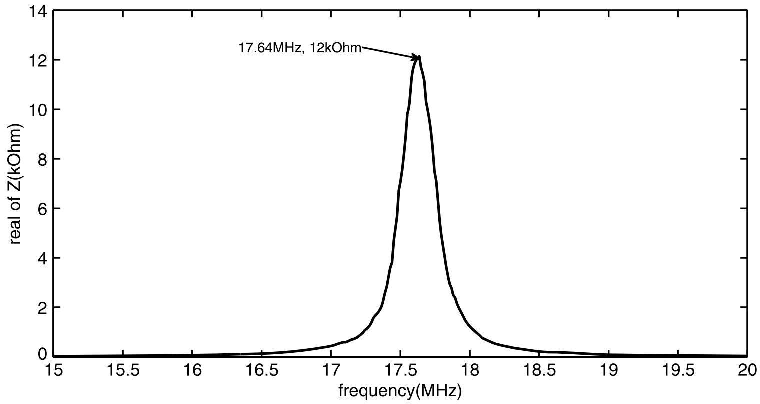


Fig. 3. Real impedance measurement of the planar coil combined with a capacitor of 51 pF.

R_p, L_p and C_p denote the electrical circuit parameters of the planar coil. Usually, the resistance R_p and the capacitance C_p are calculated precisely, while the inductance L_p is estimated through usual empirical formulas [17,18]. For example, Eq. (4) gives the inductance formula for a rectangular planar coil in nH.

$$L = 2l\left(\ln\left(\frac{l}{W}\right) - k\right)N^p \tag{4}$$

where l is the winding length (in cm), W the width of the rectangular cross section of the conductor (in cm), N the number of windings, while k and p are specific factors whose values depend on the geometry and the adopted technology used in the planar coil fabrication.

These formulas give approximate values, as they do not take into account the subtract effect and are only valid for a specific geometry and specific technology. It has shown recently that it is possible to use specific measurement impedance based on the use of a differential probe [19] to extract the device electric parameters. In our paper, an extraction method is proposed based on the use of two impedance measurements obtained using a vector network analyzer (VNA). The first measurement concerns the planar coil impedance and the second one concerns the planar coil impedance combined with a micro cms capacitor C_{add} of 51 pF. As shown in Fig. 3, the resistance value drops from 29.68 kΩ to 12 kΩ at the resonance. This effect is due to the Ohmic loss due to the weld of the cms capacitor, which is estimated to be 20 kΩ.

Thus, the intrinsic electrical parameters of the planar coil can be extracted according to the following expressions:

$$C_p = \frac{C_{add}}{(f_r/f_0)^2 - 1} \tag{5}$$

$$L_p = \frac{1}{C_p(2\pi f_r)^2} \tag{6}$$

$$R_p = \text{real}\big(Z(f_r)\big) \tag{7}$$

where f_r, f_0 denote respectively the resonance frequency of the planar coil (namely 123.8 MHz according to Fig. 2), and the resonance frequency of the planar coil with the miniature cms capacitor C_{add} of 51 pF (namely 17.64 MHz according

Table 1
Extracted electrical parameters values.

C_p	L_p	R_p
1.05 pF	1.56 µF	29.6 kΩ

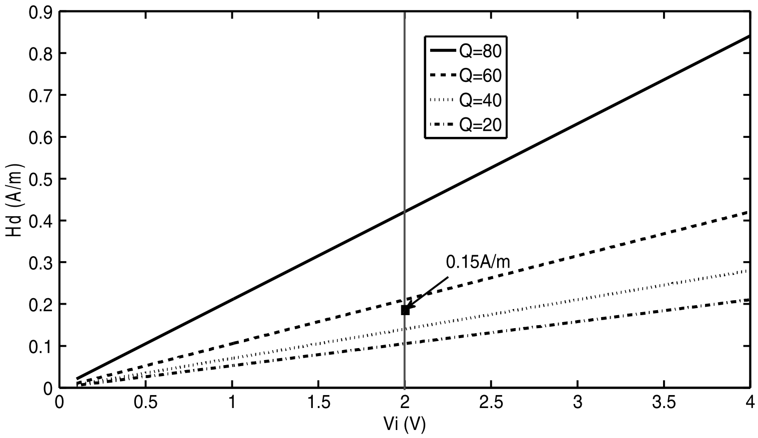


Fig. 4. Magnetic field versus V_i according to different values of Q_T.

to Fig. 3). The impedance $Z(f_r)$ represents the impedance value of the planar coil at f_r (29.6 kΩ here). Table 1 summarizes the values of the extracted electrical circuit parameters of the planar coil. The associated quality factors [calculated by (3)] are 24.5 and 69.4 for the planar coil only and for the case with the added capacitor of 51 pF, respectively.

Consequently, to work approximately at $f_1 = 13.56$ MHz, the intrinsic electrical capacitance C_i of the micro pressure sensor must be equal to 87 pF. Even if the pressure sensor has a capacitor with a smaller value, it is possible to add a micro variable capacitor to adjust the desired frequency. In the ENDOCOM project, the communications between the reader and the transponder, composed of the planar coil and the micro pressure sensor, have to respect the ISO 15693 standard for vicinity contactless cards. Thus, the operational range must be smaller than 60 cm and the minimal magnetic field applied on the active pressure sensor, $H_{d\,min}$, must be equal to 0.15 A/m. Fig. 4 shows the evolution of the induced magnetic field for different values of Q_T according to the following expression [12,15]:

$$H_d = \frac{V_i}{2\pi \mu_o Q_T f_1 N S} \tag{8}$$

where μ_0 denotes the vacuum permeability and S is the cross sectional area of the planar coil, namely, 2.775 cm^2.

It can be noticed that the magnetic field value of 0.15 A/m, which gives a bias voltage V_i of 2 V, is reached for a quality factor close to 60. With our planar coil dimensions and a suitable utilized capacitor, this quality factor value can be achieved. Fig. 5 shows the estimated values of R_i and the total input resistance R_T versus the quality factor for an operating resonance frequency of 13.56 MHz. To reach the desired qualify factor of 60, these values must be equal to 11 and 8 kΩ, respectively.

The presented results enabled us to propose in Fig. 6 an electrical circuit for an RFID telemetry system operating at 13.56 MHz. Fig. 7 compares the real part of the RFID impedance simulated using the Agilent's Advanced Design System (ADS)© to the measured ones. To achieve a capacitance value of 87 pF, two normalized cms capacitors of 68 pF and 22 pF have been connected in parallel to the planar coil in order to obtain an equivalent capacitor of 90 pF. An equivalent resistance R_i close to 10 kΩ has been added to represent the micro welds on both cms capacitors.

Another condition imposed by the ISO 15693 standard concerns the maximal magnetic field H_{max} of 5 A/m authorized for the reader (in the base station) to collect data from the sensor. In most cases, the reader is an external magnetic loop composed of one winding and whose diameter is chosen according to a desired communication distance; namely 60 cm here. Fig. 8 shows the magnetic field in the planar coil as a function of communication distance R_{com} obtained from (9) [11,14] for different diameters D of the coil reader starting from 10 to 40 cm. The diameter of the reader coil must be at least 40 cm, since the minimal desired communication distance of 60 cm is obtained while maintaining the necessary minimal magnetic fields of 0.15 A/m.

$$R_{com} = D\frac{\sqrt{(H_{max}/H_d)^{2/3} - 1}}{2} \tag{9}$$

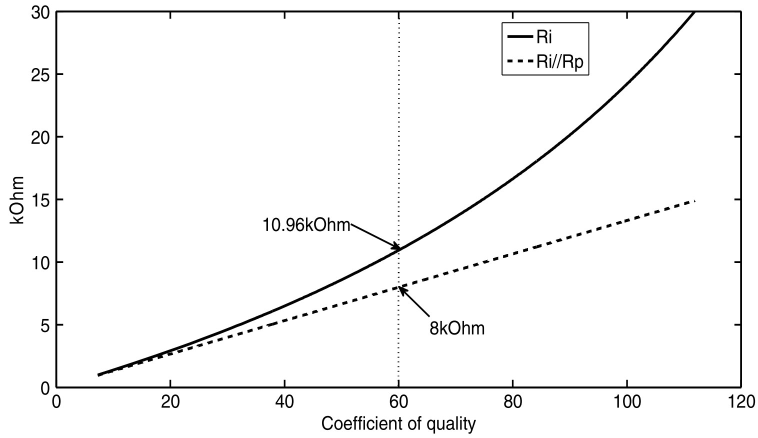


Fig. 5. R_i, R_T values for a given quality factor.

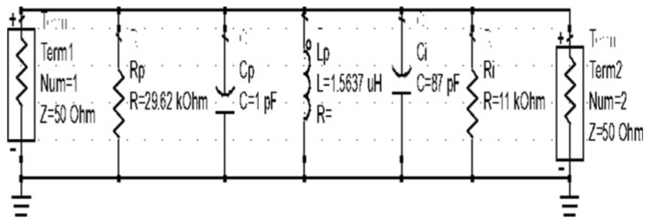


Fig. 6. Electrical circuit of an RFID telemetry system operating at 13.56 MHz.

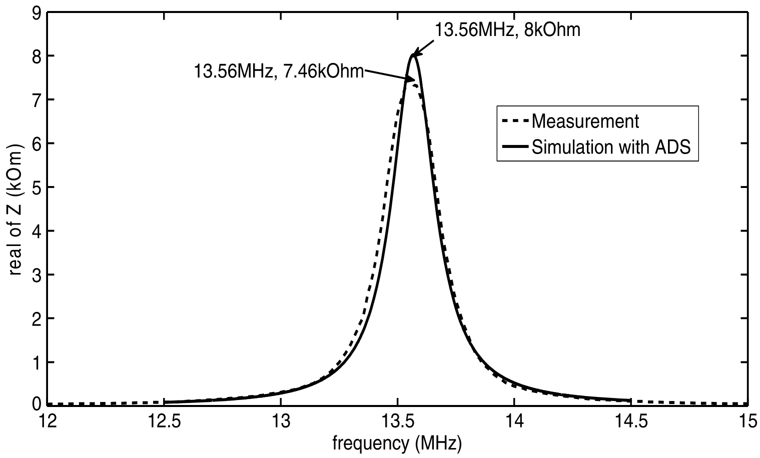


Fig. 7. Measured and simulated real impedance of the planar coil with $R_i//C_i$.

3. Water effect on the coil performances

The energy transfer between the transponder and the reader is ensured by an inductive coupling supported by a magnetic field. Water, which represents an important portion of the human body, is considered as a quasi non-magnetic media. Thus, the signal absorption at 13.56 MHz is naturally insignificant, which, as mentioned in [9], is true only if the planar coil is not in contact with water. A contact between a planar coil and water (deionized or saline) changes the effective permittivity of the coil. The resonance frequency is shifted and affects the required magnetic field sufficient to induce V_i since the quality factor is deteriorated. To illustrate this issue, different measurements according to the configurations presented in Fig. 9 have been performed. The measurements results are represented in Fig. 10. These results confirm the experimental observations mentioned in [16], namely a frequency increase more than 3 MHz and a decrease of the quality factor more than 40.

As expected, the change of the effective medium in deionized and saline water had led to a shift of the resonance frequency and to the inclusion of a resistance representing losses, whose values depend on the salinity (i.e. the conductivity).

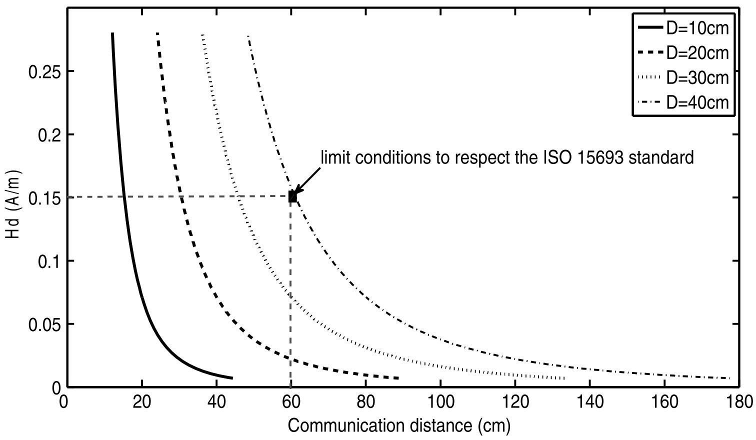


Fig. 8. Magnetic field versus communication distance for different D.

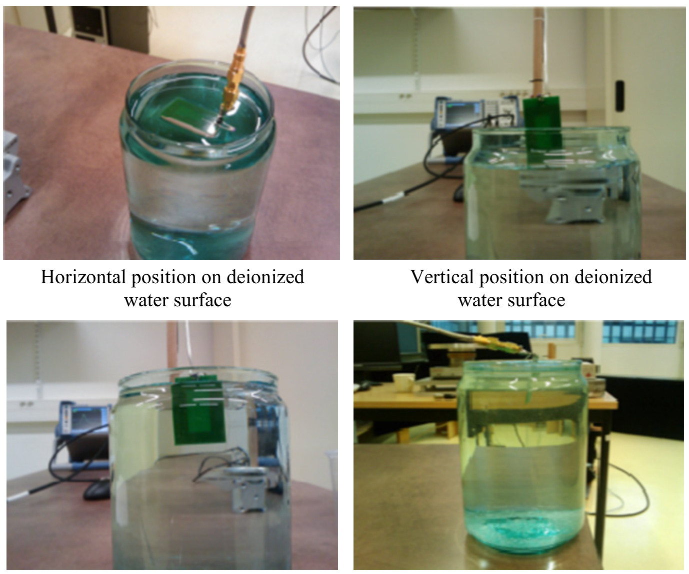


Fig. 9. Water effect on coil parameters for different studied configurations.

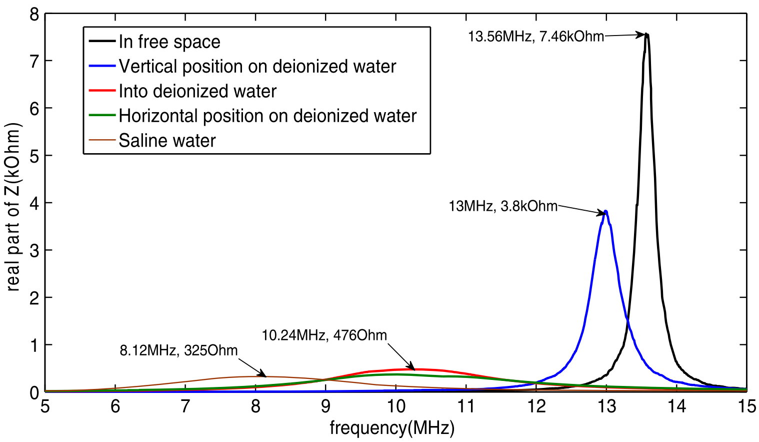


Fig. 10. Water effect on coil impedance real part according to different configurations.

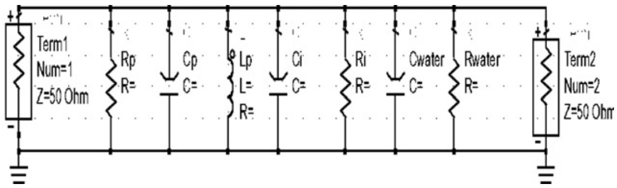


Fig. 11. Equivalent electrical circuit for an RFID planar coil into water.

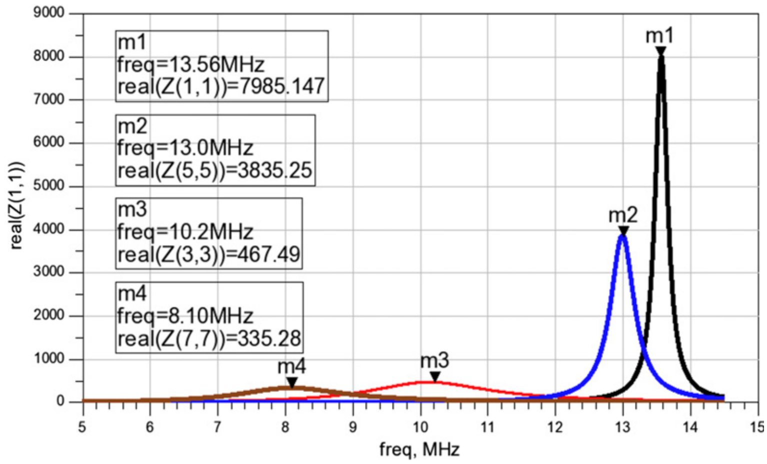


Fig. 12. Water effect on coil input resistance simulated with ADS.

Table 2
Measurement of coil parameters linked to water effect.

Configurations	Vertical position	Into deionized water	Into saline water
R_{water} (kΩ)	7.5	0.5	0.35
C_{water} (pF)	8	70	160
Quality factor	29.8	4.76	4.2

Consequently, to take into account the water effect, a parasitic capacitor and resistance are added in the previous electrical model, as presented in Fig. 11.

The electrical parameters of the water effect on the coil equivalent electric circuit are represented by the following admittance expression:

$$Y_{water} = j\omega\left(\varepsilon_0\varepsilon_r - j\frac{\sigma}{\omega}\right)\Delta$$
$$C_{water} \equiv \varepsilon_0\varepsilon_r\Delta \quad \text{and} \quad R_{water} \equiv 1/(\sigma\Delta) \tag{10}$$

where ε_0 is the dielectric constant, ε_r and σ are respectively the water relative permittivity and conductivity. The parameter Δ is a factor (in meters) that depends on the utilized geometry and technology for the planar coil.

From a parametric study performed using the ADS simulator, values of R_{water} and C_{water} corresponding to each case have been extracted. The simulation results are presented in Fig. 12 and the circuit parameters values and Q factor at each resonance frequency are given in Table 2.

To overcome this perturbation, the planar coil must be encapsulated in a biocompatible material as silicon or polyethylene. According to [9], a 300-μm-thick layer of silicone involves a shift of the frequency lower than 0.7 MHz and a 3- to 5-fold drop of the quality factor. To determinate the optimal thickness, for which resonance frequency and the quality factor do not change, when silicon or polyethylene layers are used, electromagnetic simulations have been performed using the full-wave software CST Microwave Studio. The obtained optimal thickness is about 1 mm for silicone ($\varepsilon_r = 4.7$) and 3 mm for polyethylene ($\varepsilon_r = 2.2$). To confirm this last value, we immersed, as displayed in Fig. 13, the planar coil into saline water after adding an encapsulation with a protective 3 mm-thick plastic layer having a relative permittivity similar to that of polyethylene. The resonance frequency has been slightly shifted by 0.1 MHz and the resistance has only decreased by 0.46 kΩ (Fig. 14). Then, the specified communication distance is not affected when the planar coil is placed inside a human body.

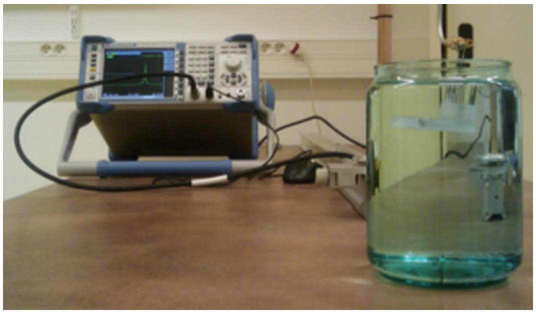


Fig. 13. Measurements set-up with the planar coil encapsulated into saline water.

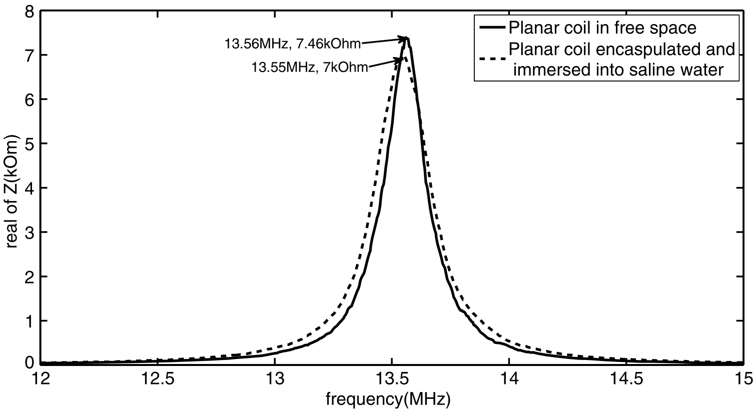


Fig. 14. Measured impedance real part of the planar coil encapsulated and immersed into saline water.

4. Conclusion

In this paper, a potential RFID biomedical planar coil respecting the ISO 15693 standard is proposed. The measurement results show that this kind of planar coil can be used with a pressure sensor to survey an AAA evolution. An electrical circuit taking into account the effect of its contact with water has been proposed to evaluate the communication distance. Thus, the influence of biological tissues containing water can be predicted. Our experimental results have confirmed that the planar coil must be necessarily coated by a biocompatible protecting layer to conserve the same performance at the required working frequency as in free space condition.

Acknowledgements

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References

[1] J.T. Farrar, V.K. Zworykin, J. Baum, Pressure-sensitive telemetering capsule for study of gastrointestinal motility, Science 126 (Nov. 1957) 975–976.
[2] C. Browning, R. Valori, D.L. Wingate, D. MacLachlan, A new pressure-sensitive ingestible radio-telemetric capsule, Lancet 318 (Sep. 1981) 504–505.
[3] M. Soma, C.D. Galbraith, R. White, Radio-frequency coils in implantable devices: misalignment analysis and design procedure, IEEE Trans. Biomed. Eng. 34 (4) (Apr. 1987) 276–282.
[4] W.J. Heetderks, RF powering of millimeter and submillimeter sized neural prosthetic implants, IEEE Trans. Biomed. Eng. 35 (5) (May 1988) 323–327.
[5] W. Liu, K. Vichienchom, M. Clements, S.C. DeMarco, C. Hughes, E. McGucken, M.S. Humayun, E. De Juan, J.D. Weiland, R. Greenberg, A neuro-stimulus chip with telemetry unit for retinal prosthetic device, IEEE J. Solid-State Circuits 35 (10) (Oct. 2000) 1487–1497.
[6] L. Ho, M. Melody, Z. Walker, T. Hamada, C.-F. Su, A prototype on RFID and sensor networks for elder healthcare: progress report, in: Proceedings of the 2005 ACM SIGCOMM Workshop on Experimental Approaches to Wireless Network Design and Analysis, Series E-WIND '05, Philadelphia, PA, USA, 2005, pp. 70–75.
[7] S. Wi Wanga, W.H. Chenb, C.S. Onga, L. Liuc, Y.W. Chuangb, RFID applications in hospitals: a case study on a demonstration RFID project in a Taiwan hospital, in: Proc. of the 39th Hawaii International Conference on System Sciences, Hawai, 2006.
[8] A. Lahtela, M. Hassinen, Requirements for radio frequency identification in healthcare, in: Medical Informatics in a United and Healthy Europe, IOS Press, 2009, pp. 720–724.
[9] M.A. Fonseca, M.G. Allen, J. Kroh, J. White, Flexible wireless passive pressure sensors for biomedical applications, in: Solid-State Sensors, Actuators, and Microsystems Workshop, Hilton Head Island, SC, June 2006.

[10] U. Schnakenberga, P. Walterb, G. Vom Bogelc, C. Krugera, H.C. Ludtke-Handjeryd, H.A. Richterd, W. Spechta, P. Ruokonenb, W. Mokwaa, Initial investigations on systems for measuring intraocular pressure, Sens. Actuators A Phys. 85 (1) (Aug. 2000) 287–291.
[11] W. Mokwa, Medical implants based on microsystems, Meas. Sci. Technol. 18 (5) (2007) R47–R57.
[12] K. FinKengeller, RFID Handbook Fundamentals and Applications in Contactless Smart Cards and Identification, 2nd ed., John Wiley and Sons, Inc., 2003.
[13] LIP6, L2E, IJLRA, INRA, INRIA, IRPHE, UTC, APHP, Orange Labs, ENDOCOM: implantable wireless pressure sensor for the follow-up of abdominal aortic aneurysm stented, J. IRBM (Ingénierie et Recherche Biomédicale/BioMedical Engeneering and Research) 32 (3) (June 2011) 163–168.
[14] E.Y. Chow, A.L. Chlebowski, S. Chakraborty, W.J. Chappell, P.P. Irazoqui, Fully wireless implantable cardiovascular pressure monitor integrated ow, with a medical stent, IEEE Trans. Biomed. Eng. 57 (6) (June 2010) 1487–1496.
[15] C.C. Colin, Miniature passive pressure transensor for implanting in the eye, IEEE Trans. Biomed. Eng. BME-14 (2) (April 1967) 74–83.
[16] A. Ginggen, Y. Tardy, R. Crivelli, T. Bork, P. Renaud, A telemetric pressure sensor system for biomedical applications, IEEE Trans. Biomed. Eng. 55 (4) (June 2008) 1374–1381.
[17] V.G. Welsby, The Theory and Design of Inductance Coils, John Wiley and Sons, Inc., 1960.
[18] D. Paret, RFID at Ultra and Super High Frequencies, John Wiley and Sons, Inc., 2009.
[19] S.K. Kuo, S.L. Chen, C.T. Lin, An accurate method for impedance measurement of RFID tag antenna, Prog. Electromagn. Res. 83 (2008) 93–106.