Integrated Bidirectional Inductive-Array Design for Power Transfer in Implantable BioMEMS †

Natiely Hernández-Sebastián 1, Francisco Javier Renero-Carrillo 1, Daniela Díaz-Alonso 2 and Wilfrido Calleja-Arriaga 1,∗

1 CD-MEMS INAOE, Puebla 72840, Mexico
2 Microtechnologies Department, Center for Engineering and Industrial Development, CIDESI, Queretaro 76125, Mexico
∗ Correspondence: wcalleja@inaoep.mx
† Presented at the 7th International Symposium on Sensor Science, Napoli, Italy, 9–11 May 2019.
Published: 8 July 2019

Abstract: This work presents a novel design of a bidirectional Inductive Power Transfer (IPT) system capable of continuous monitoring of cardiac pressure. The proposed system results from a robust electromagnetic coupling between an external reading coil and an implanted two-level (3D approach) inductor array. In this design, each coupling module follows a 13.56 MHz operating frequency, where both passive RCL networks are near field tuned. Among our main results, we obtained a Power Transfer Efficiency (PTE) of 94.1% across the 3.5 cm-thick composed biological tissue whereas the implanted coil array is about 50% of its conventional size. Since the resulting PTE efficiency is 40% higher, based on the optimized L and Q parameters, this novel approach could be used in other medical applications. This IPT system design is based on a low-cost thin film fabrication technology.

Keywords: Wireless Power Transfer; Implanted Medical Devices; Integrated Sensors; BioMEMS

1. Introduction

Recently, Implanted Medical Devices (IMD) have evolved in such extent that have given place to more possibilities for new analysis schemes in modern medical equipment, where the capability for monitoring vital signs and stimulation techniques are continuously progressing [1]. As is reported, normally these IMD’s are energy supplied using wiring lines across the body patient or by means of internal batteries. As is clearly seen, these energizing methods show poor performance because batteries require of bulky modules and the wires could lead to infection events [2]. New alternatives to supply energy consider some wireless methods, usually called Wireless Power Transfer (WPT) systems. For each implanted device a proper WPT must be carefully selected, according to some medical criteria.

IPT is one of the most popular techniques proposed for energizing a reliable IMD [3]. In this particular technique, two inductive-coupled devices are linked according to a transformer-like approach, where bidirectional power transmission is attained. Currently, IPT modules are utilized as a main technique because they present both, a high-power transmission and reduced efficiency losses caused by some misalignment [1,2]. However, most of the systems reported are composed of very complex structures and large area inductors that make them useless for implantation applications [1,4]. Also, looking for a reduction in the inductor size, some alternatives have been proposed as the use of a higher operating frequency; nevertheless, the use of higher frequencies increases the energy coupling with the biological tissue which can lead to deleterious effects in the patient [2].

Combining a low-cost thin-film surface micromachining technique and the use of polymers for flexible electronics containing an inductor-capacitor array, a new integrated L-C approach for the
design of an IPT module is proposed, offering an optimized implantable continuous monitor of the cardiac pressure for a wide 5–300 torr range. This technique considers a modular capacitor set as a pressure sensor and a two-level inductor structure for obtaining a reliable IPT system. This approach results in a lower area and maximum energy transfer across the biological tissue for obtaining a precise and low-cost IMD device, which additionally, can be adaptable for implanting in other biological application. This IPT system is capped with a robust 1.5 µm-thick polyimide film, which is part of the capacitor diaphragm and at the same time provides the required biocompatibility.

2. Wireless Pressure Sensor Description

2.1. Capacitive Pressure Sensor

The cardiac implanted device includes a capacitive pressure sensor whose touch-mode (TMCPS) dynamic sensing operation is based on an L-C passive network [3]. The TMCPS is modelled as a variable capacitor, whose capacitance is
\[ C = \varepsilon A/d, \]
where \( \varepsilon \) is the dielectric permittivity, \( A \) is the area of the contact diaphragm, and \( d \) is the composed air-solid dielectric thickness. In this approach, the diaphragm is touching the isolated lower plate according to the pressure sensed which gives the corresponding capacitance. Since in this regime \( \varepsilon \) is constant, the overall capacitance depends only on the diaphragm area contacting the lower surface. Now, according to our electromechanical model, the physiological pressure determines directly the overall mechanical characteristics for the capacitive array. After an exhaustive mechanical analysis, we observe that a single capacitor is not able to respond for the measurement of a wide pressure range; hence for a wider pressure range a higher number of capacitors is required. For this particular case, a two parallel-connected capacitors set is proposed, where the dielectric thickness \( d \) is in the 0.1 micron-thick range, using a thin composed polyimide/aluminum diaphragm, and the added capacitor area is around \( 555 \times 555 \mu \text{m}^2 \), leading to a very small two-capacitors set for sensing a 5–300 Torr pressure range. This pressure range is considered because the system is proposed for a continuous monitoring of the ventricular pressure.

2.2. Integrated Wireless Transfer System

Two L-C passive circuits, named implanted- and reading-modules are tuned to a fixed frequency, \( f_s \), in order to achieve an efficient bi-directional IPT system. Once the implanted and reading modules have been tuned, they are modelled as a two-port network. The input impedance at the reading coil can be represented according to the electrical parameters of the implanted port. Considering this two-port circuit, taking \( s = j\omega = j2\pi f \) and following the Kirkoff’s law, we obtain
\[ V_r = sL_r I_r + sL_m I_s \]
and
\[ V_s = sL_m I_r + sL_s I_s. \]
Where \( V_r \) and \( I_r \) are voltage and current across the reading module; \( V_s \) and \( I_s \) are voltage and current across the implanted coil; \( L_s \) and \( L_r \) are the inductances for the implanted and reading coils, respectively, and finally \( L_m \) is the mutual inductance. Starting at the reading module, after applying a mesh analysis the resulting equivalent impedance \( Z_{eq} \) is:
\[ Z_{eq} = \frac{V_r}{I_r} = \frac{2\pi f L_r}{1 + \frac{k^2}{1/(Q)(f/f_s)^2}} \]

In this equation, \( f \) is the excitation frequency, \( k \) is the coupling factor, \( Q = (2\pi f_s L_s)/R_2 \) is the quality factor, and \( f_s = 1/(2\pi \sqrt{LC}) \) is the resonance frequency when \( R^2 > L/C \). After analyzing this system, and according to the variations of \( Z_{eq} \), every impedance change could be induced by changes in \( k \) or \( f_s \), at the implanted module. Because this system is designed for measuring the physiological pressure, where the implanted capacitor is the dynamic sensor and the coil is a structure whose parameters are fixed, hence the only variable is \( f_s \), which depends on the capacitance variations.

Finally, every change at the sensor capacitance \( C_s \), which is influenced by the pressure changes on the surrounding biological media, determines the frequency \( f_s \). According to a bidirectional magnetic coupling, the \( f_s \) variations are sensed as changes on the equivalent impedance \( Z_{eq} \) at the reading module.

3. Inductive Coupling Link Design
Both L-C modules were physiology and electromagnetically analyzed, the coils were physically calculated considering the power transmission requirements and the critical distance separation. Next, due to the need for circular planar coils, a proper concentric alignment was considered. The electromagnetic coupling design is based on a central frequency of 13.56 MHz, according to the existing regulations to prevent tissue damage [3]. Additionally, for assuring an effective transmission signal, we considered a physiological real nucleus composed by 3.5 cm of biological tissue, see Figure 1a.

3.1. Implantable Coil

The system was designed to meet the space restrictions around the site for a secure implantation and anchoring. Hence, this restricted area conditions the use of a vertical coil approach, where a two-level winding based on the same materials used for the capacitor configuration, was required for achieving an optimum high inductance. At this site (left ventricle) the implanted module is finally composed by a capacitor-array and a double layer inductor, monolithically-integrated. They are mechanically configured with the same metal-insulator-metal stacking, which are fabricated using a low-cost and low-temperature thin-film fabrication technique [3]. An inter-digitated, aluminum-based, with two-level inductor develops a localized and effective mutual inductance whose magnitude is determined by 

\[ L = L_1 + L_2 + \frac{2M}{L_1 L_2} \]

where \( M = 2 \times k \times \sqrt{L_1 L_2} \), where \( M \) is the mutual inductance and \( L \) is the self-inductance for each winding [5,6]. After calculations under the consideration of a 2 x 2 cm² for implantation area, the vertical coil is defined by an internal diameter of 2 mm, external diameter of 18 mm, winding width of 160 µm, a selfinductance of 21 µH, electric resistance of 172 Ω, quality factor of 11.5, coupling coefficient of 0.16, and mutual inductance of 3.8 µH.

3.2. External Coil

The external coil is fabricated on a PCB FR-4 plate, attending the Finkenzeller criteria for circular windings [6], which specify \( D_{out} \leq 2X\sqrt{2} \) for the external radius and considering \( X \) as the radiation distance between both inductors across the biological tissue. After calculations, the external coil is geometrically defined by an internal diameter 2 mm, external diameter 80 mm, and winding width 700 µm; for achieving a selfinductance 21.2 µH, electric resistance 5.6 Ω, quality factor 362, and mutual inductance 3.8 µH.

4. Simulation and Results

The inductive coupling was simulated and analyzed considering the near field approximation using the mathematical tools of ANSYS HFSS software. The simulation model considers a 3.5 cm-thick of biological tissue separating both radiating ports and a 13.56 MHz operating frequency for this system. Figure 1b, shows the quality factor for a wide frequency range; as can be seen Q shows higher magnitudes at the lower frequency range, which is suitable for medical applications.

![Figure 1. (a) Scheme showing the inductive coupling across the biological tissue and (b) quality factor over a wide frequency range for both designed inductive modules.](image-url)
Figure 2a shows the magnetic flux density for this IPT system, considering a 10 cm² area and a 3.5 cm radiation distance and Figure 2b shows the transmission efficiency as a function of operation frequency; we observe that for the worst case, when $\eta = 80 \mu T$, the system shows a field with an intensity sufficiently high for an effective transmission of energy to the implanted coil.

![Magnetic flux density and transmission efficiency](image)

**Figure 2.** (a) Magnetic flux density of the external coupling module and (b) power transmission efficiency as a function of operating frequency.

Finally, the simulation scheme for the coupling across the biological tissue shows the following results: first, they allow a 94.1% PTE under the lower pressure range; second, under the higher pressure range the efficiency decreases to 72.8%. This robust and controlled coupling attenuation comes from the novel implanted L-C array where a two-capacitor array is dynamically following a wide 5–300 torr pressure range presented by the left ventricle.

5. Conclusions

A novel approach for the design of an Inductive Power Transfer (IPT) integrated system was proposed using a modular L-C array that considers optimum dimensions and adequate materials in order to cover a wide pressure range. Disregarding the L-C size, any array could be fabricated with our biocompatible low-cost and low-temperature surface micromachining fabrication process.

For implanting capacitive-based pressure sensor systems, some guidelines were proposed. For instance, the pressure range to be monitored is wide, a single capacitor is mechanically surpassed and this can be solved by combining series of complementary area capacitive-arrays for sensing ever-wider pressure range. The chosen technology is critical to achieve this goal. This work is particularly considering a very narrow physiological space such as the presented by the left ventricle. Thus, a complementary size-restricted two-level metallic inductor was developed for obtaining the highest L and Q parameters. Among the results of the simulations for the coupling through the biological tissue, we found a 94.1% PTE for the low-pressure range, and a decreased efficiency of 72.8% for the high-pressure range. This reliable bidirectional IPT integrated system can be easily adapted for several other physiological applications.

**Acknowledgments:** Natiely Hernandez Sebastián acknowledges Conacyt program scholarship #549792.

**Conflicts of Interest:** The authors declare no conflict of interest.

**References**


© 2019 by the authors. Licensee MDPI, Basel, Switzerland. This article is an open access article distributed under the terms and conditions of the Creative Commons Attribution (CC BY) license (http://creativecommons.org/licenses/by/4.0/).