On-chip implantable antennas for wireless power and data transfer in a glaucoma monitoring SoC

L. Marnat, M. H. Ouda, M. Arsalan, K. Salama, A. Shamim

Abstract—For the first time separate transmit and receive on-chip antennas have been designed in the eye environment for implantable IOPM application. The miniaturized antennas fit on a 1.4 mm^3 CMOS ($0.18 \text{ }\mu\text{m}$) chip with the rest of the circuitry. A 5.2 GHz novel inductive fed and loaded receive monopole antenna is used for wireless powering the chip and is conjugately matched to the rectifier in the energy harvesting and storage unit. The 2.4 GHz transmit antenna is an octagonal loop which also acts as the inductor of the VCO resonant tank. To emulate the eye environment in measurements, a custom test setup is developed which comprises plexiglass cavities filled with saline solution. A transition, employing a balun, is also designed which transforms the differential impedance of on-chip antennas immersed in saline solution to a $50 \text{ }\Omega$ single-ended microstrip line. The antennas on a lossy Si substrate and eye environment provide sufficient gain to establish wireless communication with an external reader placed 10 cm away from the eye.

Index Terms—on-chip antenna, implantable, intraocular pressure, sensor, wireless powering, balun

I. INTRODUCTION

GLAUCOMA, an eye disease that causes damage to the optic nerves due to high intraocular pressure (IOP), leads to progressive, irreversible vision loss. It is the second leading cause of blindness [1]. Regular eye pressure monitoring can identify the patients at risk and help start early preventive measures to avoid further eye damage. Current clinical devices such as Goldman tonometry and Pulsair pneumo tonometry do not provide continuous monitoring [2]. Additionally, in-clinic monitoring through these devices is not sufficient for at-risk patients because the IOP varies significantly throughout the day and can be substantially greater at times such as during intense physical activity or sleeping [1]. As such, a device that enables continuous IOP monitoring (IOPM) is highly desirable.

The size of the system is crucial to reduce invasive surgical procedures. Due to the heavy absorption of signals in human body, generally longer wavelengths (frequencies below 1 GHz) are used. This results in large antenna sizes, which are required for wireless communication from the implant to the external world. One way to miniaturize antennas is to use higher frequencies at the cost of larger attenuation of the signal in human body. Another way to achieve compactness is integrating the antennas with the RF circuits on the same chip,

but this also enhances losses due to the low resistivity of the typical silicon substrates ($10 \text{ }\Omega\cdot\text{cm}$) [3]. The lossy environment of the eye makes the wireless communication and the subsequent antenna design even more challenging. The only relief is the short communication range required for this particular application (10 cm) [4].

Implantable IOPM is a new area of research and not much work has been reported on this. An implantable IOPM reported in [1], has a 27 mm long antenna, unsuitable for the implant process. In [5], a wireless implantable IOPM is demonstrated but relies on inductive coupling for data transfer. Recently, a passive IOPM system has been proposed which also utilizes inductive coupling [6]. The main drawback of passive systems is that they have very limited functionality and low accuracy level. The limitation with inductive coupling is that it requires perfect alignment with an external coil [1]. This work reports two separate implantable antennas for transmit (Tx) and receive (Rx) functions in a 1.4 mm^3 active IOPM system-on-chip (SoC). The antennas have been optimized in eye environment and a custom test setup has been created to emulate the eye for measurements.

II. DESIGN CONSIDERATIONS

The block diagram of the active IOPM SoC employing $0.18 \text{ }\mu\text{m}$ CMOS technology is illustrated in Fig. 1. The chip is designed to harvest and store the energy from the incoming 5.2 GHz RF signals. The sensor's data is transmitted from the implanted chip to the external reader through an oscillator transmitter at 2.4 GHz [3]. Separate antennas are integrated on the chip to provide Tx and Rx functions. Two different operating frequencies are used to reduce complexity and achieve good isolation between the Rx and Tx paths.

Since the chip is for eye implant, due considerations must be taken for the design and subsequent characterization. A biocompatible package is necessary and must be added around the chip for a safe implant. The chip must be placed 5 mm inside the anterior chamber of the eye, as shown in Fig. 2 [4]. Due to the size limitations, integration of the two antennas on the chip is quite challenging. However, the high permittivity aqueous humor inside the eye ($\epsilon_r = 68$) aids in antenna miniaturization. On the other hand, the loss of this medium ($\tan \delta = 0.2677$) seriously affects the antenna efficiency.

Fig. 1: Block diagram of the wireless IOPM SoC

Fig. 2: Position of the IOPM SoC in the anterior chamber of the eye

According to [4], with an EIRP of 4 W, a wireless IOPM device must be able to communicate with an external reader placed 10 cm away from the eye. The gain required for the Rx antenna can then be calculated placed using Friis equation [7]. The minimum power level required for the rectifier operation in the energy harvesting unit must be $-21 \mu\text{W}$, so the gain of the Rx antenna must be equal to or larger than -20 dBi . Requirements on the Tx side are less stringent as a large gain external antenna can be used to receive the data.

III. ANTENNA DESIGN

The performance of an antenna in air is very different from the case when it is implanted in the eye. It is therefore necessary that the simulation model incorporates the eye environment. The simulation model emulating the chip embedded in the eye is depicted in Fig. 3. The eye is modeled with a plexiglass cavity filled with a saline solution and supported by a substrate. Another plexiglass cavity is placed underneath the substrate. A hole in the substrate permits the solution to be filled in the bottom cavity and ensures that the chip is completely immersed in the solution.

The CMOS stackup used to design the on-chip antennas is illustrated in Fig. 4. It comprises a low resistivity Si substrate, six metal layers embedded in a thick oxide and a passivation layer on top of M6 metal layer. An additional $10 \mu\text{m}$ film of parylene C is coated all around to make the chip biocompatible. In order to optimize the space, antennas are designed on two different metal layers, namely M4 and M6.

Fig. 3: Simplified simulation model of the chip in the eye environment

Parylene C ($\epsilon_r = 2.9$)	10 μm
Passivation ($\epsilon_r = 6, \tan \delta = 0$)	1.13 μm
SiO_2 ($\epsilon_r = 4, \tan \delta = 0$) M6	2 μm
SiO_2 ($\epsilon_r = 4, \tan \delta = 0$) M4	2.76 μm
SiO_2 ($\epsilon_r = 4, \tan \delta = 0$)	580 nm
Si ($\epsilon_r = 11.9, \text{conductivity} = 10 \Omega\cdot\text{cm}$)	5.44 μm
Si ($\epsilon_r = 11.9, \text{conductivity} = 10 \Omega\cdot\text{cm}$)	300 μm
Parylene C ($\epsilon_r = 2.9$)	10 μm

Fig. 4: $0.18 \mu\text{m}$ CMOS process stackup

The Tx and Rx antenna designs are shown in Fig. 5. The Rx antenna is designed on M6 layer while the Tx antenna is on the M4 layer. It worth mentioning here that a large area of the layer M6 is also utilized for on-chip MOSCaps to store the energy. This means that the Rx antenna cannot employ large continuous metal areas.

Fig. 5: Tx (black) and Rx (gray) antennas design on M4 and M6 respectively (dimensions in mm)

A. Transmit antenna

The sensors data is communicated through a 2.4 GHz transmitter comprising a voltage controlled oscillator (VCO). In order to optimize space, the inductor of the VCO resonant tank has been optimized to act as the Tx antenna [3]. The octagonal loop, shown in Fig. 5, is designed to provide the required inductance “L” for the tank as well as an antenna gain of higher than -20 dBi . The loop diameter and the metal width affects the L and the quality factor “Q” of the loop and are utilized to optimize VCO performance along with the radiation efficiency. It worth mentioning here that larger antenna radiation efficiency results in a lower Q and L.

The input impedance of the loop antenna is $Z_{\text{Tx}} = 7.2 +$

$j67.5 \Omega$ which corresponds to an L of 4.5 nH with a Q of 9 at 2.4 GHz. The 2D radiation pattern of the Tx antenna in the E- and H-planes is shown in Fig. 6 (a). As expected for an electrically small loop antenna, the radiation pattern is omnidirectional in the plane of the loop. A gain of -29 dBi at 2.4 GHz is obtained, primarily due to the losses of the Si substrate and lossy saline solution. However, this gain is sufficient for establishing communication at a distance of 10 cm with a high gain external antenna.

B. Receive antenna

The Rx antenna is designed to receive 5.2 GHz RF signal from an external source and feed it to the energy harvesting and storage unit. The Rx antenna needs to be matched to the rectifier circuit, which has a complex input impedance of $Z_{\text{rect}} = 11.1 - j233.3 \Omega$ at 5.2 GHz. The design of the Rx antenna must provide an input impedance of Z_{Rx} that is the conjugate of Z_{rect} . In addition, the antenna geometry should leave enough space for storage MOSCaps in M6 layer. The design must minimize the coupling with the Tx loop antenna.

A custom monopole is designed in metal layer M6, as shown in Fig. 5. An inductive feeding is used to provide the required large inductance for matching with a highly capacitive impedance of the rectifier circuit. Furthermore, an inductive load is added at the other end to reduce the antenna size. A parasitic element near the monopole helps increase the gain. The input impedance of the monopole is $Z_{\text{Rx}} = 27.4 + j60 \Omega$ at 5.2 GHz. Despite considerable inductive antenna impedance, the large capacitive part of the rectifier could not be matched completely. A matching network composed of an on-chip inductor and two capacitors is thus utilized to provide the required conjugate match. Fig. 6 presents the 2D radiation pattern and gain of the Rx antenna. It can be observed that a maximum gain of -14.5 dBi is obtained along the plane of the substrate.

(a) Tx antenna at 2.4 GHz (b) Rx antenna at 5.2 GHz
Fig. 6: Simulated radiation patterns in the E-/H-planes

IV. FABRICATION AND MEASUREMENTS

The fabricated chip wire-bonded to a printed circuit board (PCB) is shown in Fig. 7 (a). After the wire-bonding process, the chip is placed in a custom plexiglass cavity realized to emulate the eye environment, as shown in Fig 7 (b) and (c). As discussed in section III (Fig. 3), the test fixture comprising two plexiglass cavities filled with a saline solution, and separated with a PCB (duroid with $\epsilon_r = 2.2$), enables complete characterization of the implanted antennas. The saline solution

is used to emulate the aqueous humor present in the anterior chamber of the eye. To get the same permittivity and losses ($\epsilon_r = 68$ and $\tan \delta = 0.2677$), 13.45 g of sodium chloride is mixed with one liter of de-ionized water. A sticky tape is put on both cavities to seal them.

The arrangement mentioned above emulates the eye environment well, however poses a real challenge to the characterization of the antennas with conventional methods. To access the Tx/Rx on-chip antennas in the cavities, a special arrangement is required. Furthermore, to test the differential antennas with single ended SMA connectors, baluns are required.

Fig. 7: (a) Fabricated chip wire-bonded to the PCB, (b) chip in the plexiglass cavity, (c) cavity filled with saline solution and sealed with tape

A. Chip to SMA Transition (including Balun design)

The goal is to transform the complex differential antenna impedance ($Z_{\text{ant-dC}}$) into a single ended 50Ω impedance ($Z_{\text{ant-s}50\Omega}$). It is a challenging design as this transition passes through three different mediums: saline solution, plexiglass and air. The $Z_{\text{ant-dC}}$ is first transformed into real differential impedance ($Z_{\text{ant-dR}}$) through a CPS line, as shown by the section L1 in Fig. 8. The impedance $Z_{\text{ant-dR}}$ is then converted into a single ended real impedance ($Z_{\text{ant-sR}}$) through a microstrip balun, as shown by the section L2. A quarter wavelength microstrip transformer is then utilized to the targeted $Z_{\text{ant-s}50\Omega}$ (section L3).

Fig. 8: Chip to SMA transition including Balun design

Balun at these low frequencies can be large, so the best way to miniaturize it is to put it inside the high permittivity saline solution. At the same time, the balun must be designed in a single medium (either air or saline solution) to avoid reflections and impedance mis-match as it is more sensitive due to its non-symmetric nature. The size of the balun is large for the 2.4 GHz band, so it cannot fit in the plexiglass cavity. However, the 5.2 GHz balun can be accommodated in the saline solution as shown in Fig. 8.

B. Measurements

The initial antenna simulation model, shown in Fig. 3, does not include the transition from the chip to the SMA. So in order to match the measurements with this transition, the simulation model must be updated. The measured s-parameters are compared with the ones obtained from the updated simulated model in Fig. 9. A good agreement between the simulated and the measured results can be observed, especially for the Rx antenna. For the Tx loop antenna operating at 2.4 GHz, there is a shift in the resonant frequency by 400 MHz. The discrepancy can be attributed to the random metal fill done in the layers by the CMOS foundry to fulfill the metal density requirements. It can be seen that the baluns do not affect the resonance of the on-chip antennas.

Fig. 9: Simulated and measured S_{11} of on-chip antennas versus frequency

The effect of the baluns on the radiation of the on-chip antennas must be studied. When the chip is wire-bonded on the PCB, its radiation characteristics are different than the chip alone, as can be seen in Figures 6 and 10. The baluns use a ground plane that can affect the radiation of the on-chip antennas. As can be seen in Fig. 10, both the gain and the radiation patterns are deteriorated. However, this is not a concern for the design, as the transition with balun is not required in the practical application.

Fig. 10 will be updated with the measured gain and radiation patterns in the final manuscript.

Fig. 10: 3D radiation patterns of the on-chip antennas simulated when the chip is in saline solution and wire-bonded to the PCB

V. CONCLUSION

The design of on-chip antennas suitable for implantable wireless intraocular pressure monitor (IOPM) is presented in this paper. Two antennas operating at 2.4 GHz and 5.2 GHz have been designed on a 1.4 mm³ chip and optimized to be able to communicate in the lossy mediums (chip and eye) to an external reader. A specific test fixture has been fabricated to measure on-chip antennas with simple SMA connectors when it is in the eye environment. A good agreement between simulated and measured results for the impedance measurement can be observed. The simulated and measured results indicate the suitability of this design for implantable IOPM application.

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