Miniature Coplanar Implantable Antenna on Thin and Flexible Platform for Fully Wireless Intracranial Pressure Monitoring System

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Minimally invasive approach to intracranial pressure monitoring is desired for long-term diagnostics. The monitored pressure is transmitted outside the skull through an implant antenna. We present a new miniature (6 mm × 5 mm) coplanar implant antenna and its integration on a sensor platform to establish a far-field data link for the sensor readout at distances of 0.5 to 1 meter. The implant antenna was developed using full-wave electromagnetic simulator and measured in a liquid phantom mimicking the dielectric properties of the human head. It achieved impedance reflection coefficient better than $-10$ dB from 2.38 GHz to 2.54 GHz which covers the targeted industrial, scientific, and medical band. Experiments resulted in an acceptable peak gain of approximately $-23$ dBi. The implant antenna was submerged in the liquid phantom and interfaced to a 0.5 mW voltage controlled oscillator. To verify the implant antenna performance as a part of the ICP monitoring system, we recorded the radiated signal strength using a spectrum analyzer. Using a half-wavelength dipole as the receiving antenna, we captured approximately $-58.7$ dBm signal at a distance of 1 m from the implant antenna which is well above for the reader with sensitivity of $-80$ dBm.

1. Introduction

According to Monroe–Kellie doctrine, the cranial cavity is a fixed space with three components, blood, cerebrospinal fluid (CSF), and brain, and they are in volume equilibrium [1–3]. An abnormal increase in the volume of any of the components beyond the limits of intracranial compliance leads to a pathological rise in ICP. This is a medical condition known as intracranial hypertension (IH) and requires immediate medical intervention. In adults, normal ICP can be up to 15 mmHg and, in critically ill patients, the goal is to maintain ICP below 20 mmHg [1–3]. Thus, the monitoring of ICP and management of IH are both frequent and life-saving procedures.

Sustained IH can rapidly be fatal and damage the brain. Overall, IH can be induced by diverse causes. They can be categorized as intracranial (e.g., brain tumor, nontraumatic hemorrhage, ischemic stroke, and hydrocephalus), extracranial (e.g., hypoxia, hypertension, liver failure, and Reye syndrome), or postoperative complications of neurosurgery (e.g., brain lesion and edema) [1–4]. Finally, IH can also be idiopathic with unknown etiology [5].

In critical care setting, ICP is commonly monitored from a catheter inserted into the ventricular system of the brain through a burr hole (ventriculostomy). This invasive method is considered a gold standard, because it is accurate and allows on-site recalibration as well as drainage of CSF to manage IH. However, due to its invasiveness (patient discomfort) and risks of infection and hemorrhage [1, 2, 6, 7], it is nonoptimal for repeated and/or prolonged use. However, long-term ICP monitoring would improve the safety of people predisposed to IH due to an illness such as hydrocephalus [6, 8] managed with an implanted cerebral shunt valve [9] or as a result of a previous neurocritical event.

To address the need, an array of noninvasive approaches has been studied [7, 10]. However, due to limitations in...
stability and accuracy, to our best knowledge, they have not yet entered the clinical routine. A focus in the recent research on medical technology has been wireless ICP monitoring based on capacitive pressure sensors [11–13]. Generally, they do not need to be wirelessly powered, but due to the zero-power operation, the sensor readout is challenging. As a result, the monitoring systems have entailed a sensor in the cranial cavity tethered to a subcutaneous unit through the skull [11, 13]. This intrusive approach was also taken in an inductively powered piezoresistive ICP monitoring device [8, 14]. Finally, authors of [15] presented a battery-assisted ICP system based on a capacitive pressure sensor and a battery-assisted wireless transmitter integrated into a transcranial bolt.

In contrast, our approach is a fully wireless and cranially concealed implantable system based on a remotely powered piezoresistive pressure sensor on a thin and flexible substrate [16, 17]. We have already demonstrated pressure readout from the sensor through wires (without the data transmission unit) [16]. Figure 1 presents the architecture of the system. The implant has four main parts. The first of these is a 2-turn coil antenna for wireless powering, the second is a rectifier for RF-to-DC conversion, the third is a piezoresistive pressure sensor, and the forth is a data transmission unit. As a part of the data transmission unit in the implant, this work focuses only on the development and testing of the implant antenna. The novelty of this work is the development of a new miniature (6 mm × 5 mm) coplanar implant antenna on a thin and flexible platform and its integration with the implant platform. The goal is to make orientation insensitive sensor readout at distances of 0.5 to 1 meter.

2. Antenna Development and Simulation Results

Fundamental challenge in the development of implantable antennas is miniaturizing the overall size and exact form factor to minimize intrusiveness. In the literature, researchers have reported several implant antenna miniaturization techniques. These include (i) using of high permittivity dielectric substrate/superstrate materials [18–20], (ii) lengthening of the current path on the radiating patch by meandering/spiraling and/or inserting the slot/line [21–25], (iii) inserting a shorting pin between the radiating patch and the ground plane (planar inverted-F antenna configuration) [26–29], (iv) vertically stacking two or more radiating patches [30–32], and (v) loading (inductive, capacitive, or split ring) for impedance matching [33–36]. Despite the small footprint and adequate electromagnetic performance combined with dual-band operation, the above-mentioned antennas are rigid with significant thickness generally measured in millimeters. Hence, these antennas are not befitting our application where thin and flexible platform is a priority.

Recently, many antennas have been reported on flexible form or polyimide substrates [37–42]. However, all of these antennas are either large in size or two-sided structure (patch on one side and ground on the other side of the substrate). Alternatively, several authors have also shown coplanar structures viable in implantable antennas [20, 43, 44]. This approach removes the requirement on the substrate thickness. However, the demonstrated antennas still exhibit size in centimeters or use of ceramic high-permittivity materials, though the biological environment itself exhibits high dielectric constant and we are able to achieve a resonant antenna at 2.45 GHz within the footprint of 6 mm × 5 mm on regular low-permittivity polyimide. Moreover, the antenna is fully flexible, thus conforming to the physiological environment.

Given the required footprint size of 6 mm × 5 mm (due to available space on the implant) and operation frequency of 2.45 GHz, we considered planar monopole configuration a possible approach in our application. To reduce the footprint further, the antenna topology converged to an inverted-F type of a structure. A spiral structure provides lower resonance frequency and higher radiation efficiency than a meander [25, 26]. We chose a spiral structure for folding the arm of the antenna. The ground plane is coplanar with the radiating arm as shown in Figure 2.

ANSYS HFSS v15 was used to simulate the antenna performance. A 4-layer tissue model (see Figure 3(a)) was used and each tissue layer was assigned frequency-dependent properties provided in [45]. A commercially available silicone was applied to cover both sides of the antenna for better antenna efficiency and lower SAR value [46]. In the prototype manufacturing, we achieved a total thickness of one millimeter for the coated structure. However, as detailed in Section 2.1, the simulations predict that total thickness of only 0.3 mm would provide equal performance. The measured dielectric properties of the silicone at 2.45 GHz were \( \varepsilon_r = 3.3 \) and \( \tan \delta = 0.007 \). Due to the VCO size, \( "Tf" \), \( "Sf" \), and \( "W" \) were kept constant at 0.35 mm, 0.3 mm, and 3 mm, respectively. Moreover, due to available space, antenna maximum dimensions were set as \( L + Lg + Sf = 6 \text{mm} \).
Table 1: Optimized dimensions of the antenna.

| Symbol          | Value [mm] |
|-----------------|------------|
| L               | 3.55       |
| W               | 3          |
| Lg              | 2          |
| Wg              | 4          |
| Ls              | 1.9        |
| Ws              | 1          |
| fT              | 0.4        |
| fF              | 0.75       |
| sT              | 0.5        |
| Fs              | 0.3        |
| Ts              | 0.35       |
| fF               | 1.05       |
| Ts              | 0.2        |

Table 2: Simulated antenna parameters at 2.45 GHz.

| Reflection coefficient [dB] | Gain [dBi] | Input impedance [ohm] | Directivity [dB] | Radiation efficiency [%] |
|-----------------------------|------------|-----------------------|------------------|--------------------------|
| −14.07                      | −19.63     | 47.47 + 19.5i         | 4.33             | 0.5                      |

Table 3: Simulated SAR and $P_{t,max}$ for bone and brain.

| Layer   | SAR$_{max}$ [W/kg] for 0.5 mW input power | $P_{t,max}$ [mW] |
|---------|----------------------------------------|------------------|
| Bone    | 0.1446                                 | 5.53             |
| Brain   | 0.0104                                 | 7.63             |

Figure 2: Top view of implant antenna.

Figure 4 shows the simulated reflection coefficient of the antenna. The antenna is matched to 50 Ω. The simulated −10 dB bandwidth of the antenna is 280 MHz (from 2.21 GHz to 2.49 GHz) and, at 2.45 GHz, the antenna reflection coefficient is −14 dB. During the simulation, we observed that shorting path distance (“F”) has significant effect in fine-tuning of the resonance frequency. Figure 5(a) presents the E-field (V/m) distribution at 2.45 GHz. It is clear from E-field (V/m) distribution that the radiating element of the antenna is acting as a λ/2 resonator at 2.45 GHz. Figure 6 presents the antenna 3D gain pattern at 2.45 GHz. The main goal in the pattern shaping was to maintain low radiation intensity toward the brain. As seen from Figure 6, the antenna directs radiation away from the brain. The simulated antenna peak gain is −19.6 dBi at 2.45 GHz which is acceptable for small antenna with size of 6 mm × 5 mm. Figure 7 shows the simulated gain in E-plane and H-planes. Moreover, Table 2 summarizes the simulated antenna parameters at 2.45 GHz. Lastly, Figure 8 shows that very thin coating decreases the antenna gain. This indicates that antenna coating could be made as thin as 0.3 mm without detriment to the electromagnetic performance of the antenna.

2.2. Specific Absorption Rate. The amount of power radiating from the antenna is limited by the Specific Absorption Rate (SAR). We followed US Federal Communications Commission (FCC) regulation which limits the SAR averaged over one gram of tissue to SAR$_{max}$ = 1.6 W/kg. In our model, the implant was placed inside the brain layer and the bone layer was next to the brain (see Figure 3(a)). During the simulation, it was observed that maximum of the local SAR occurred at the bone and the brain.

SAR estimation was done using another model where the implant was located in a 18.75 × 18.75 × 30 cm$^3$ block of a brain and with the same sized bone phantom. To ensure the numerical stability, the height of the block was set at 30 cm, thus accommodating two averaging cubes containing 1 gram of the brain and the bone [47]. Figure 5(b) shows the local SAR distribution at the bone interface. The maximum local SAR value was 0.58 W/kg. The maximum SAR-compliant transmission power ($P_{t,max}$) was determined in HFSS which generated SAR$_{max}$. Table 3 shows $P_{t,max}$ and SAR$_{max}$ for the brain and the bone with 0.5 mW input. $P_{t,max}$ is lower for bone and puts the upper limits for power which can be radiated from the antenna without violating the SAR$_{max}$ regulation.

3. Antenna Measurement and Discussion

After the design procedure, the antenna together with sensor platform was fabricated on a flexible polyimide substrate. The antenna was positioned in a liquid phantom mimicking the human head properties. We followed IEEE standard for head properties (at 2.45 GHz, $\varepsilon_r = 39.2$ and $\sigma = 1.8$ S/m). The liquid phantom was made of water, sugar, and salt using a recipe explained in [48]. Relative permittivity of water was reduced
by adding sugar, whereas adding salt increased conductivity. Figure 9 presents the measured electrical properties of the liquid phantom. The measured liquid phantom properties at 2.45 GHz were \( \varepsilon_r = 39.5 \) and \( \sigma = 2.65 \text{ S/m} \). We performed two different measurements. In the first measurement, reflection coefficient of the antenna was measured in a liquid phantom. In the second measurement, the antenna was attached to a voltage controlled oscillator (VCO) and radiating power from the antenna was measured through another external receiving antenna. We performed the second measurement after attaching VCO, because in our final implant design the antenna will be attached to the same VCO.

For the first measurement, a SMA connector was attached to the antenna through 1 cm long microcoaxial cable. The microcoaxial cable had a diameter of 0.8 mm. Then, 1 mm thick coating of silicone was applied. After fabrication, the antenna was placed inside the liquid phantom and reflection coefficient was measured through a Vector Network Analyzer (VNA). The effect of cable was removed from the measurement results by applying VNA’s port extension property and extending the port position to the cable’s end. Figure 10 presents the simulated (both human layered and liquid phantom models) and measured reflection coefficients. Figure 11 shows the measurement setup and the fabricated antenna. Simulation and measurement results are in good agreement. The measured −10 dB bandwidth is 160 MHz (from 2.38 GHz to 2.54 GHz) with 20.8 dB return loss at 2.45 GHz, although the measured bandwidth of the antenna is smaller than the simulated one but still covers the ISM band.

In the second measurement, the antenna was attached to a voltage control oscillator (VCO) and VCO was powered through an external DC power supply. VCO was tuned to generate 2.45 GHz signal. Afterward, 1 mm silicon coating was applied. Then, the antenna was placed inside the liquid phantom and VCO was activated through 3 V power supply. As the VCO was activated, it started to transmit signal at 2.45 GHz. The transmitted signal from VCO was received through an external \( \lambda/2 \) dipole receiver antenna. The external \( \lambda/2 \) dipole receiver antenna was oriented to receive maximum signal strength. The receiver antenna was connected to the spectrum analyzer to measure the received power level of the signal. The signal power level was measured at 0.5 m and 1 m distances from the transmitting antenna. Figure 12 shows the measurement setup, the measurement picture, and the fabricated antennas. Table 4 presents the average of five independent received power levels of the signal at each distance. When VCO is activated through 3 V power supply, the typical output power of the VCO is −3 dBm (0.5 mW) with 12 dB return loss from 2.4 GHz to 2.5 GHz [49]. The receiver \( \lambda/2 \) dipole antenna parameters were separately measured. The measured receiver antenna’s return loss was 15.7 dB and gain was 1.46 dB. Power levels of −53.0 dBm and −58.6 dBm were recorded at 0.5 m and 1 m distances from the implant antenna, respectively. These recorded power levels are much above for the reader with sensitivity of −80 dBm.

Antenna gain was estimated in the peak direction through Friis transmission equation [50], given as

\[
\frac{P_r}{P_t} = G_r G_t \left(1 - |r|^2\right) \left(1 - |t|^2\right) \left(\frac{\lambda}{4\pi d}\right)^2 \rho, \quad (1)
\]

Table 4: Received power level and estimated antenna gain.

| Distance (d) [m] | Received power level [dBm] | Estimated gain [dBi] |
|------------------|----------------------------|---------------------|
| 0.5              | −53.0                      | −23.25              |
| 1                | −58.6                      | −22.94              |
where $P_r$ and $P_t$ are received and transmitted powers, respectively, $\lambda$ is the signal wavelength, $d$ is distance between the transmitter and the receiver antennas, $\rho$ is polarization mismatch factor, and $G_r$ and $G_t$ are receiver and transmitter antenna gains, respectively. Both of our transmitter and receiver antennas are properly matched. Moreover, during the measurement, we aligned both antennas to get maximum power for low polarization mismatch. This would lead us to ignore impedance and polarization mismatch factor and (1) is reduced to
\[
\frac{P_r}{P_t} = G_t G_r \left( \frac{\lambda}{4\pi d} \right)^2.
\]  
(2)

Antenna gain was estimated through (2) for both distances (0.5 m and 1 m) to verify the consistency between results. Table 4 presents the estimated antenna gain. The
simulation gain in the liquid phantom was $-20.5 \, \text{dBi}$ and estimated gain is within $3 \, \text{dB}$ of simulated gain. Thus, this verifies the antenna radiation performance. The difference between simulation and measurement gains might be due to the measurement equipment/setup uncertainty (e.g., coaxial cable loss and reflections from surroundings).

4. Conclusion

An implant antenna was presented for battery-free intracranial pressure (ICP) monitoring system. The antenna was optimized within available small dimensions of $6 \, \text{mm} \times 5 \, \text{mm}$. The antenna performance was measured in a liquid phantom mimicking human head properties. The measured reflection coefficient and estimated gain of the antenna agree with simulated results. Moreover, simulation shows that maximum SAR-complaint transmit power was $5.53 \, \text{mW}$. Finally, a power level of $-58.6 \, \text{dBm}$, which is well above for the reader with sensitivity of $-80 \, \text{dBm}$, was recorded at $1 \, \text{m}$ distance from the implant antenna when submerged to $0.5 \, \text{mW}$ VCO. Our future step is to assemble all of other electronics to develop a fully wireless battery-free ICP monitoring system.

Competing Interests

The authors declare that there are no competing interests regarding the publication of this paper.
Figure 10: Simulated (both human head and liquid phantom) and measured reflection coefficient (dB).

Figure 11: Reflection coefficient measurement setup and fabricated antenna.

Figure 12: Powered level measurement and fabricated antennas: (a) measurement setup; (b) measurement picture.

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