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Polarization Reconfigurable Planar Inverted-F Antenna for Implantable Telemetry Applications

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ABSTRACT A polarization reconfigurable implantable planar inverted-F antenna (PIFA), operating at 2.45–GHz ISM band, is introduced for monitoring and biotelemetry purposes. The proposed implantable antenna featuring with the reconfigurability of two polarizations is effective to overcome the polarization mismatch and multi-path reflection in an indoor environment. By using two groups of PIN diodes as radio frequency (RF) switches along $x$-direction and $y$-direction respectively, two orthogonal linear polarizations can be switched to avoid communication link failure. The proposed antenna is based on a planar inverted-F antenna structure and the implantable environment of human body to develop. The antenna exhibits a good stability of an impedance bandwidth with the reflection coefficient $\leq -10$ dB in the environment of a human phantom. The simulated results in simple muscle-layer phantom, multi-layer phantom and 3D realistic human voxel model demonstrates the good robustness of the proposed antenna in the complex implantable environment. An in-vitro test was carried out in body tissue simulant liquid to validate our proposed design. A sleeve balun was fabricated and incorporated into the semirigid coax cable in order to eliminate the unwanted effect of the long cable during the measurement. The experimental results agree with simulation and show an overlapped impedance bandwidth of 9.1% from 2.3 GHz to 2.52 GHz in different reconfigurable states. The radiation performances, evaluation of Specific Absorption Rate (SAR), and link characteristics in different propagation scenarios are also discussed. This proposed antenna technology is suitable for implantable telemetry applications.

INDEX TERMS Biotelemetry, implantable medical devices, implantable antennas, reconfigurable antennas.

I. INTRODUCTION With the growing concern of human health and the advancement in supporting technology, wireless body area network (WBAN) is drawn much attentions by the industry in the last decade. The devices of WBAN placed close to human body can be categorized into wearable devices and implantable medical devices (IMDs). Recently, IMDs are receiving significant attentions since they can detect the human biomedical information accurately and conveniently without wired sensors outside the body [1]. As one possible scenario of WBAN applications shown in Fig. 1, a diabetic patient equips with an implantable antenna and subcutaneous glucose sensors connecting with the whole healthcare system. The information of glucose level, which needs to be controlled in a reasonable range for diabetic patients, can be transmitted to the external receivers continuously and timely [2]. Other IMDs like intracranial pressure monitoring devices [3], neural recording systems [4], a retinal prosthesis [5] can also be applied to such scenario, where the implantable antennas play an important role in the wireless communication link. However, it remains a great challenge to design a suitable implantable antenna since it would take several factors into design consideration such as robustness, size, biocompatibility and safety [6].

Due to multi-path reflections in the indoor environment between the implantable antenna and the external receiver, polarization mismatch may occur which cannot guarantee a robust communication link for biomedical telemetry. There are a few approaches to address this issue. Circularly-polarized implantable antennas have been proposed in some works [7]–[9] for providing the immunity to
multi-path reflection and undetermined orientations of the implants. However, the circular polarization property can be altered by the surrounding environment, especially when the antenna is implanted in human body. Another method is placing multiple on-body antennas outside the human body in order to achieve polarization diversity performance [10]. Obviously, this would increase the equipment size and complexity. Besides, a multi-polarization reconfigurable off-body antenna had been proposed, aiming to minimize the polarization mismatching and achieve an optimal communication link [11]. However, its bulky structure may not be suitable for some space-limited scenarios, such as implantable telemetry applications where antenna size should be small. One polarization reconfigurable PIFA had been proposed as a mobile terminal antenna [12]. Two shorting posts with two RF-MEMS switches were used to achieve different polarizations. However, the switches and DC bias circuits were placed within the top layer of the antenna, which may influence the broadside radiation performances and the complicated bias circuits would not be suitable for implantable application. Moreover, the maximum radiation direction is at $\theta = 30^\circ$ with poor front-to-back ratio (FTBR) at broadside, which is also undesirable in biotelemetry applications.

Apart from aforementioned designs, we propose an implantable and reconfigurable antenna for wireless biomedical telemetry. To the best of the authors’ knowledge, the proposed antenna may be the first implantable antenna featuring with polarization reconfigurable characteristics. The antenna structure is based on the implantable environment of human body to develop. One layer of superstrate is placed above the substrate for biocompatibility consideration, thus the PIN diodes, inductors and DC pads are all placed on the bottom layer of the substrate. The DC bias circuits includes only two DC pads and RF choke inductors, making it simple and compact for implantable application. It shows a directed radiation towards free space with a maximum radiation direction at $\theta = 0^\circ$ and acceptable FTBR, which is desirable in biotelemetry applications. By introducing two groups of PIN diodes as RF switches along the slot in the ground plane, two orthogonal linear polarizations can be switched to avoid communication link failure. Instead of using separated radiating elements, the proposed antenna employs the same structural region as the radiated elements of different polarization modes. In this way, the occupied space of antennas can be decreased. By adopting a symmetric antenna structure about the principle diagonal, a compact PIFA is formed with $x$-oriented or $y$-oriented polarization depending on which groups of PIN diodes are active. To solve the unwanted effect of the leakage currents exist on the outer surface of cables during the measurement, we added a sleeve balun in the simulation and incorporated it into the coax feedline.

This paper is organized as follows. In Section II, the geometry of the proposed antenna is introduced, together with its simulation results. In Section III, the operating principle and stability to the surrounding environment are analyzed. Section IV further evaluates the antenna performances in the three dimensional realistic human voxel model. Section V discusses the experimental results and the communication link characteristic, followed by the conclusions in Section VI.

II. ANTENNA DESIGN

A. SIMULATION MODEL

To improve the accuracy and efficiency of implantable antennas design, two different phantoms have been set up to imitate the practical environment. As shown in Fig. 2 (a), the antenna is embedded in a muscle-layer cubic phantom with a depth of $ds = 10$ mm. Besides, a multi-layer cylindrical phantom, including layers of skin, fat, muscle and bone with different thicknesses, is also constructed to imitate a 2/3 equivalent human arm, as shown in Fig. 2 (b) and (c). In this case, the antenna is implanted in the muscle layer with a depth of $dp = 15$ mm. As listed in Table 1, the dielectric properties of the tissues at the center frequencies of 2.45 GHz are adopted to simplify simulation [13]. The related parameters are listed in Table 2.
TABLE 1. Dielectric properties of different tissues.

| Layer     | Relative permittivity | Conductivity (S/m) |
|-----------|-----------------------|--------------------|
| Skin (5 mm) | 38.0                  | 1.45               |
| Fat (4 mm)  | 5.28                  | 0.10               |
| Muscle (31 mm) | 52.7                  | 1.74               |
| Bone (5 mm)  | 11.4                  | 0.39               |

TABLE 2. Optimized antenna dimensions and parameters.

| Parameter | Value | Parameter | Value | Parameter | Value |
|-----------|-------|-----------|-------|-----------|-------|
| l         | 13.5 mm | l5        | 1.8 mm | w5        | 2.3 mm |
| w         | 11.5 mm | l6        | 2 mm  | h1        | 1.27 mm |
| l1        | 10.5 mm | w1        | 0.4 mm | h2        | 0.635 mm |
| l2        | 9.1 mm  | w2        | 1 mm  | h3        | 1 mm   |
| l3        | 1.8 mm  | w3        | 2 mm  | g         | 0.5 mm |
| l4        | 1 mm    | w4        | 1.1 mm | ds        | 10 mm  |
| lm        | 200 mm  | Rmuscle   | 36 mm | Rskin     | 45 mm  |
| Rbone     | 5 mm    | Rfat      | 41 mm | dp        | 15 mm  |

B. GEOMETRY OF THE RECONFIGURABLE PIFA

Fig. 3 depicts the structure of proposed antenna which is symmetric about the principle diagonal except for the DC pads and RF choke inductors. The prototype is etched on a substrate of Rogers RT/duroid 6010 ($\varepsilon_r = 10.2$, tan $\delta = 0.0023$) with a thickness of $h1$. For the sake of convenience, the center of the top-layer patch is set as the original point of the coordinate system. Two edges of the top-layer patch are shorted to the ground plane by closely placed shorting posts with diameter of 0.7 mm, forming equivalent shorting walls as a significant part of PIFA. Note that when the number or diameter of the shorting pins is large enough, the closely placed shorting pins can be equivalent to a solid shorting wall, due to the electromagnetic leakage from the gap between nearby pins is negligible. Here 5 shorting pins are used. Six PIN diodes, two RF choke inductors and the DC pads are all placed on the bottom layer of the substrate. To achieve switchable polarization modes, two groups of PIN diodes (Group A and Group B) are placed along $x$-direction and $y$-direction respectively. One open-end slot, with bending angle of 90 deg, is loaded in the ground plane in order to provide space for the installation of PIN diodes. Additionally, the cathodes placements of the two groups of diodes, with respect to the two copper strips separated by the slot, are different to each other. To provide necessary DC biasing voltage for PIN diodes, two DC pads are etched near the ground plane and two inductors (56 nH) are used as RF choke. One layer of superstrate with a thickness of $h2$ is placed above the substrate as shown in Fig. 2. The material of the superstrate is the same to the substrate for simplicity and convenience of fabrication and assembly. Below the substrate, it is one layer of sensor system which is used to gather the biomedical information of human body, such as a glucose sensor system for glucose level monitoring [14], and in this work, we initially set this layer as vacuum with a thickness of $h3$ in simulation.

For proof-of-concept demonstration purposes, one type of PIN diodes (Bar50-02L, Infineon Technologies) [15] and Murata inductors (LQW18AN56N00) [16] are adopted for simulation and measurement. The PIN diode equivalently acts as a parasitic inductance (0.4 nH) in series with a forward resistance (2 ohm) at ON state, and performs as a reverse capacitance (0.08 pF) in parallel with a reverse resistance (5K ohm), then in series with a parasitic inductance (0.4 nH) at OFF state [17]. After optimization with the assistance of ANSYS HFSS v.16, the location of the feeding points is $(x = -1, y = 1)$ and a compact size of 13.5 mm $\times$ 11.5 mm $\times$ 1.905 mm is obtained. The top-layer patch of the antenna is slightly bigger than the bottom layer. The other related parameters are listed in Table 2.

C. SIMULATION RESULTS

As shown in Fig. 4, for the muscle-layer phantom, an impedance bandwidth of 200 MHz (2.33 – 2.53 GHz) is obtained in state 1, which is almost the same to state 2 (2.34 – 2.54 GHz), thus the proposed antenna exhibits a wide overlapped impedance bandwidth of 7.8% (2.34 – 2.53 GHz), covering the frequencies of interest (2.4 – 2.48 GHz). Fig. 5 shows the simulated radiation patterns. The antenna has a directed radiation towards free space, which is desirable in biotelemetry applications. The gains in the muscle-layer phantom are $-23$ and $-23.2$ dBi in state 1 and state 2, respectively. Whereas, for the antenna in the multi-layer phantom, they are around 3 dB lower than that of muscle-layer phantom, due to the fact that the implanted depth in multi-layer phantom is larger, causing more energy dissipation in human tissues. The XPD at boresight can be above 10 dB in different states. From the above simulation results, there exists only a little difference between the two simulation
III. OPERATING PRINCIPLE AND DISCUSSION

A. DUAL-POLARIZED PROPERTY OF THE PROPOSED ANTENNA

Based on the symmetric characteristic of the antenna structure and the different cathodes placement of the two groups of PIN diodes described above, two orthogonal polarization modes can be switched by applying different status of DC biases. For example, the PIN diodes (Group A) located along y-axis would be turned on when DC#1 = 1.5 V and DC#2 = 0 V. In this case, a PIFA with x-oriented polarized operating mode (State 1) is formed. The current distributions on the patch within a period can be seen in Fig. 6. It exhibits a good direct consistency along x-axis. The current near the “un-function” shorting wall is small, thus the existence of this edge has little effect on the antenna performances. Similarly, another PIFA with y-oriented polarized operating mode (State 2) can be formed when the PIN diodes (Group B) located along x-axis are forward biased. In this case, comparing to state 1, the main current paths just rotates 90 deg, meanwhile occupying the same radiating area of the antenna, thus a compact reconfigurable implantable PIFA is achieved without additional reconfigurable feeding network. Table 3 presents the polarization states of the reconfigurable PIFA and the corresponding active groups of PIN diodes. Reconfigurable polarizations can be obtained through the on and off states of groups of PIN diodes.

B. PARAMETRIC STUDY

Based on the operating principle discussed above, the feeding point needs to be located along the principle diagonal of the radiator in order to fulfill the requirement of same radiating area and similar performance of XPD in different polarization states. From our simulation, we found that the performance of impedance matching would deteriorate when the feeding point deviates from the optimized location while the XPD may be greater, thus it needs to seek a balanced between the performances of impedance matching and XPD. The optimized location was determined when x = −1 and y = 1 after final optimization. The related curves are not shown here for brevity.

C. EFFECT OF THE PIN DIODES

In order to evaluate the diodes effect on the antenna performances and to provide design guidelines, a study of the
effect of different numbers of PIN diodes on the resonant frequencies and XPD was carried out. Fig. 7 shows simulated reflection coefficient for three cases with different numbers of diodes. It can be seen that the resonant frequency shifts upward when the number of the diodes increases from 2 to 4. This feature can be used to tune the antenna to operate at the desired frequency bands if needed. On the one hand, with less diodes, the resonant frequency can be shifted downward without changing the antenna size, thus it can achieve a greater miniaturization in constrained space for implantable application. However, the gain at corresponding band would be lower due to its reduced electrical size. On the other hand, with more diodes, the antenna would operate at a higher frequency band with relatively higher gain. Nonetheless, the antenna with more diodes requires more DC power to support the operation. Therefore, the choice of the number of the diodes needs to take the limited space reserved for the antenna, operating bands, acceptable gain and power dissipation into account. To reduce the fabrication complexity and achieve relatively acceptable radiation performance, the case with three diodes was chosen for the final design. Considering the radiation efficiency degradation caused by the diodes, we set the forward resistance as zero in simulation to evaluate the antenna performance for the case with lossless diodes, and we found that the radiation efficiency of the proposed antenna is around 2 dB lower than the case with lossless diodes.

Due to the power consumption introduced by the PIN diodes, it needs to evaluate the total power consumption of the IMD system. Two button cell batteries (1.55 V and 42 mA·h) are used to initially evaluate its operating lifetime as introduced in [18]. As it points out that the system consumes about 1.86 mJ (20 mA × 1.55 V × 30 ms) per emission. In other words, the operating power is 62 mW. For one PIN diode, if there are no additional current choke resistance to provide a lower dissipation, then the maximum power dissipation, under operating current of 100 mA, would be 250 mW according to [15]. In this case, the total power dissipation would be 750 mW when three diodes are turned on at a time. For some biomedical applications such as subcutaneous real-time glucose monitoring, the patient with Type 1 diabetes in intensive therapy needs to test the blood glucose concentrations at least four times a day according to the suggestions of the American Diabetes Association [19]. Here we suppose a maximum power dissipation of 850 mW for the whole IMD system and the emission occurs six times a day. The batteries can supply 468 720 mJ (42 mA × 1.55 V × 3600 s × 2), then the calculated lifetime of the batteries can be up to 8 years theoretically.

**D. ANTENNA STABILITY TO THE VARIATION OF TISSUE PROPERTIES AND IMPLANTED DEPTH**

Due to the fact that the electrical properties of human tissue vary from organ to organ, person to person and may even change with ages, it needs to evaluate the antenna stability when the tissue properties vary in a reasonable range. Table 4 lists the impedance bandwidths, gains and XPD when the relative permittivity varies from 30 to 70, and the conductivity varies from 0.5 to 3. There is only a little fluctuation of the impedance bandwidth, exhibiting the merit of impedance stability. This feature makes the proposed antenna available to be implanted into a variety of human tissues. With larger conductivity, the gain at 2.45 GHz would decrease. This is because the larger the conductivity is, the more energy dissipation will be absorbed by the human body. The XPD at boresight can still remain above 10 dB except for the case when \( \varepsilon_r = 30 \) and \( \sigma = 0.5 \).

| Relative Permittivity | Conductivity [S/m] | | | Gain (dB) | XPD (dB) |
|----------------------|-------------------|-------------|-------------|-------------|-------------|
| 30                   | 0.5               | 2.38-2.55   | -21.5       | 7           |
|                      | 1.74              | 2.36-2.55   | -24.5       | 12          |
|                      | 3                 | 2.37-2.57   | -28.5       | 11          |
| 52.7 (proposed)      | 0.5               | 2.34-2.55   | -19.3       | 20          |
|                      | 1.74              | 2.33-2.53   | -23         | 17          |
|                      | 3                 | 2.36-2.56   | -27         | 15          |
| 70                   | 0.5               | 2.32-2.54   | -21.7       | 11          |
|                      | 1.74              | 2.33-2.56   | -22.5       | 15          |
|                      | 3                 | 2.35-2.57   | -25.7       | 15          |

In practical scenario, the position of the antenna in human tissue will move and differ to the design case. We evaluate the effect of different implanted depth (5-15 mm) on the antenna performance and it has been found that there is almost no change of the reflection coefficient. However, the gain is more sensitive to the variation of implanted depth. It shows a tendency that the larger the implanted depth, the lower the
gain value, due to more energy dissipation caused by human body.

In conclusion, the proposed antenna shows a good stability to the surrounding environment in terms of the variation of tissue properties and embedded depth, especially exhibiting the merit of stable impedance bandwidth.

### IV. REALISTIC HUMAN MODEL AND DISCUSSION

Except for the muscle-layer and multi-layer phantoms discussed above, in order to further evaluate the antenna performances in a more realistic human body which is not homogeneous and composed of different heterogeneous tissues, a male Zubal phantom is used in CST simulation environment. The Zubal phantom is shared by Yale University [20], [21] and the required dielectric parameters of different tissues and organs can be set according to Cole-Cole formulation [13].

Fig. 8 shows four different implant positions including human arm (Position A: implanted in tissue of muscle), stomach (Position B), small intestine (Position C) and colon (Position D). The corresponding antenna performances, as listed in Table 5, remain good for all cases except for the case of small intestine, whose gain value is relatively small which is mainly due to the fact that the conductivity of small intestine and the implanted depth are larger than the other cases. Meanwhile, the simulated 1-g averaged SAR value of small intestine is relatively larger since it is proportional to the conductivity which can be seen from the following SAR formulation:

$$\text{SAR} = \frac{\int \sigma(r) \left| \bar{E}(r) \right|^2 \rho(r) \, dr}{\rho(r)},$$  \hspace{1cm} (1)

where $\sigma(r)$ is the conductivity of the tissues, $\bar{E}(r)$ is the electric field and $\rho(r)$ is the density of the tissues. Considering the issue of human body safety, the maximum 1-g averaged SAR values are included in Table 5. According to the IEEE standard C95.1-1999 [22] which requires the SAR averaged over any 1 g of tissues should be less than 1.6 W/Kg, the calculated allowable maximum input power should not exceed 22 mW (13.4 dBm) in order to satisfy the safety regulation for these four implant positions. This value is larger than the implantable transmitter power which is on the order of $-10$ dBm [23], thus the safety of human body can be guaranteed.

Fig. 9 shows the radiation patterns of the proposed antenna implanted in different organs. They all show the main radiation towards the outside human body. Overall, when the antenna is embedded at different positions in a three-dimensional realistic human voxel model in CST, the performances are acceptable and similar to the simulated results in HFSS, further demonstrating that the proposed antenna has the good robustness.

### TABLE 5. Antenna performances of the proposed antenna in different organs in reconfigurable state 1.

| Biological Tissues  | Arm (muscle layer) | Stomach | Small Intestine | Colon |
|---------------------|--------------------|---------|----------------|-------|
| Relative permittivity | 52.7               | 62.2    | 54.5           | 53.9  |
| Conductivity [S/m]   | 1.74               | 2.16    | 3.13           | 1.99  |
| $|S11|$ (GHz)          | 2.33               | 2.33    | 2.30           | 2.33  |
| Gain (dBi)           | -24.5              | -36.0   | -43.6          | -25.8 |
| XPD (dB)             | 12.7               | 10.5    | 11.4           | 17.7  |
| 1-g Avg. SAR (W/Kg)  | 38.2               | 66.3    | 72.6           | 61.2  |
| Allowable Maximum Input Power (mW) | 41.9 | 24.1 | 22.0 | 26.1 |

### FIGURE 8. Three dimensional realistic human voxel models in CST for verifying antenna performances when the proposed antenna is implanted in different organs. (a) Arm, (b) Stomach, (c) Small intestine, (d) Colon.

### FIGURE 9. Simulated normalized radiation patterns of the proposed antenna in reconfigurable state 1 when implanted in different organs. (a) Arm, (b) Stomach, (c) Small intestine, (d) Colon.
shown in Fig. 10, in order to eliminate the unwanted effect of the long cable. A plastic container of $100 \text{ mm} \times 100 \text{ mm} \times 100 \text{ mm}$, whose size is the same as the muscle-layer phantom used in the simulation setup, would be filled with body tissue simulant liquid in the experimental setup. The used liquid has a measured relative permittivity of 53.2 and a conductivity of 1.97 at the center frequency of 2.45 GHz. Fig. 11 shows the measured reflection coefficient of the proposed antenna in all operating states. The measured bandwidths are 11.3% (2.26 – 2.53 GHz) and 9.1% (2.3 – 2.52 GHz) for the reconfigurable state 1 and state 2, respectively. The overlapped impedance bandwidth is 9.1% from 2.3 GHz to 2.52 GHz. The minor discrepancy between the measured and simulated results could be due to the fabrication and measurement tolerances, also due to the soldering effect, the thin air gap between the superstrate and substrate, and the inconsistency of the PIN diodes.

![Figure 10. Photographs of the fabricated prototype.](image)

![Figure 11. Measured reflection coefficient of the proposed antenna.](image)

**B. COMMUNICATION LINK SETUP AND MEASUREMENT**

In order to verify the polarization characteristic of the proposed antenna in real implanted configurations, the communication link between the implantable antenna and the external dipole was set up on a platform as shown in Fig. 12. The antenna was immersed into the liquid with a gap of 10 mm relative to the surface of the container. Besides, one linearly polarized dipole was placed right below the implanted antenna with a distance of 250 mm between them. A 1.5 V AAA battery was used for DC biasing and the reconfigurable polarization states can be switched by simply interchanging the two DC lines connected to the two poles of the battery according to the assignment shown in Table 3. The measurement was conducted at the temperature of $23^\circ\text{C}$ and humidity of 66%.

Fig. 13 shows the measured $S_{21}$ between the implantable antenna and the external dipole with different orientations. It can be seen from Fig. 13 (a) that for the reconfigurable state, a maximum transmission coefficient of $-52.4$ dB at 2.45 GHz appears when the dipole is aligned along x-axis. As for state 2, the maximum transmission occurs when the dipole is aligned along y-axis, as shown in Fig. 13 (b). It indicates that the orientation of the co-polarization of the implantable antenna operating at state 1 and state 2 are aligned along x-axis and y-axis, respectively. Besides, the large discrepancy between different states, for case#1, case#3, case#5 and case#6, confirms the polarization reconfigurable characteristic of the proposed antenna. Overall, the maximum transmission coefficients for all cases vary from $-52$ to $-59.5$ dB, exhibiting that the polarization mismatching can

![Figure 12. Communication link measurement setup between the polarization reconfigurable implantable antenna and the external dipole, and diagram of different orientations of the dipole in the measurement.](image)

![Figure 13. Transmission coefficients between the polarization reconfigurable implantable antenna and the external dipole with different orientations. (a) Reconfigurable state 1, (b) Reconfigurable state 2.](image)
be compensated through switching proper operating states of the implantable antenna when the orientation of the dipole is uncertain.

Fig. 14 illustrates the measurement setup of radiation performance. One elbow pipe was incorporated into the liquid container in order to provide a smooth path for the coaxial cable and it shows no big influence on the radiation performance from simulation. Fig. 15 shows the measured radiation patterns. The measured gains are $-24.5$ and $-24.2$ dBi respectively for state 1 and state 2, which are around 1.5 dB lower than the simulation. The deviation is mainly due to the relatively larger conductivity of the body tissue simulant liquid compared to the simulated one, insertion losses of the six PIN diodes, losses of the two inductors and the inaccuracy of the implanted depth in measurement experiment. Based on the obtained results, the antenna features stable impedance matching and dual-polarization reconfigurable properties which can mitigate the polarization mismatching problems in biotelemetry application.

![Measurement setup of radiation performance.](image1)

**FIGURE 14.** Measurement setup of radiation performance.

![Measured normalized radiation patterns for different reconfigurable states (a) state 1, (b) state 2.](image2)

**FIGURE 15.** Measured normalized radiation patterns for different reconfigurable states (a) state 1, (b) state 2.

C. LINK BUDGET CALCULATION AND STUDY

To evaluate the communication range and the quality of the biotelemetry link, link budget ($LM$) of the proposed antenna has been analyzed. In this study, the transmitted power ($P_t$) is $-4$ dBm which is the same as introduced in [24], and different from that, not only the free-space propagation channel is analyzed, but the indoor multi-path propagation scenarios are also taken into consideration. In indoor communication environment, as pointed out in [25], the mean gain value is equal to the value of radiation efficiency and is used for $LM$ calculation. As for the proposed antenna in the muscle-layer phantom, the radiation efficiency is around $-30$ dB, thus in this work, we set the transmitting antenna gain as $-30$ dBi. Besides, the smallest gain of $-43.6$ dBi in case of small intestine implantation, given in Table 5 in Section IV, is also used for initially evaluating the communication capacity of the whole biotelemetry system.

Due to the polarization reconfigurable properties of the proposed antenna, it allows a reliable communication except for a potential maximum polarization mismatch of 3 dB which can be improved by designing a multi-polarization reconfigurable antenna as [11]. This maximum polarization mismatch loss ($LP$) is also included in the calculation. According to [25], [26], $LM$ can be expressed by

$$LM = \text{Link C}/N_0 - \text{Required C}/N_0 = (P_t + G_t - L_p - L_m + G_r - N_0) - (E_b/N_0 + 10\log_{10} B_r - G_c + G_d). \quad (2)$$

where

$$L_m = 10\log_{10} (d/d_0) + 20\log_{10} (4\pi d_0/\lambda) + \chi_\sigma, \quad (3)$$

$$L_f = L_{m=2} = 20\log_{10} (4\pi d/\lambda), \quad (4)$$

$$N_0 = 10\log_{10} (kT_0 (NF - 1)), \quad (5)$$

and $m$ is the path-loss exponent which is propagation scenarios dependent, $d$ is the distance between the transmitter and receiver, $d_0 = 1$ m is the reference distance, $\lambda$ is the wavelength in free space, $\chi_\sigma$ is the shadowing factor and is set as zero here, and the other related parameters are shown in Table 6. Three different scenarios are considered, including the free space propagation model ($m = 2$), indoor propagation with line-of-sight (LOS, $m = 1.5$) and indoor propagation with non-line-of-sight (NLOS, $m = 3$).

The calculated $LM$ with varying distance is shown in Fig. 16. For a determined transmitter power, the lower the transmitter antenna gain, the smaller $LM$ for all propagation scenarios, reflecting a worse communication quality. It can be seen that for indoor LOS environment ($m = 1.5$), the $LM$ is larger than the case of free space, which means a slower level of degradation with increased received power resulted from the reflections from the surrounding indoor environment. Whereas, it shows a fast degradation for the case of indoor NLOS scenario since the obstacles between the implanted antenna and external base station can impose a destructive effect on the propagation channels. In this case, a decreased communication distance can be observed. For all different cases as shown in Fig. 16, the data communication with a positive $LM$ can occur up to 10 m which is desirable for indoor biotelemetry application.
TABLE 6. Parameters for link budget.

| Parameter                        | Value     |
|----------------------------------|-----------|
| Operating frequency (GHz)        | 2.45      |
| Tx power \( P_t \) (dBm)         | -4        |
| Tx antenna gain \( G_t \) (dB)   | -30       |
| \( EIRP \) (dBm)                 | -34       |
| Max. polarization mismatch loss \( L_p \) (dB) | 3        |

Propagation scenarios:
- Free space loss \( L_s (m = 2) \) (dB)
- Indoor LOS loss \( L_{los} (m = -1.5) \) (dB)
- Indoor NLOS loss \( L_{nlos} (m = 3) \) (dB)

Path-loss exponent \( m \) and distance dependent:

Receiver:
- Rx antenna Gain \( G_r \) (dB)
- Ambient temperature \( T_a \) (K)
- Receiver noise figure \( NF \) (dB)
- Boltzmann constant \( k \)
- Noise power density \( N_0 \) (dB/Hz)

Signal quality:
- Bit rate \( B \) (kb/s)
- Bit error rate
- \( E_b/N_0 \) (ideal-BPSK) (dB)
- Coding gain \( G_c \) (dB)
- Fixing deterioration \( G_f \) (dB)

FIGURE 16. Calculated link margin for three different propagation scenarios.

VI. CONCLUSION

This paper has introduced a new polarization reconfigurable PIFA for implantable telemetry applications. It aims to solve the issue of polarization mismatching in complex indoor environment and focus on the implantable antenna design. The proposed antenna is based on PIFA structure and is evaluated under three different simulation setups, exhibiting a good characteristic of insensitivity to the variation of surrounding environment. Measurement of communication link further demonstrates the effectiveness of the proposed antenna to avoid polarization mismatching. A study of LM in different propagation scenarios verifies its effective operation in complex indoor environment. The overall performances indicate that the proposed implantable antenna is very promising for numerous potential biotelemetry applications, e.g., transcutaneous glucose monitoring. This work is useful for researchers who are engaged in designing implantable antennas.

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