Antenna Design for Wireless Electrocorticography

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***Abstract*–We present a design methodology for implantable antennas patterned on a flexible substrate for electrocorticography. The antennas are designed for maximum link gain across the human skull to efficiently power the implant. Simulation results show that optimizing the transmission frequency together with antenna geometries can provide sufficient power and voltage to the implant.**

I. INTRODUCTION

The recent advent of brain-machine interfaces has caught wide attention in the fields of technology and medicine. Electrocorticography (ECoG) is a neural recording technique in which an array of electrodes is placed on the cerebral cortex to record the electrical activity generated by the cortical neurons. This level of invasiveness avoids penetrating and potentially scarring the cortex and has been shown to enable high enough resolution to locate the focus of a seizure and potentially to control prosthetic limbs. This provides hope for patients with spinal cord injuries and neurological conditions. Powering and communicating with the miniature, implanted recording devices remotely via electromagnetic (EM) fields has proven to be a compelling approach. It enables fully integrated long-term recording solutions and removes the need for risky and expensive surgeries to replace discharged batteries. This greatly improves the patients' comfort. However, the optimization of the transcranial wireless link for remote powering remains a challenge. In particular, at the implant side the antenna size is highly constrained making it a challenge to achieve sufficient voltage at the implant input for efficient RF-to-DC conversion. [1-2]

This paper presents a design methodology for implantable antennas for wireless EcoG system The seamless integration of the antenna, electrode array, and microelectronics is considered as the top design priority. The voltage and power provided to the implanted electronics are evaluated while conforming to human safety limits on EM field exposure.

II. DESIGN REQUIREMENTS AND CONSTRAINTS

To activate the implanted microelectronics, the implant antenna needs to present sufficient voltage amplitude at its terminals and provide the required continuous power for the circuitry. However, the power that can be safely supplied to an outside-body transmitter is limited by the regulated specific absorption ratio (SAR). Since the antenna-body EM interaction is predominantly due to the electric field, power transfer via magnetic field in the reactive field region is often preferred. Moreover, compared with a conventional loop, a segmented loop has been proposed to enable higher SAR compliant transmitted power [3]. For the link power efficiency, maximizing the implanted antenna performance is, however, most important.

For recording neural signals from the cortex, an 8×8 electrode array with 0.5 mm electrode pitch is chosen. The array needs to conform to the cortex and is therefore fabricated on a flexible and biocompatible Parylene C in a custom Parylene MEMS process at wafer level [4]. The thickness of the conductive layer produced is 260 nm with gold as the main component. As the minimal device size and structural complexity are top priorities in the design, the implant antenna is patterned on the same platform in the same process. However, in the range of 100s of MHz, which has been found a suitable frequency range for wireless powering of mm-size implants [2-3], this results in conductor thickness of less than 10% of the skin depth. Therefore, additional loss due to the current crowding phenomenon is to be expected and consequently the degrees of freedom in the antenna design are greatly reduced. In particular, long and narrow antenna traces are to be avoided. Since a simple single-layer structure, capable of efficiently extracting energy from magnetic field is required, a single-turn loop with the outer dimensions of 6.5×6.5 mm^2 was initially chosen as the implant antenna. To minimize the device size, the loop is designed to enclose the electrode array (see Fig. 1). As a result, there is an unavoidable EM interaction between the two, so that electrode array needs to be considered as a part of the antenna throughout the design process.

III. WIRELESS LINK SIMULATION

All simulations were conducted with ANSYS HFSS v13, which is a full-wave EM field simulator based on finite element method. The simulation model is illustrated in Fig. 1. A layered tissue model with frequency dependent dielectric properties given in [5] was used to model the transcranial radio channel. To account for additional loss sources, the the series resistance of the segmenting capacitors (40 mΩ) and the implant antenna bonding resistance (0.3 Ω) were included in the simulation model. To characterize the wireless link, the maximum available link power gain (G_{ma}) obtained from the two-port simulation, power available from the implant antenna ($P_{av.A}$), and voltage across various antenna load impedances were simulated. The power and voltage values are reported corresponding to the maximum SAR compliant transmitted

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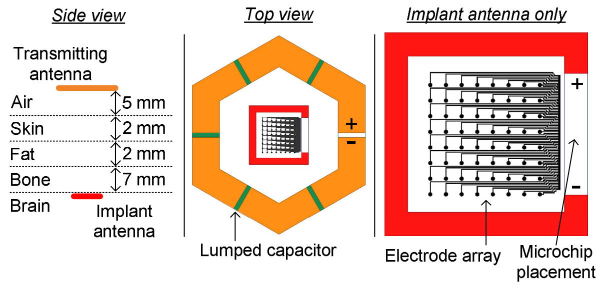

Figure 1. Simulation model.

power ($P_{tx,max}$) following the U.S. regulations (max SAR=1.6 W/kg).

Fig. 2, shows the simulated G_{ma} versus frequency for a wireless link composed of the 6.5×6.5 mm^2 implanted loop with 0.75 mm trace width and segmented loop antennas [3] (capacitors adapted at each frequency) with different diameters (D). The results suggest that D=15 mm provides close to optimal transmit antenna size – frequency combination in terms of power efficiency, but there is, still room to adjust the loop trace width (w) to alleviate the current crowding in the sub-skin-depth antenna. According to the results shown in Fig. 3, choosing w appropriately is actually a major design choice: with w<0.5 mm, G_{ma} drops sharply, while letting w>0.75 mm yields little improvement. Since the maximal voltage amplitude (V_L) across conjugate matched load is achieved around w=0.75 mm, this value was chosen for the final system performance evaluation.

Based on the results shown in Fig. 3, with w=0.75 mm, 2.8 mW is available for RF-to-DC conversion at 1 V, under conjugate-matched conditions, while we expect 0.5 mW to be sufficient for the considered application with conversion efficiencies above 50%. However, at voltages below 0.3 V, the

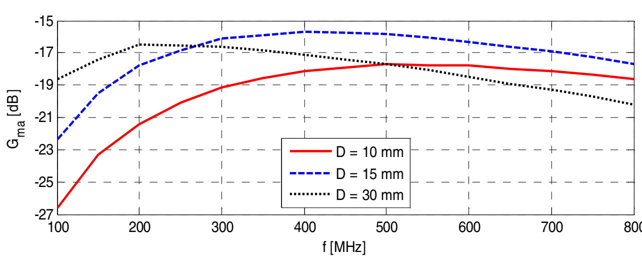

Figure 2. Maximum achievable link power gain for different transmitting antennas.

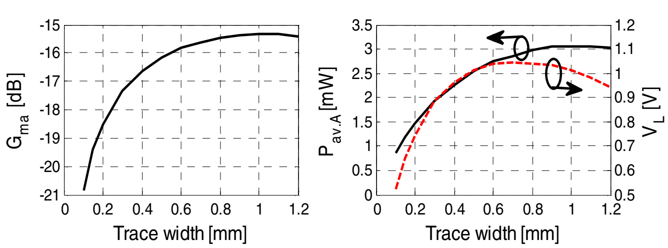

Figure 3. Effect of implant antenna trace width on the overall system performance at 400 MHz.

conversion efficiencies drop [2]. Therefore, we also analyzed the voltage at the implant input with various antenna-to-load power transfer efficiencies (τ, the ratio of the power delivered to the load to $P_{av.A}$). Given the antenna impedance (Z_A), the load impedances (Z_L) accepting at least $\alpha \times 100\%$ of $P_{av.A}$ are contained in a circle $C_\tau(\alpha)$ in the complex plane. Thus, we computed the lower bound for the load voltage amplitude:

$$V_{L,\alpha} = \min_{Z_L \in C_\tau(\alpha)} \sqrt{\tau}\sqrt{2\frac{|Z_L|^2}{\mathrm{Re}(Z_L)} G_{ma} P_{tx,max}}, \quad \tau = \frac{4\mathrm{Re}(Z_A)\mathrm{Re}(Z_L)}{|Z_A+Z_L|^2}, \qquad (1)$$

with various mismatch loss factors α. Table 1 shows the results with other relevant system performance data. In conclusion, even in the case of 50% mismatch loss, 1.4 mW is expected to be available for RF-to-DC conversion at 0.4 V, leaving a clear margin to the considered minimum requirement of 0.5 mW. This could be exploited to shrink the antenna further.

TABLE I. SYSTEM PERFORMANCE AT 400 MHZ.

$P_{tx,max}$ [mW]	G_{ma} [dB]	$P_{av.L}$ [mW]	V_L [V]	$V_{L,0.9}$ [V]	$V_{L,0.75}$ [V]	$V_{L,0.5}$ [V]
104	−15.6	2.8	1.04	0.72	0.58	0.41

IV. CONCLUSIONS

The integration of an antenna with an ECoG electrode array has enabled an implantable transcranial wireless link which can power a 64-channel microelectronic system for neural signal acquisition and data transmission. Using full-wave EM simulations we demonstrated how to optimize the operation frequency, and antenna geometries for overall power efficiency, while guaranteeing SAR compliance and sufficient voltage swing at the implant.

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