31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Miniaturization of Implantable Wireless Power Receiver

Ada S. Y. Poon
Electrical Engineering, Stanford University
adapoon@stanford.edu
(*Invited Paper*)

Abstract

Implantable medical devices will play an important role in modern medicine. To reduce the risk of wire snapping, and replacement and corrosion of embedded batteries, wireless delivery of energy to these devices is desirable. However, current autonomous implants remain large in scale due to the operation at very low frequency and the use of unwieldy size of antennas . This paper will show that the optimal frequency is about 2 orders of magnitude higher than the conventional wisdom; and thereby the power receiving coils can be reduced by more than 100 fold without sacrificing either power efficiency or range. We will show that a mm-sized implant can receive 100's μW of power under safety constraints. This level of power transfer is sufficient to enable many functionalities into the micro-implants for clinical applications.

1. Introduction

Implantable medical devices will play an important role in modern medicine for preventive and post-surgery monitoring, drug delivery, local stimulation, and biomimetic prosthesis. To reduce the risk of wire snapping, and replacement and corrosion of embedded batteries, wireless delivery of energy to these devices is desirable. Low-frequency electromagnetic field as the carrier in conjunction with inductive coupling as the transmission mechanism is a commonly used approach. In the past fifty years, analyses, circuit design techniques, link optimization, and prototype implementations [1–4] tended to operate at frequencies below 10 MHz. This requires the diameter of receive antennas to be a few cm and hence limits the clinical application of implantable devices.

In this paper, we use a four-step approach to show that the optimal frequency for wireless power transmission over dispersive tissue is at least 2 orders of magnitude higher. We begin with closed-form analysis to derive the optimal frequency for transmission over a homogeneous medium where the dielectric properties of tissue are parameterized by the Debye relaxation model. Then, we take into account the body-boundary effect and perform numerical analysis over planarly layered medium. Afterwards, we include the antenna effect by performing electromagnetic simulation. Finally, we include safety constraints and show that a mm-sized implant can receive 100's μW of power.

2. Homogeneous Dispersive Tissue

Let us first review the reasoning behind the popular use of low-frequency carrier in wireless power transmission. As tissue absorption increases with frequency, most analyses assume that lower frequency would yield better transfer efficiency. They therefore omit the displacement current. The propagation of electromagnetic field is then governed by a diffusion equation which is a quasi-static approximation to Maxwell's equations. Solving the diffusion equation reveals that electromagnetic fields decay exponentially inside tissue and the length of diffusion is inversely proportional to the square root of frequency. That is, higher frequency decays faster which agrees with the initial assumption and reinforces the use of low-frequency carrier. However, the diffusion equation is a valid approximation for good conductors, and tissue is better modeled as a low loss dielectric in which displacement current is significant. In this section, we include the displacement current and model the frequency variation of the relative permittivity by the Debye relaxation model [5]:

$$\varepsilon_r(\omega) = \varepsilon_\infty + \frac{\varepsilon_{r0} - \varepsilon_\infty}{1 - i\omega\tau} + i\frac{\sigma}{\omega\varepsilon_0} \tag{1}$$

where τ is the relaxation time constant, ε_{r0} is the static relative permittivity, ε_∞ is the relative permittivity at frequencies where $\omega\tau \gg 1$, and σ is the static conductivity. Each type of tissue is characterized by 3 parameters: ε_{r0}, ε_∞, and τ.

We model the transmitted field from the source by

Table 1: Approximate optimal frequency assuming $d = 1$ cm.

Type of tissue	f_{opt} (GHz)
Blood	3.54
Bone (cancellous)	3.80
Bone (cortical)	4.50
Brain (grey matter)	3.85
Brain (white matter)	4.23
Fat (infiltrated)	6.00
Fat (not infiltrated)	8.64
Heart	3.75
Kidney	3.81
Lens cortex	3.93
Liver	3.80
Lung	4.90
Muscle	3.93
Skin (dry)	4.44
Skin (wet)	4.01
Spleen	3.79
Tendon	3.17

the field emanated from the set of lowest order magnetic multipoles, and assume that the scattered field from the receiver is negligible. Then the electromagnetic fields at a point $\mathbf{r}$ in the medium are:

$$\mathbf{H}(\mathbf{r}) = \frac{iI_tA_t}{4\pi}\left[\alpha_x\psi_x(\mathbf{r}) + \alpha_y\psi_y(\mathbf{r}) + \alpha_z\psi_z(\mathbf{r})\right] \quad (2a)$$

$$\mathbf{E}(\mathbf{r}) = -\frac{\omega\mu_0 I_tA_t}{4\pi}\left[\alpha_x\xi_x(\mathbf{r}) + \alpha_y\xi_y(\mathbf{r}) + \alpha_z\xi_z(\mathbf{r})\right] \quad (2b)$$

where $k = \omega\sqrt{\mu_0\varepsilon_0\varepsilon_r}$, I_tA_t is the transmit magnetic moment, $\psi_x(\mathbf{r})$ and $\xi_x(\mathbf{r})$ give the respective magnetic and electric fields due to a magnetic dipole pointing in the x direction, and $(\alpha_x, \alpha_y, \alpha_z)$ is the orientation of the transmit dipole. Suppose the receive dipole is at $-\hat{\mathbf{z}}d$ and is pointing in the direction $\hat{\mathbf{n}}$, and the tissue region is defined by $z < d_1$. Then, we define the power transfer efficiency as the ratio of received power to total tissue absorption:

$$\eta := \frac{\omega^2\mu_0^2A_r^2\,\mathrm{Re}\,\frac{1}{Z_L}\left|\mathbf{H}(-\hat{\mathbf{z}}d)\cdot\hat{\mathbf{n}}\right|^2}{\omega\varepsilon_0\int_{z<-d_1}\mathrm{Im}\,\varepsilon_r(\omega)\left|\mathbf{E}(\mathbf{r})\right|^2 d\mathbf{r}} \quad (3)$$

where A_r is the area of the receive dipole and Z_L is the load impedance.

For a given $\hat{\mathbf{n}}$, we derive the frequency that maximizes the efficiency over all possible orientations of the transmit dipole α_m's. The optimal frequency satisfies

$$\omega_{opt} \approx \sqrt{\frac{c\sqrt{\varepsilon_{r0}}}{d\tau\Delta\varepsilon}} \quad (4)$$

Based on the measured data in [6], Table 1 lists the approximated optimal frequencies for 17 different kinds of tissue assuming $d = 1$ cm. All approximated optimal frequencies are in the GHz-range. They are above 1 GHz even for $d = 10$ cm. Furthermore, the transfer efficiency at the optimal frequency is approximately proportional to $\frac{1}{d^3}$. This implies that the regime for optimal power transmission is in between the far field and the near field.

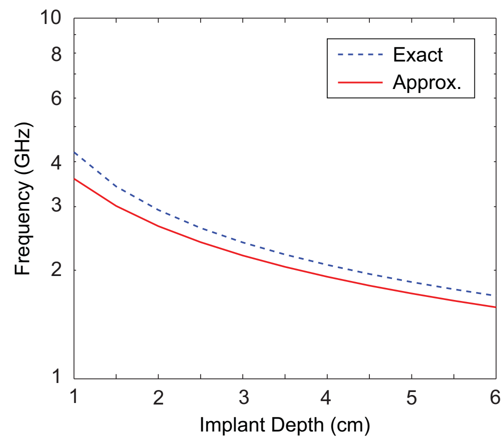

Figure 1: Approximated vs. exact optimal frequency in homogeneous medium

As muscle is the most widely reported tissue, let us take muscle as an example. We consider a receive dipole tilted 45° with respect to the z-axis. The area of the receive dipole is (2 mm)^2. We first compute the efficiency that optimizes the orientation of the transmit dipole for the given receive orientation. Then, we find the frequency that yields the maximum efficiency for different implant depth, and compare it with the approximation in (4). Fig. 1 shows these curves. The approximated and exact optimal frequencies are very close.

3. Planarly Layered Body Model

The analyses in homogeneous medium reveal that the effect of tissue absorption on the power transfer efficiency is not as worse as we used to believe. Higher efficiency can be achieved by operating at higher frequency until reaching the optimal frequency. In this section, we include the air-tissue interface and study its effect on the optimal frequency. We model human body as a planarly layered medium, as illustrated in Fig. 2. We numerically compute the optimal frequency for three different models of the source region: (1) a magnetic dipole with orientation optimized for the given receive orientation, (2) a magnetic dipole pointing in the z direction, and (3) a square current loop with uniform current distribution. Fig. 3 plots the optimal frequency versus the implant depth. The transmit source is 1 cm above the tissue in-

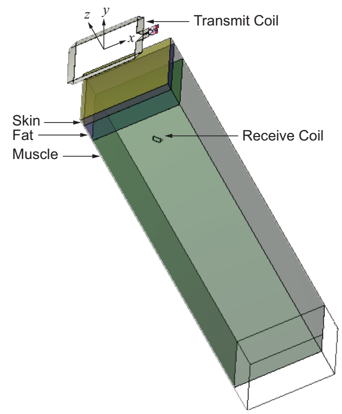

Figure 2: 3D view of the transmit coil, the receive coil, and the tissue model in IE3D

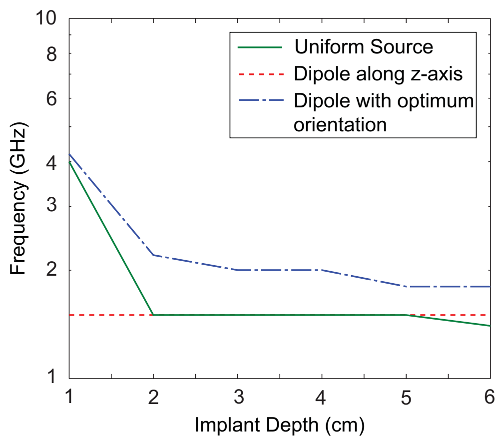

Figure 3: Optimal frequency vs. depth of implant in planarly layered medium

terface and the receive dipole is tilted 45° with respect to the z axis. The thickness of skin and fat are 2 mm and 5 mm respectively. The width of the square current loop is 2 cm. All optimal frequencies are above 1 GHz. The uppermost curve in Fig. 3 follows closely to the curves in Fig. 1. This implies that the air-tissue interface does not have significant effect on the optimal frequency.

4. Matching at Receiver

The received power depends on the load impedance Z_L. In the inductively coupled link, the load impedance is chosen to achieve a given output voltage. Various tuning configurations – series vs. shunt tuning – are chosen to maximize the power transfer efficiency. Shunt tuning at receiver and series tuning at transmitter is commonly used. At 1 GHz or above, the mutual impedance between the transmit and the receive antennas is no longer purely imaginary as in the inductive link. We should

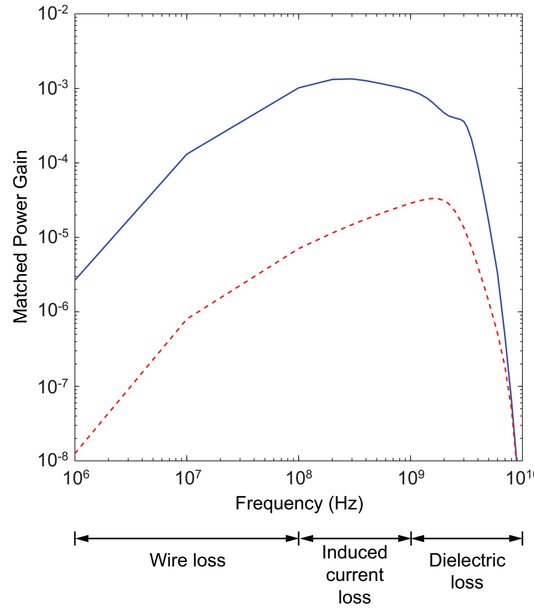

Figure 4: Simulated matched power gain vs. frequency for transmit coil of width 2 cm (solid line) and 2 mm (dotted line).

use a different approach and model the power link as a two-port network. As our objective is to maximize the power transfer efficiency, we should perform simultaneously conjugate matching at both transmitter and receiver. The output voltage requirement is then fulfilled by the choice of different forms of matching network. As a result, the load impedance Z_L in the definition of η is related to the self-impedance of the receive antenna. At conjugate matching, the total impedance of port 2 is approximately equal to 2 times the real part of the self-impedance and half of the power absorbed by port 2 goes to the load. Therefore, the load impedance Z_L can be chosen to be 4 times the real part of the self-impedance. In the last section, we choose $Z_L = 5.6\ \Omega$ which is 4 times the resistance of a 2-mm side square copper loop.

As the self-impedance of the receive antenna changes with frequency, the optimal choice of Z_L changes with frequency as well. To understand the effect of load impedance on the optimal frequency, we perform electromagnetic simulations. We use Zeland IE3D full-wave electromagnetic field solver to obtain the S-parameters for the 2-port system in Fig. 2. The tissue model and orientation of the coils are the same as in last section. The transmit coil is a 2-cm side square copper loop with a trace width of 2.00 mm and trace thickness of 0.04 mm. The receive coil is a 2-mm side square copper loop with a trace width of 0.20 mm and trace thickness of 0.04 mm. Fig. 4 plots the variation of matched power gain with frequency. The loss in the matched power gain includes not only tissue absorption

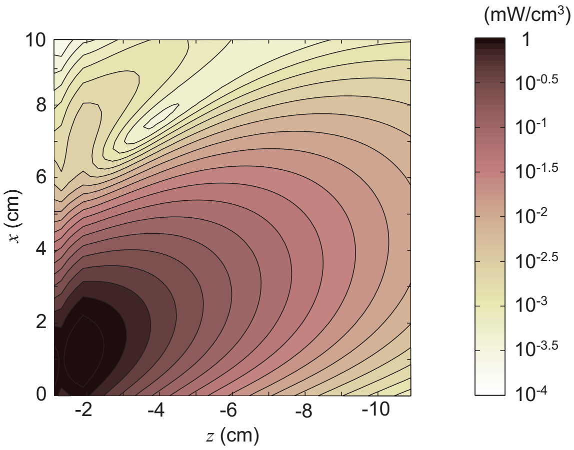

Figure 5: Local SAR distribution at 1 GHz.

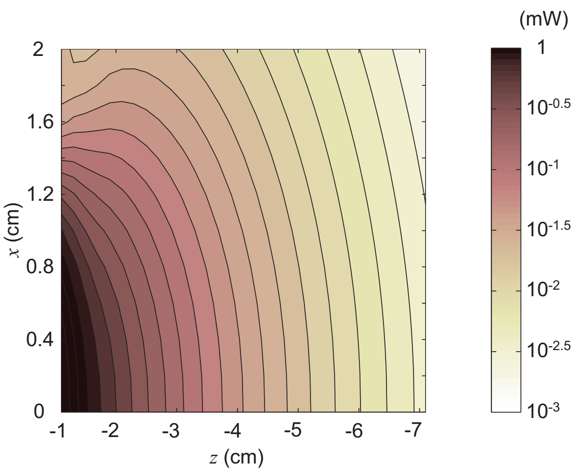

Figure 6: Received power distribution at 1 GHz.

but also ohmic loss in both coils and radiation loss from the transmit coil. At low frequency, the ohmic loss dominates but its rate of increase with frequency is less than that of the received power. As a result, the power gain increases with frequency. The rate of increase reduces when loss from induced current dominates. Finally, it decreases steeply with frequency when dielectric loss dominates.

The optimal frequency for the 2-cm side transmit coil shifts to the sub-GHz range due to the unoptimized design of the transmit antenna. The wavelength at 1 GHz is about 4 cm and the dimension of the transmit antenna is in this range. Therefore, future work will focus on designing the transmit antenna such that the optimal frequency will shift back to the GHz-range.

5. How Much Power Can Be Delivered?

As the transmit source is close to the tissue, we adopt the safety guidelines for specific absorption rate (SAR). In the IEEE guideline, the SAR limits are 1.6 W/kg for any 1 g of tissue and 4 W/kg for any 10 g of tissue. In the ICNIRP guideline, the SAR limit is 4 W/kg for any 10 g of tissue in hands, wrists, feet, and ankles, and 2 W/kg for any 10 g of other tissue. In general, 1 g of tissue approximately occupies 1 cm^3 and 10 g of tissue occupies 2.15 cm^3. That is, the tissue absorption should not exceed 1.6 mW in 1 cm^3 and 20 mW to 40 mW in 2.15 cm^3.

We consider a square loop of width 2 cm at a distance 1 cm above the air-tissue interface and an air-skin-fat-muscle multi-layered medium. We first assume $I_t A_t$ equal to 1 Am^2 and compute the SAR distribution. Then, we scale the SAR distribution by a factor such that the maximum value of the scaled distribution equals to 1.6 mW/cm^3. This scaling factor also gives the transmit current which is 280 mA. Fig. 5 plots the SAR distribution at 1 GHz after scaling. The corresponding received power distributions are plotted in Fig. 6. At an implant depth of 2 cm ($z = -3$ cm), the received power is 150 μW for a dipole pointing in the z direction. If the less stringent SAR limit is used, the received power will be increased by approximately 10 times. As a result, sub-mW to mW of power can be safely delivered to a mm-sized receive antenna from a cm-sized transmit antenna with a separation of a few cm at carrier frequency of 1 GHz.

References

[1] T. Akin, K. Najafi, and R. M. Bradley, "A wireless implantable multichannel digital neural recording system for a micromachined sieve electrode," *IEEE J. Solid-State Circuits*, vol. 33, no. 1, pp. 109–118, Jan. 1998.

[2] W. Liu, et al., "A neuro-stimulus chip with telemetry unit for retinal prosthetic device," *IEEE J. Solid-State Circuits*, vol. 35, no. 10, pp. 1487–1497, Oct. 2000.

[3] C. Sauer, et al., "Power harvesting and telemetry in CMOS for implanted devices," *IEEE Trans. Circuits Syst. I*, vol. 52, no. 12, pp. 2605–2613, Dec. 2005.

[4] M. W. Baker and R. Sarpeshkar, "Feedback analysis and design of RF power link for low-power bionic systems," *IEEE Trans. Biomed. Circuits Syst.*, vol. 1, no. 1, pp. 28–38, Mar. 2007.

[5] A. V. Vorst, A. Rosen, and Y. Kotsuka, *RF/Microwave Interaction with Biological Tissues*. New Jersey, USA: Wiley-IEEE Press, 2006.

[6] S. Gabriel, R. W. Lau, and C. Gabriel, "The dielectric properties of biological tissues: III. parametric models for the dielectric spectrum of tissues," *Phys. Med. Biol.*, vol. 41, no. 11, pp. 2271–2293, Nov. 1996.