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Physical Insights on Wireless Powering of Deep-Body Ultraminiature Bioelectronics

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Abstract—The development of wireless communication and miniaturized electronics has resulted in the emergence of deep-body bioelectronics, which has the potential to significantly enhance the diagnosis, treatment, and monitoring of medical conditions. However, the assimilation of these devices into clinical practice hinges on optimal wireless connectivity and powering. Wireless connectivity is a key feature for real-time data transmission and device regulation within the body. In addition, wireless powering allows for further miniaturizing the devices by removing batteries. However, it is still a challenge due to electronic constraints and the inherent lossy nature of the human tissues. This work analyzes the physical concepts behind wireless connectivity and powering deep-body bioelectronics to address the fundamental bounds on their efficiency and propose novel approaches to enhance their performance.

Keywords—bioelectronics, deep-body implants, physical bounds, wireless connectivity, wireless powering.

I. INTRODUCTION

The rapid advancements in wireless communication and miniaturized electronics have revolutionized the healthcare field, enabling the development of deep-body bioelectronics. These innovative devices, ranging from implantable sensors [1]–[3] to therapeutic implants [4]–[6], have the potential to significantly improve the diagnosis, treatment, and monitoring of various medical conditions. However, the successful integration of deep-body bioelectronics into clinical practice critically depends on two main factors: optimal wireless connectivity and efficient powering.

Wireless connectivity enables real-time data transmission, remote monitoring, and seamless communication between bioelectronic devices and external systems. Establishing reliable and robust wireless connections deep within the human body ensures accurate and timely data collection and facilitates device control and reconfiguration. However, challenges such as signal attenuation and limitations in wireless propagation pose significant obstacles that must be overcome to achieve optimal connectivity for deep-body bioelectronics [7].

Additionally, the powering of deep-body bioelectronics presents unique challenges due to the constraints of miniaturized devices and the lossy and heterogeneous nature of the human body [8]. Furthermore, traditional power sources, such as batteries, have limitations in terms of size, capacity, and the need for periodic replacement or recharging. Therefore, developing efficient wireless powering solutions is crucial for the long-term, reliable operation of bioelectronic devices, reducing the need for invasive procedures and ensuring patient safety and comfort.

Therefore, this contribution aims to overview the physical aspects of wireless connectivity and powering of deep-body bioelectronics, evaluate the fundamental bounds on their

efficiency, and discuss different approaches that allow us to reach performance close to the optimal values.

II. PROBLEM FORMULATION

The two-dimensional-axisymmetric model presented in Fig. 1 is used to assess the maximum efficiency of an implantable antenna. The antenna is represented by a cylindrical surface with length L and radius R_c , surrounded by a lossless capsule-shaped region filled with a lossless dielectric perfectly matched to the wave impedance in the surrounding medium. This implant is positioned at the center of a spherical phantom with radius R_p and electric permittivity $\epsilon_r(f)$ equivalent to the human muscle [9]. The external medium is also considered lossless and matched to the wave impedance in the muscle phantom to eliminate the losses due to the reflection. Therefore, the efficiency obtained with this model is limited only by intrinsic losses (reactive near-field dissipation and power absorption in the lossy medium) and corresponds to the maximum efficiency values. In addition, the radiating source is modeled by its equivalent current distributions \mathbf{J}_s on the cylindrical encapsulation, as shown in Fig. 1(b) for an electric and (c) for a magnetic source.

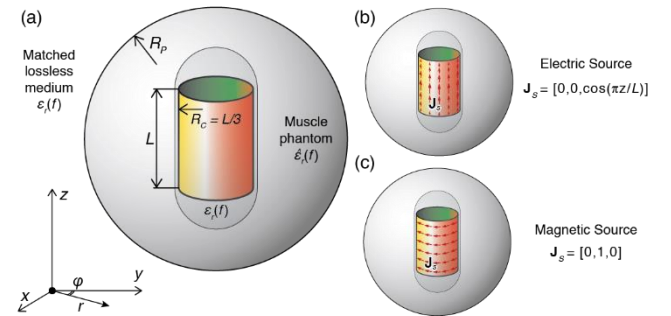


Fig. 1. Two-dimensional-axisymmetric formulation of the problem: (a) a capsule-shaped source is positioned at the center of a spherical muscle-like phantom. An (b) electric and (c) magnetic source are modeled based on their equivalent current distribution in cylindrical coordinates (r, ϕ, z) .

This problem can be mathematically formulated by employing energy conservation. In this way, the power P_t transmitted by the implanted antenna is given by the sum of the power flow around the phantom interface P_r , the power dissipated in the lossy tissue P_d , and the time-average energy stored in the magnetic W_m and electric W_e fields:

$$P_t = P_r + P_d + i2\omega(\overline{W_m} - \overline{W_e}). \quad (1)$$

Therefore, the antenna efficiency η can be defined as the ratio between the real-valued received and transmitted powers:

$$\eta = \frac{\Re[P_r]}{\Re[P_t]} = \frac{\Re\left[\oint_{\Omega_p} \left(\frac{1}{2} \mathbf{E} \times \mathbf{H}^*\right) \cdot d\mathbf{s}\right]}{\Re\left[\oint_{\Omega_s} \left(\frac{1}{2} \mathbf{E} \times \mathbf{H}^*\right) \cdot d\mathbf{s}\right]} \quad (2)$$

where \mathbf{E} and \mathbf{H} are the time-harmonic electric and magnetic fields, respectively, and the power flow integrals are evaluated around the phantom Ω_p and the implant encapsulation Ω_s . In this work, the fields and the numerical integration were evaluated using COMSOL Multiphysics.

III. IN-BODY RECEIVERS

The design of in-body antennas is constrained by limitations on their shape and dimensions; however, they can be properly tailored in order to achieve performance close to the optimal values. Firstly, the antenna type (electric or magnetic) that leads to maximum efficiency depends on the implantation depth and operating frequency, as it is shown in Fig. 2. As can be noticed, a magnetic-type receiver (e.g., a loop antenna) has been found to have higher efficiencies than electric-type antennas at lower frequencies. However, as it approaches the sub- and low-GHz bands, the high permittivity of human muscle increases the effective aperture size of an electric source, enhancing its efficiency. Note that this analysis does not consider the ohmic losses of realistic antennas. Specifically, miniature loop antennas can exhibit high losses at lower frequencies reducing their applicability for body-implanted applications [10].

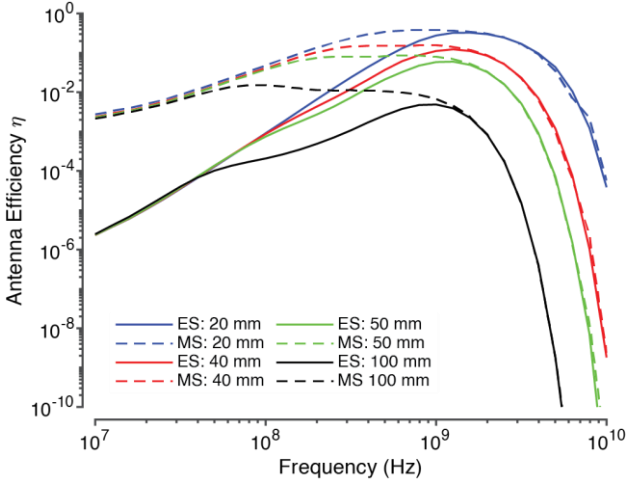


Fig. 2. Maximum implantable receiver efficiency as a function of the operating frequency and implantation depth considering electric (continuous lines) and magnetic (dashed lines) receivers.

Generally, the losses related to near-field coupling are dominant at lower frequencies, whereas tissue attenuation causes most of the power loss at higher frequencies. Hence, it is plausible to assume that there is a frequency that optimizes the trade-off between these two mechanisms of loss, which is confirmed in Fig. 2. Besides, even though this optimal frequency decreases as the implantation depth increases, it remains within the range of ISM bands between 915 MHz and 2.45 GHz. A cautionary remark: the wave-impedance mismatch losses on the surface of the body are absent in this analysis since we assume that the on-body wireless power source would be perfectly matched to tissues. Therefore, the results in Fig. 2 differ from the bounds on implantable antenna efficiency reported in [7], [11].

Moreover, engineering techniques can improve the performance of in-body antennas, despite the limitations imposed on their design. For example, one way to enhance the gain of implantable antennas is by using high-permittivity encapsulations [12]. The dielectric loading increases the aperture size and reduces near-field losses, thus increasing the

efficiency. Other options to enhance their performance are filling the encapsulation with a low-loss dielectric material and using the thickest possible substrate [13]. Finally, reconfigurable receiver antennas can achieve near-optimal radiation performance across a broad range of conditions [14].

IV. ON-BODY TRANSMITTERS

Once the medium is linear and considering that there is no polarization mismatch, the physical problem of wireless powering a bioelectronic implant can be understood as reciprocal to the previously analyzed case (this approach has already been explored and validated in [8]). Therefore, the transmitter is the on-body source entirely conformal to the phantom interface, whereas the implanted antenna is the receiver. In this way, the results presented in Fig. 2 also correspond to the maximum achievable WPT efficiency.

The development of battery-free bioelectronics requires efficient WPT techniques to power miniaturized implants. However, current approaches in the literature do not perform optimally. One reason for that is their reliance on single-antenna transmitters. The physical bounds are obtained considering a body-conformal radiating structure; therefore, conformal transmitter arrays are preferable to mono-antenna configurations. In addition, conformal transmitter arrays can be reconfigured to focus the electromagnetic fields toward the receiver, covering a wider implantation range. Another way to increase the WPT efficiency is by designing a transmitter that establishes an optimal wave path from the transmitter to the receiver. Therefore, a well-designed transmitter can mitigate losses associated with reflection on the air-skin interface and the reactive near-field.

Implantable WPT applications typically position the transmitter antenna directly on the skin. Even though, in this case, the transmitter is closer to the receiver, the near-field is located in the lossy tissues, as can be seen in Fig. 3, which shows the wave impedance $|Z_w|$ profile shown across several layers of human tissue, considering a normally impinging plane wave. In this case, the far-field region is located after the last muscle layer and inside the heart, where $|Z_w|$ converges to approximately 50Ω . In this way, part of the energy stored in the near-field is dissipated, thus reducing the power delivered to the implanted device.

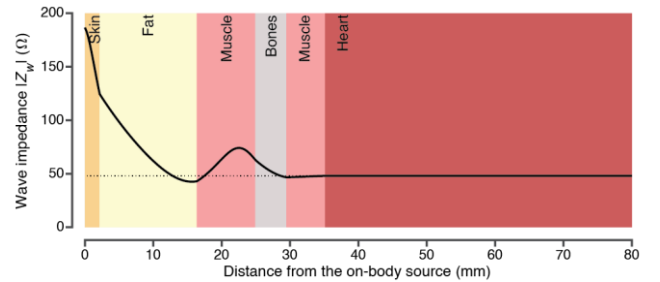


Fig. 3. Wave impedance $|Z_w|$ across a planar stratified model of the human torso, considering a normal plane wave incidence.

One way to mitigate these losses and improve the WPT efficiency is by using a buffer structure that interfaces the transmitter and the skin. A lossless superstrate can confine the near-field in the buffer, reducing losses associated with the reactive near-field. It can be achieved by employing low-loss substrates with optimized permittivity and thickness to reduce wave impedance mismatching at the air-skin interface. For instance, the wave impedance at the buffer-skin boundary is presented in Fig. 4, considering a buffer with $\lambda/4$ thickness,

with λ being the guided wavelength in the superstrate. As can be seen, the wave impedance converges to the far-field values when the permittivity is higher than ϵ_{skin} . Once most of the layers in Fig. 3 are composed of high-permittivity tissues, the effective permittivity ϵ_{eff} is in the range $\epsilon_{skin} < \epsilon_{eff} < \epsilon_{heart}$, which corresponds to the range in which $|Z_w|$ converges.

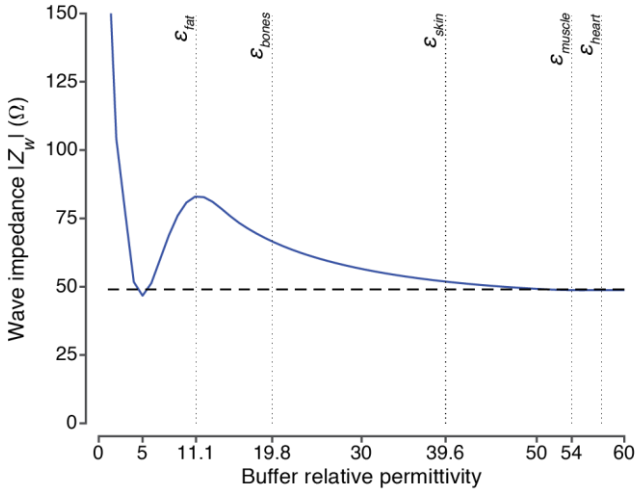


Fig. 4. Wave impedance at the interface buffer-skin as a function of the buffer permittivity, considering a quarter-wavelength buffer.

Buffered transmitters not only increase efficiency but also reduce electromagnetic exposure. This is because the highest electric field values are restricted to the superstrate area, preventing them from reaching human tissues. Therefore, by utilizing conformal transmitter arrays and buffer structures, WPT performance can be maximized while minimizing power loss and electromagnetic exposure [8].

V. CONCLUSION

The application of wireless communication and powering techniques to deep-body, mm-sized bioelectronics have paved the way for promising treatments and diagnostics that can revolutionize healthcare. However, successfully integrating these devices into clinical practice relies heavily on achieving performance close to optimal values. Mitigating losses such as power attenuation, near-field dissipation, and wave reflection is crucial for fully realizing the potential of bioelectronic devices. Besides, it reduces electromagnetic exposure, thus ensuring user safety for the long-term operation of bioelectronic devices within the human body. By exploring the physical concepts regarding in-body wireless communication and powering, this paper proposes novel solutions to achieve optimal performance.

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