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A Comprehensive Comparative Study on Inductive and Ultrasonic Wireless Power Transmission to Biomedical Implants

Ahmed Ibrahim [Student Member, IEEE],

Electrical Engineering Department at the Pennsylvania State University, University Park, PA 16802, USA

Miao Meng [Student Member, IEEE], and

Electrical Engineering Department at the Pennsylvania State University, University Park, PA 16802, USA

Mehdi Kiani [Member, IEEE]

Electrical Engineering Department at the Pennsylvania State University, University Park, PA 16802, USA

Abstract

This paper presents a comprehensive comparison between inductive coupling and ultrasound for wireless power transmission (WPT) to biomedical implants. Several sets of inductive and ultrasonic links for different powering distances (d_{12}) and receiver dimensions have been optimized, and their key parameters, including power transmission efficiency (PTE) and power delivered to the load (PDL) within safety constraints, have been compared to find out which method is optimal for any given condition. Two design procedures have been presented for maximizing the PTE of inductive and ultrasonic links by finding the optimal geometry for the transmitter (Tx) and receiver (Rx) coils and ultrasonic transducers as well as the optimal operation frequency (f_p). Our simulation and measurement results showed that the ultrasonic link transcends the inductive link in PTE and somewhat in PDL for a small Rx of 1.1 mm³ (diameter of 1.2 mm), particularly when the Rx was deeply implanted inside the tissue ($d_{12} = 10$ mm). However, for a larger 20 mm³ Rx (diameter of 5 mm), the inductive link achieved higher PTE and PDL, particularly at shorter distances ($d_{12} < 30$ mm). The optimal loading condition is shown to be quite different in inductive and ultrasonic links. Despite higher performance for small Rx and large d_{12} , the ultrasonic link is more sensitive to Rx misalignments and orientations. This led us to propose a new design procedure based on the worst-case misalignment scenario. The simulation results have been validated by measurements. The inductive and ultrasonic links, operating at 30 MHz and 1.1 MHz, achieved measured PTEs of 0.05% and 0.65% for the 1.1 mm³ Rx located 30 mm inside tissue and oil environments with optimal load resistances of 295 Ω and 3.8 k Ω , respectively.

Index Terms

Wireless power transmission; inductive links; ultrasound; biomedical implants; efficiency; design methodology

I. Introduction

WIRELESS power transmission (WPT) is widely used in many applications with different power requirements to eliminate power cords [1]-[8]. WPT to biomedical implants can eliminate the need for bulky batteries with a limited lifetime, and reduce the implant size [9]. WPT is vital in today's implantable medical devices (IMDs) that interface with the central and peripheral nervous systems (CNS and PNS) for recording and stimulating neuronal activity using different modalities such as electricity, light, and ultrasound [10]-[15].

IMDs can potentially be utilized in a wide range of applications such as cochlear, retinal, cortical, and PNS implants [16]-[19]. The required power range for IMDs varies significantly due to widely varying system requirements and constraints. For example, IMDs are used with various media—such as air, bone, and tissue—and with different implantation depths from several millimeters to centimeters, as well as different functions, such as monitoring and/or modulation of neural activity with different numbers of channels and modalities (e.g. electrical, optical, and ultrasound) [20]. Cochlear and retinal implants require tens of mW transferred across sub-cm to cm distances [16], [17]. The power consumption of cortical implants ranges from sub-mW to tens of mW, which highly depends on the number of recording/stimulation channels and their function (recording and/or stimulation) [18]. State-of-the-art recording front ends consume several μW of power, however, the data transmitter often dominates the total power consumption in IMDs, particularly if the number of channels is high [21]. PNS implants for restoring the dysfunctional motor and sensory functions in the limbs require transmitted power levels of tens to hundreds of mW across several millimeters due to the large stimulation current [22]. There are also cardiac and gastric slow-wave recording implants that only require hundreds of μW or less, but across several centimeters [15], [19]. Achieving robust, efficient, and safe WPT to small implants located deep inside the body is of great importance in all applications, regardless of the challenges posed by the wide range of system requirements.

Fig. 1 shows the generic block diagram for a wirelessly powered implant with emphasis on its main blocks for WPT. In the transmitter side (Tx), it includes an external energy source that supplies an efficient power amplifier (PA) to drive the primary element, which, in this paper, is either a series resonant LC-tank ($L_1 C_1$) or an ultrasonic transducer (U_1). The energy source drives the primary element at the power carrier frequency (f_p). In the receiver side (Rx), it includes a secondary element (parallel resonant $L_2 C_2$ -tank or U_2) that receives the AC power carrier, and a power management block to convert the incoming AC signal to a constant DC supply for the implant core. In this paper for simplicity, the power management and implant core are modeled as an AC load, R_L in Fig. 1.

Until recently, inductive coupling was the conventional method for WPT to biomedical implants, particularly for those with centimeter (cm) dimensions. This is due to its high

power transmission efficiency (PTE) defined as the delivered power to R_L divided by the Tx element's ($L_1 C_1$ or U_1) input power as well its minimal dissipation of the magnetic field inside the tissue, and safety considerations [23]-[26]. Recently, several efforts have been made towards the miniaturization of implants to millimeter (mm) and even sub-mm dimensions in order to reduce the tissue damage and inflammation, and improve the implant's longevity [27]-[33]. For improving PTE of mm-sized coils, a common theme has been the significant increase of f_p to hundreds of MHz.

In [28]-[30], it has been suggested that the optimal f_p to deliver power to an implant coil with an outer diameter (D_{o2}) of 2 mm is within the GHz/sub-GHz range. An f_p of 120 MHz has been proposed in [31] for powering an on-chip coil with D_{o2} of 1 mm. An optimal f_p around 300 MHz has been used to deliver power to a single-loop square coil with D_{o2} of 1 mm and 5 mm in [21],[32]. The geometry and f_p of the inductive link have been optimized in [33] for D_{o2} of 1 mm to maximize PTE for a fixed R_L of 5 k Ω , resulting in the optimal f_p = 200 MHz. We have also presented the design and optimization of inductive links for powering mm-sized implants by proposing a new figure-of-merit (FoM) to achieve not only high PTE but also maximum delivered power to the load (PDL) within safety constraints [34], [35]. At the powering distance (d_{12}) of 12 mm with D_{o2} = 1 mm, we have demonstrated two inductive links at f_p s of 20 MHz and 100 MHz with the PTE and maximum PDL of 1.4% and 3.3%, and 2.2 mW and 0.72 mW, respectively.

Although these new designs and optimizations have partially improved the PTE and PDL of inductive/RF-based links, small PTE and stringent safety constraints have recently led to the emergence of ultrasound-based WPT links for mm-sized implants [36]-[41]. The ultrasonic link in [36] has achieved a PTE of 0.002% in delivering power to an ultrasonic Rx with a size of 0.127 mm³ at d_{12} = 3 cm using a commercial Tx transducer (U_1) operating at f_p = 5 MHz. An ultrasonically powered implant has been reported in [37], delivering a regulated power of 100 μ W to an ultrasonic Rx with a diameter (D_{ou2}) of 1 mm at d_{12} = 3 cm using a 1 MHz commercial U_1 . For optimal WPT to mm-sized implants, we have also presented the design and optimization of ultrasonic links, based on which an ultrasonic link at f_p = 1.8 MHz has been demonstrated with the measured PTE of 0.66% at a d_{12} of 3 cm for D_{ou2} = 1.2 mm [39].

A comprehensive study of inductive and ultrasonic links for different conditions in terms of implant size, depth, misalignment, and orientation can help the designers of WPT for biomedical implants to choose the optimal modality. In [40], simulated PTEs of inductive and ultrasonic WPT links for D_{o2} of 2–10 mm have been compared for different d_{12} 's, however, the conclusions have been made based on non-optimized inductive and ultrasonic links, which can be misleading. In [41], the measured and simulated PTEs of an ultrasonic link has been compared with the simulated PTEs of a comparably sized inductive link with identical Tx and Rx diameters of 4.4 mm. Again, both inductive and ultrasonic links have not been optimized in this work. In addition, the maximum PDLs of inductive and ultrasonic links under safety constraints have not been compared in the past.

Recently, we have also briefly compared the simulated PTEs of inductive and ultrasonic links for small and large implants [42]. However, optimal link parameters for different d_{12} 's

were not reported, and no comparisons were provided for different d_{12} 's or for the rotation of both inductive and ultrasonic implants, and maximum PDL under safety constraints was not considered. In addition, no detailed optimization procedure, measurement results, or design guidelines for ultrasonic links based on their misalignment were provided in [42]. Therefore, the literature lacks a comprehensive, accurate, and validated (through measurements) comparison between key parameters (both PTE and PDL) of inductive and ultrasonic links for WPT to biomedical implants with different size, depth, alignment, orientation, and power requirements in order to clarify which WPT method is more efficient and robust for a given application.

In this paper, we present a performance comparison in terms of PTE and PDL between several sets of inductive and ultrasonic links. The links have optimal geometries and f_p for two different Rx sizes of 1.1 mm^3 (diameter = 1.2 mm) and 20 mm^3 (diameter = 5 mm) at various d_{12} 's of 5–50 mm. The PTE and PDL of these links vs. Rx misalignment and orientation will also be discussed. A design guideline for ultrasonic links will also be presented based on worst-case misalignment scenarios. This paper focuses on providing design guidelines and clarity for the designers of IMDs as to which modality is more suitable for any given application. For applications that involve bone (e.g. cortical implants) or air (e.g. retinal implants), ultrasound by itself is not suitable due to the large ultrasound loss and reflection. However, we have recently demonstrated a hybrid inductive-ultrasonic link, which can be used for such applications [43]. To demonstrate the proposed links in a real-world application, we will also provide a design example for a 1-channel gastric slow-wave recording implant with $10 \mu\text{W}$ of power consumption at an implantation depth of 5 cm [15].

Section II summarizes the design parameters and procedures for both inductive and ultrasonic links. The comparative simulation and measurement results will be discussed in Sections III and IV, respectively. Finally, discussions and concluding remarks will be presented in Sections V and VI, respectively.

II. Design and Optimization of Inductive and Ultrasonic WPT Links

A. Inductive WPT Link

The WPT link in Fig. 1 can be converted to a conventional 2-coil inductive link by replacing the primary and secondary elements with two resonant LC-tanks, i.e., series $L_1 C_1$ and parallel $L_2 C_2$, respectively. The PTE of the inductive link, as its model is shown in Fig. 2, can be found from,

$$\eta = \frac{k_{12}^2 Q_1 Q_{2L}}{1 + k_{12}^2 Q_1 Q_{2L}} \cdot \frac{Q_{2L}}{Q_L}, \quad (1)$$

where $Q_1 = \omega L_1 / R_1$, $Q_2 = \omega L_2 / R_2$, $\omega = 2\pi f_p = (L_1 C_1)^{-0.5} = (L_2 C_2)^{-0.5}$, $Q_{2L} = Q_2 Q_L / (Q_2 + Q_L)$, and $Q_L = R_L / \omega L_2$ is referred to as the load quality factor [44]. The term Q_{2L} / Q_L

represents the loss inside Rx and any R_L mismatch. In order to find the optimal R_L that maximizes PTE, i.e., $R_{L,PTE}$, the optimal $Q_{L,PTE} = R_{L,PTE}/\omega L_2$ can be found from

$$Q_{L,PTE} = \frac{Q_2}{\sqrt{1 + k_{12}^2 Q_1 Q_2}}, \quad (2)$$

which can be further simplified to Q_2 for loosely coupled coils with small k_{12} , particularly in WPT to mm-sized implants [34]. In such small k_{12} and optimal-loading conditions, PTE in (1) can be simplified to $k_{12}^2 Q_1 Q_2 / 4$ [34]. Therefore, for optimal load condition in (2), k_{12} , Q_1 , and Q_2 should all be maximized to optimize PTE. As we have discussed in detail in [34], any given R_L smaller than the optimal $R_{L,PTE}$ can be transformed to $R_{L,PTE}$ by adding one off-chip capacitor, but matching any given R_L larger than $R_{L,PTE}$ requires an inductor, which is not feasible in mm-sized implants.

The IEEE standard for maximum human exposure to RF electromagnetic fields states that the tissue specific absorption rate (SAR) should not exceed a certain value for safe operation. The peak SAR limit, averaged over 1 g of tissue, for RF exposure to the human head is 1.6 W/kg, which will be considered in the rest of this paper without the loss of generality [35]. The maximum allowable PDL under SAR constraints (P_L) can be calculated by multiplying the power provided by the PA with the PTE from (1),

$$\begin{aligned} P_L &= 0.5(R_1 + R_{ref})I_{1,SAR}^2 \times \eta \\ &= 0.5R_1(1 + k_{12}^2 Q_1 Q_{2L})I_{1,SAR}^2 \times \eta \end{aligned} \quad (3)$$

where $I_{1,SAR}$ is the maximum Tx coil current under the SAR constraint and R_{ref} is the reflected resistance to L_1 [44]. For small k_{12} , P_L in (3) can be simplified to $0.5R_1 I_{1,SAR}^2 \times \eta$ [34].

In order to optimize the inductive links we proposed a detailed design procedure for WPT to mm-sized inductive links, made with wire-wound coils (WWCs), in [34]. But that design procedure optimizes a figure of merit for mm-sized implants as opposed to PTE. Fig. 3 shows our modified design procedure to maximize PTE with a matched load condition for cm- and mm-sized implants. We will provide a summary of this procedure in this paper, but we would like to refer the reader to [34] for a more detailed procedure as well as 3D surfaces of optimized parameters. We choose WWCs because they can achieve higher Q and, consequently, higher PTE [45]. The optimization should be done in a full-wave electromagnetic field simulator, such as HFSS (Ansoft, Pittsburgh, PA), as shown in Fig. 2. The design procedure is as follows. Step-1: design constraints are imposed by the application and fabrication technology, including the maximum values for the implant size, i.e., outer diameter (D_{o2}) and height of L_2 , as well as L_1 and L_2 wire widths (w_1, max , w_2, max), the nominal value for the implant depth inside the tissue (d in Fig. 2), and finally the minimum value for the wire spacing between turns (s_{min}) as twice the thickness of the wire

insulation. Step-2: the initial values for L_1 and L_2 geometries, such as w_1 , the number of turns (n_1 and n_2), w_2 , and s_1 , and L_1 spacing from tissue (d_s in Fig. 2) are chosen. Step-3: Q_2 is maximized in an iterative process by sweeping n_2 and w_2 for different f_p s using $s_{2,min}$. This step finds the optimal L_2 geometry. Step-4: PTE in (1) will be maximized in an iterative process by sweeping n_1 , D_{o1} , w_1 , and s_1 for different f_p s. This step leads to optimal L_1 and f_p . Step-5: the optimal d_s that increases Q_1 and consequently maximizes PTE in (1) is found. This leads to a slight increase in the powering distance (d_{12}) to $d + d_s$. Finally, the design is further validated and fine-tuned through measurements in step 6 [34].

It should be noted that Q_1 also depends on the medium surrounding L_1 (particularly its conductance), which affects PTE and PDL in (1) and (3). As shown in Fig. 2, separation of L_1 from the skin (d_s) can help to improve Q_1 at the cost of lowering k_{12} . Therefore, to strike a balance between Q_1 and k_{12} in (1), optimal d_s will be in the range of several millimeters, suggesting that L_1 should always be carried by the subject.

B. Ultrasonic WPT Link

An ultrasonic WPT link can be realized in Fig. 1 by replacing the primary and secondary elements with two disc-shaped ultrasonic transducers (U_1 and U_2); its model is shown in Fig. 4. In this paper, symmetric disk-shaped piezoelectric transducers, made with lead zirconate titanate (PZT), will be used to achieve high electromechanical coupling and less complexity in 3D modeling and computation in finite-element method (FEM) simulation tools [39]. It should be noted that similar design methodologies can be generalized to other types of ultrasonic transducers. As shown in Fig. 4, the ultrasonic transmitter (U_1) should be in perfect contact with the skin to eliminate any acoustic reflections at this interface. In practice, some ultrasonic gels can also be applied at the interface. We have already discussed in detail the theory behind ultrasonic WPT using disk-shaped piezoelectric transducers in [39]. The PTE of such links can be found from,

$$PTE = \frac{I_o \times A_2 \times \eta_2}{P_{in}}, \quad (4)$$

where η_2 is the mechanical-electrical power conversion efficiency of U_2 including any R_L mismatch within the Rx, I_o is the acoustic intensity of the U_1 beam, A_2 is the effective cross-section area of U_2 , and P_{in} is the delivered power to U_1 from the energy source. Any impedance mismatch at the Tx-transducer/skin interface can significantly reduce the PTE due to acoustic reflections that drastically reduce I_o in (4). As discussed in [39], disk-shaped unfocused transducers have a natural focal point where the beam is the narrowest and I_o is maximum. The focal point distance of U_1 depends on the outer diameter of U_1 (D_{ou1}) as well as f_p . The thickness-mode resonance frequencies of U_1 and U_2 should be matched to f_p to increase their mechanical-electrical power conversion efficiencies. In addition, A_2 in (4) is a function of D_{ou2} . Finally, for maximum power transfer in the Rx, any given R_L should be matched to the U_2 impedance at resonance, depending on D_{ou2} and t_2 . Therefore, U_1 and U_2 geometries, including $D_{ou1,2}$ and $t_{1,2}$ in Fig. 4 as well as f_p , need to be co-optimized in a matched-load condition to achieve the highest PTE in (4).

It should be noted that the optimal geometries of the ultrasonic transducers can be accurately predicted if the aspect ratio (D_{ou}/t) is large [39]. However, for most applications involving mm-sized implants, the aspect ratio is restricted due to the small Rx dimension requirement. Therefore, FEM simulation tools such as COMSOL Multiphysics (COMSOL, Burlington, MA), need to be used in simulating and optimizing ultrasonic links for WPT to mm-sized implants. Fig. 4 shows the ultrasonic WPT link model in COMSOL, in which U_1 is air-backed to improve its conversion efficiency and U_2 is mounted on a silicon substrate, integrating the power management and implant-core circuitry, and is located inside castor oil with an attenuation coefficient of 0.8 dB/cm/MHz to mimic the soft tissue. In ultrasonic WPT, since U_1 needs to be in contact with the skin, d_{12} and d are the same ($d_s = 0$). In order to minimize reflections at U_1/U_2 interfaces due to a mismatch in acoustic impedance of two media, matching layers with two materials are used [39]. A perfect matching layer (PML) has also been considered at the boundaries of the medium to avoid acoustic reflections.

For ultrasonic WPT links, the maximum achievable PDL under FDA safety constraints (P_L) is limited by the maximum spatial-peak temporal-averaged intensity (I_{SPTA}) of 720 mW/cm² [46]. In order to find P_L in COMSOL, the link is simulated in 2D Axial Symmetric configuration, and the intensity magnitude, which is equivalent to I_{SPTA} , is plotted. Then the input power is increased by increasing the AC input voltage until I_{SPTA} of 720 mW/cm² is observed, at which P_L can be calculated from the AC voltage across R_L .

Fig. 5 shows the simulated sound-wave intensity of the optimized 1.1 mm³ ultrasonic link (shown later in Table III) at $d = 10$ mm, when the maximum I_{SPTA} of 720 mW/cm² was achieved at the vicinity of U_1 , i.e., in the near field region of ~1 mm. Although matching layers were added to Rx, there are still some reflecting waves from the Rx, which can partially either reinforce or cancel out the transmitted waves depending on the Rx location (d) and f_p . As shown in Fig. 5, the ultrasound intensity within the whole medium (not necessarily at the implant location) needs to be found for safety constraints.

Our design procedure in [39] can only maximize ultrasonic link PTE for a fixed R_L . A revised design procedure is presented in Fig. 6 that maximizes ultrasonic link PTE for matched-load condition by optimizing the U_1 and U_2 geometries, f_p , and R_L . While a summary of the design procedure is presented here, we would like to refer the reader to our previous work in [39] for more details as well as 3D surfaces of optimized parameters. The optimization flowchart starts with the design constraints imposed by the application, including the maximum implant size, i.e., outer diameter (D_{ou2}) and overall thickness of U_2 ($t_{2ov} = t_2 + t_{m2}$, where t_{m2} is the matching-layer thickness), as well as the nominal value for d_{12} in step 1.

In step 2, initial values for f_p , D_{ou1} , the thicknesses of U_1 piezoelectric transducer and its matching layer (t_1 and t_{m1}), t_{m2} , and R_L (several k Ω) are chosen. The detailed guidelines for selecting initial values of f_p , D_{ou1} , t_1 , t_{m1} , t_2 , and t_{m2} can be found in our previous work [39]. Rx geometry is optimized in step 3 to maximize PTE by sweeping both t_2 and D_{ou2} , resulting in a 3D surface as shown in [39], followed by sweeping R_L in an iterative process until R_L changes less than 1%. This leads to the optimal U_2 geometry with optimal R_L ($R_{L,PTE}$) for the given U_1 geometry.

In step 4, U_1 geometry is optimized by sweeping both t_1 and D_{ou1} to maximize PTE, using the values for t_2 , D_{ou2} , and R_L from step 3. Steps 3 and 4 are repeated iteratively until t_1 and D_{ou1} change less than 1%. This leads to optimal U_1/U_2 geometries and $R_{L,PTE}$ for the selected f_p . In step 5, f_p is changed slightly and steps 1-4 are repeated for the new f_p in an iterative process to find the optimal f_p that leads to the maximum PTE. The procedure in Fig. 6 should be repeated for several f_p s until the new PTE values for higher f_p s are at least two times smaller than the peak PTE. Step 5.1 determines the optimal U_1/U_2 geometries, R_L , and f_p that achieve the highest PTE which are further validated and fine-tuned through measurements.

C. Design Examples

In order to compare the performance of inductive and ultrasonic WPT links for different conditions including implant size and depth, different sets of inductive and ultrasonic links were optimized for PTE considering the same design constraints: 1) implant size was limited to 1.1 mm³ and 20 mm³ by limiting its thickness to 1 mm and outer diameter to 1.2 mm and 5 mm, respectively. This covers small and relatively large implant sizes. 2) In order to compare these links at different implant depth, d was set to 5 mm, 10 mm, 30 mm, and 50 mm. 3) In optimizations, optimal an load condition was considered for all links. In our optimization setups as shown in Fig. 4, U_2 was mounted on and L_2 was wound around a silicon substrate with 0.3 mm thickness, mimicking the implant core circuitry. Finally, coils and ultrasonic transducers were made of single conductor copper wire and lead zirconate titanate (PZT), respectively.

III. Inductive and Ultrasonic WPT Links Performance Comparison

Tables I and II summarize the optimal coils and transducers geometries and f_p for 16 sets of inductive and ultrasonic links for the Rx size of 1.1 mm³ ($D_{o2} = D_{ou2} = 1.2$ mm) and 20 mm³ ($D_{o2} = D_{ou2} = 5$ mm) at four different d within 5-50 mm range, based on our design procedures in Sections II.A and II.B, respectively. For inductive links in Table I, the optimal f_p of the small 1.1 mm³ Rx was increased to 100 MHz at d of 5 mm and 10 mm to increase Q_2 and consequently PTE. At $d = 30$ mm, f_p was reduced to 30 MHz and 20 MHz, because D_{o1} was increased by ~2 times to improve k_{12} at large distances, and larger D_{o1} reduced the self-resonance frequency of L_1 . Therefore, f_p was reduced to increase Q_1 and PTE. For the larger 20 mm³ Rx, optimal f_p of 20 MHz was achieved for all d , because a large Q_2 could be achieved at lower f_p considering large D_{o2} of 5 mm. In all inductive links, an optimal d_s of 2 mm further improved Q_1 and PTE.

All ultrasonic links in Table II were optimized in castor oil medium with the mass density of 956 kg/m³ and sound velocity of 1474 m/s [47]. Schott glass and araldite glue were used as the first and second layers for acoustic matching in both U_1 and U_2 [48]. The optimal f_p for the ultrasonic links were close to each other, within 1.6-3 MHz range for both small and large Rx's. As d was increased to 50 mm, D_{ou1} was also increased to extend the focal zone and improve PTE, particularly for the 1.1 mm³ Rx, in which achieving a narrow focal zone was key [39]. Table II also lists the thicknesses of each two-material matching layer (t_{m1} and t_{m2}), which are added to t_1 and t_2 . Since U_2 overall thickness should be limited to 1 mm and

U_2 is mounted on a silicon die with a thickness of 0.3 mm, $t_2 + t_{m2}$ was limited to 0.7 mm in our design procedure.

Figs. 7a and 7b compare the simulated PTEs and maximum PDLs within safety limits (P_L) vs. Rx implant depth (d) for optimal inductive and ultrasonic links in Tables I and II for the Rx sizes of 20 mm³ and 1.1 mm³, respectively. For the 20 mm³ Rx, as shown in Fig. 7a the inductive link PTE and P_L were significantly higher at shorter d of 5 mm and 10 mm. At $d = 10$ mm, the inductive link achieved 1.7 times more PTE (47.8% vs. 28%) and 17.8 times more P_L (131.8 mW vs. 7.4 mW) compared to the ultrasonic link. At $d = 30$ mm, although the ultrasonic link achieved higher PTE (8.5% vs. 4.7%), the inductive link P_L was still significantly higher (5 mW vs. 2.2 mW). At larger $d = 50$ mm, the ultrasonic link achieved higher PTE and P_L of 2.5% and 2 mW compared to the inductive link PTE and P_L of 0.45% and 0.55 mW, respectively.

For the smaller 1.1 mm³ Rx, as shown in Fig. 7b the ultrasonic link outperformed the inductive link with significantly higher PTEs at all distances. The PTE for the ultrasonic link was 1.4 times higher than the inductive link (19.4% vs. 13.5%) at small $d = 5$ mm, and 109 times higher (0.7% vs. $6.4 \times 10^{-3}\%$) at larger d of 50 mm. Although the inductive link P_L at short $d = 5$ mm was higher (4.4 mW vs. 1.5 mW), the ultrasonic link achieved much higher P_L at $d = 10$ mm, particularly at large d of 50 mm, in which the P_L for the ultrasonic link was 178.6 times higher than the inductive link (1 mW vs. 5.6×10^{-3} mW). Several lessons can be learned from Fig. 7. 1) For large Rx sizes, the inductive link P_L is much larger than that of the ultrasonic link because magnetic fields at MHz frequencies are quite safe [35]. 2) At large d , particularly when d is significantly larger than D_{o2} or D_{ou2} , the ultrasonic link achieves higher PTE and P_L . 3) For small mm-sized Rx's, the ultrasonic link can achieve much higher PTE and P_L , particularly at larger distances. 4) The ultrasonic link P_L within the safety constraint is not significantly higher than that of the inductive link. As shown in Fig. 7, there are conditions for which the ultrasonic link achieves higher or comparable PTE but less P_L . Indeed, the main reason for 178.6 times higher P_L in the ultrasonic link for 1.1 mm³ Rx at $d = 50$ mm is the 109 times improvement in PTE.

It should be noted that the ultrasonic link P_L for 1.1 mm³ Rx in Fig. 7b peaked at d of 30 mm because at larger distances D_{ou1} was increased by the optimization procedure to provide a more distant focal zone, resulting in less acoustic intensity for the same input power. As shown in [39], D_{ou1} mainly determines the beam diameter. In addition, the overlapping sound waves generated by the reflections from Rx are more dominant at shorter distances. The ultrasonic link P_L decreased at $d > 30$ mm due to a PTE drop at larger distances.

Fig. 8 shows the optimal loadings ($R_{L,PTE}$) that were found in the optimization of inductive and ultrasonic links for the 1.1 mm³ and 20 mm³ Rx's at different d . For the large 20 mm³ Rx, the inductive link $R_{L,PTE}$ is within k Ω -range (PDL of several mW), while the ultrasonic links require much smaller $R_{L,PTE}$ of hundreds of ohms (PDL of tens of mW). Therefore, for small mW-range, the PDL ultrasonic links suffer from mismatch to the Rx, further decreasing their PTE. If more PDL is needed, large $R_{L,PTE}$ in inductive links can easily be realized by adding one capacitor or employing multi-coil links, since the Rx size is quite large [34], [43]. Therefore, inductive links are superior for large Rx's. For a small 1.1 mm³

Rx, the inductive link $R_{L,PTE}$ is smaller, particularly at large distances. Since less power (mW-range and below) can practically be delivered to mm-sized implants due to safety issues, ultrasonic links are superior, particularly for deeply-implanted mm-sized devices.

Figs. 9a and 9b show the simulated values of PTE and P_L of optimized inductive and ultrasonic links for a 1.1 mm^3 Rx at $d = 10 \text{ mm}$ vs. Rx lateral misalignment in the Y direction, as shown in Fig. 2, and angular rotation (θ), respectively. Although the ultrasonic link achieved higher PTE (10.6% vs. 3.65%) when Tx and Rx were perfectly aligned, it is drastically prone to Rx misalignment and rotation due to its focused nature. As shown in Fig. 9a, the ultrasonic link PTE and P_L reduced by 88 and 95 times from 10.6% to 0.12% and 2 mW to $210 \mu\text{W}$ for only 3 mm of misalignment, respectively. For the same condition, the inductive link PTE and P_L slightly reduced by 1.3 times from 3.65% to 2.9% and 0.88 mW to 0.6 mW, respectively.

A similar trend can be seen in Fig. 9b for Rx rotation around the Y-axis. For a small θ of 30° , the ultrasonic link PTE and P_L reduced by 29 and 32 times to 0.35% and $62 \mu\text{W}$, respectively. However, the inductive link PTE and P_L only dropped by ~ 1.3 times to 2.8% and 0.6 mW for $\theta = 30^\circ$, respectively. Fig. 9 clearly shows that inductive coupling is a more robust technique for WPT to implants that involve misalignment and rotations compared to ultrasound because both PTE and P_L of the inductive link are significantly higher for $> 1 \text{ mm}$ of misalignment and $> 20^\circ$ of rotation. Therefore, ultrasonic links need to be designed such that they become less prone to misalignment and rotation.

IV. Measurement Results

The accuracy of our HFSS and COMSOL simulations for inductive and ultrasonic links, respectively, were validated through measurements particularly at large d and small D_{o2}/D_{o1} . To reduce the fabrication complexity, we optimized an ultrasonic link without matching layers at $d = 30 \text{ mm}$ and $D_{o2} = 1.2 \text{ mm}$ and used it in measurements. Our optimization resulted in optimal $D_{o1} = 15.9 \text{ mm}$, $t_1 = 1.95 \text{ mm}$, $t_2 = 0.95 \text{ mm}$, $f_p = 1.1 \text{ MHz}$, and $R_{L,PTE} = 3.8 \text{ k}\Omega$. Figs. 10a and 10b show the experimental setups for measuring the PTE for the optimized inductive and ultrasonic links, respectively, for a 1.1 mm^3 Rx at $d = 30 \text{ mm}$. We first measured S-parameters using a network analyzer and then calculated PTEs using the equations in our previous works [34], [39].

Inside the inductive link measurement setup in Fig. 10a, the L_1 and L_2 coils with the specifications in Table I ($D_{o2} = 1.2 \text{ mm}$, $d = 30 \text{ mm}$) were connected to a pair of SMA connectors, and held in parallel and perfectly aligned using a Plexiglas/plastic frame. In order to mimic the lossy tissue environment, L_2 was wrapped in a 3-cm thick layer of chicken breast. Similar to Fig. 2, L_1 was spaced 2 mm from the tissue. Our proposed de-embedding method in [34] was used to improve the accuracy of measured Q_1 , Q_2 , and k_{12} from the S-parameters.

Inside the ultrasonic link measurement setup in Fig. 10b, U_1 and U_2 (without matching layers) were connected to a pair of SMA connectors and held perfectly aligned inside an oil tank, mimicking soft tissue. U_2 was mounted on a small printed circuit board (PCB) and

held above the bottom of the tank using a 3D-printed stand which was fixed at the center of the oil tank. U_1 was glued on a 3D-printed holder which could move in the Y and Z directions as shown in Fig. 10b.

Fig. 11a shows the simulated and measured PTEs of the optimized inductive links vs. f_p for the Rx's size of 1.1 mm^3 ($D_{o2} = 1.2 \text{ mm}$) at $d = 30 \text{ mm}$ and $d_s = 2 \text{ mm}$, i.e., $d_{12} = 32 \text{ mm}$. The measured PTE at the optimal f_p of 30 MHz was 0.053%, which was slightly smaller than the simulated PTE of 0.078%. This discrepancy could be due to the geometry variations of the hand-fabricated L_2 , and any possible L_2 misalignments in measurements. Nonetheless, the measured PTEs were close to the simulated ones, particularly at smaller $f_p < 20 \text{ MHz}$.

Fig. 11b shows the simulated and measured PTEs of the same inductive link vs. Rx lateral misalignment in the Y direction (Fig. 2). The simulated and measured PTEs at the Rx lateral misalignment of 10 mm were 0.047% and 0.024%, respectively. The measured results in Fig. 11b further confirms that the inductive link is quite robust to misalignment.

Figs. 12a and 12b show the simulated and measured PTEs for the optimized ultrasonic link for a 1.1 mm^3 Rx with a matched load of $3.8 \text{ k}\Omega$ vs. (a) f_p at $d = 30 \text{ mm}$, and (b) d at the optimal f_p of 1.1 MHz. Close simulated and measured PTEs of 0.69% and 0.62% were achieved at $d = 30 \text{ mm}$ and $f_p = 1.1 \text{ MHz}$, respectively, validating the accuracy of our simulations.

V. Discussion

Our measurement setups in Fig. 10 use a homogenous medium for WPT. To verify the effects of the multilayer propagation medium, which is more practical, on the PTE of the inductive and ultrasonic WPT, we provide a real-word example of PNS interfacing in the following. Our current active research is on recording gastric slow-wave signals by implanting an array of mm-sized implants inside the sub-mucosa space of the stomach to study the electrophysiology behind stomach dismotility [15]. Fig. 13 shows the abdominal multilayer structure of a human for such an application, in which a 1.1 mm^3 inductive/ultrasonic Rx, optimized for $d = 50 \text{ mm}$ as listed in Tables I and II, is located at $d = 50 \text{ mm}$. Table III shows the key electrical and acoustic parameters of these layers [49]-[53].

The inductive and ultrasonic links were simulated in both a homogeneous medium (muscle for inductive link and soft tissue for ultrasonic link) as well as the multilayer structure in Fig. 13. Figs. 14a and 14b compare the simulated PTEs of inductive and ultrasonic links, respectively, in single-layer and multilayer media. As shown in Fig. 14a, the inductive link PTE was slightly increased in the multilayer medium because the conductivity of the skin layer is less than that of muscle (as shown in Table III) and, therefore, the quality factor of the Tx coil, which faces the skin and muscle in multilayer and single-layer media, respectively, was increased in the multilayer medium. Fig. 14b also shows that the ultrasonic PTE changes were negligible, because the calculated reflections at the interface of layers in the multilayer media were as small as 3.28% at the skin-fat interface, 0.96% at the fat-muscle interface, and 0.76% at the muscle-soft tissue interface.

Although ultrasonic links achieve high PTE and P_L , particularly for WPT to deeply-implanted mm-sized devices, they are drastically prone to any Rx misalignment due to their focused nature, as shown in Fig. 9a. This may cause ultra-low PTE and even malfunction in powering implants that involve dynamic movements, notably PNS interfaces for recording from or stimulating actively moving organs, such as the stomach and heart [15], [54]. Using ultrasonic beamforming to steer the ultrasound beam towards the implant can partially mitigate the misalignment issue [55]. However, this requires knowledge about the implant's location at any moment as well as complex and power-consuming external hardware. In addition, beam-forming can be a suitable choice when a network of implants is wirelessly powered. Therefore, we believe that the ultrasonic WPT link for a single implant needs to be optimized for the worst-case scenario of the misalignment.

As shown in [39], the ultrasound beam is narrowest in the focal zone, which is an optimal location for the Rx when it is perfectly aligned. However, by placing the Rx inside the deep far-field region, PTE can be improved for a wider range of misalignments at the cost of PTE reduction in perfectly aligned conditions. This can be done by reducing D_{oul} , which decreases the focal length [39].

Due to the limited aspect ratio of the Tx transducer, t_1 should also be optimized along with D_{oul} to maintain U_1 resonance at f_p . It should be noted that due to the side beams, increasing D_{oul} might also improve PTE for a certain misalignment condition, as shown in Fig. 5. However, this approach places Rx in the near-field region with several maxima and minima in the sound intensity, which can lead to drastic PTE reduction if the Rx moves slightly [39].

Our proposed optimization of ultrasonic links for worst-case misalignment scenarios is as follows. Step 1: the design procedure in Fig. 6 with fully-aligned Rx will be followed first to find optimal U_1 and U_2 geometries as well as f_p . Step 2: the optimal D_{oul} in step 1 is reduced by 0.1 mm and the corresponding t_1 for resonating U_1 at f_p is found by sweeping t_1 around its optimal value from step 1. Step 3: using the new D_{oul} and t_1 values, PTE will be found for all practical misalignments. It is clear that for the new D_{oul} and t_1 , PTE for the fully-aligned condition will be reduced. However, steps 2 and 3 will be repeated in a recursive process until the PTE at the worst-case misalignment keeps decreasing. Finally, among the U_1 geometries that lead to the maximum or close-to-maximum PTEs at the worst-case misalignment, the one with relatively higher PTEs for lower misalignments, including the aligned condition, will be chosen.

For a worst-case misalignment of 3 mm, we re-optimized the ultrasonic link with a 1.1 mm^3 Rx at $d = 10 \text{ mm}$ by changing D_{oul} and t_1 . The specifications of the ultrasonic link, including U_1 and U_2 geometries, for the perfectly-aligned condition, are listed in Table II. Fig. 15a shows the simulated 3-D surface of PTE vs. t_1 and D_{oul} for the perfectly aligned condition. It can be seen that the maximum PTE of 10.6% was achieved at $D_{oul} = 6.4 \text{ mm}$ and $t_1 = 1.15 \text{ mm}$ (similar to the results in Table II), which validates our design procedure in Fig. 6 for fully aligned implants. In Fig. 15a, for each D_{oul} there is an optimal t_1 that corresponds to U_1 resonance at an optimal f_p of 1.8 MHz. As D_{oul} was decreased from 6.4 mm to 2 mm, the peak PTE also decreased from 10.6% to 1.45% due to widening the ultrasound beam.

Fig. 15b shows the simulated 3-D surface of PTE vs. D_{oul} and Rx misalignment in the Y direction, as shown in Fig. 4, using the optimal t_1 in Fig. 15a for each D_{oul} . It can be seen that at 3 mm of misalignment, the highest PTE of 0.36% was achieved with a relatively small D_{oul} of 2.4 mm. At similar misalignment of 3 mm, the links with D_{oul} of 3 mm and 2.2 mm achieved slightly lower PTEs of 0.32% and 0.34%, respectively. We prefer D_{oul} of 3 mm, because 1) at 3 mm of misalignment, its PTE is very close to the maximum achievable PTE, 2) in fully-aligned condition, it achieves a higher PTE of 4% compared with the PTEs of 2.85% and 2.58% for D_{oul} of 2.4 mm and 2.2 mm, respectively, and 3) as shown in Fig. 15b it achieves relatively higher PTE for a wide range of misalignments. As shown in Figs. 15a and 15b, for the misalignment of 0 to 3 mm, the link with the relatively small D_{oul} of 3 mm shows only 12.5 times reduction in PTE from 4% to 0.32%, as opposed to 88 times decrease in PTE for the larger D_{oul} of 6.4 mm from 10.6% to 0.12%. In other words, by reducing D_{oul} from 6.4 mm to 3 mm, PTE was increased by 2.7 times (0.32% vs. 0.12%) for 3 mm of misalignment at the cost of 2.6 times less PTE without misalignment (4% vs. 10.6%). Such smaller variation in PTE, i.e., flatter PTE surface, for a smaller D_{oul} as shown in Fig. 15b can also highly benefit the external power driver and battery by reducing their output power range [56].

Fig. 16 compares the measured PTEs of two ultrasonic links with different U_1 geometries and identical 1 mm³ Rx's for $d = 30$ mm and Rx misalignment of 0–5 mm in the Y direction as shown in Fig. 10b. The “large U_1 ” in Fig. 16 refers to the optimal U_1 geometry without a matching layer ($D_{oul} = 15.9$ mm, $t_1 = 1.95$ mm) for the aligned condition, while the “small U_1 ” refers to a transducer with a reduced D_{oul} of 9.5 mm ($t_1 = 1.95$ mm) to improve robustness against misalignment. It can be seen in Fig. 16 that the PTE for the large U_1 is 3.1 times higher than the PTE for the small U_1 (0.62% vs. 0.2%) when U_2 is perfectly aligned. However, for 5 mm misalignment in the Y direction, the link with the small U_1 outperformed the link with the large U_1 by 2.6 times (0.12% vs. 0.047%) due to the wider ultrasonic beam for the small U_1 at $d = 30$ mm. When U_2 was misaligned by 5 mm, the PTE for the link with large U_1 was significantly reduced by 13 times, while the PTE for the link with small U_1 only reduced by 1.7 times,

This paper mainly focused on the performance (PTE and PDL) comparison between inductive and ultrasonic links. There have also been works that have used GHz/sub-GHz-range RF links for WPT to mm-sized implants [28]–[33]. Table III compares the specifications and performance of our inductive and ultrasonic links with the state-of-the-art ultrasonic and RF links for WPT to deeply implanted mm-sized devices. Since the helix Rx coil in [30] has a D_{o2} of 2 mm and height of 3 mm, i.e., a total volume of 9.4 mm³, for a fair comparison we have also optimized an inductive link for similar Rx volume at $d = 50$ mm. The specifications of this inductive link are as follows: $D_{o1} = 60$ mm, $w_1 = 700$ μ m, $n_1 = 12$, $s_1 = 350$ μ m, $D_{o2} = 2$ mm, $w_2 = 100$ μ m, $n_2 = 23$, $s_2 = 30$ μ m, $d_s = 2$ mm, optimal $f_p = 11$ MHz, and $R_{L,PTE} = 3.6$ k Ω .

As shown in Table IV, the ultrasonic link with the small Rx size of 1 mm³ has still achieved higher PTE of 0.7% at $d_{12} = 50$ mm compared with its inductive and RF counterparts, which have an even larger Rx size of 9.4 mm³, with the PTEs of 0.13% and 0.04%, respectively. Despite smaller Rx size, the ultrasonic link P_L is ~ 1 mW, which is also higher than the RF

link in [30] with $P_L = 0.2$ mW under the SAR limit of 2.5 W/kg averaged over 10 g of tissue. This clearly shows that ultrasonic links are superior for powering deeply implanted mm-sized devices if they are properly optimized for worst-case misalignment scenarios.

Since the PTE and P_L of inductive and ultrasonic links vary significantly with d_{12} and implant size (as shown in Fig. 7), a designer can utilize the FoM formulas, which we have proposed in [34] and [57], to strike a balance between these two key parameters. Our design procedures in Section II can easily be extended to optimization of FoM by replacing the PTE equation with the FoM equation in each step.

The changes in the implant's position over time highly depend on its size and application. For instance, cochlear implants utilize a pair of magnets to fully align the Tx and Rx coils and fix the implant's position. For cm-sized implants, this can be a practical solution. For mm-sized implants, particularly in the PNS with active movements, the implant is supposed to be light enough to freely float inside the tissue, reducing tissue damage [27]. Therefore, the implant will move, which is the main reason we proposed a revised design procedure in this section that compensates for the implant's movements. For our design example of a gastric implant similar to [58], the implant will be endoscopically placed into the submucosa space of the stomach by making small submucosal pockets.

The above-mentioned IMD applications require the delivery of tens of mW of power or below, which requires only several Watts in the worst-case scenarios (with low PTE). The power rating of electronic and acoustic components is well above this range. Thus the inductive and ultrasonic WPT link reliability against the exposure of electronic and acoustic components to high power levels is not an issue here. Regarding the reliability of the WPT links against distance, misalignment, orientation, and loading variations, we have extensively discussed them in both Sections III and V.

VI. Conclusion

The performances of different scenarios for inductive and ultrasonic WPT to biomedical implants for different Rx sizes and implantation depths were compared to clarify which modality is more suitable for any given biomedical application. Both PTE and maximum PDL under safety constraints were compared, suggesting that 1) for short distances and large Rx size, inductive coupling is the suitable choice, 2) for mm-sized devices, particularly those deeply implanted (several cm) inside the body, ultrasonic links achieve higher performance, 3) optimal load condition is very different for inductive and ultrasonic links and plays an important role in their performances, 4) ultrasonic links cannot necessarily deliver much higher power due to high sound intensity close to the Tx transducer, and 5) if the application involves implant movements, ultrasonic links are drastically prone to misalignments and should be designed for the worst-case misalignment scenarios. These conclusions were made based on our study on several sets of optimized inductive and ultrasonic links with the Rx volumes of 1.1 mm³ and 20 mm³ at different powering distances of 5-50 mm. Our simulation results have been validated by measurements.

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Biographies

Ahmed Ibrahim (S'18) received the B.Sc. and M.Sc. degrees in electrical engineering from Ain-Shams University, Cairo, Egypt, in 2008 and 2014, respectively. During his Master's studies, he designed inverter-based low-voltage continuous-time sigma delta ADCs. From 2008 to 2011, he was working as an RF Design Engineer at Sysdsoft, Cairo, Egypt, where he designed ADCs. From 2011 to 2014, he worked as an Analog Design Engineer at Si-Ware Systems, Cairo, Egypt, where he designed sensor interfaces circuits for consumer MEMS based accelerometers, gyroscopes, and spectrometer including clocking and phase-locked loop circuits. In August 2014, he joined the Penn State University Integrated Circuits and Systems Lab (ICSL), where he is working toward the Ph.D. degree. His research interests are integrated circuit design for bio and wireless applications and wireless power transfer. He serves as a reviewer for the IEEE SENSORS JOURNAL, IEEE TRANSACTIONS ON BIOMEDICAL CIRCUITS AND SYSTEMS, IEEE TRANSACTIONS ON CIRCUITS AND SYSTEMS I & II.

Miao Meng received his B.S. in Electrical and Computer Engineering from University of Wisconsin Madison, Wisconsin, in May 2011, and M.S. in Electrical Engineering from Columbia University, New York, NY in December 2012. He interned as an IC Designer and worked on level shifter optimization for I/O interface at IBM, China. In Aug. 2014, he joined the integrated circuits and systems lab (ICSL) as a PhD student. His research interest is integrated circuits for bio applications and ultrasound-based wireless power transfer and data communication to/with miniature implantable devices.

Mehdi Kiani (S'09-M'14) received the B.S. degree from Shiraz University, Shiraz, Iran, and the M.S. degree from Sharif University of Technology, Tehran, Iran, in 2005 and 2008, respectively. He received his M.S. and Ph.D. degrees in Electrical and Computer Engineering from the Georgia Institute of Technology in 2012 and 2013, respectively. He is currently an assistant professor in the department of Electrical Engineering at the Pennsylvania State University. His research interest is integrated circuits and systems design for bio application. He was the recipient of the Georgia Tech Sigma Xi Best PhD Thesis Award, and Georgia Tech Chih Foundation Research Award for excellent research in the fields of engineering and health sciences. He is an Associate Editor of the IEEE Transactions on Biomedical Circuits and Systems and has served on the subcommittees of the IEEE Sensors and IEEE Custom Integrated Circuits Conference.

References

1. Finkenzeller K. RFID-Handbook. 2nd. Hoboken, NJ: Wiley; 2003.
2. Kim C, Seo D, You J, Park J, Cho B. Design of a contactless battery charger for cellular phone. IEEE Trans Ind Electron. Dec; 2001 48(6):1238–1247.
3. Hatanaka K, Sato F, Matsuki H, Kikuchi S, Murakami J, Kawase M, Satoh T. Power transmission of a desk with a cord-free power supply. IEEE Trans Magn. Sep; 2002 38(5):3329–3331.
4. Lee B, Kiani M, Ghovanloo M. A smart wirelessly powered homecare for long-term high-throughput behavioral experiments. IEEE Sensors Journal. Sep; 2015 15(9):4905–4916. [PubMed: 26257586]
5. Lee S, Lee B, Kiani M, Mahmoudi B, Gross R, Ghovanloo M. An inductively-powered wireless neural recording system with a charge sampling analog front-end. IEEE Sensors Journal. Jan 15; 2016 16(2):475–484. [PubMed: 27069422]
6. Kilinc EG, Conus G, Weber C, Kawkabani B, Maloberti F, Dehollain C. A system for wireless power transfer of micro-systems in-vivo implantable in freely moving animals. IEEE Sensors Journal. Feb; 2014 14(2):522–531.
7. Yilmaz G, Atasoy O, Dehollain C. Wireless energy and data transfer for in-vivo epileptic focus localization. IEEE Sensors Journal. Nov; 2013 13(11):4172–4179.
8. Badr B, Somogyi-Gsizmazia R, Delaney K, Dechev N. Wireless power transfer for telemetric devices with variable orientation, for small rodent behavior monitoring. IEEE Sensors Journal. Apr; 2015 15(4):2144–2156.
9. Zhou D, Greenbaum E. Implantable neural prostheses 1. New York, NY: Springer; 2009.
10. Chen K, Yang Z, Hoang L, Weiland J, Humayun M, Liu W. An integrated 256-channel epiretinal prosthesis. IEEE J Solid State Cir. Sep.2010 45:1946–1956.
11. Yin M, Borton D, Aceros J, Patterson W, Nurmikko A. A 100-channel hermetically sealed implantable device for chronic wireless neurosensing applications. IEEE Trans Biomed Cir Syst. Apr.2013 7:115–128.
12. Nicolelis M. Actions from thoughts. Nature. 2001
13. Lee H, Kwon K, Li W, Ghovanloo M. A power-efficient switched capacitor stimulating system for electrical/optical deep brain stimulation. IEEE Int Solid-State Cir Conf. Feb.2014 :414–416.

14. Deisseroth K. Optogenetics. *Nature*. Jan.2011 8
15. Ibrahim A, Farajidavar A, Kiani M. A 64-channel wireless implantable system-on-chip for gastric electrical-wave recording. *IEEE Sensors Conf*. Oct.2016
16. Zeng F-G, Rebscher S, Harrison W, Sun X, Feng H. Cochlear implants: System design, integration, and evaluation. *IEEE Rev Biomed Eng*. 2008; 1:115–142. [PubMed: 19946565]
17. Weiland J, Humayun M. Visual prosthesis. *Proc IEEE*. Jul; 2008 96(7):1076–1084.
18. Lebedev MA, Nicolelis MA. Brain–machine interfaces: Past, present and future. *Trends Neurosci*. 2006; 29(9):536–546. [PubMed: 16859758]
19. DiMarco J. Implantable cardioverter-defibrillators. *New England J Med*. 2003; 349(19):1836–1847. [PubMed: 14602883]
20. Agarwal K, Jegadeesan R, Guo Y, Thakor N. Wireless power transfer strategies for implantable bioelectronics. *IEEE Rev Biomed Eng*. Dec.2017 10:136–161. [PubMed: 28328511]
21. Muller R, Le H, Li W, Ledochowitsch P, Gambini S, Bjorninen T, Koralek A, Carmena J, Maharbiz M, Alon E, Rabaey J. A minimally invasive 64-channel wireless μ ECOG implant. *IEEE J Solid State Cir*. Jan.2015 50:344–359.
22. Borton DA, Yin M, Aceros J, Nurmikko A. An implantable wireless neural interface for recording cortical circuit dynamics in moving primates. *J Neural Eng*. Apr.2013 10(2)
23. Harrison R. Designing efficient inductive power links for implantable devices. *IEEE Int Symp Cir Syst*. May.2007 :2080–2083.
24. Agarwal K, Jegadeesan R, Guo YX, Thakor NV. Wireless power transfer strategies for implantable bioelectronics. *IEEE Reviews in Biomedical Engineering*. 2017; 10:136–161. [PubMed: 28328511]
25. Baker MW, Sarpeshkar R. Feedback analysis and design of RF power links for low-power bionic systems. *IEEE Trans Biomed Cir Syst*. Mar; 2007 1(1):28–38.
26. Jow UM, Ghovanloo M. Design and optimization of printed spiral coils for efficient transcutaneous inductive power transmission. *IEEE Trans Biomed Cir Syst*. Sep.2007 1:193–202.
27. McConnell G, Rees H, Levey A, Gutekunst C, Gross R, Bellamkonda R. Implanted neural electrodes cause chronic, local inflammation that is correlated with local neurodegeneration. *J Neural Eng*. Oct.2009 6
28. Pivonka D, Yakovlev A, Poon A, Meng T. A mm-sized wirelessly powered and remotely controlled locomotive implant. *IEEE Trans Biomed Cir Syst*. Dec; 2012 6(6):523–532.
29. Kim S, Ho J, Poon A. Wireless power transfer to miniature implants: transmitter optimization. *IEEE Trans Ant Propag*. Oct; 2012 60(10):4838–4845.
30. Ho J, Yeh A, Neofytou E, Kim S, Tanabe Y, Patlolla B, Beygui R, Poon A. Wireless power transfer to deep-tissue microimplants. *Proc Natl Acad Sci*. Jun.2014 111:7974–7979. [PubMed: 24843161]
31. Zargham M, Gulak P. Maximum achievable efficiency in near-field coupled power-transfer systems. *IEEE Trans Biomed Cir Syst*. Jun.2012 6:228–245.
32. Mark M, Bjorninen T, Ukkonen L, Sydanheimo L, Rabaey J. SAR reduction and link optimization for mm-size remotely powered wireless implants using segmented loop antennas. *Biomed Wireless Techn Networks Sensing Syst (BioWireless)*. Jan.2011 :7–10.
33. Ahn D, Ghovanloo M. Optimal design of wireless power transmission links for millimeter-sized biomedical implants. *IEEE Trans Biomed Cir Syst*. Feb; 2016 10(1):125–137.
34. Ibrahim A, Kiani M. A figure-of-merit for design and optimization of inductive power transmission links for millimeter-sized biomedical implants. *IEEE Trans Biomed Cir Syst*. Dec; 2016 10(6): 1100–1111.
35. IEEE standard for the safety levels with respect to human exposure to radiofrequency electromagnetic fields, 3 KHz to 300 GHz. 2006IEEE Standard C95.1
36. Seo D, Carmena J, Rabaey J, Maharbiz M, Alon E. Model validation of untethered, ultrasonic neural dust motes for cortical recording. *J Neurosci Methods*. Apr.2015 244:114–122. [PubMed: 25109901]
37. Charthad J, Weber M, Chang T, Arbabian A. A mm-sized implantable medical device (IMD) with ultrasonic power transfer and a hybrid bi-directional data link. *IEEE J Solid-State Cir*. Aug.2015 50:1–13.

38. Song S, Kim A, Ziaie B. Omni-directional ultrasonic powering for millimeter-scale implantable devices. *IEEE Trans Biomed Eng.* Nov; 2015 62(11):2717–2723. [PubMed: 26080376]
39. Meng M, Kiani M. Design and optimization of ultrasonic wireless power transmission links for millimeter-sized biomedical implants. *IEEE Trans Biomed Cir Syst.* Feb; 2017 11(1):98–107.
40. Denisov A, Yeatman E. Ultrasonic vs. inductive power delivery for miniature biomedical implants. *International Conf Body Sensor Networks.* 2010:84–89.
41. Chou T, Subramanian R, Park J, Mercier P. A miniaturized ultrasonic power delivery system. *IEEE Biomed Cir Syst Conf.* Oct.2014 :440–443. M.
42. Ibrahim A, Meng M, Kiani M. Inductive and ultrasonic wireless power transmission to biomedical implants. *Intern Symp Cir Syst.* May.2017
43. Meng M, Kiani M. A hybrid inductive-ultrasonic link for wireless power transmission to millimeter-sized biomedical implants. *IEEE Trans Cir Syst II.* Oct; 2017 64(10):1137–1141.
44. Kiani M, Jow U, Ghovanloo M. Design and optimization of a 3-coil inductive link for efficient wireless power transmission. *IEEE Trans Biomed Cir Syst.* Dec.2011 5:579–591.
45. Ibrahim A, Kiani M. Inductive power transmission to millimeter-sized biomedical implants using printed spiral coils. *IEEE 38th Eng Med Biol Conf.* Aug.2016
46. Food and Drug Administration (FDA). Information for manufacturers seeking marketing clearance of diagnostic ultrasound systems and transducers. Guidance for Industry and FDA Staff. Sep 9.2008
47. Fyke F, Greenleaf J, Johnson S. Continuous wave measurements of acoustic attenuation in an oil/polymer mixture. *Ultrasound Med Biol.* 1978; 5:87–90.
48. Callens D, Bruneel C, Assaad J. Matching ultrasonic transducer using two matching layers where one of them is glue. *NDT&E International.* Dec.2004 37:591–596.
49. Kim J, Lim H, Lee S, Kim Y. Thickness of rectus abdominis muscle and abdominal subcutaneous fat tissue in adult women: correlation with age, pregnancy, laparotomy, and body mass index. *Archives of Plastic Surgery.* 2012; 39(5):528. [PubMed: 23094250]
50. Akkus O. Evaluation of skin and subcutaneous adipose tissue thickness for optimal insulin injection. *Journal of Diabetes & Metabolism.* 2012; 03(8)
51. Edelman S. *ESP Ultrasound.* 4. Jul, 2012 Understanding ultrasound physics.
52. Azhari H. *Basics of biomedical ultrasound for engineers.* John Wiley & Sons; Apr, 2010
53. Gabriel S, Lau R, Gabriel C. The dielectric properties of biological tissues: III. Parametric models for the dielectric spectrum of tissues. *Phys Med Biol.* Nov; 1996 41(11):2271–2293. [PubMed: 8938026]
54. Timko B, Cohen-Karni T, Yu G, Qing Q, Tian B, Lieber C. Electrical recording from hearts with flexible nanowire device arrays. *Nano letters.* Jan.2009 9:914–918. [PubMed: 19170614]
55. Seo D, Tang H, Carmena J, Rabaey J, Alon E, Boser B, Maharbiz M. Ultrasonic beamforming system for interrogating multiple implantable sensors. *IEEE 37th Eng Med Biol Conf.* Aug.2015
56. Kiani M, Ghovanloo M. An RFID-based closed loop wireless power transmission system for biomedical applications. *IEEE Trans Cir Syst-II.* Apr; 2010 57(4):260–264.
57. Kiani M, Ghovanloo M. A figure-of-merit for designing high performance inductive power transmission links. *IEEE Trans Indus Elect.* Nov.2013 60:5292–5305.
58. Paskaranandavadivel N, Wang R, Sathar S, O'Grady G, Cheng L, Farajidavar A. A wireless multichannel system for in-vivo monitoring of gastric activity propagation. *Neurogastroenterology & Motility.* Apr.2015 27:580–585. [PubMed: 25599978]

Fig. 1.

Generic WPT link across the skin for powering biomedical implants using a pair of primary and secondary elements. In the inductive and ultrasonic links, these elements are LC-tanks (series L_1C_1 and parallel L_2C_2) and ultrasonic transducers (U_1 - U_2), respectively. For simplicity, the power management and implant core have been modeled as an AC load, R_L . Depending on the application, the propagation medium can include one or more layers of air, skin, bone, soft tissue, fat, muscle, etc. It should be noted that the ultrasonic link is not suitable for air and bone media.

Fig. 2.

(a) The 2-coil inductive link model in HFSS to find the optimal geometries of L_1 and L_2 , i.e., $D_{o1,2}$, $w_{1,2}$, $n_{1,2}$, $s_{1,2}$, as well as the optimal f_p based on our design procedure in [34]. L_2 is located inside muscle tissue. The total powering distance (d_{12}) is $d + d_s$.

Fig. 3. Iterative optimization flowchart for designing efficient inductive WPT links. The optimal values for L_1/L_2 geometries, f_p , and R_L that maximize PTE can be found.

Fig. 4.

The ultrasonic WPT link model in COMSOL to find the optimal geometries for U_1 and U_2 , i.e., $D_{ou1,2}$, $t_{1,2}$, as well as the optimal f_p . Matching layers are added on both U_1 and U_2 . The soft tissue is mimicked by the castor oil with similar acoustic attenuation of 0.8 dB/cm/MHz. A perfect matching layer (PML) has been considered at the boundaries of the medium to avoid acoustic reflections.

Fig. 5. Simulated sound-wave intensity of the 1.1 mm³ ultrasonic link optimized at $d = 10$ mm, where the maximum spatial-peak temporal time-averaged intensity (I_{SPTA}) of 720 mW/cm² was achieved at $d = 1$ mm.

Fig. 6. Iterative optimization flowchart for designing efficient ultrasonic WPT links in COMSOL. The optimal values for U_1/U_2 geometries, f_p , and R_L that maximize PTE can be found.

Fig. 7.

Comparison between simulated values for PTE and maximum PDL within safety limits (P_L) vs. d for the optimal inductive and ultrasonic links with the Rx sizes of (a) 20 mm^3 and (b) 1.1 mm^3 . The links specifications have been listed in Tables I and II.

Fig. 8.

The optimal loading ($R_{L,PTE}$) vs. d for maximizing PTE in optimized inductive and ultrasonic links listed in Tables I and II.

Fig. 9.

Simulated PTE and P_L of optimized inductive and ultrasonic links for 1.1 mm^3 Rx at $d = 10 \text{ mm}$ vs. (a) Rx lateral misalignment in Y direction as shown in Figs. 2 and 4, and (b) Rx angular rotation (θ).

Fig. 10. Measurement setup for measuring S-parameters of (a) inductive link and (b) ultrasonic link using a network analyzer to calculate PTE for 1.1 mm^3 Rx at $d = 30 \text{ mm}$.

Fig. 11.

Comparison between simulated and measured PTEs of optimized inductive link for 1.1 mm³ Rx and $d = 30$ mm vs. (a) f_p and (b) lateral misalignment in the Y direction.

Fig. 12. Comparison between simulated and measured PTEs of optimized ultrasonic link (without matching layers) for 1.1 mm³ Rx vs. (a) f_p at $d = 30$ mm, and (b) d at the optimal f_p of 1.1 MHz.

Fig. 13.

Human abdominal multilayer structure for a mm-sized inductive/ultrasonic gastric implant, located at $d = 50$ mm. The key parameters of these layers for simulations have been listed in Table III.

Fig. 14.

Comparison between simulated PTEs of optimized WPT links for 1.1 mm^3 Rx and $d = 50 \text{ mm}$ in single-layer medium and multilayer medium in Fig. 12 for (a) inductive link and (b) ultrasonic link, showing close PTE results.

Fig. 15. Simulated 3D PTE surfaces of the ultrasonic link for 1.1 mm^3 Rx at $d = 10 \text{ mm}$ vs. (a) different t_1 and D_{oul} values in perfectly aligned condition, and (b) Rx lateral misalignment in Y direction (as shown in Fig. 2) and D_{oul} , demonstrating that reducing D_{oul} helps to improve PTE for misaligned links.

Fig. 16.

Measured values of PTE at $d = 30$ mm vs. lateral misalignment in Y direction for two different U_1 transducers with D_{oul} of 15.9 mm and 9.5 mm.

TABLE I

Optimized Coil Geometries and Operation Frequencies for 8 Design Examples

	Parameters	Symbols	1.1-mm ³ Rx				20-mm ³ Rx			
L_1	Inductance (nH)	L_I	359	409	3816	4426	912	2396	3152	6285
	Outer Diameter (mm)	D_{oI}	18	20	44	43	32	38	46	50
	Number of Turns	n_I	5	5	13	12	7	9	8	12
	Wire Width (μm)	w_I	644							
	Wire Spacing (μm)	s_I	644							
L_2	Quality Factor	Q_I	236		136	214	282	248	186	147
	Inductance (nH)	L_2	42	42	147	210		298		
	Outer Diameter (mm)	D_{o2}	1.2				5			
	Number of Turns	n_2	7				6			
	Wire Width (μm)	w_2	100				100			
	Wire Spacing (μm)	s_2	30				30			
	Quality Factor	Q_2	55		36	30.7	59.3			
	Optimal Operation Frequency (MHz)	f_p	100	100	30	20	20			
	Implant Depth Inside Tissue (mm)	d	5	10	30	50	5	10	30	50
	Optimal L_I and Tissue Spacing (mm)	d_s	2	2	2	2	2			
Optimal Load (Ω)		$R_{L,PTE}$	1550	1550	295	167	2168			
PTE (%)		η	13.5	3.65	0.078	0.0064	67.4	47.8	4.71	0.45
Max PDL Within Safety Limits (mW)		P_L	4.4	0.88	0.068	0.0056	346.4	131.8	5	0.55

Grayed cells indicate the design constraints.

TABLE II

Optimized Ultrasonic Transducer Geometries and Operation Frequencies for 8 Design Examples

	Parameters	Symbols	1.1-mm ³ Rx				20-mm ³ Rx			
			5.8	6.4	10.8	13.3	5.7	5.2	7	8.9
U_1	Diameter (mm)	D_{out1}	5.8	6.4	10.8	13.3	5.7	5.2	7	8.9
	Thickness (mm)	t_1	1.05	1.15	1.05	1.13	0.6	0.64	0.63	0.65
	Backing	–	Air							
U_2	1 st + 2 nd Matching Layers Thickness (mm)	t_{m1}	0.25 + 0.08	0.26 + 0.08	0.30 + 0.09	0.16 + 0.05	0.16 + 0.05	0.16 + 0.05	0.17 + 0.05	0.17 + 0.05
	Diameter (mm)	D_{out2}	1.1	1.2	1.2	1.2	4.7	4.9	4.8	5
	Thickness (mm)	t_2	0.32	0.31	0.3	0.3	0.45	0.47	0.48	0.47
	Backing	–	Silicon							
	1 st + 2 nd Matching Layers Thickness (mm)	t_{m2}	0.25 + 0.08	0.26 + 0.08	0.30 + 0.09	0.16 + 0.05	0.16 + 0.05	0.16 + 0.05	0.17 + 0.05	0.17 + 0.05
Optimal Operation Frequency (MHz)			1.9	1.8	1.8	1.6	3	2.9	2.9	2.8
Implant Depth Inside Tissue (mm)			5	10	30	50	5	10	30	50
Optimal Load (Ω)			4700	4500	4800	2000	140	130	140	100
PTE (%)			19.4	10.6	2.3	0.7	40.4	28	8.5	2.5
Max PDL Within Safety Limits (mW)			1.5	2	2.1	1	8.1	7.4	2.2	2

Grayed cells indicate the design constraints.

TABLE III

Key electrical and Acoustic Parameters of the Human Abdominal Layers

Layer	Sound Speed (m/s)	Density (kg/m ³)	Attenuation (dB/cm/MHz)	Acoustic Impedance (MRayls)	Conductivity [S/m]	Relative Permittivity
Skin	1730	1150	1.8	1.99	0.0004	4
Muscle	1575	1065	1	1.68	0.2	4
Fat	1450	950	0.6	1.38	0.035	2.5
Soft Tissue	1474	956	0.8	1.41	0.02	4

TABLE IV
Benchmarking of Recent WPT Links for Deeply-Implanted mm-Sized Devices with Different Modalities

Parameters	This work			2014, [30] RF	2015, [36] Ultrasound	2015, [38] Ultrasound		
	Inductive	*Inductive	Ultrasound					
Tx	43	60	13.3	60	6.35	20	30	
Rx	Outer Diameter (mm)							
	Outer Diameter/Cross-Section Area (mm/mm ²)	2/3.14	1.2/1.1	2/3.14	0.127×0.127	1×5	2×2	2×4
	Volume (mm ³)	1.1	1.1	9.4	0.002	5	8	16
	Powering Distance (mm)	52	52	50	30	200		
	Optimal Operation Frequency (MHz)	20	11	1600	5	2.3	1.15	
	PTE (%)	0.0064	0.13	0.04	0.002	0.4	1.7	2.7
Max PDL Within Safety Limit (mW)		0.0056	1.4	±0.2	–	–		

* Link specs: $w/ = 700 \mu\text{m}$, $n/ = 12$, $s/ = 350 \mu\text{m}$, $w2 = 100 \mu\text{m}$, $n2 = 23$, $s2 = 30 \mu\text{m}$, $RL, PTE = 3.6 \text{ k}\Omega$.

⁺ For peak SAR limit of 2.5 W/kg averaged over 10 g of tissue.