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Ultrasonic vs. Inductive Power Delivery for Miniature Biomedical Implants

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***Abstract*—In this paper we compare two methods of wireless power delivery to implanted microdevices: ultrasonically and via inductive coupling. We build models for both methods and compare them in terms of power transmission efficiency, for different separations and receiver sizes. The simulation results show that at small distances between source and receiver (1 *cm*) the inductive system outperforms the ultrasonic one (efficiency of 81% vs. 39% for a receiver of 10 *mm* diameter). At larger distances (10 *cm*) the efficiencies of both systems reduce significantly, but the ultrasonic system demonstrates much better performance (0.2% vs. 0.013% for a 10 *mm* receiver). As the receiver gets smaller this gap increases drastically (0.02% vs. $0.02{\cdot}10^{-3}$% for a 2 *mm* receiver) while the distance after which the ultrasonic system outperforms the inductive one reduces (from 2.9 *cm* for a 10 *mm* receiver to 1.5 *cm* for a 5 *mm* receiver).**

Keywords-acoustic waves; implantable microdevices; inductive coupling; inductive powering; power transmission efficiency; ultrasonic powering; wireless power delivery

I. INTRODUCTION

Advances in MEMS (microelectromechanical systems) and in low power electronics are creating ever-increasing opportunities in miniature sensing devices. These opportunities include new implantable medical devices for monitoring biological parameters of the human body, as well as devices aimed at improving human body treatment, reducing discomfort and promoting better health and wellbeing.

Biomedical implants have been powered mainly by batteries so far. However, batteries frequently dominate the size and the cost of the device and have to be replaced or recharged occasionally. As the devices get smaller these facts make them less attractive as the primary power source. Therefore alternative techniques of energizing implantable microdevices are needed. Energy harvesting is an attractive alternative to batteries in low-power biomedical implants and has received increasing research interest in recent years [1]. Ambient motion is one of the main sources of energy for harvesting, and a wide range of motion powered energy harvesters have been proposed or demonstrated, particularly at the microscale.

Another alternative to batteries is wireless power delivery, where a receiver, instead of harvesting ambient energy, is energized wirelessly by sending a signal of a particular frequency. One of the most widespread and well established methods of wireless power delivery is via inductively coupled coils. However this tends to have low efficiency at larger distances, and as the size of the receiving device decreases this becomes an ever greater problem. For inductive powering the coil size to separation ratio is the key issue. Its efficiency can be improved up to a point by increasing the operating frequency, but this is limited by higher cost, higher tissue attenuation and increased radiation for a given coil size.

To date power delivery by ultrasound has been used mainly in the fields of non-destructive testing and remote sensing. Several groups reported systems for delivering power through a steel wall [2-5]. Acoustic waves, due to their lower speed, have much smaller wavelengths than radio waves for a given frequency, which means that more directional transmitters and receivers can be achieved at reasonable frequencies. For powering embedded sensors, in the body or in structures, acceptable attenuation levels can also be reached.

Ultrasound in the human body is primarily used for medical imaging. Typical frequencies for this application are in the range 3–6 *MHz*, and with acoustic velocities in human tissues from 1500–2000 *m/s*, this results in wavelengths of 0.3–0.7 *mm*. With transducers of overall dimensions of at least a few millimeters, in either single element or array form, this allows reasonably directional transmission. This makes ultrasonic power delivery an attractive method of energizing implanted microdevices wirelessly and it receives increased attention today.

The idea of using acoustic waves to transmit energy was proposed by Cochran *et al.* as early as 1985 [6]. The system they built is an internal fixation plate that contains a piezoelectric element generating current when excited mechanically by external ultrasound. This current is then delivered to the electrodes at a bone fracture site in order to stimulate healing or prevent nonunion. Using an ultrasonic transducer (with input voltage of 10–20 *V* at a frequency of 2.25 *MHz*) the authors performed external excitation of piezoceramic samples (of $5 \times 5 \times 0.9\,mm$ size) deeply implanted in living soft tissues (near femur site of beagles) [7]. It was shown that the system is able to generate direct (rectified) currents of 20 μA providing power output of approximately 1.5 mW/cm^2.

Suzuki *et al.* reported a combined system for delivering power and data to an implantable medical device [8]. The optimal operating frequency of 1 *MHz* was found for piezo

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oscillators of 30 *mm* diameter, giving a maximum efficiency of 20%. During the experiment the distance between source and receiver varied in the range 7–100 *mm*. Philips *et al.* designed a peripheral nerve stimulator powered by externally applied ultrasound [9]. The piezoceramic test chips of 3.5 *mm* diameter were implanted inside a living tissue (near sciatic trunk of large American Bullfrogs) and excited externally by a 6 *mm* transducer at a frequency of 2.25 *MHz*. With the output power of up to 1.5 mW/cm^2 the ultrasonic system was capable of providing enough energy to the peripheral nerve microstimulator.

Shih *et al.* presented a subcutaneously implantable device which receives energy though externally applied ultrasound [10]. It has a bulk piezoelectric resonator packaged by biocompatible cohesive gel (in cubic, spherical and irregular shapes) and incorporating an acoustic antenna which can receive refracted waves propagating inside the package. During the experiment the spherical package of 7.8 *mm* radius implanted inside the soft tissue (streaky pork) at 0.5–8 *cm* separations was excited externally by an ultrasonic transducer at frequencies in the range of 5–100 *kHz*. The maximum power transmission efficiency of -40 *dB* (0.01%) was obtained at a frequency of 75 *kHz* and 2.5 *cm* separation. Arra *et al.* built and tested an ultrasonic powering system with the potential to be used in implantable applications [11]. Their acoustic link operates in degassed water at a frequency of 840 *kHz* and the transducer diameters are about 25–30 *mm*, giving efficiencies of 21–35 % at transmitter-receiver distances between 5 *mm* and 105 *mm*.

Recently Ozeri *et al.* investigated an ultrasonic transcutaneous energy transfer for wireless power delivery to implanted microdevices [12]. The authors proposed a system consisting of two piezoelectric transducers of 15 *mm* diameter and 3 *mm* thickness with a 1.3 *mm* thick acoustic matching layer (made of graphite). Experimental measurements were carried out for soft tissues (slices of pork of 5–35 *mm* thickness) immersed in a test water tank. Operating at a frequency of 673 *kHz* the system demonstrated the overall power transmission efficiency of 27% (at 5 *mm* separation). The authors analyzed in detail such important design considerations for maximum power transfer as selection of operating frequency, acoustic impedance matching, circuit design for excitation of transducers, and output power conditioning. They also built a finite element model in order to study the pressure intensity profile generated by a transducer and define the preferred receiver location in its different zones (near/far field and the focus).

However, none of the groups build any models to account for the high attenuation in human tissue (especially at large distances) and obtain the overall power transmission efficiency. The reported results for the efficiency are based solely on experimental measurements. Results in [8] and [11] are obtained for pure water (Suzuki *et al.* experimented with a slice of pork as an alternative to the human body, but no quantitative results were given). The attenuation of ultrasound in human tissue is more than two orders of magnitude higher than in water (0.9 $dB/cm{\cdot}MHz$ for soft tissues vs. 0.002 $dB/cm{\cdot}MHz$ for water [13]), making unclear these systems' performance in real biomedical applications.

In this paper we investigate ultrasonic and inductive power delivery for miniature receiving devices. We build models for both systems and perform their analysis in terms of power transmission efficiency. The systems operate at optimal frequencies for separations of 1–10 *cm*, powering receivers of 2–10 *mm* diameter, embedded in human tissue. Conclusions on the applicability of the systems for various separations are drawn. The important trend of miniaturizing implantable microdevices and its effect on the choice of wireless power supply mechanism are studied.

II. ULTRASONIC POWER DELIVERY

An analytical solution for wireless power transmission through acoustic waves was proposed by Hu *et al.* as early as 2003 [2]. However, apart from a few simplified cases, their model is very cumbersome to solve analytically because it is based on the wave equation and the equations of piezoelectricity. Mason suggested a new method of one-dimensional analysis of the problem based on network theory [14]. He proposed an equivalent electrical circuit where the piezoelectric material is separated into one electrical (representing the electrodes where input power is applied) and two acoustic ports (front and back faces of the material).

The Mason's equivalent circuit for the one-dimensional problem is depicted in Fig. 1. At the electrical port the components represent standard electrical elements and the voltage (in V) is related to the current (in A) via $V = Z{\cdot}I$ where Z is the electrical impedance (in Ω). At the acoustic ports the force (in N) is related to the longitudinal velocity on the piezoelectric surface v (in m/s) via $F = Z_a{\cdot}v$ where Z_a is the acoustic impedance (in kg/s). The ideal electromechanical transformer converts electrical voltage V to mechanical force F via $F = V{\cdot}N$, current I to velocity v via $v = I/N$ and electrical impedance Z to acoustic one Z_a via $Z_a = Z{\cdot}N^2$.

Thorough analysis and comparison of Mason's and other models [15] showed that although it has some problems like negative capacitance at the electrical port, it gives exactly the same results as an analytical solution based on the wave equation. Sherrit *et al.* successfully applied it to study power delivery through a steel wall [16]. They built a very flexible model which allows taking into account additional acoustic elements as well as all possible loss mechanisms.

Our model of acoustic power delivery into the human body is based on the system shown in Fig. 2. It is similar to the one of [16] except that it does not have front and tail masses and uses matching layers to reduce acoustic

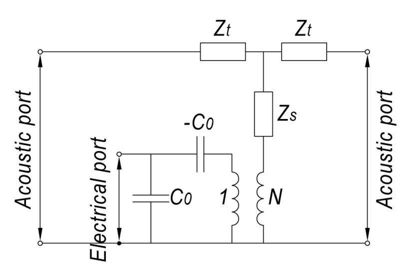

Figure 1. Mason's equivalent circuit for piezoelectric material (after [15])

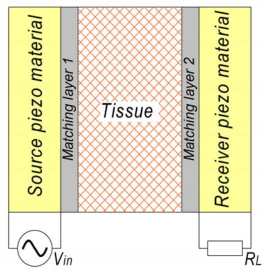

Figure 2. Ultrasonic system for wireless power delivery (after [16])

impedance mismatch between the piezoelectric layers and tissue. A sinusoidal voltage is applied to the source piezoelectric material, which emits an acoustic wave travelling through the tissue towards receiver. When the wave reaches the receiver, its mechanical energy is converted back to electrical form, generating a voltage which is then supplied to the load. This load can be a small sensor, actuator or other component like a power pre-conditioning circuit for a medical implant.

The equivalent circuit for the described system is presented in Fig. 3. All the circuit parameters are thoroughly described in [16]. For our model we optimize the load impedance in terms of frequency by compensating the parallel capacitance C_{02} (which is the internal static capacitance of the piezoelectric material), giving $R_L = 1 / \omega \cdot C_{02}$. No power losses due to possible lateral misalignment of the layers are taken into account, making the model one-dimensional. The overall power transmission efficiency of the system η is determined as the ratio between the power delivered to the load resistor and the input power [16]:

$$\eta = |V_L \cdot I_L / Re(V_{in} \cdot I_{in})|.$$

We use *CTS 3203HD* (formerly *Motorola 3203HD*) piezoelectric material for our simulations. In order to reduce the acoustic impedance mismatch between piezoelectric materials and tissue, quarter wavelength matching layers (silver) are introduced. The losses due to high attenuation of ultrasound in tissue are taken into account by using complex quantities as was proposed by Holland [17].

Material properties and geometric parameters used in our simulation are listed in Table I. We assume that both piezoelectric materials have the shape of a disk, and we sweep the diameter of the receiver in the range 2–10 *mm* during the simulation. Another crucial parameter is the distance between the piezoelectric disks (the thickness of the tissue layer in our model), which is varied in the range 1 to 10 *cm*. We do not limit the operating frequency of a system to a certain value but change it in a wide range (up to 1 *MHz*). During the simulation it was found that for the whole frequency range the geometric parameters of the source piezoelectric have approximately the same effect on the system efficiency, and therefore can be fixed to their optimal values. The same was done with all other parameters from Table I except the thickness of the receiver disk, which is limited by the implant overall size.

III. Inductive Power Delivery

Energy transfer by inductive coupling is another method of delivering wireless power to implanted microelectronic devices. A simplified equivalent circuit for the inductively coupled coil system is shown in Fig. 4. The primary coil L_1 is attached to the skin and driven by sinusoidal current which creates alternating magnetic flux. This flux penetrates the turns of the implanted secondary coil L_2 creating a voltage across it due to electromagnetic induction, which is then provided to the load. The highest voltage gain and therefore efficiency of the inductive link is achieved when both LC-

TABLE I. Material Properties and Geometric Parameters Used for Simulation of Ultrasonic Power Delivery

Material property / geometric parameter	**Piezoelectric (*CTS 3203HD*) [16]**	**Soft tissue [13, 18, 19]**	**Matching layer (silver)**
Density (kg/m^3)	7500	1058	10490
Velocity of sound (*m/s*)	4821 · (1+0.00575i)	1540 · (1+0.019i)	3650
Diameter (*mm*)	source: 20 receiver: 2 – 10	= source piezo	= source/ receiver piezo
Thickness (*mm*)	source: 3 receiver: 3	10 – 100	λ/4

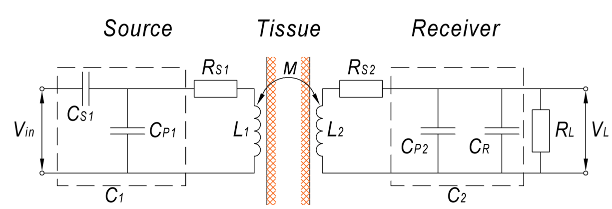

Figure 4. Equivalent circuit for the inductively coupled coil system (after [20])

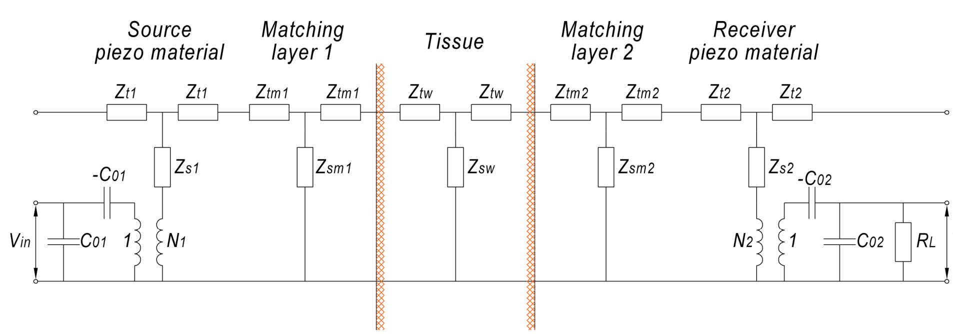

Figure 3. Equivalent circuit for the ultrasonic system shown in Fig. 2

tanks are tuned at the resonant frequency $\omega_0 = 1/\sqrt{L_1C_1} = 1/\sqrt{L_2C_2}$. In order to work always at resonance, the values of added capacitors C_{S1} and C_R have to be adjusted for every geometric configuration of the coils. For each coil the circuit element C_P represents internal parasitic capacitance between the coil turns. Below a frequency of 20 *MHz* the power losses in the tissue can be neglected [21]. Therefore the series resistance of the coil wire R_S dominates the power losses.

Assuming ideal voltage source, optimized load and neglecting the losses in the tissue, the maximum efficiency of the inductive link η is given by [22]:

$$\eta = \frac{k^2Q_1Q_2}{\left(1+\sqrt{1+k^2Q_1Q_2}\right)^2},$$

where k is the coupling coefficient between coils L_1 and L_2, and Q_1 and Q_2 are the unloaded primary and secondary quality factors respectively. Most of these parameters are interrelated. The inductance of coils L and therefore the coupling coefficient k can be increased by raising the number of turns for each coil without changing their outer diameter. However, because of the narrower and longer wires the increased parasitic resistance R_S will result in a smaller quality factor Q. It is therefore required to find an optimal geometric configuration of primary and secondary coils in order to maximize the overall power transmission efficiency η.

One of the most attractive configurations of the coil system for energizing high power biomedical implants is printed spiral coils. Unlike traditional wire-wound technology, which requires sophisticated equipment for batch fabrication and scaling down the dimensions, printed spiral coils can benefit from the use of modern lithography processes. They can be defined in single or multiple layers on a rigid or flexible substrate, therefore offering more flexibility in terms of geometry optimization. The square-shaped printed spiral coil with its geometric parameters is illustrated in Fig. 5, where d_i and d_o are the inner and outer diameters of the coil respectively, w is the wire width and s is the spacing between adjacent turns.

Our model of wireless power delivery by inductive coupling is based on the system shown in Fig. 4. Geometric parameters used for our simulation are listed in Table II. The optimization of these parameters performed in [20] was done for two particular resonant frequencies (1 *MHz* and 5 *MHz*). We follow the same procedure, but for a resonant frequency of 13.56 *MHz* (generic frequency for RFID applications in the megahertz range). Results of the optimization process are described below in the order the optimal values are derived.

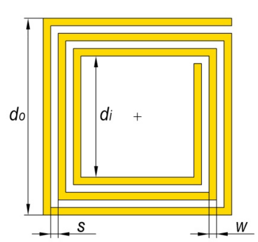

Figure 5. Square-shaped printed spiral coil and its geometric parameters (after [20])

TABLE II. GEOMETRIC PARAMETERS USED FOR SIMULATION OF INDUCTIVE POWER DELIVERY

Geometric parameter [20]	Primary coil (source)	Secondary coil (receiver)
Outer diameter d_o (*mm*)	20	2 – 10
Filling factor φ	0.32	1
Wire width w (μm)	1000	d_{o2} / 10
Wire thickness (μm)	38	38
Spacing between turns s (μm)	150	150
Resonant frequency (*MHz*)	13.56	13.56
Distance between coils (*cm*)	1 – 10	

Analysis of the outer diameter of the primary coil showed that increasing its value always gives better efficiency. Therefore for fair comparison we limit it to the same value as used in acoustic system (20 *mm*, see Table I). The optimal value of filling factor of the primary coil is 0.32. For the secondary coil the optimal value of filling factor is 1. Because the outer diameter of the secondary coil is increasing during the simulation (from 2 to 10 *mm*), its wire width is the most difficult parameter to optimize. We found that the maximum efficiency is obtained when it is 1/10 of the outer coil diameter, giving at least two turns even for the smallest coil (of 2 *mm* diameter). For the largest outer diameter (10 *mm*) we restricted the width to 1000 μm because of increasing parasitic resistance and the reduction of quality factor. The same value was found during the optimization of the wire width of the primary coil. All other geometric parameters are listed in [20]. The model of the inductively coupled system was validated by FEM simulation as well as experimental measurements in [20], where good agreement between calculated, simulated and measured results was demonstrated.

IV. RESULTS

The results of the optimization of ultrasonic and inductive systems show that there are only three major parameters affecting overall power transmission efficiency: operating frequency, diameter of the receiver and the source-receiver distance. This makes comparison of the two systems reasonable and almost independent of other geometric parameters (since for both systems their values are optimized for maximum efficiency). For each configuration of the acoustic system it is required to find the optimal frequency (not necessarily the first resonance) which gives the maximum overall efficiency. This frequency also depends on the distance between source and receiver, therefore during the simulation we sweep it in the range of up to 1 *MHz* recalculating the efficiency every time. The inductive system demonstrates maximum performance when operating at resonant frequency, which we fixed to 13.56 *MHz*.

We performed simulations of ultrasonic and inductive power delivery systems described above and compared their overall efficiency in terms of the distance between source and receiver and the receiver diameter. The simulation results for the efficiency as a function of the receiver diameter are shown in Fig. 6 and Fig. 7. As expected, the overall efficiencies of both systems increase as the receiver gets larger. Fig. 6 clearly indicates that at small distances the inductive system performs significantly better than the

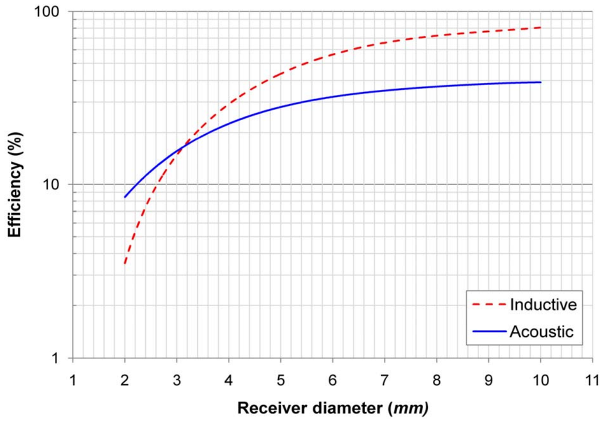

Figure 6. Efficiency as a function of receiver diameter (at 1 *cm* distance)

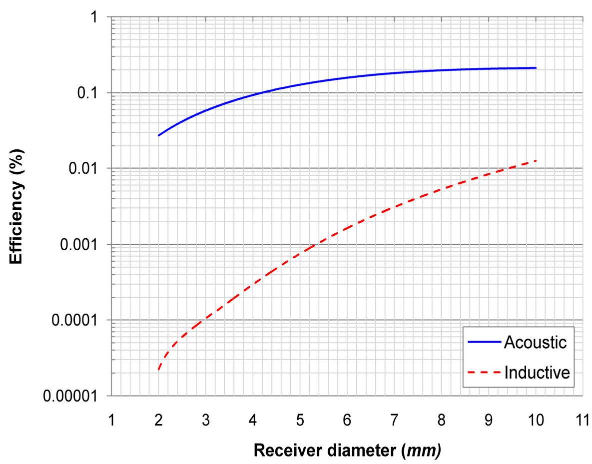

Figure 7. Efficiency as a function of receiver diameter (at 10 *cm* distance)

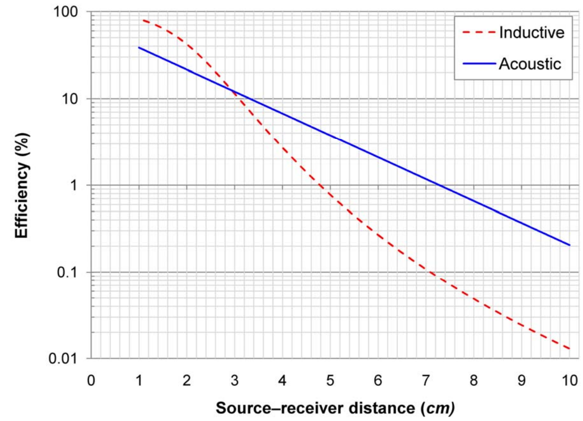

Figure 8. Efficiency as a function of source-receiver distance (for a 10 *mm* receiver)

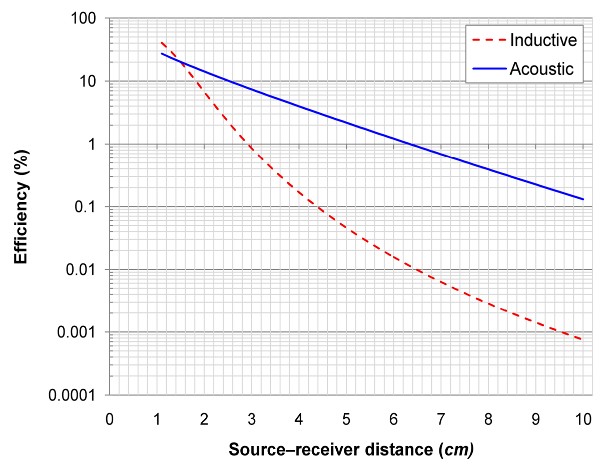

Figure 9. Efficiency as a function of source-receiver distance (for a 5 *mm* receiver)

ultrasonic one for a large receiver size (81% vs. 39% for a 10 *mm* receiver). However, even at such small distances this advantage vanishes as the receiver gets much smaller. For very small receivers with the diameter not exceeding 3 *mm* the ultrasonic system demonstrates better results (8.8% vs. 3.4% for a 2 *mm* receiver). Therefore acoustic power delivery becomes more attractive for energizing biomedical implants as they get smaller.

Fig. 7 illustrates the efficiency of acoustic and inductive power delivery systems when the distance between source and receiver is 10 *cm*. As the distance increases, the acoustic field stabilizes and the variation between different phases of the wave becomes smaller [11]. When ultrasound propagates in a lossless medium this effect results in a larger average of received power and therefore of the overall efficiency. However, due to high attenuation of ultrasound in tissue the efficiency of ultrasonic power delivery reduces significantly at larger distances. Nevertheless it outperforms the inductive system even for larger receivers (0.2% vs. 0.013% for a 10 *mm* receiver). This advantage becomes more evident as the receiver gets smaller (0.02% vs. $0.02 \cdot 10^{-3}$% for a 2 *mm* receiver) making the ultrasonic system very attractive not only for subdermal but also for deeply implanted microdevices.

The overall efficiency as a function of the distance between source and receiver is shown in Fig. 8 and Fig. 9. The efficiency in Fig. 8 is the maximum possible since it is calculated for a receiver of 10 *mm* diameter (the largest in our simulation). The results in Fig. 9 are calculated for a 5 *mm* receiver. As we can see from the graphs, for each size of the receiver there is a certain distance after which the acoustic power delivery provides better efficiency. This distance gets smaller as the receiver size reduces (from 2.9 *cm* for the largest receiver to 1.5 *cm* for a 5 *mm* receiver).

V. Conclusion

We have built detailed models for two wireless power delivery methods: an ultrasonic, and an inductively coupled coil system. Geometric parameters of each system were optimized in order to maximize the power transmission efficiency, leaving only three major parameters (operating frequency, diameter of the receiver and the source-receiver distance) affecting the performance. The optimal operating frequency of the acoustic system was calculated iteratively during the simulation while the inductive system was always operating at the pre-defined resonant frequency of 13.56 *MHz*. The comparison of the two systems was performed for different source-receiver distances (1–10 *cm*) and sizes of the receiving device (of 2–10 *mm* diameter).

All the conclusions drawn in this paper are based only on simulation; the results are not confirmed experimentally. Although our acoustic model accounts for the losses in tissue as well as in piezoelectric materials, it does not take into account spreading losses due to the acoustic beam divergence. The model is one-dimensional since we neglect the misorientation and lateral misalignment of the piezoelectric disks. However some authors already studied the disks mutual orientation and its effect on the overall system efficiency. Arra *et al.* showed that the efficiency can drop by as much as 90% or more when receiver and source disks are not parallel (at misorientation angle of 5°) [11], but this was for more directional transducers ($ka \sim 90$, where k is the acoustic wavenumber and a is the transducer diameter) than considered here ($ka \sim 50$ for a 20 *mm* source and ~ 5 for a 2 *mm* receiver at a frequency of 645 *kHz*). Ozeri *et al.* found that the efficiency reduces significantly when the transducers are laterally misaligned (by as much as 70% at small separations when the lateral non-overlapping is about 30% of the transducers' diameter) [12]. In this case $ka \sim 40$, but the effect will be lower for the small transducers (particularly the receivers) we have analyzed.

In order to simplify our inductive model we assume that the coils operate in air, neglecting their surrounding environment. Taking into account increased parasitic capacitances (because of the high permittivity tissues and fluids surrounding the coils) as well as series resistance (affected by skin effect and eddy currents) Jow *et al.* showed that the overall efficiency will reduce significantly when operating inside the body (more than by a factor of 2) [23]. However this result only emphasizes the points we made from our simulation: the ultrasonic power delivery outperforms the inductive one at larger distances and for smaller implants.

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