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Optimal Design of a Wireless Charging System for an Implanted Hearing Device

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Abstract

Wireless power transfer (WPT) medical systems are progressively recognized as practical solutions for powering advanced micro-electronic devices, particularly in biomedical implants. These systems offer the potential to power devices such as cochlear implants without requiring invasive procedures to replace internal batteries, significantly enhancing patient comfort and device lifespan. Designing WPT systems with high power transfer efficiency presents several challenges, including the compact size of the WPT coils, the airgap distance between the external power source and the implanted device, the operating frequency, and ensuring tissue safety by minimizing power dissipation. This paper explores several strategies to optimize WPT coils and electronic converters for implantable medical devices (IMDs), focusing on designing efficient charging circuits for cochlear implants. Conventional design procedures for both square and circular coils are first considered, where each coil's geometry impacts the magnetic field distribution and, consequently, the efficiency of power transfer. Optimal design concerns are then applied, including the use of a Z-source inverter topology, which significantly improves efficiency. For the square coil, the Z-source inverter improved efficiency from 72.75% to 79.23%, while for the circular coil, efficiency improved from 74.55% to 82.31%. These improvements highlight the effectiveness of the Z-source inverter in enhancing power transfer efficiency by superior matching impedance and minimizing energy losses. The performance of these coil designs (in terms of power transfer efficiency, received power, and operating frequency) is demonstrated, providing insights into the trade-offs between coil geometry and overall WPT system performance. The findings offer valuable guidelines for the development of more effective WPT solutions in biomedical applications, with the potential to improve the functionality and longevity of implantable medical devices.

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Keywords: Cochlear Implants, Wireless Charging Implant Device, Wireless Power Transfer, Optimal Design.

1 Introduction

A further development aimed at overcoming the drawback of answering a specific biomedical problem and conceived a general technique of synthesis of a wireless power transmission system for a Class-E amplifier to minimize the cost, the weight, or the power dissipation associated with the solution (Biten, 2023; Madugalla & Perera, 2024). The method, based on electromagnetic modelling and the recognition of the nonlinear character of the problem, proposed the minimization of an overdetermined cost function (Taghvaie et al., 2023). Before starting with the entire experimental characterization, we focus on a necessary step to accomplish the synthesis of a truly optimized wireless power transmission system, namely, the investigation of an efficient and generalized design process (Salih & Nangir, 2024). In particular, debating a further example of practical interest (Zavrel et al., 2024), the study and implement the numerical model necessary to embrace the theory of optimization towards the synthesis of very-low-power applications, far below the resonance and the carrier frequency for which power losses usually dwarf (Nazarova & Bobomuratov, 2023).

A stringent condition for the approval of any implanted device is the safety and reliability of the power supply. These requirements are of particular importance in long-term implants, where the removal of the device for recharging the battery or replacing the battery is both arduous and painful (Gao et al., 2024). To avoid these issues, stimulation systems benefiting from a wireless power transmission have been proposed, which release the patient from the need to dutifully charge a power supply and provide a longer lifetime for the implant (Ghazi et al., 2021). Recently, the problem of the design of a wireless power transmission system for a biomedical device like an implanted hearing device was posed (Odeh & Taleb, 2023; Mathew & Asha, 2024). The objective was to test the possibility of substituting an external battery for an implanted anti-epileptic vague nerve stimulator. The device was assumed to work in the same power range as the carrier-frequency system approved for external transmission, and the investigation showed that the system took advantage of the significant reduction in the size of the receiving coil and the increase in the transmission quality factor. However, a vehicle for wireless power transfer is not by itself a system. At least two coils are indeed necessary to accomplish the power transfer (Brovont et al., 2023).

2 Background and Significance

In hearing device applications, the WPT system has its challenges due to size-reduction requirements, the implantable hearing device's battery charging requirements like the minimum level of charge, variable charging cycle time, low sensitivity to misalignment, and tissue heating.

Implantable hearing devices are powered by small and long-life batteries (Brochier et al., 2022). However, the need for regular renewal and the possibility of leaking hazardous materials make the current battery technology problematic. Wireless power transfer (WPT) arises for this point and attracts research interest in hearing device applications (Kumar, 2024). The WPT is based on the magnetic field coupling of the transmit and receive coils (Xu et al., 2023). Power is provided via the magnetic field generated in the transmit coil by an electric current (Leema et al., 2024). The receive coil captures a portion of this magnetic field, and the power is transformed into electricity (Sowmya et al., 2023) (Vishnuram et al., 2023). Parents must supervise new born hearing testing for multiple reasons, including but not limited to hearing aid fitting, which is often needed within the first month of life. Also, older children and adults with hearing loss often use hearing devices for communication. Implantable hearing devices may result in improved hearing ability over time and serve a larger user community, including patients with mild to moderate high-frequency sensorineural hearing loss and tinnitus. The work presented herein is focused on design methodologies and considerations relevant to the implementation of a wireless power transfer process for a hearing device, and it is limited by this focus. However, it is also important to note that a generalizable approach is described. A method useful for improvements in the wireless power design for devices besides the hearing device is described (van Nunen, 20230).

3 Wireless Power Transfer Technologies

Two typical wireless power transfer (WPT) technologies are usually used in the hearing device developing physical domain, including electromagnetic coupling technology and ultrasonic power transfer technology. Electromagnetic resonance coupling technology with high transmission efficiency when performing at such a relatively large distance (about thirty or more mm) has been introduced to develop hearing implant devices (Gao et al., 2023). The transmitting antenna must still be close to the head skin or at a relatively far distance to transmit low-frequency electromagnetic energy, typically using a large loop coil to generate the magnetic field. The receiving antenna of the implanted device is located or tuned just right to perform a prominent energy harvesting efficiency (Essa et al., 2024). as illustrated in Figure 1. Compensation capacitors are added to achieve maximized power transfer efficiency. A general electromagnetic force generating a single coil is used to deliver the power from the transmitting antenna (Vishnuram et al., 2023).

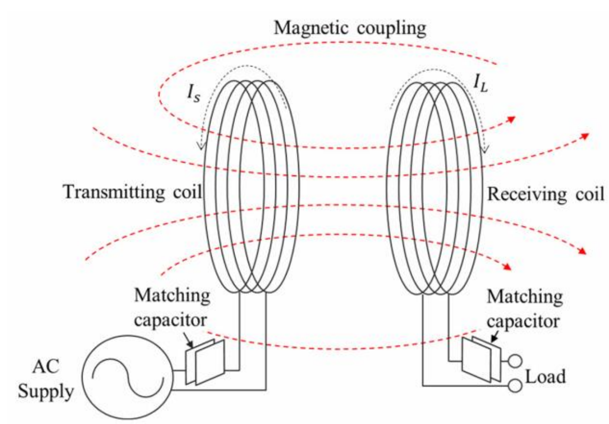


Figure 1: Conceptual Diagram of the Wireless Power Transfer System. Magnetic Coupling Enables Electrical Energy to be Transferred Wirelessly (Kim et al., 2020)

Ultrasonic power transmission has the advantage of being non-electromagnetic, benefiting from providing a clean energy source with no need to consider other electronic devices that could interfere with electromagnetic fields (Razek, 2023). Therefore, ultrasound energy is particularly attractive for rapid and effective energy delivery. It has been widely used in wireless electricity (Sathish Kumar, 2023). There are no basic restrictions that prevent the transmission range from being too close or too far; because of the strong forward directionality, there should be no potential harm to the human body or other animals (Mehri et al., 2016). Also, because the transmission distance is relatively close, a small transducer is enough to perform excellent energy harvesting capabilities for wireless charging receiver with its simple single coil generating structure (Li et al., 2024), which significantly simplifies the design of the receiver, shortens its design cycle, and reduces the difficulty of adjusting its implant position.

4 Implanted Hearing Devices (CI)

There are an estimated 278 million people suffering from severe to profound sensorineural hearing loss, and most of them are in developing countries. The hearing damage usually cannot be recovered, and the patient needs to use hearing enhancement devices (HEDs) to help them listen to the world. HEDs can be divided into external hearing devices and implanted hearing devices (Virzob et al., 2023). as illustrated in Figure 2.

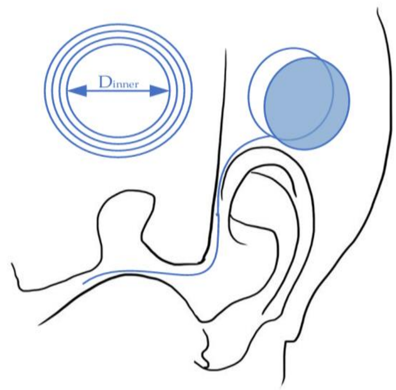


Figure 2: Wireless Power Transfer for a Cochlear Implant System (Zhou et al., 2020)

The external hearing device generally includes the hearing aid, which consists of several types, such as BTE (behind-the-ear), ITE (in-the-ear), ITC (in-the-canal), CIC (completely-in-the-canal), and the body-worn hearing device. The main purpose of the external hearing device is to enlarge the sound pressure level in the ear canal, which leads to high-frequency sound amplification, full-frequency sound amplification, or speech sound amplification. Great technological progress has been achieved for the external hearing device in the last several decades (Ramachandra & Nalina, 2024).

The implanted hearing device includes the cochlear implant (CI), the auditory brainstem implant (ABI), the mechanical middle ear device, and the BCD (bone conductive device). The principle of mechanism for the CI and the ABI is similar. A sound detector is placed behind the ear and collects the sound that is going to the speech processor (Schilk et al., 2023). The processor converts the sound into a control signal and sends the control signal to the RF link converter, where the control signal is converted into RF and transmits the RF signal to the inside sub-stage implant. The RF signal is intercepted by the recovery coils and converted from RF to the control signal, which is then sent to the electrode or the electrode array. This stimulates the neurons of the cochlea along or at the cochlea's inner wall. Then, the cochlea's inner wall stimulates the clear channel and transmits the signal to the brain. A wire connects the CI to the microphone that is connected to the human aural organ. The sensorimotor hearing cells of the inner ear are activated by the auditory brainstem implant (Ojha, 2023).

The definition of A cochlear implant is considered as a small electronic device that operates electrically to ensure the stimulates of the cochlear nerve in the skull (nerve for hearing). The implantation includes external and internal parts (Rayes et al., 2019). The device of Cochlear Implant (CI) bypasses acoustic hearing through direct electrical stimulation to the auditory nerve. Through everyday listening and auditory therapy training, the cochlear implants give the chance to both children and adults to train and learn to interpret all received signals as speech after speech therapy courses and sound after training (Naples & Ruckenstein,2020).

Regarding to external part sitting behind the patient's ear by specialized sergeant uses special tools like ESU and drill to set the implanted piece inside the patient's head (Mohammed et al., 2023). The external part sits behind the ear which picks up sounds with one of the small components in the external part called a microphone. Then it starts processing the sound and transmits it to the second internal part of the implant the small component called the coil. As the internal part sits under the skin behind the patient's ear during surgery done by experienced surgeons. A small electrode placed in a thin wire inserted into the cochlea by surgeons, which is an anatomic part of the inner ear (Aebischer et al., 2023). The small wire sends signals by the electrodes to the cochlear nerve, as it sends sound data to the brain so it can produce a sensation of hearing after receiving the data through the coil from the external part (Li et al., 2024). As a matter of fact, normal hearing is not restored, but with appropriate therapy training and practice, the hearing experience can be improved which means an increased awareness of sounds in the surrounding environment, as well a better communication through reading lips easier and listening (Heutink et al., 2021), as shown in Figure 3.

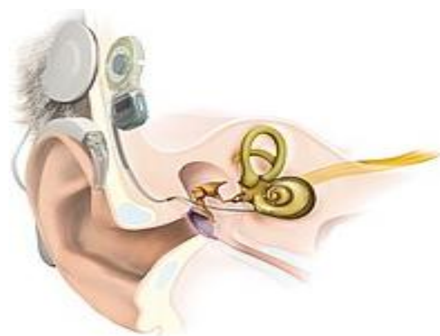


Figure 3: Cochlear Implant System

In the early days of implants which started in the 1970s and the 1980s, speech perception via an implant was steadily increased. More than 200,000 people around the United States, which had received a Cochlear Implant in 2019 (Tessler et al., 2023). Many of these users implanted with modern implants gain reasonable to good hearing and perfect speech perception skills with post-implantation, especially when combined with lip reading. There are some challenges one of these challenges that remain with the implants is the skill of speech and hearing understanding after implantation which shows a wide range of differences across individual implant patients. There are some facts that need to be considered such as parental involvement, education level and age of implantation, cause of hearing loss and duration, how the implant is placed in the cochlea, the overall health condition of the cochlear nerve, also individual skills of re-learning are measured to contribute to this difference. The auditory nerve delivers impulses to the brain, where they are translated as sound (Yap et al., 2020). A cochlear implant system has useful blocks, as shown in Figure 3. It has an external unit, known as the sound processor which consists of a digital signal processing (DSP), also a power amplifier, and a radio frequency transmitter for sending the power and stimulus data through the skin. The internal unit contains a Radio Frequency receiver to demodulate and rebuild the digital signal containing encoded sound information. Then, the microcontroller starts to decode the sound information and propose this to the stimulator. As the stimulator begins to apply stimuli to the electrode arrays inside the cochlear implant. The readout system also reads out the responses of the neural (Pirim et al., 2024).

An additional type of implant, that still in the process of study phase, is an implantable device of a cochlear implant, as shown in Figure 3, in this type of device, everything is implanted inside the body of the patient which means external and internal part of the device and the HFIC coil only use when an external device control and battery of the device system is charging needed (Mahdi et al., 2023). Figure 4 shows the useful block diagram of an implanted cochlear implant.

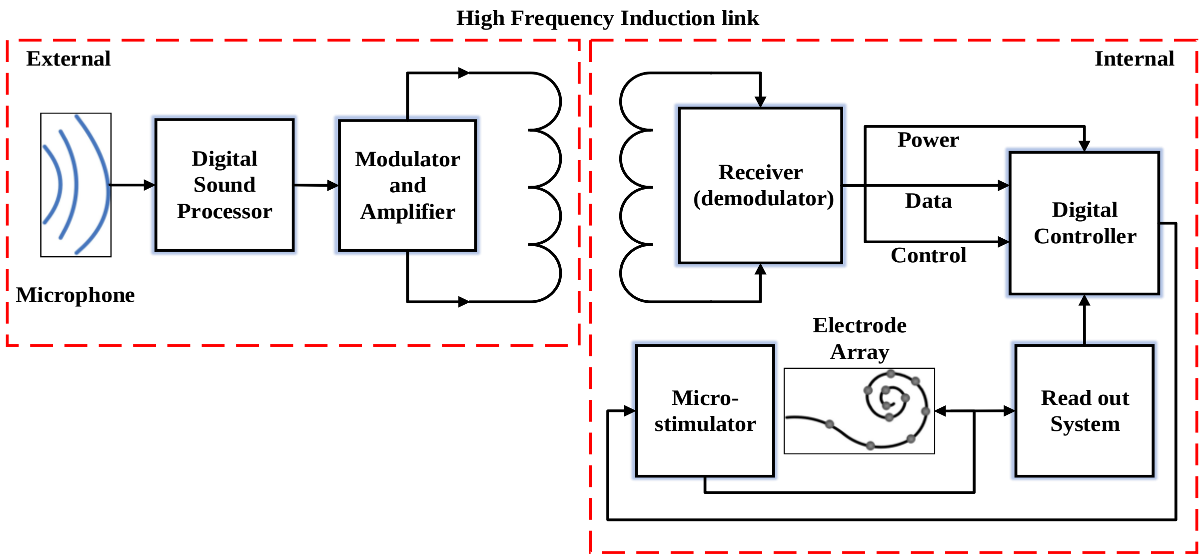


Figure 4: Block Diagram of a Modern Cochlear Implant System

To pick up a signal that is converted into sound, a microphone is attached within the auditory canal or another component called an accelerometer is attached to the middle ear bone structure to translate bone vibration into an electrical impulse representing the unique acoustic information, therefore acting as a microphone. The sound processor of the device can be implemented inside the body by using either technology analog or digital methods as it consumes as tiny power as possible (Park et al., 2024). Cochlear implants (CI) offer a large majority of receivers with a significant grade of speech understanding. Though CI systems depend on external parts, which are named Behind-the-Ear (BTE) or audio processors with coils, this can have numerous disadvantages: for example, hardware can be exposed to external disturbance and the effects of both head movement and gravity. The external device is also put at risk by dust, humid, or dirty surroundings as well as by any physical activities that can lead to water contact such as swimming or other sports in general (Dhanasingh et al., 2024). In addition, there are some patients afraid of the cosmetic entrance of the external parts that are visible (behind-the-ear hearing aids), which might not be desirable to so many potential candidates (Karrenbauer et al., 2023).

5 Cochlear Implant Manufacturers

There are currently three manufacturers of CIs that have FDA approval, Cochlear Corporation (Australia), Advanced Bionics Corporation (USA), and Med-El GmbH (Austria) (Figure 5).

Figure 5: (A) AB HR90 K (B) Cochlear Nucleus 5 (C) Med-El Sonata ti100

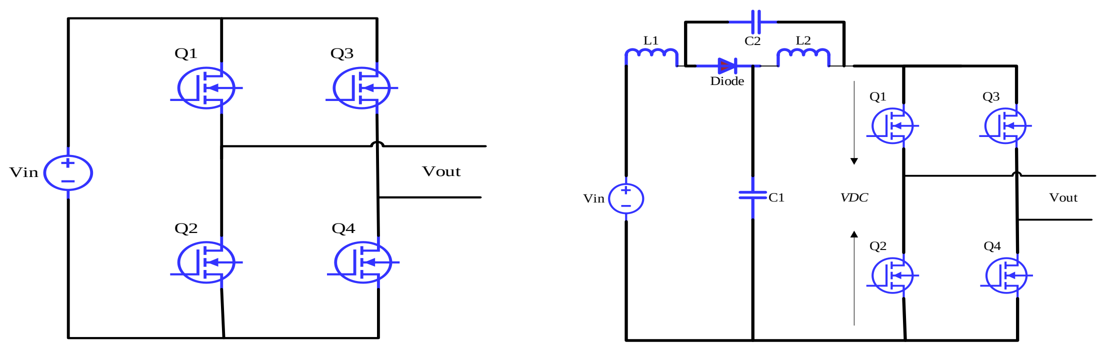
Regardless of differences in component design and also sound-processing method or strategies, device performance is more comparable between all three implant manufacturers when assessing present-day designs. Practically all modern CIs share a set of common useful components (Yaar-Soffer et al., 2024). Recent devices and electrode designs are available from all manufacturers (Table 1), discuss some of the notable advances in electrode design and use (Aljazeeri & Hagr, 2024), and finally most discuss the potential future of the electrode design options. While several driving ideas continually influence and figure out the evolution of electrode design, authors focus their conversation on the electrode-nerve interface and the combination of trauma-minimizing strategies for the sake of brevity (Lambriks et al., 2023).

Table 1: Cochlear Implant Manufacturers (Dhanasingh et al., 2024)

Category	Subcategory	Manufacturers		
		Bionics (HR90K)	Cochlear (Nucleus5)	Med-EI (Sonata)
Dimensions (mm)	Receiver Stimulator	L: 56, W: 28, Th: 5.5	L: 50.5, W: 30.5, Th: 3.9	L: 45.7, W: 24.8, Th: 5.9
	Titanium Housing	L: 20, W: 20, Th: 5.5	L: 22.3, W: 23.5, Th: 3.9	L: 17.4, W: 24.8, Th: 5.9
	Telemetry Coil	L: 28, W: 28, Th: 3.0	L: 28.2, W: 30.5, Th: 3.3	L: 28.3, W: 24.8, Th: 3.7
Weight (g)	Receiver Stimulator	12	8.8	8.6
where L is the maximum length, W is the width and Th is the thickness.				

6 Electronic Circuits

The primary coil of an inductively coupled wireless transmission system has to be excited by high-voltage AC. An amplifier is needed for both voltage amplification and effective DC-to-AC conversion since batteries generate significantly lower voltages. In this design, the converting process is implemented by DC-AC single-phase traditional and quasi-Z-Source inverters . Two inverter circuit types are depicted in Figure 6 (Mo et al., 2024; Bharatiraja et al., 2017).



(a) (b)

Figure 6: DC-AC Single Phase (a) Traditional and (b) Quasi-Z-Source Inverters

Figure 6 (a) depicts a single-phase full-bridge inverter with four power switches: Q1–Q4. The switch pairs (Q1, Q4) and (Q2, Q4) operate sequentially. An inverter is a circuit that converts DC power into AC power at desired output voltage and frequency.

The second method proposed to improve the efficiency system is the quasi-Z-source inverter shown in Figure 3(b). L1, L2, C1, and C2 for the quasi-Z-Source inverter in Figure 6 (b) can be calculated by equations blow (Bharatiraja et al., 2017; Endiz & Akkaya, 2022):

$$L = \frac{V_{C1} D V_{in}}{P f_{sw} r_C} \quad (1)$$

$$C = \frac{2PD}{V_{in} V_{DCpeak} f_{sw} r_v} \quad (2)$$

Where: L, C, the inductance of each inductor and capacitance of each capacitor in the quasi-Z-Source inverter, respectively. V_{C1}, D, Vin, P, f_{sw}, r_C, r_V, the operating dc voltage of the capacitor C_1, duty cycle, input voltage, power rating, switching frequency, output current ripple, and output voltage ripple respectively.

7 Methodology

In this paper, the Optimization of the coupling coefficient and investigation of the system efficiency are demonstrated at a distance of 10 mm, optimizing and Coupling coefficient as shown in Figure 7.

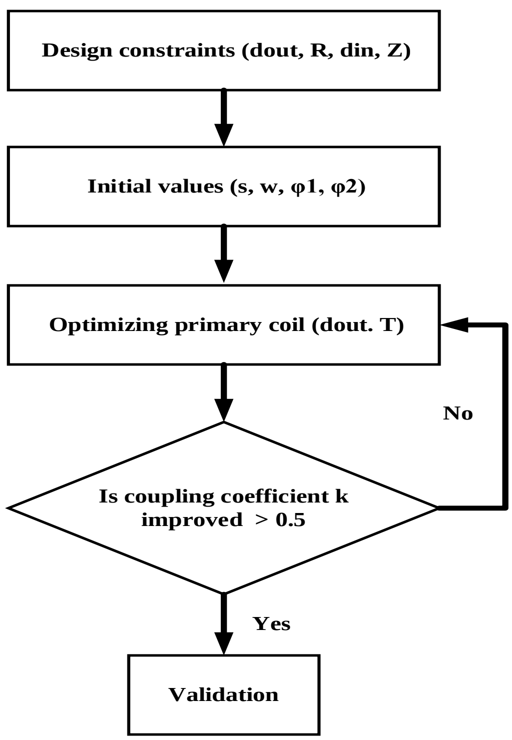


Figure 7: Flowchart of Optimization Coupling Coefficient

There is a set of controlled parameters that are executed by factors which are associated with the chosen cochlear implant system. The parameters are usually defining the size constraints as very limited by where the internal device is located. Some other factors are associated with fabrication technology. It consists of the minimum features that result in a very acceptable manufacturing. As in this case, the constrained limits and their standards are listed in Tables 2 and 3. The initial Values of the primary coils have an outer diameter defined as $dout$ the spacing between conductors ($s1$), as the width of conductors defined as ($w1$) for the primary coils can select variables to be improved. Some set of the initial values for this numerus is defined before to improve the coupling coefficient. The external coil is only modified and the internal coil size is kept small. First, AS changes the width of the Inductor (w) and keeps the initial values set except for the outer diameter of the coil change with the change in Inductor width and measuring the coupling coefficient. The inductance can be calculated according to the formula below (Mutashar et al., 2013):

$$L = N^2 \frac{c_1 D_{avg} \mu_0}{2} \left[\ln\left(\frac{c_2}{\varphi}\right) + c_3\varphi + c_4\varphi^2 \right] \quad (3)$$

where: -
(L): self-inductance in Henry,
(μ_0): Absolute permeability of free space,
(d_{in}): outer diameter of the coil (m),
(d_{out}): inner diameter of the coil (d),
(N): number of Turns,
(D_{avg}): coil average diameter:

Table 2: The Parameter Initial Values of the External and Internal Coils for Circular Coil (Mo et al., 2024; Mutashar et al., 2013)

Definition	**Symbol**	**External Coil**	**Internal Coil**
Outer diameter	d_{out}	56 mm	11.6 mm
Inner diameter	d_{in}	10 mm	0.5 mm
Average diameter	D_{avg}	33 mm	8.25 mm
Number of turns	N	30	8
Inductor width	w	0.5 mm	0.3 mm
Turn spacing	s	0.3 mm	0.1 mm

Table 3: The Parameter Initial Values of the External and Internal Coils for Square Coil (Bharatiraja et al., 2017; Mehri et al., 2016)

Definition	**Symbol**	**External Coil**	**Internal Coil**
Outer diameter	d_{out}	32 mm	11.6 mm
Inner diameter	d_{in}	6.1 mm	0.5 mm
Average diameter	D_{avg}	-	-
Number of turns	N	16	7
Inductor width	w	0.55 mm	0.25 mm
Turn spacing	s	0.25 mm	0.25 mm

$$D_{ave} = \frac{d_{out} + d_{in}}{2} \quad (4)$$

(φ): filling factor:

$$\varphi = \frac{d_{out} - d_{in}}{d_{out} + d_{in}} \quad (5)$$

1) Quality Factor

The inductor superiority factor is a significant parameter that affects the connection power efficiency, which is related to the capacitance and parasitic resistance of the inductor. Take in mind the skin effect, in order to calculate the total parasitic resistance, the formula below can be used (Mutashar et al., 2013):

$$R_s = R_{dc} \frac{t_c}{\delta \,.\, (1 - e^{-\frac{t_c}{\delta}})} \quad (6)$$

Where:
(R_{dc}): DC resistance:
(ρ_c): Metal resistivity (Ω)
(l_c): Total length of the conductor (m)

(t_c): Conductor thickness (mm)
(δ) : Being the skin metal depth:
(f) : Operating frequency (MHZ)
(μ) : Permeability constant (m/H):

Pay no attention to the parasitic capacitance of the circuit, the superiority factor of the Printed Coil Circuit (PSC) as it can be expressed as follow $Q = \omega L/Rs$ (Mutashar et al., 2013).

2) Coil Mutual Inductances

Mutual inductance can be a key parameter for inductive power transmission. Assuming the perfect alignment, which has the total common inductance between two coils (Khan et al., 2019) it can be expressed by (Mutashar et al., 2013):

$$M_{T.R} = \frac{\mu_0 N_T d_{out.T}^2 N_R d_{out.R}^2 \pi}{2\sqrt{(d_{out.R}^2 + Z^2)^2}} \quad (7)$$

where:
(d_{out}): Coil outer diameter,
(N_T): Number of turns of transmitter coil,
(N_R): number of turns of the receiver coil,
(Z): Distance between coil.

8 Result and Discussion

To improve the coupling coefficient, the external coil is modified to keep the internal coil size small. First, we changed the width of the Inductor (w) and we kept the initial values set except for the outer diameter of the coil change with the change in inductor width and we measured the coupling coefficient (shown in Tables 4).

Table 4: Change the Width of the Inductor (w) for the Square and Circular Coils

w **(mm)**	***L*** **(uH)**	***K*** **(*L*)**	d_{0ut_T}**(mm)**
Square Coil			
0.55	5.04	0.0789	32
0.75	5.04	0.0789	38
0.95	6.22	0.1342	44
1.15	6.93	0.1708	51
1.35	7.75	0.2057	57.2
1.55	8.27	0.2463	64
1.75	8.91	0.2848	70.1
1.95	9.57	0.3272	76.5
2.15	10.25	0.3722	83
2.55	11.58	0.4655	95.7
Optimal Case			
2.75	**12.25**	**0.5151**	**102.1**
Circular Coil			
0.5	25.6	0.2152	58
0.7	28.924	0.3121	70
0.9	32.317	0.4045	82
1.1	35.75	0.5047	94
Optimal Case			
1.3	**39.209**	**0.6121**	**106**

Second, set the initial values as well, but change Turn spacing and measure the coupling coefficient. Now, after optimizing, the Coupling coefficient is 0.5151 for a square coil and 0.6121 for a circular coil. Table 5 lists the optimized parameters of the designed external coils.

Table 5: The Parameter of the External Coil After Optimization

Symbol	External coil (circular)	External coil (square)
d_{out}	106 mm	102.1mm
d_{in}	10 mm	6.1 mm
N	30	16 mm
W	1.3 mm	2.75 mm
S	0.3 mm	0.25 mm

After applying the above-proposed method to improve the efficiency of the implanted hearing system, it's found that the operating efficiency is 72.75% for the square coil and 74.55% for the circular coil, Figure 8 shows the efficiency ratio for tow coils. Based on the results obtained, the system efficiency ratio for both coils has somewhat low values, as we are dealing with an auditory system that is in direct contact with the patient’s vital parts.

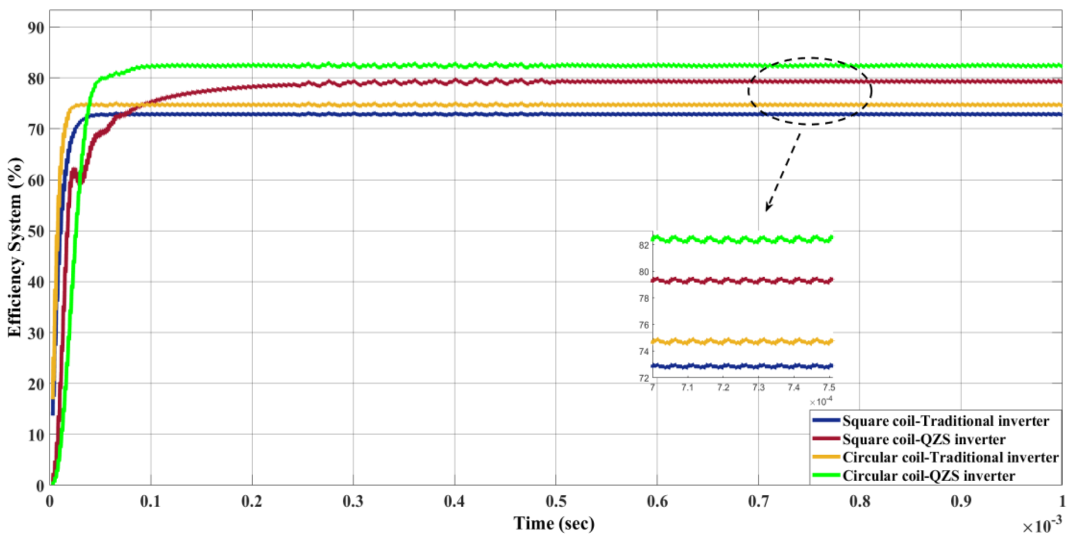
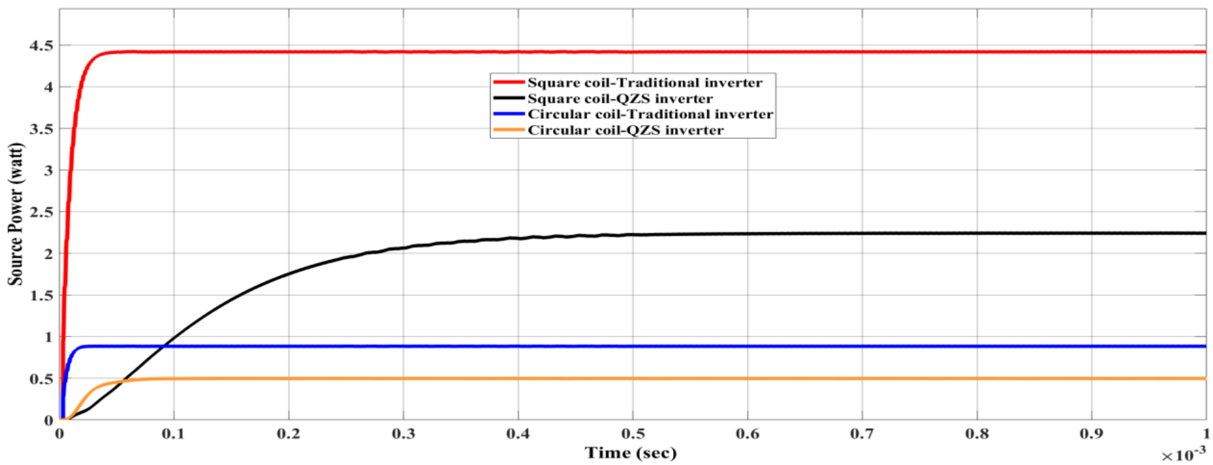


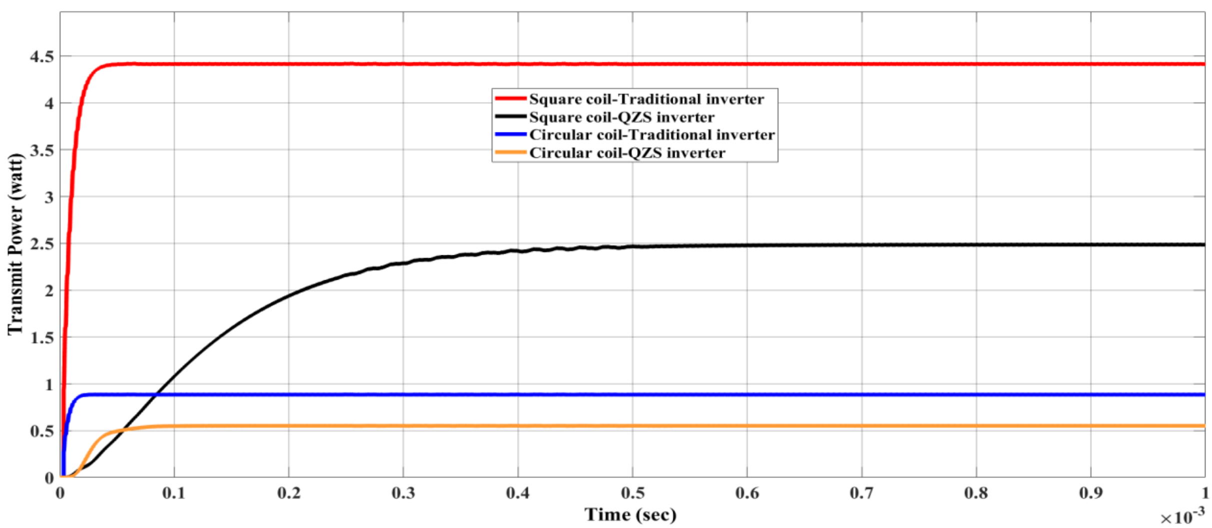
Figure 8: Efficiency Ratio for the Hearing System

Accordingly, this study resorts to applying the other proposed method, which is to replace single phase inverter with a quasi-Z-source inverter as shown in Figure 6. According to Figure 6 and from Eq 1, 2 can be finding L_1=L_2= 190μH, and C_1=C_2=0.86μF. The design process included applying a quasi-Z-source inverter circuit with a wireless power transmission circuit (using Table 5 for the wireless power transmission parameters design). After applying the design method founding that the efficiency for the square coil 79.16% and 82.29% for the circular coil, for more details see Figure 8.

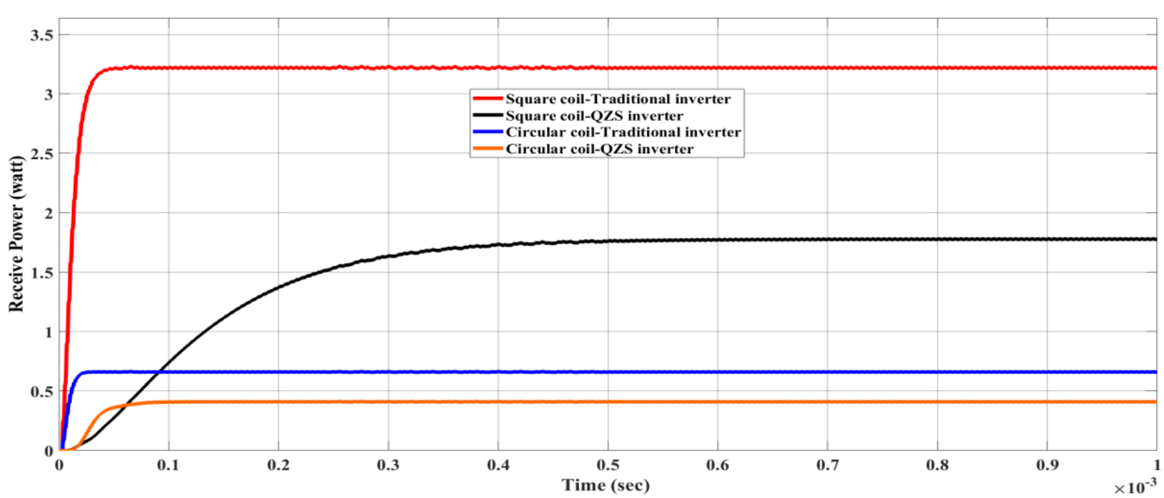
Figure 8 apparent that the efficiency ratio in the quasi-Z-source inverter circuit is higher than when traditional inverter circuits for two types of coils, also, the efficiency ratio for the circular coil in the case of traditional and quasi-Z-source inverter circuits is higher than the square coil. Figure 9 shows source, transmit, and receive power. From Figure 9 (a) it can be noted that the source power decreased when using a quasi-Z- source, this decrease led to a decrease in the transmission and receive power of the coils and thus reduced the amount of power losses. Table 6 contains details design for two coils and two types of inverters.



(a)



(b)



(c)

Figure 9: (a) Source, (b) Transmit, and Receive Power

Table 6: Design Parameters for the Implanted Hearing System

Inverter Type	Source Power (watt)	Output Power (watt)	Loss Power (watt)	Efficiency (%)
Square Coil				
Traditional	4.419	3.215	1.204	72.75
Quasi-Z- Source	2.243	1.776	0.467	79.16
Circular Coil				
Traditional	0.8852	0.6599	0.2253	74.55
Quasi-Z- Source	0.4982	0.4099	0.0883	82.29

9 Conclusion

In this research, the improvement efficiency of the implanted hearing system is discussed by improving the design of the electronic circuits for the system, and this is achieved through proposed methods for the improvement process, which are represented in both improving the coil parameters of the wireless transmission circuit method and quasi-Z-source inverter circuit method. Based on the practical results (see Table 6), it can be said that using the quasi-Z-source inverter circuit type in building the system helps improve the performance of the device when compared to improving the coil parameters method. We also find through the practical ratios that circular coil gave better results than square coil, which means that it is better to use it in the process of designing the wireless transmission circuits for implanted hearing devices.

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