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Survey of near-field wireless communication and power transfer for biomedical implants

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HIGHLIGHTS

- Review of wireless power transfer techniques and their classifications in biomedical applications.
- Comparison of WPT methods for biomedical devices based on performance metrics.
- Introduction to major near-field wireless power transfer topologies and related mathematical models.
- Comparison of data transmission schemes in WPT-based modulation techniques.
- Investigation of the main applications of biomedical devices, then discussion of the challenges and solutions.

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ABSTRACT

Bio-implanted medical devices with electronic components play a crucial role due to their effectiveness in monitoring and diagnosing diseases, enhancing patient comfort, and ensuring safety. Recently, significant efforts have been conducted to develop implantable and wireless telemetric biomedical systems. Topics such as appropriate near-field wireless communication design, power use, monitoring devices, high power transfer efficiency from external to internal parts (implanted), high communication rates, and the need for low energy consumption all significantly influence the advancement of implantable systems. In this survey, a comprehensive examination is undertaken on diverse subjects associated with near-field wireless power transfer (WPT)-based biomedical applications. The scope of this study encompasses various aspects, including WPT types, a comparative analysis of WPT types and techniques for medical devices, data transmission methods employing WPT-based modulation approaches, and the integration of WPT into biomedical implantable systems. Furthermore, the study investigates the extraction of research concerning WPT topologies and corresponding mathematical models, such as power transfer, transfer efficiency, mutual inductance, quality factor, and coupling coefficient, sourced from existing literature. The article also delves into the impact of the specific absorption rate on patient tissue. It sheds light on WPT's challenges in biomedical implants while offering potential solutions.

1. Introduction

Nowadays, the energy of wireless power transfer (WPT) is widely employed to replace bio-implantable batteries in implanted systems or devices such as implantable microsystems for nerve and muscle stimulators, deep brain stimulators, and retinal and cochlear implanted devices [1]. As a result of problems caused by batteries for living tissue, such as their size, chemical effects, and lifetime, an addition is to reduce surgical procedures with WPT [2,3]. The system comprises two primary parts [4]: an exterior component that transfers power and data inductively to an internal (implanted) portion positioned within the body. Because the two parts are weakly linked, the system requires an efficient design for its external and internal coils to avoid tissue damage [5,6]. Mobile and wearable devices, medical devices, and wireless sensors are commonplace battery-powered electronic devices used in daily life [7]. Implantable medical devices (IMDs) are a group of microelectronic equipment that monitor or replace sensory functions and are increasingly improving quality of life. The aim of employing an IMD to help patients originated in the 1950s when the development of transistors made it possible to create wholly implanted

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pacemakers [8,9]. Various such systems have since been upgraded, such as cochlear implants, brain-computer interfaces, heart rate monitors, and retinal implants [10].

Most IMDs require a constant and sufficient power supply to operate appropriately, so researchers have extensively explored multiple power sources for IMDs in previous decades [11,12]. Different power techniques can enable IMDs to be used autonomously by generating electrical energy to supplement or replace battery power [13]. Significant challenges have included implant size limitations, the need to operate continuously, biocompatibility, modulation techniques, coil geometry, data rate, efficiency, and tissue safety [14,15]. Regarding power transmission distance, WPT is classified into two broad categories, far-field and near-field transmission [16,17], as depicted in Figure 1. Far-field transmission prioritizes low-power applications over transmission efficiency. Laser or microwaves typically accomplish the far-field along a direct line-of-sight communication path [16]. Considering the need for safety in human exposure and its efficiency [18], far-field transmission cannot be regarded as a viable alternative for power transfer in daily life [19].

Near-field transmission is superior to microwave or laser transmission due to its high efficiency and lower radio frequency (RF) exposure safety limit [20]. The most popular near-field WPT techniques used are inductive power transfer (IPT), permanent magnet coupling (PMC), and capacitive power transfer (CPT) [21]. Furthermore, due to the resonance of the electric and magnetic fields in the LC circuit, the usage of magnetic resonance coupling (MRC) for IPT has gained popularity in the WPT system [16,22]. Near-field transmissions could be further categorized into mid-range and short-range transmissions. In the former type, the receiver and transmitter are generally separated by a few millimeters due to the two-coil technique [23]. For short-range uses, the resonant circuit's working frequency generally falls between 10 kHz and several megahertz [24]. Typically, the energy dissipated by a power inverter increases directly with the operating frequency. The best approach for data transmission and wireless power in IMDs is near-field magnetic systems since they can effectively transmit both without harming the human body [25,26].

However, there are several drawbacks to near-field magnetic systems as well, including poor data rates, limited transmission distances, and sensitivity to the relative positions of the transmitter (Tx) and receiver (Rx) [27,28]. Existing applications, which need short-range, low-data-rate communications, may not require solving these problems urgently. Based on the application, the magnetic system's limited transmission range might not be a drawback. Researchers mentioned that the magnetic communication system's short-range feature helps ensure security and frequency reuse [27, 29]. When selecting an operating frequency for wireless transfer, the maximum safe dose must also be considered. According to IEEE’s safety requirements for RF exposure, the specific absorption rate (SAR) must not surpass 1.6 W/kg [30, 31]. Thermal impacts on biological tissues are the primary safety concern at frequencies greater than 5 MHz. Both dose and exposure limits are specified in the safety standard [32]. Dose limits define the maximum power density tissue can absorb; exposure limits define the maximum strength and power density of an incoming electromagnetic field produced by an RF source [33]. The maximum allowable dosage for whole-body exposure in an open environment within the frequency band of 100 kHz to 6 GHz is 0.08 W/kg; the dose limit for the torso and head is 2 W/kg, and the dose limit for the pinnae and limbs is 4 W/kg [34]. When exposed to radiation with a frequency range between 6–300 GHz, the body's surface is subject to a dosage cap of 20 Wm^2 in an unconstrained environment. The maximum safe exposure limits differ depending on factors, including body part, frequency, and duration of exposure. However, they cannot exceed 10 mW cm^2 [34,35]. Consequently, near-field is the most popular technique for biomedical applications [36].

In contrast to previous survey/review papers that concentrated on data transfer and wireless power transfer, this article distinguishes itself by delivering a comprehensive survey of various WPT types, specifications, and advantages. Furthermore, it examines the use of WPT-based modulation techniques for data transmission, focusing on demonstrating the prevalence of amplitude shift keying (ASK) modulation in biomedical applications. Additionally, the article explores a range of implantable biomedical devices, including brain optoelectronic implants, implantable cardiac pacemakers, and biomedical capsule endoscopy.

This paper aims to provide an in-depth analysis of near-field WPT technology and its challenges in facilitating the design of biomedical implants using WPT. The following are the review's contributions:

1) A general review of wireless power transfer techniques and their classifications supported by equivalent circuits in biomedical applications is introduced.
2) A comparison of WPT methods for biomedical devices regarding several performance metrics was extracted.
3) The major near-field wireless power transfer topologies with related mathematical models are introduced.
4) Data transmission schemes in WPT-based modulation techniques were reviewed and compared regarding modulation type, data rate, carrier frequency, and power delivered to the load.
5) The main applications of biomedical implantable device-based WPT were investigated, and the selected articles were compared.
6) The challenges and solutions for WPT in biomedical implantable devices have been discussed, and the future trends in the intended application are presented.

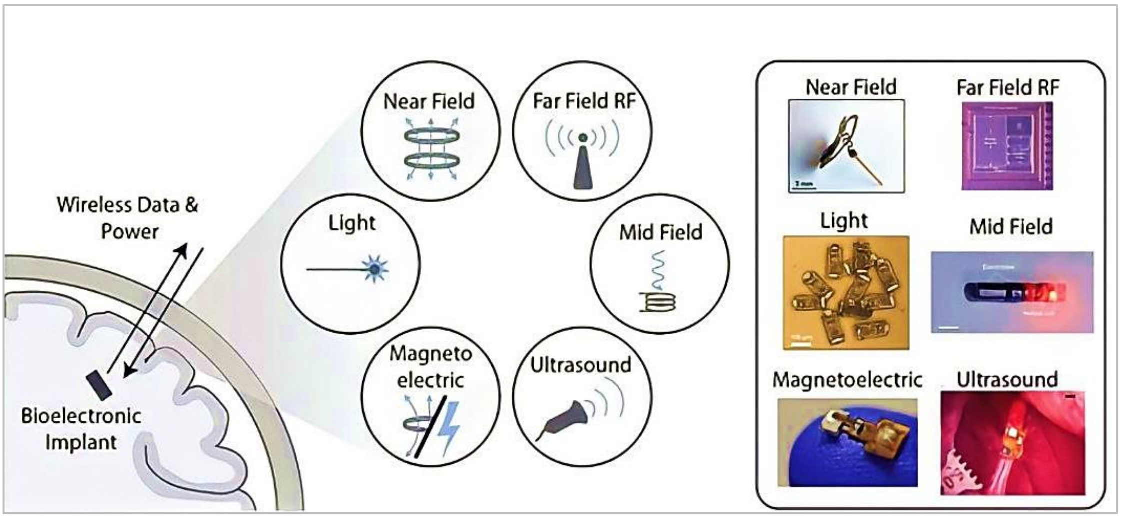

Figure 1: Types of wireless power transfer [37]

2. Review of wireless power transmission

WPT technology allows the transmission of power over a media, from an energy source to an electrical load, without electrical cables [38,39]. This technology is widely used in many applications, from high-powered electric vehicles to sophisticated low-power healthcare implants [40-42] and white goods such as cell phones and electric toothbrushes [43]. Providing sufficient energy without requiring large batteries continues to be a critical task, and researchers have attempted to overcome this issue since the inception of IMDs. A cardiac pacemaker was the first device implanted for an extended period using WPT; through an external power transmitter, WPT provides an alternative power delivery method for IMDs [44]. In general, WPT techniques can be classified as radiative or non-radiative. Non-radiative approaches include inductive coupling [45], capacitive coupling [46], and magnetic resonance coupling [47].

WPT techniques using inductive coupling are widely employed in implanted microelectronic systems to improve patient comfort and minimize infection hazards [48]. A crucial criterion for bio-implants is that they are as tiny as feasible. An implanted device's size determines how far into the body's biological tissue it can work. Typically, these devices are inserted into patient tissue at a depth of less than 10 mm [49]. Microsystems are often implanted at a deepness of 1–4 mm [50]. Cochlear implants are placed in the temporal bone 3–6 mm deep [51], while retinal implants are placed 5 mm deep [52]. Khan and Choi [53] examined a fully planar 4 coils MRC-based WPT employing circuit-based modelling. They found that the MRC–WPT system's efficiency decreased as the receiver location changed due to transmission angle, axis misalignment, and distance changes. Notably, because the implanted system was placed within the tissue layer, it was not easy to detect the precise location of the receiving device. Several studies have enhanced the WPT's efficiency by utilizing a ceramic-filled chamber, while others have tracked the system's ideal efficiency. Nevertheless, the power transfer efficiency (PTE) of the implanted system-based WPT remains small but has increased slightly under various situations [54].

Yazdi et al. [55] developed a two-coil inductive WPT technique with three distinct coil shapes (circular, square, and elliptical) at a resonance frequency of 13.56 MHz to beat low PTE using six different coil configurations. Furthermore, the highest PTE for a square–square coil was 56.2% when the coils were close together. Recent advancements in WPT design aim to maximize PTE across several tissue layers while retaining an implant's reliability. At low-MHz operating frequencies, conventional WPT techniques employ a receiver–transmitter (Rx–Tx) coil couple for inductive coupling [56].

3. Types of wireless transfer energy systems

WPT may be roughly divided into radiative and non-radiative categories. Laser power transfer (LPT) and Microwave power transfer (MWPT) use radiative power transmission. In contrast, inductive power transfer (IPT), capacitive power transfer (CPT), magnetic resonance power transfer (MRPT), and ultrasonic power transfer (UPT) use non-radiative power transmission methods, as demonstrated in Figure 2. The non-radiative power methods are the most promising techniques for implanted devices. UPT has outstanding performance characteristics for single deep implants. Nonetheless, it typically requires off-chip components to be connected via an elastic printed circuit board, which adds to the implant's full size.

Inductive power transfer is appropriate for medium-shallow implantation depths and provides the critical feature of allowing many chips to be powered simultaneously. CPT exhibits excellent characteristics for short-range implantation deepness. However, it also offers the essential feature of permitting inclusion on flexible substrates (like IPT) while preventing electromagnetic intervention (like UPT). IPT has been extensively studied in preceding decades and, as a result, is the most frequently utilized WPT approach in the scientific world [57]. Even though UPT and CPT are relatively emerging methods in this area, they have already demonstrated encouraging results in developing wireless implanted equipment for biomedical applications and neurological prosthetics [58]. This article describes the basic concepts of each WPT method and then discusses the most innovative approaches with the potential for application in implantable devices. Furthermore, it compares three strategies in its concluding remarks.

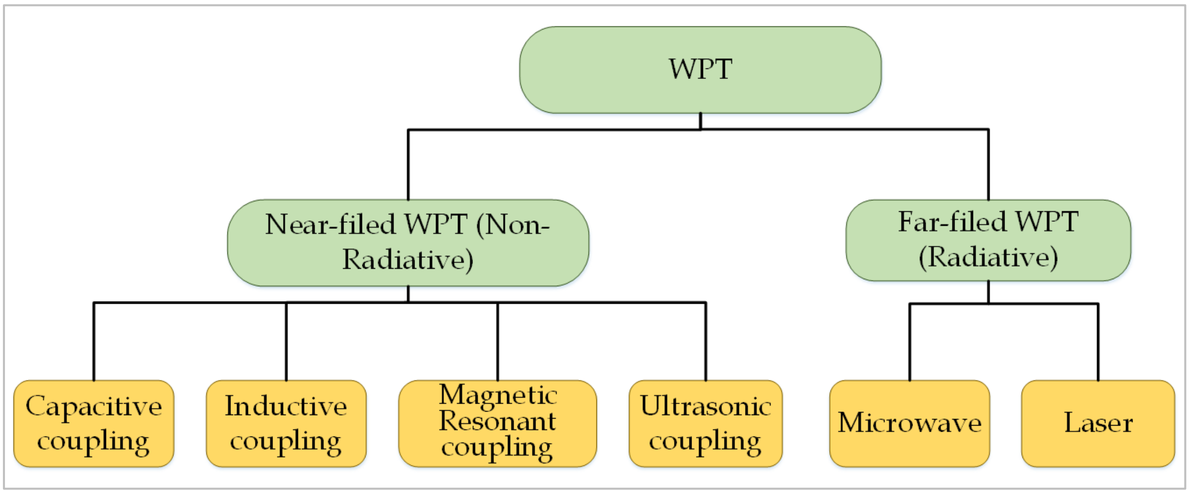

Figure 2: Types of wireless transfer energy systems

3.1 Capacitive coupling system

CPT is a rapidly growing technology in the WPT topic. CPT employs an electrical field to achieve energy transfer. The capacitive connection of metal electrodes enables energy transmission via electric fields. Non-radiative fields are those in which the energy is limited to a small area around the transmitter. No power leaves the source (transmitter) if no destination (receiving) system or absorbent material is found within the designated range of the pair. When two antennas' distances are more significant than their combined diameters, only a negligible quantity of power may be received, as the intensity of created electric fields declines exponentially with distance. The distance between the coupling plates is crucial for CPT. As depicted in Figure 3, the CPT coupler comprises two sets of metal sheets interconnected in series [20].

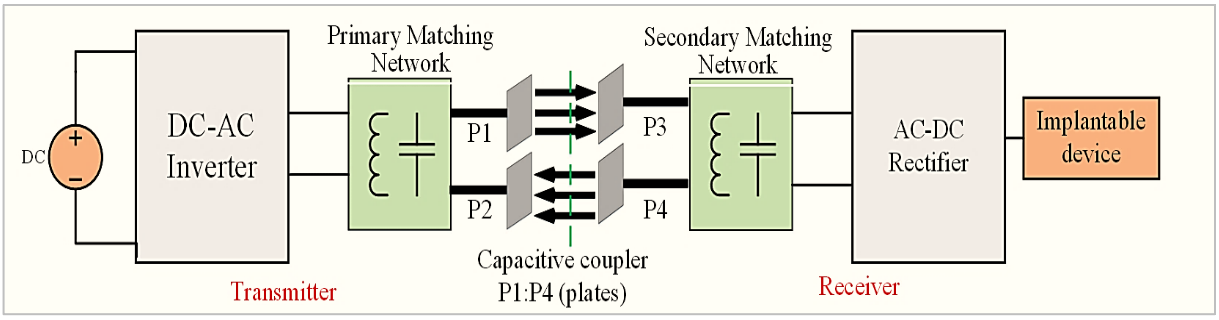

Figure 3: The CPT structure diagram

The CPT employs a displacement current to transmit energy between two contactless conductor plates without a conducting medium. Equations (1, 2) give the displacement and conduction [59]:

$$I_{displacement} = \epsilon_o \epsilon_r(w) A \frac{\partial \vec{E}}{\partial t} \quad (1)$$

$$I(t)_{conduction} = \frac{V(t)\sigma(w)A}{D} \quad (2)$$

where $\vec{E}$ is the electric field intensity, σ is the conductivity of the medium, A is the area of the plate, ϵ_r , ϵ_o relative and free space permittivity, w=2πf angular frequency, v(t) is the applied voltage, t denotes time, and D is the distance between the plates. The displacement current must be raised to enhance the energy transfer, but the conduction current must be lowered as much as possible to minimize the power lost in the tissue. The electric field intensity $\vec{E}$ is the most significant crucial factor determining the displacement current. The transfer efficiency can be improved by reducing the distance between the coupling plates and raising the transmitter excitation voltage. The rate of variation of the electric field $\vec{E}$ which affects the displacement current value [60]. The CPT technique has many advantages, including its minimal eddy current loss between adjacent metals and low cost [61]. CPT is suitable for short-range and low-power uses, such as those for biomedical devices [62], USB and mobile device charging [63], and, most notably, for wirelessly charging unmanned aerial vehicles (UAVs) [64].

Sedehi et al. [65] have presented a capacitive-linked, inductive power transfer technology for deep-implanted biomedical devices that enables the safe flow of energy into the body whilst utilizing the least possible amount of implant volume. Electrodes with parallel insulated capacitive insulation conveyed a constant current flow into the tissue and implants. The final power supply operates at 6780 kHz and provides a power of 10 mW deep within the body whilst conforming to the IEEE C95.1 fundamental restriction standard.

3.2 Inductive coupling system

IPT is a near-field WPT technique that is based on the idea of magnetic induction [66]. In Figure 4, an IPT transmitter generates an alternating-current (AC) magnetic field by delivering an AC voltage to a linked coil. Simultaneously, the magnetic induction principle generates an alternating current on the IPT receiver's wiring coil. This method uses two coils— a secondary for the receiver and a primary for the transmitter—similar to a voltage transformer. In this case, the air gap in the transformer produces an inductive magnetic environment. The IPT approach consumes more energy for the resistors in the primary coil and is ineffective over extended distances [67].

IPT has both pros and drawbacks: it has modest topology and high-power transfer efficiency, even over relatively short distances, is non-radiative and, therefore, safe for human tissue, and is easy to implement. The drawbacks are that it has a low transmission distance, requires alignment between receiver and transmitter coils, and is unsuitable for portable use [68].

Inductively coupled systems have traditionally used rectangular coils. However, Duan et al. [69] have developed a new approach for optimizing and characterizing them. A PTE of 46.4% was achieved using two rectangular coils separated by 10 mm and running at a frequency of 3MHz. The suggested system was proven through simulations and experimentation. Wang et al. [70] designed a retinal prosthesis-based IPT with a carrier frequency of 1 and 20-MHz for data and power, which could deliver approximately 250 mW of power at an air-gap of 7 mm to 1.5 cm. As measured at 7 mm, Wang et al.'s system achieved a power transfer efficiency of 30%. Yazdi et al. [71] applied the inductive WPT system to two coils of different shapes (circular, elliptical, and square) with six coil combinations. The results indicate that the maximum PTE for a square–square coil at 13.56 MHz and 0 mm between coils is 56.2%. As stated by the authors, efficiency decreases as distance increases, but PTE increases as the number of turns increases.

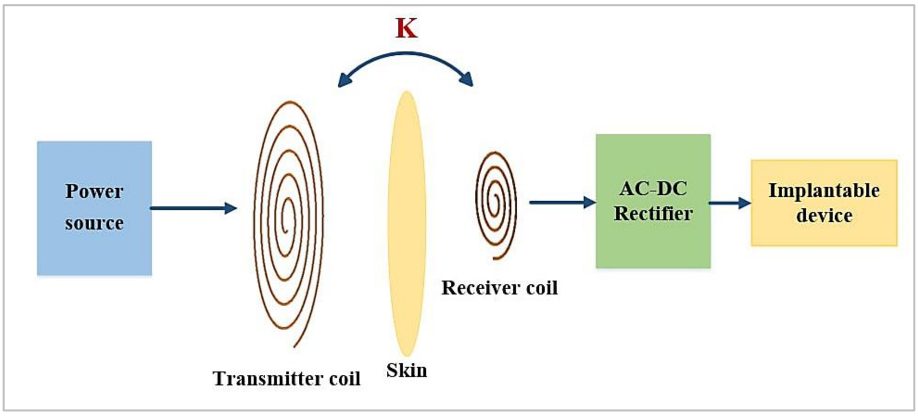

Figure 4: Inductive power transfer system

3.3 Magnetic resonance coupling system

A research team at the Massachusetts Institute of Technology suggested magnetic resonance. Coupling–WPT system (MRC–WPT) technology in 2007. The proposed MRC-WPT technique improves the limited transmission range and inefficient power transfer based on inductive WPT. The transmitting and receiving terminals are synchronized to a similar resonant frequency and use a solid MRC to transfer power efficiently without wires. This is accomplished by connecting the inductance of the transmitting coil to a compensation circuit to create a resonant tank. The magnetically coupled theory describes the process of energy transfer between the coils. MRC–WPT is obtained by operating at a coil resonance frequency. This technique might be accomplished with two or more coils. MRC-WPT was established in recent years. Also, these systems have numerous advantages, including eliminating power lines, charging several devices simultaneously, and having a broad power range. Therefore, the WPT system was used to supply many items with power, including IMDs, electric vehicles (EVs), and consumer electronics [72].

In MRC-WPT, there are four possible configurations: parallel–parallel (P/P), series–series (S/S), series-parallel (S/P) and parallel–series (P/S). As magnetic coupling is relatively weak at greater air-gap, transmitted power is inversely related to transfer air-gap. MRC-WPT employs the strong coupling condition to transmit energy wirelessly across an evaporating near-field and does not emit radiation [73]. To optimize power transmission and minimize losses caused by weak coupling among the system's receiver and transmitter coils, coil inductance should be adjusted by adding parallel or series capacitances to ensure the resonance frequency with the coil inductance [74]. Because a resonator is required in both the transmitter (primary circuit) and receiver (secondary circuit), a capacitor in series or parallel with an inductive coil may be used to achieve resonant topologies in a basic WPT system. The MRC–WPT system has four different topologies that may be used to increase power transmission. The procedure of selecting a compensation topology differs depending on the functions of the transmitter and receiver. The required transmission range or voltage is the critical factor influencing the adoption of various topologies [75]. The selection of the compensation topology depends on the requirements of each application. By placing a capacitor in series, the actual portion of the impedance is not altered, and only a negative reactance is introduced. However, inserting a parallel capacitor modifies the imaginary and real portions of the impedance. The design of the compensating network might be chosen based on load constraints, application requirements for wirelessly powered devices, and source impedances. Figure 5 (a-d) demonstrates the various compensating topologies for PS, PP, SS, and SP.

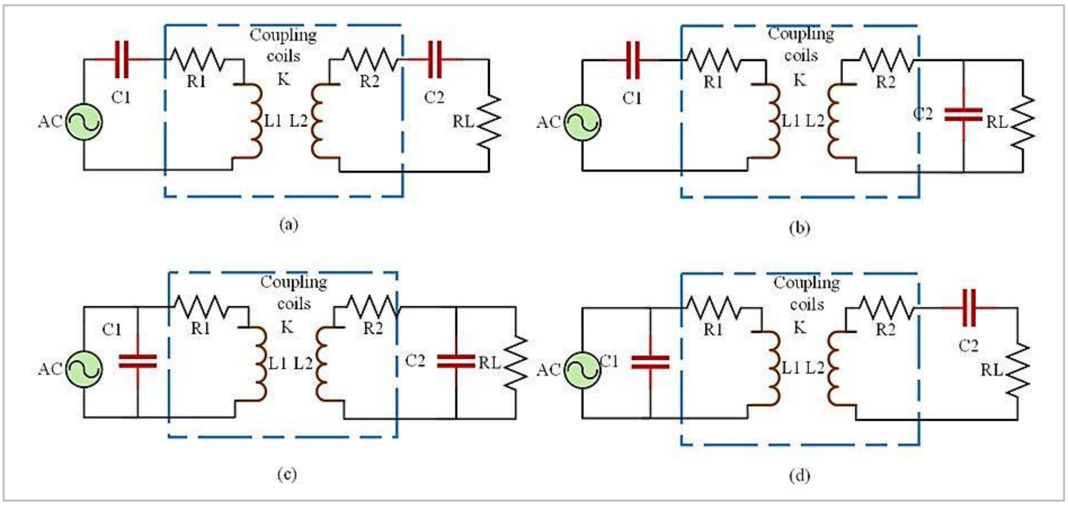

Figure 5: The four different topologies of the magnetic resonance coupling system are (a) series-series compensation, (b) series-parallel compensation, (c) parallel-parallel compensation, and (d) parallel-series compensation

Usually, the fundamental compensation topology has a resonance frequency, as shown in Equation (3):

$$\omega = \frac{1}{\sqrt{L_1C_1}} = \frac{1}{\sqrt{L_2C_2}} \tag{3}$$

The reflection impedance Z_r is derived in Equation (4), and Z_2 is reflected from the receiving coil to the transmitting coil.

$$Z_r = \frac{\omega^2M^2}{Z_2} \tag{4}$$

where Z_2 is the impedance of the receiver side, the transmitter impedance Z_1 stands for the equivalent impedance on the primary side with a value dependent on the compensation topology, which can be represented for series compensation [76] and as shown in Equation (5):

$$Z_1 = j\omega L_1 + \frac{1}{j\omega C_1} + Z_r \tag{5}$$

and for parallel compensation as shown in Equation (6):

$$Z_1 = \frac{1}{j\omega C_1 + {}^1/_{(j\omega L_1 + Z_r)}} \tag{6}$$

The receiver impedance is dependent on the compensation topology, which can be represented for series compensation as Equation (7):

$$Z_2 = j\omega L_2 + \frac{1}{j\omega C_2} + R_L \tag{7}$$

and for parallel compensation as Equation (8):

$$Z_2 = j\omega L_2 + \frac{1}{j\omega C_2 + 1/R_L} \tag{8}$$

The link impedance (Z_{link}) at the primary side may be calculated as Equation (9):

$$Z_{link} = Z_{L1} + Z_r \tag{9}$$

Z_{L2} and Z_{L1} the inductive impedances on the secondary and primary sides of the WPT circuit, respectively ,and as shown in Equations (10 and 11), can be determined by ignoring the parasitic capacitance of the Tx and Rx coils.

$$Z_{L1} = R_1 + j\omega L_1 \tag{10}$$

$$Z_{L2} = R_2 + j\omega L_2 \tag{11}$$

For various topologies, link impedance (Z_{link}), link gain (A), and PTE can be calculated as shown in Table 1.

Table 1: The link impedance (Z_{link}), PTE, and link gain (A) for various topologies

Topology	Link impedance (Z_{Link})	Link Gain (A)	Power transfer efficiency η
SS	$Z_{L1}+\frac{1}{j\omega C_1}+\frac{(\omega M)^2}{Z_{L2}+\frac{1}{j\omega C_2}+R_L}$	$\frac{-j\omega M R_L}{Z_{Link}(Z_{L2}+\frac{1}{j\omega C_2}+Z_{out}}$	$\frac{R_L}{R_2+R_L+R_1(\frac{R_2+R_L}{\omega M})^2}$
SP	$Z_{L1}+\frac{1}{j\omega C_1}+\frac{(\omega M)^2}{Z_{L2}+\frac{1}{j\omega C_2}+\frac{1}{R_L}}$	$\frac{-j\omega M R_L}{Z_{Link}(Z_{L2}+(j\omega C_2+\frac{1}{R_L})+1)}$	$\frac{R_L}{R_2+R_L+\frac{R_2R_L^2}{(\omega L_2)^2}+\frac{R_1R_2^2}{(\omega M)^2}+\frac{R_1(L_2\omega^2+\frac{R_L}{\omega^2}}{(\omega M)^2}}$
PS	$\frac{1}{j\omega C_1+\frac{1}{Z_{L1}+\frac{(\omega M)^2}{Z_{L2}+\frac{1}{j\omega C_2}+}}}$	$\frac{-j\omega M R_L Z_{Link}}{(Z_{L1}+Z_r)\,(Z_{L2}+\frac{1}{j\omega C_2}+R_L}$	$\frac{R_L}{R_2+R_L+R_1(\frac{R_2+R_L}{\omega M})^2}$
PP	$\frac{1}{j\omega C_1+\frac{1}{Z_{L1}+\frac{(\omega M)^2}{Z_{L2}+\frac{1}{j\omega C_2}+}}}$	$\frac{-j\omega M Z_{Link}}{(Z_{L1}+Z_r)\,(Z_{L2}+(j\omega C_2+\frac{1}{R_L})+}$	$\frac{R_L}{R_2+R_L+\frac{R_2R_L^2}{(\omega L_2)^2}+\frac{R_1R_2^2}{(\omega M)^2}+\frac{R_1(L_2\omega^2+\frac{R_L}{\omega^2}}{(\omega M)^2}}$

Compensation capacitors C_1 and C_2 are used to increase the energy transmitted from an alternating current source to output with loading resistance R. The electrical properties may be determined using the Equations (12 and 13) [77].

$$C_1=\frac{1}{\omega L_1} \tag{12}$$

$$C_2=\frac{1}{\omega L_2} \tag{13}$$

The WPT system's efficacy is measured as Equation (14):

$$\eta=\frac{P_{out}}{P_{in}} \tag{14}$$

P_{out} is the output power and P_{in} is the input power of the system.

For superior quality and optimum PTE, the primary coil should be adjusted for series resonance, while the secondary coil should be tuned for parallel resonance [78]. In general, series compensation results in the tuned circuit functioning at a higher voltage and a lower current, allowing greater efficiency. For this reason, the SS topology is frequently used in wireless charging systems. Power transmission capacities are maximized in SP and SS topologies instead of PS and PP. The SS topology is preferred over the SP for biomedical implant applications at higher frequencies. As a result, an SP topology provides excellent WPT performance at lower frequencies [79]. Wang et al. [80] studied the four MRC circuit topologies and their distinctions and selected the most suitable type depending on system needs. They found that the series-parallel (SP) circuit provided the highest PTE and load power among the four tested topologies, making it the most suitable for use in a WPT system with a high load impedance. Kamarudin et al. [81] presented a strongly coupled magnetic resonance technique for wireless power transfer. The antennas are designed in circular and square shapes with 1-turn coils with a frequency of 3.4–3.5 GHz. The circular antenna has the highest efficiency at 31.58% when the distance is 2 mm, 31.26%, and 31.02% at 3 and 4 mm, respectively.

Therefore, the MRC–WPT technique has been developed to overcome the efficiency loss when a significant distance separates the receiver and transmitter. As a subset of the magnetic induction technique, the transmitter and receiver resonators must have the same resonance frequency for this configuration to work. This is a more detailed system design, but it allows for simultaneous charging and powering of several devices from a single transmitter and for WPT across a greater distance or area. Furthermore, MRC–WPT focuses in addition to the coupling coefficient K but on the quality factor Q. By constructing coils with a high-quality factor, and it is possible to overcome distance issues and achieve high transfer efficiency even though the separation between the transmitter and receiver increases, K's value drops.

3.4 Ultrasonic coupling system

Ultrasound is a periodic sound pressure with a frequency that exceeds the human hearing limit. Certain animals, like bats and dolphins, rely on ultrasonic vibrations to locate prey and barriers. The ultrasound applications available to humans have expanded significantly in recent years, including in the industrial, chemical, medical, and military areas [82]. Using ultrasonic transmission to gather energy is a relatively recent technique, where "ultrasonic" refers to frequencies above 20,000 Hz, which is the upper limit of what can be heard. Frequencies of 10 Hz and higher are standard in medical diagnostic ultrasound scanning. This procedure does not adversely affect the human body or electronic equipment [83]. The UPT module comprises a transmitter that converts electrical energy into ultrasonic energy and a receiver that converts ultrasonic energy back into electrical energy. The ultrasonic concept is depicted in Figure 6 as a transmitter (Tx) utilizing the piezo-electric effect to create ultrasonic waves from electrical power. The produced wave might be transmitted wirelessly across the tissue, water, and air. Also employing the piezoelectric effect, the receiver (Rx) changes the wave into electrical power. With a relatively low operating frequency and a short wavelength, an ultrasonic WPT device can achieve long-distance power transfer while keeping a relatively small size [84].

Zhu et al. [85] used ultrasonic waves to power implanted biosensors. Ozeri and Shmilovitz [86] have proven that ultrasonic WPT can transmit power with 100 mW at a 40-mm distance. They designed their piezo transducer with a diameter of 15 mm and a thickness of 3 mm for this application. They observed a PTE of 27% running at a frequency of 673 kHz in an environment with transmission loss. Moreover, a rectification efficiency of 88.5% was attained. Chang et al. [87] manufactured a 30.5mm3 implant with bidirectional ultrasonic power and 95 kbps data connections at a depth of 8.5 cm and a frequency of 1MHz. Meng and Kiani [54] constructed an ultrasonic connection for a 1-mm^3 WPT implant. They used a piezoelectric disc transducer to insert a silicon die. Simulations demonstrated a PTE of 2.11% for an RL 2.5 k at 1.8 MHz and 3 cm spacing. To validate the technology, they constructed an ultrasonic link to a 1-mm3 implant on the PCB; it demonstrated 0.65% PTE at simulation and 0.666%. Mahmood et al. [82] presented a way of delivering electricity to a portable heart rate sensor for biomedical applications created with a 40-kHz ultrasonic transducer. System efficiency was 69.4 percent at 4 cm, with an output power of 0.316 mW. 97% heart rate measurement accuracy was achieved compared to the standard equipment. Moreover, this research demonstrates that capacitor charging time increased with distance for various super-capacitor values, notably for a value such as 8 F. Therefore, it is recommended to utilize the UPT with the air gap for 4 cm of generating the appropriate voltage in a short amount of time.

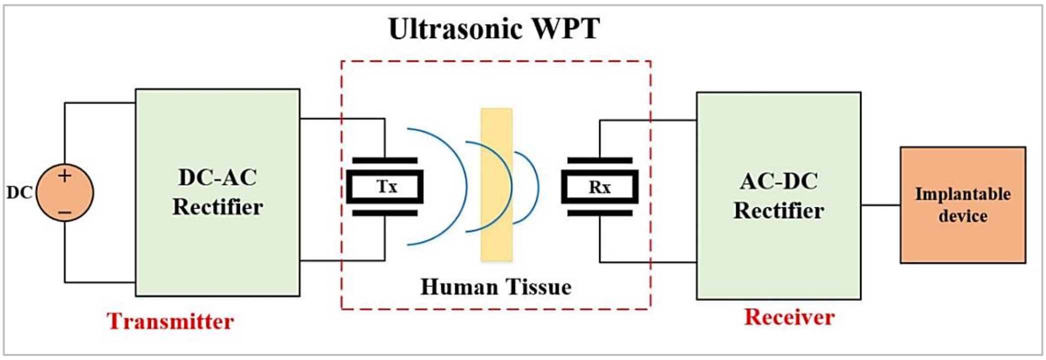

Figure 6: Ultrasonic power transfer system

4. Comparison of wireless power transfer methods of medical devices

The most promising WPT approaches for implanted devices are IPT, CPT, MRC, and UPT. In MRC, energy is transferred within a few meters, and arrangement between a transmitting and receiving coil is not required. MRC has disadvantages, including lower efficiency at greater distances and complex implementation. UPT has exceptional performance qualities for deep implants and is capable of deep penetration and high-input power; it is suitable for medium-shallow implantation depths and enables the simultaneous powering of several devices. CPT functions effectively at short-range implantation depths, allowing integration on flexible substrates without EMI. CPT offers low-cost production flexibility and stability for ultra-small ICs. Misalignments and angular rotations affect it less than UPT. One shortcoming of the capacitive method is that the link's efficiency is frequency-dependent. While increasing the frequency is required for efficient power transfer, it also develops losses and decreases penetration.

Furthermore, recent studies have indicated that resonant transfer techniques can attain high efficiencies at low frequencies. Compared to other methods, the inductively coupled method can send power outputs from a few microwatts to a few watts with high efficiency in the near-field range. IPT has been the subject of significant research for decades and, as a result, is the most extraordinary and often employed WPT technique in the scientific world. Although CPT and UPT are comparatively new approaches, they have already shown promise in creating wireless implanted devices for biomedical applications.

Table 2 highlights previous reviews and covers near-field WPT methods (magnetic resonance, inductive, capacitive, and ultrasonic coupling). It also discusses WPTs' power, frequency, efficiency, distance, and transmitter and receiver geometries. According to the results shown in the table, inductive and capacitive coupling produce greater power than alternative coupling methods; their transfer efficiency is generally considerable but decreases as the distance between the primary and secondary coils increases. The primary and secondary coils must be appropriately aligned to achieve a high efficiency. MRC may accomplish a transmission distance of up to 200 mm and a frequency of up to 400 MHz. This method has a lower transmission efficiency than other techniques (i.e., inductive, capacitive, and ultrasonic coupling). The inductive coupling technique may provide intermediate-range energy delivery to the load. Unfortunately, the efficiency of the transfer is lowered using this approach. UPT is suitable for low-power applications and achieves long-distance power transfer.

Table 2: Comparison of previous research works for wireless power transfer system types

	Power (mW)	Frequency (MHz)	Efficiency (%)	Distance (mm)	Transmitter geometric	Receiver geometric	Ref.
Inductive	68	3.25	67	20	60 T 10	20 T 2	[88]
Power	57–447	13.56	5.7–44.7	20–50	75 × 75	20 × 30	[89]
	N/A	13.56	44.2	3~4	DoutT = 28	DoutR= 8	[90]
					DinT = 5.5	DinR= 5.1	
	0.095	915	0.019	8	N/A	0.25	[91]
	18	13.56	7.7, 11.7	10	DoutT = 30	doutR =10	[92]
	1.3	60	2.4	16	DoutT = 45	DoutR= 1.2	[93]
	N/A	6.78	74.47	15	DoutT = 44	DoutR= 10.5	[94]
					DinT = 7.92	DinR= 6.61	
	1,450	2	27	80	DoutT = 140	DoutR= 65	[95]
	165	13.56	N/A	22	DoutT = 56	DoutR= 11.6	[96]
					DinT = 10	DinR= 5	
Capacitive	53	N/A	2.6	15	80×30	4×40	[97]
Power	30.6	0.2-20	51.9	3	20×20	20×20	[98]
	90	1	70	3	20x20	20x20	[99]
	290	0.21	38.4	5	20×20	20×20	[100]
	108.4	190	66.4	5	40×40	10× 20	[62]
Magnetic	70	13.56	35	6	DoutT = 20	DoutR = 20	[101]
resonance	47.55	433.9	1.21	50	13.3, 16.8	15 ×7 ×6	[102]
	1,015	0.2	N/A	240	DoutT = 110	DoutR= 110	[103]
	26	16.47	0.7	70	DoutT = 220	DoutR= 9	[104]
Ultrasonic	~2	1.8	2.11	30	DoutT= 10.8,	DoutR=1.1, 1.2	[54]
Power					15.9		
	2.48	1.15	0.4	200	20	2×2×2	[105]
	8.7	2.3	1.7		30	2×4×2	
	2.6	0.28	18	18	DoutT = 20	DoutR= 20	[106]
	28	1	1.6	105	29.6 × 72	1 × 5	[107]

T: thickness of coil; N/A: not available; DoutT & DoutR: outer diameter for transmitter and receiver coils; DoutR & DinR: inner diameter for transmitter and receiver coils

5. Data transmission in WPT-based modulation techniques

Critical criteria for selecting the modulation schemes are based on the application’s simplicity, system efficiency, power, and bandwidth. The standard modulation techniques utilized in implanted devices include amplitude shift keying (ASK), frequency shift keying (FSK), and phase shift keying (PSK). A common argument for continuous amplitude modulation is its requirement for constant power flow due to the impossibility of on-chip energy storage. While PSK and FSK and their derivatives offer fast data speeds and excellent noise immunity, they are cumbersome to set up and use much power [108]. ASK is the most distinctive approach because it is easy to operate and has a low power consumption for the demodulation process. The ASK system, on the other hand, is more likely to be affected by carrier amplitude disturbances, such as interference and changes in coupling [109]. Increased resistance to these disturbances can be achieved by deploying ASK with an amplitude modulation index of 100% when one of the amplitudes in ASK is zero volts, the resulting modulation is known as on-off keying (OOK) [110].

Additionally, in ASK with pulse-width modulation (ASK-PWM) or with pulse-position modulation (ASK–PPM), data values can be stored in duration (rather than amplitude) [111]. This is considered to be the simplest and most energy-efficient modulation method. Many designers, therefore, prefer PSK or FSK modulation over amplitude modulation. Hannan et al. [112] studied these techniques and found that ASK modulation is the most suitable technique for implantable biomedical devices. Which modulation technique is chosen is highly dependent on the application's requirements. PSK modulation and its variants (quadrature, differential, and quasi-coherent) are used in systems that require constant power transfer, high data speeds, and better noise protection. However, these systems are complicated to implement and consume excessive energy [113]. FSK modulation offers similar advantages and disadvantages to PSK but needs a wider bandwidth to transmit both modulated frequencies [114]. Mazzilli and Dehollain [115] describe an OOK-ASK demodulator. Using OOK-ASK modulation, with 4 and 10% modulation index, a data rate equal to 50 kb/s and a carrier frequency f_o = 1 MHz. The input power to the analogue front end is identical to −55 dBm, and the gain is 45 dB. Using 1.5 V and 184 W, a low-power OOK-ASK demodulator was created for biomedical implants. Trigui et al. [109] proposed carrier-width modulation and quad-level CWM. Off-widths and modulation timing are minimal. They were done around the zero crossing for a primary coil current. This allows effective PTE and high modulation bandwidth. The modulated signal has significant amplitude resistance since the carrier is time-modulated. Due to its high data rate and bandwidth, Lu and Sawan [116] used offset quadrature PSK modulation. The system was simple to build, used little power, took up little space, and sent data at 8 Mbps and 13.56 MHz. The modulator and demodulator both used 1.8 V and 16 W of electricity.

Cirmirakis et al. [117] have created a passive phase shift keying (PPSK) modulator integrated into an IMD for uplink communication. The modulator is designed to use a single pair of coils to transmit both power and data. At 13.56 MHz carrier frequency, the data link can achieve data speeds of up to 847.5 kbps (1/16 of the carrier). The fastest data rate (four times quicker) is obtained by a wireless link utilized simultaneously for power delivery and uplink communication in IMDs. Gozalpour and Yavari [118] used a biomedical inductive link for frequency-shift keying (FSK) modulation, suitable for biomedical implants with a winding diameter of 3.22 mm. The secondary side has two resonance frequencies, improving efficiency. The power delivered to load and power transmission efficiency are 57.4 mW and 48.4%, respectively, with a measured bit error rate of less than 10^{-5}. Table 3 summarizes the research on WPT-based modulation techniques. Also highlighted are the modulation method, frequency, data rate, and maximum power delivered to load (PDL). As shown in Table 3, both ASK and FSK are examples of effortless data transfer approaches. ASK modulation is the most commonly used in biomedical applications because it is simple and easy to use; it has a frequency range of 1–13.6 MHz and a maximum PDL range of 2–100 mW.

Table 3: Modulation techniques based on WPT

Modulation scheme	fc (MHz)	Data rate (Mbit/s)	Maximum PDL (mW)	Ref.
CWM	10	1.66	29.66	[109]
ASK	13.56	0.15	9.2	[119]
ASK-FSK	1	0.166	90	[99]
PWM OOK	6.78	0.02	520	[120]
ASK	10	2	10	[121]
PWM ASK	5	0.168	13	[122]
ASK	13.56	N/A	2	[123]
FSK	0.2	0.00976	125	[124]
PSK	13.56	4.16	N/A	[125]
ASK	2.5	0.128	100	[126]
OOK	8	0.000125	7	[116]

6. Biomedical implantable device-based wireless power transfer

Various health-related networking and wireless communication technologies have been developed recently, ranging from low-power medical radios that capture body energy to wireless sensor networks for in-home monitoring and diagnostics [127]. Today, such wireless systems are integral features of many modern medical devices. Wireless communication is particularly prevalent in IMDs [128], including pacemakers, cardiac defibrillators, insulin pumps, and neurostimulators [129]. Wireless connectivity makes it possible to monitor patients' vital signs from afar and treat patients quickly, which allows the healthcare system to run more smoothly [130]. Medical implants are designed to replace missing biological structures, sustain damaged ones, or augment existing ones [131]. In contrast to transplants, transplanted biomedical tissues and medical implants are manufactured devices [132]. The implant surfaces that come into contact with the body may comprise biomedical materials such as titanium, silicone, or apatite. Certain implants, such as artificial pacemakers and cochlear implants, contain electronics. Certain implants are bioactive, such as implanted tablets or drug-eluting stents. Implants can be approximately classified according to their intended use [133]. Adding wireless and networking to IMDs has numerous advantages. It allows continuous monitoring of the patient's physiological data and other symptomatology acquired by the device, reducing the time required to track medical problems regularly and interfering with the patient's everyday activities. Improved oversight and management of IMD operations enables speedier resolution of issues and the implementation of corrective actions. The two preceding criteria indicate that overall patient monitoring and IMD operation costs will drop [134].

Implanted devices are categorized as passive or active based on whether they require electricity. Passive implants, such as simple screws or artificial valves, are utilized for support or mobility. By contrast, active implants, such as neurological or cardiac implants, replace organ functions or treat associated disorders. The power needs of most active implants range from a few microwatts to tens of milliwatts, depending on the application [135, 136]. The required power level is application-dependent and might be very low (0.1 mW for pacemakers) or very high (> 100 mW for neurostimulators and medication pumps) [137].

6.1 Brain optoelectronic implant

Optogenetics is a medical technique that uses lasers or µLED to turn on genetically changed neurons in the brain. The most challenging part of making an optoelectronic implant is ensuring that the µLED receives power. Biswas et al. [138] proposed a new method based on a WPT module. Using the FR4 substrate and Cu trace as a prototype, a miniature 10.5 mm x 10.5 mm receiver coil was manufactured. The power was sent to the implant at a frequency of 7.2 MHz, using inductive power transmission with magnetic coupling and at a depth of 5 mm. The results were acceptable, but the performance must be enhanced by optimizing the size of the receiver coil and implementing it on a biocompatible, flexible substrate. Figures 7(a–e) show several designs of flexible wireless optoelectronic implants for optogenetic devices.

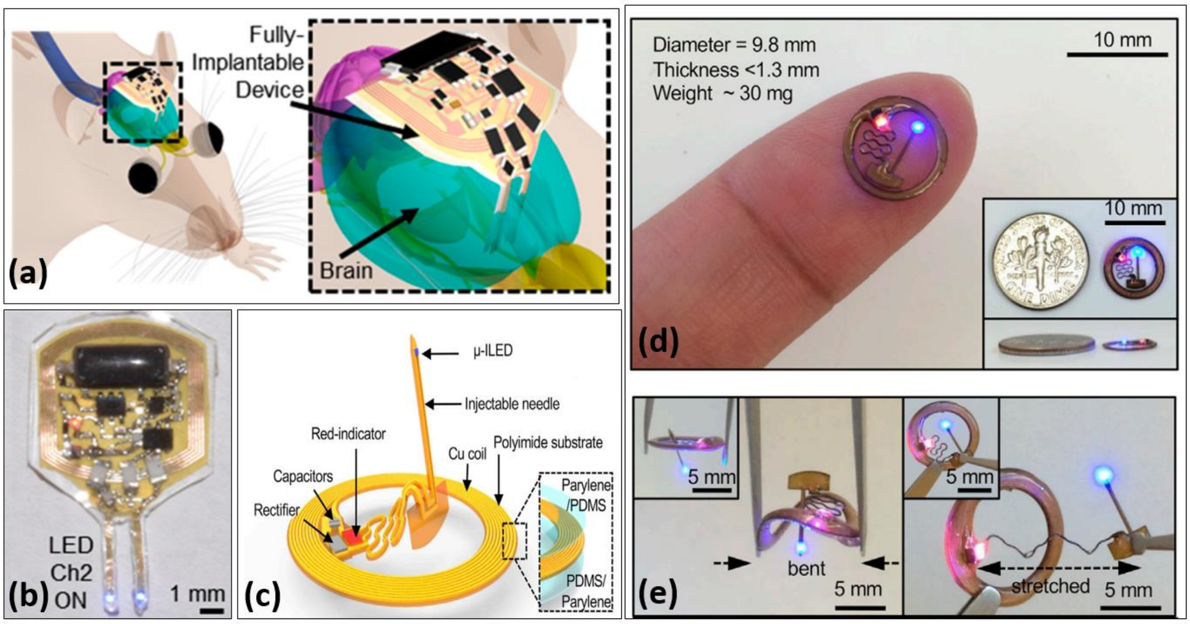

Figure 7: A wireless power transfer design of optoelectronic brain implants (a) the device location relative to the brain [139] (b) dual-channel wireless operation [139], and (c,d,e) Thin, Flexible Wireless Optoelectronic Implants for Optogenetic Experiments [140]

Samineni et al. [141] designed a µLED device with a thin, flexible form (overall dimensions length 10 mm, width 5 mm, thickness 0.2 mm) and a rectangular coil for wireless power harvesting comprised of seven planar loops with a 50 mm pitch. As shown in Figure 8, an updated device possesses the same features and dimensions (11.9 x 5.07 x 0.15 mm, copper traces, seven turns, 60 m width, 50 m spacing) [142]. Modifications include rounded edges and mass-production-optimized printed circuit boards.

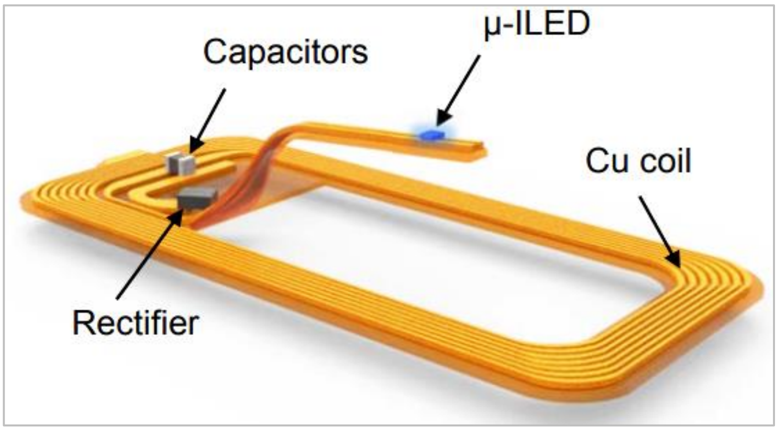

Figure 8: WPT for a flexible optoelectronic system in battery-free wireless optoelectronic devices [141]

6.2 Implantable cardiac pacemakers

Implantable cardiac pacemakers are small medical devices that are inserted beneath the chest. They are critical in extending the lives of people suffering from cardiovascular illnesses. They deliver electrical impulses to the heart if it deviates from its typical, regular beating, as in bradycardia, in which the heart beats too slowly [143]. The batteries in today's pacemakers typically last 7–13 years. However, this was not always the case. Ruben–Mallory zinc mercuric oxide batteries were initially employed, and entire units were encased in epoxy. Many experiments were conducted with various power sources; in the 1970s, lithium-compound batteries were discovered to be the most suitable and least harmful in the long term [144].

Xiao et al. [145] proposed a pacemaker WPT electromagnetic model. The pacemaker’s charging system successfully charged the lithium-ion battery from 3.98 V (80% residual capacity) to 4.2 V within 30 minutes when powered by a 5.66 V/300 kHz power source via a 4-mm subcutaneously implanted coil of 0.15 mm thickness. Regarding charging system safety, electromagnetic and thermal simulation findings suggested that the maximum SAR and temperature increases in tissues were 36.8 W/kg and 0.66 °C, respectively. These findings offer theoretical and practical support for developing a wireless charging system. Ko and Feng [146] demonstrated the feasibility of using an implantable, wireless, two-channel electromyography (EMG) recording system for three months in a rodent. The system consists of an implanted device and an external reading module. The implanted device consists of a multichannel brain amplifier chip linked to inductive power and data-transfer circuitry. The external reader module includes a transcutaneous power supply, electronics for EMG data recovery, and a USB-to-computer interface for display and storage. The implanted device was attached to stainless-steel EMG electrodes and then enclosed in biocompatible, FDA-compliant silicone.

Ho et al. [147] optimized the field pattern by incorporating a transmitter with dynamic focusing capacity. The transmitter was designed to power a tiny pacemaker or cortical implant (2–3.5 mm) in the midfield (5 cm spacing). The field was concentrated at several locations using phase control of the antenna feeds. The focus was changed for the experiment using an optical indication of received power as feedback. The study did not discuss using back telemetry to alter the emphasis practically. A newly designed inductive WPT system was proposed by Cruciani et al. [148] to recharge the pacemaker's battery. This research design a coils and pacemakers of the WPT system (parallel, planar, stacked coils) that are separated by a distance d. The primary coil measures 22 mm externally and 18 mm inside. The secondary coil's dimensions are 30 mm Dout by 12 mm Din. The primary coil is external, while the secondary coil is implanted within the pacemaker case. Coils had 0.49 mutual inductance. Using a magnetic shield improved power transfer while lowering magnetic field maximums. Campi et al. [149] utilized the electromagnetic field safety of a WPT system based on MRC between two coils to recharge pacemaker batteries at two different frequencies (300 kHz and 13.56 MHz). This study tested several WPT topologies and determined that the SP configuration at 300 kHz and the SS configuration at 13.56 MHz provide the best WPT capacitance compensations. The safety of electromagnetic fields has been examined by numerical dosimetry studies using anatomically accurate human body models, which have found no unique issues with this use. Figures 9a–c show an example of a microtubular pacemaker used for wireless cardiac electrotherapy.

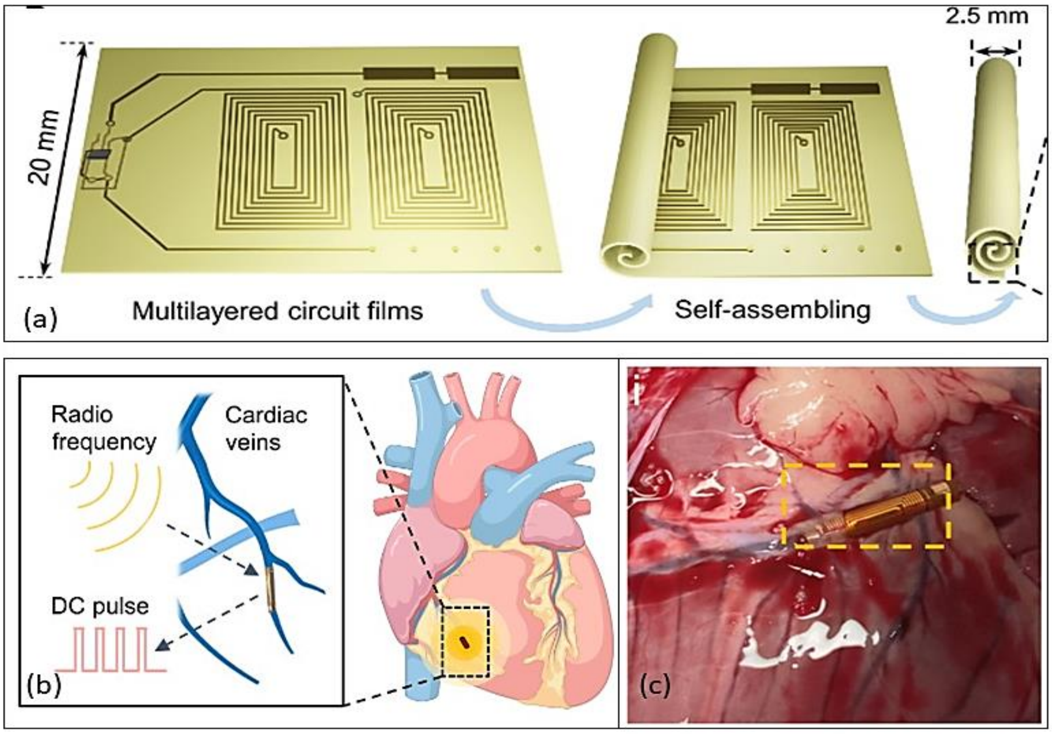

Figure 9: Pacemaker with WPT coils (a) the microtubular pacemaker’s schematic design uses the self-adhesiveness of the polymer layer (b) the implantation of the microtubular pacemaker to the cardiac anterior vein (c) Implantation of the microtubular pacemaker in animal body [150]

Mahmood et al. [151] presented a new way of designing and implementing a wireless charging system for cardiac pacemakers based on a series-parallel spiderweb coil, as shown in Figure 10 (a-d). In this study, the goal voltage was 5 V, which is sufficient to charge a pacemaker. Three source voltages were tested: 5, 15, and 25 V. Six operational frequencies ranging from 1.78–6.78 MHz were utilized to evaluate the design. The optimal results were obtained at 1.78 MHz. At a 10 mm air gap, the power transfer efficiencies for 5 V and 15 V were 95.75% and 92.08%, respectively.

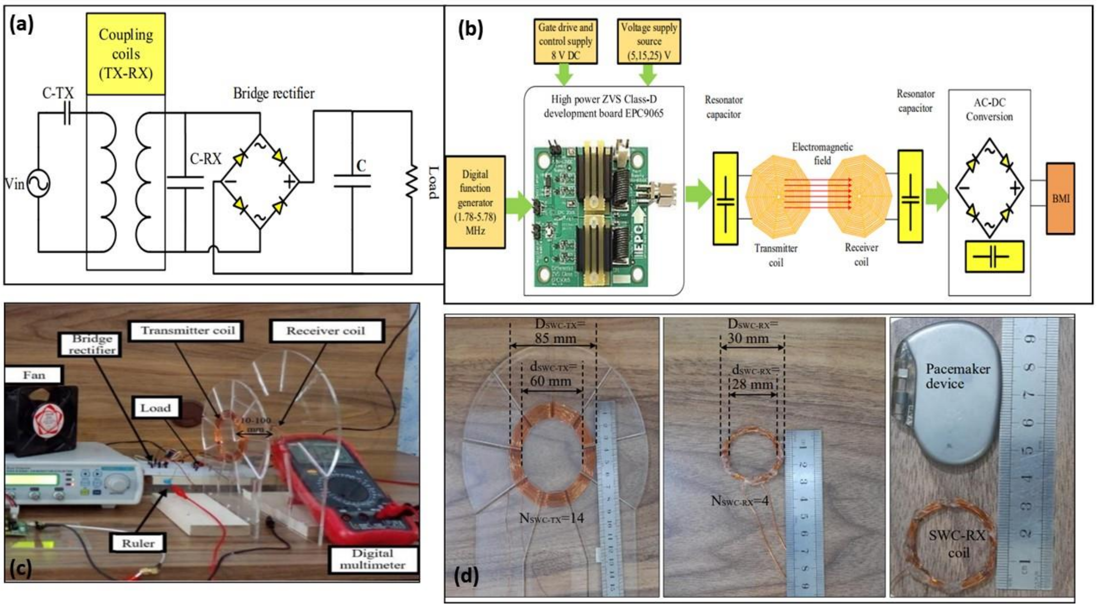

Figure 10: Cardiac pacemakers with a series-parallel spider web coil (a) Model for coupling coils of wireless power transfer circuits (b) Overall system design with a class D operational amplifier (c) Experimental setup for measuring WPT system efficiency (d) Transmitter and Receiver coil dimensions [151]

6.3 Biomedical capsule endoscopy

Wireless capsule endoscopy (WCE) is a significant advance in gastrointestinal (GI) tract diagnosis because it is painless and can visualize the whole digestive tract. Currently, endoscopic capsules are powered by two coin-shaped batteries that deliver 25mW of power for 6–8 hours [152]. WPT is a viable technique for extracorporeal WCE powering because it provides more readily available and sustained energy. With the development of capsule endoscopy technology, Patients now have access to a procedure that can be used in place of endoscopy, gastroscopy, and colonoscopy of the small intestine, all of which have historically been uncomfortable because they use flexible cables with large diameters [153].

Na et al. [104] improved a technique to determine the optimal performance of the biomedical capsule endoscopy MR-WPT system. A receiver with a 9 mm for a 16.47 MHz resonance frequency was used to implement a sufficiently small system to fit inside the capsule endoscope already in use. A receiver and the load coils experienced a depth of 7 cm inside a tissue block with a low-efficiency degradation of about 0.380 dB and transferred energy of 300 mW to the load. A proposed system was observed to have a meagre SAR of 1.74 W/kg, indicating that it is safe for human use. As shown in Figure 11(a and b), a sub-gigahertz (GHz) WPT technology for capsule endoscopes designed in [102] achieved results at a distance of 5 cm, with an operating frequency of 433.9 MHz, and gained a 1.21% PTE through duck intestine tissue and multilayered porcine tissues, corresponding to 47.55 mW of PDL at the SAR, with simulations indicating a SAR of less than 2.54 W/kg.

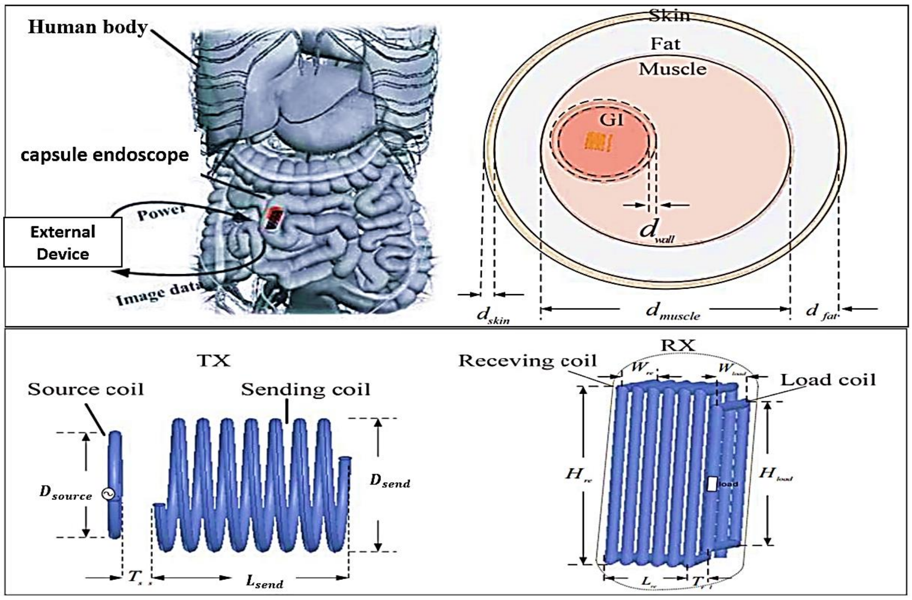

Figure 11: A power transfer system for an implanted device in a human body at a sub-GHz frequency (a) A wireless capsule endoscope inside the human body, (b) Geometry of transmitter and receiver coils [102]

Basar et al. [154] developed a new wearable WPT system for WCE applications.the researcher design a 3-coil inductive connection by a small wearable power-transmitting coil (PTC-I) and a 2–3-dimensional power-receiving coil to increase a PTE. The researchers used a new power-transmission-coil III (PTC-III) configuration and power-combining technology on the receiving side to improve stability. A tested result confirmed that the suggested PTC-I-based system transferred 758 mW of power with received power stability (RPS) of 68.7% and a PTE of over 8%. However, the proposed PTC-III-based system achieved an overall RPS of 79.2%. Its PTE was 5.4%, meaning 570 mW of the power was transferred. The efficiency of the new design is still far superior to previous designs.Figure 12 shows several types of examples of capsule endoscopy wireless systems.

Table 4 provides a comparison of the three different categories of biomedical implantable applications discussed in sections (6.1) through (6.3) and details the differences between them in terms of the type of WPT, power consumption (mW), efficiency (%), transfer distance (mm), and size of the implanted device (mm).

Table 4: Implantable device applications with their specific parameters

Implantable device	**Type of WPT**	**Power consumption (mW)**	**Efficiency (%)**	**Transfer distance (mm)**	**Size of the implanted device (mm)**	**Ref.**
Brain optoelectronic implant	Inductive and magnetic resonance coupling	0.0138	3.16	5	10 x 10.5	[138]
	Inductive coupling	12000	NA	80	0.22 × 0.27	[141]
Implantable cardiac pacemakers	magnetic resonance coupling	NA	NA	12	31.4 × 47.4 × 8.1	[145]
	midfield powering	0.01	NA	50	Din= 2	[147]
	Inductive coupling	1500	NA	10	30 × 12	[148]
	Magnetic resonance coupling	1000	95	10	17.65	[149]
	Magnetic resonance coupling	6.72×10^{-6}	95.75	10	Dout = 30 Din= 28	[155]
Biomedical capsule endoscopy	Magnetic resonance coupling	26	NA	70	Dout =9	[104]
	Magnetic resonance coupling	47.55	1.21	50	15.9 × 7.7	[102]
	Magnetic resonance coupling	758 570	8.21 5.40	35	L= 27.9 d=13	[154]

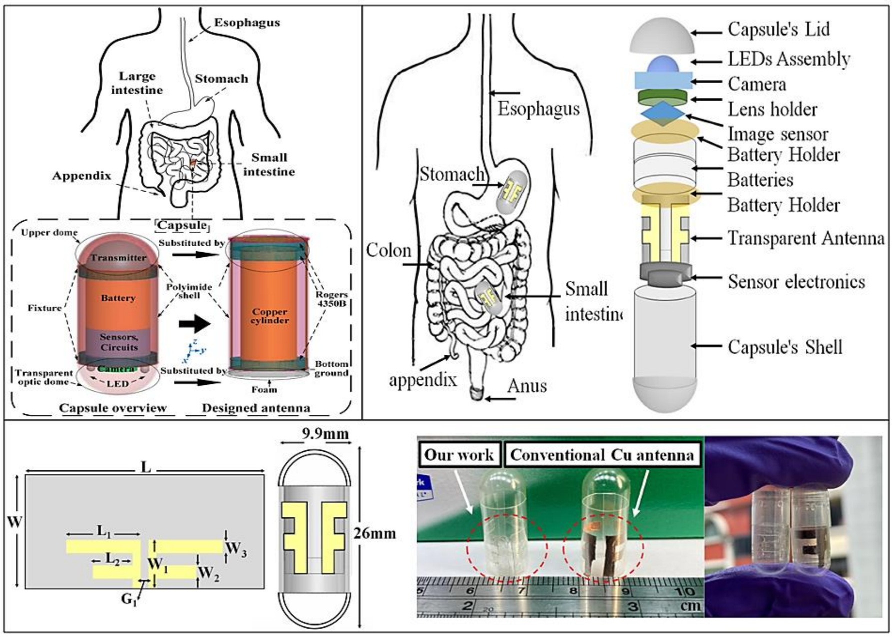

Figure 12: Designs for a wireless power transfer system for capsule endoscopy [156,157]

7. Specific absorption rate effect on tissue

specific absorption rate (SAR) is the primary indicator of tissue safety in WPT applications for MIDs [158]. This standard explains the quantity of radio frequency energy the structure absorbs using radio system devices. The SAR must be less than 2 W/kg for 10 grams of tissue, as the IEEE standard identifies [159]. SAR is the time derivative of the electromagnetic energy absorbed or consumed by the mass equal to one kilogram included in the human body scale. It can be computed using Equation (15):

$$\mathrm{SAR}=\frac{d}{dt}\left[\frac{dW}{dm}\right]=\frac{d}{dt}\left[\frac{dW}{\rho\,.\mathrm{dv}}\right] \quad (15)$$

where *dW* is defined as the time derivative of the incremental energy, *dm* is a cumulative mass, *dV* is a volume element, ρ is a mass density. SAR is often estimated for the whole body or a small piece (usually 1 g or 10 g of tissue), and it calculates the radio wave effect [160].

Internationally, scientific, industrial, and medical frequency channels are allocated for WPT system technology used in MIDs. The lower kilohertz range (6.78, 13.56, and 27.12 kHz), the lower MHz range (6.78, 13.56, and 27.12 MHz), the higher MHz range (433.9 and 915 MHz), and the GHz range (2.45, 5.8, and 24.125 GHz) are included in that groups. Patient tissue safety is a critical consideration in designing WPT for MIDs. Tissue safety is primarily determined by the amount of electromagnetic field absorbed by the human body [161]. Dielectric permittivity is a constant and actual value at sufficiently low frequencies. Because low-frequency waves penetrate deep in the tissue, which has limited absorption at these frequencies, and because deep portions of the body have fewer nerve endings, the impacts of the radio-frequency waves (and the resulting harm) may not be felt immediately by the patient. With increasing frequencies, human tissue's relative permittivity and conductivity drop and rise, resulting in increased tissue absorption. Microwaves penetrate the body more slowly and heat the tissue more quickly [162]. However, it is necessary to consider low frequencies to avoid a significant increase in SAR. As a result, one of the most difficult challenges to overcome in achieving a higher-quality factor for an implant coil to produce satisfying PTE. In [163], The authors demonstrated that the greatest measured peak gain is 26.54 dBi. In addition, the specific absorption rate (SAR) values for 1 g and 10 g were calculated and met safety standards. Shah et al. [164] evaluated the effects of implants on the SAR in a head model positioned 30 mm above the WPT system. Even small implants significantly impacted the SAR distribution and the local peak and mass-averaged SAR values. Compliance with international safety limits was investigated, and it was determined that the maximum allowable transmit power (MATP) for the designed WPT system was 43 W and 6.8 W in the presence of the skull plate.

Saidi et al. [165] , the suggested antennas' specific absorption rates are computed using the layer model. The input power levels for the two MTM resonators (E-shape and E-interdigital) are 1.5 mW for HFSS and 2.5 mW for CST. For the MTM E-shape resonators, the SAR values are 1.51 W/kg and 0.16 W/kg in HFSS and 1.55 W/kg and 0.15 W/Kg in CST for 1 and 10 g, respectively. The values for the MTM E-interdigital are 1.13 W/kg and 0.12 W/Kg in HFSS and 1.21 W/Kg and 0.12 W/Kg in CST, averaged over 1 g and 10 g, respectively. In [166], the researchers used miniaturized and wirelessly powered pacemakers to perform leadless biventricular (BiV) pacing. Two pacemakers with flexible form factors were inserted epicardially on a porcine model's right and left ventricles and were inductively powered at 13.56 MHz and 40.68 MHz, respectively. An external power source of 0.3 W and 0.8 W at frequencies of 13.56 MHz and 40.68 MHz yields SARs significantly below the safety regulatory limit by 2-3 orders of magnitude.

8. Challenges and solutions for WPT

Ensuring simultaneous WPT power in biomedical implants is challenging. It is a challenge to simultaneously ensure high data rates, lower power consumption, constant transfer power, high strength against noise, and ease of implementation. Therefore, there are some solutions to challenges in implanted medical devices.

1) To ensure proper protection for implant users, particular norms for current, E-filed, and SAR evaluations in the human body are required. Combining data with geometric models and experimental tests is best for getting the most accurate results [167, 168].
2) Eddy current is the primary contributor to electrical fields in tissue, and SAR can be used to evaluate these fields. Conductivity and permittivity properties should be prioritized for tissue deemed nonmagnetic. Tissue absorption and electric field distribution vary according to quality, impacting SAR. Because the human body is so complex, various methods exist to build and make in vitro models [169].
3) Adverse health implications of implanted WPTs are characterized as nonthermal or thermal. Exposure to an electromagnetic field of less than 10 MHz causes nonthermal effects. Thermal consequences caused by tissue heat absorption are explicitly constrained. Current laws generally restrict active implantable medical devices to a temperature increase of less than 1 degree Celsius above average body temperature [170].
4) Regarding implanted devices, the packing material is critical for biosafety. For WPT, the desired metals may have harmful consequences after being implanted. The system must be encapsulated in a biocompatible material to avoid these undesirable consequences. In general, encapsulating materials such as polydimethylsiloxane, NuSil, and others have been chosen under the original design of implanted biomedical devices. Different encapsulations, on the other hand, would affect PTE [171].
5) For highly developed medical devices like pacemakers and cochlear implants, it is essential to have a high level of safety and dependability.

6) The overall PTE for the movable implants currently in development is still relatively low despite various coil combinations being proposed to improve misalignment tolerance. It is possible to request the positioning strategy to improve PTE further.

9. Conclusion

This article aimed to provide an overview and discussion of the use of WPT in implantable biomedical devices, emphasizing the parameters that influence WPT output. The paper examines the various WPT systems proposed in the literature, emphasizing crucial performance characteristics for medical implants. Considerations include the implant location, the transmission medium, and safety warrants; it investigated their qualities and influences on WPT system parameters, such as efficiency, input power, and transfer distance. Safety evaluations are required before WPT can be used in devices such as cochlear implants and pacemakers. When designing devices, notably endoscopic and drug administration systems, it is necessary to optimize coil design to maximize load fluctuation tolerance. In this study, the power provided to medical implants in the examined works ranged from a few µW to 12 W, with distance ranges varying from 3 mm to 240 mm and a maximum efficiency of up to 95%. The most promising approaches are inductive and capacitive coupling, which produce greater power than alternative coupling methods; their transfer efficiency is generally considerable but decreases as the distance between the primary and secondary coils increases. As for modulation techniques, ASK modulation is the most commonly used in biomedical applications because it is simple and easy to use; it has a frequency range of 1–13.6 MHz and a maximum PDL range of 2–100 mW.

Future research in the field may concentrate on proposing new systems or techniques and improving the efficiency, reliability, and security of communication and power transfer systems for biomedical implants. This could involve developing sophisticated signal processing methods to mitigate interference and enhance data transfer rates, optimizing coil designs to maximize power transfer efficiency, and exploring novel materials for implantable devices to improve biocompatibility and longevity. Furthermore, it would be beneficial to evaluate the performance of various topologies and determine the most efficient and secure topology in biomedical applications. Any proposed technique or design should also adhere to a specific safety standard, maintaining a SAR of less than 2 W/kg for 10 grams of tissue.

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Author contributions

Conceptualization, S. Mutashar, S. Gharghan, and H. Abduljaleel; methodology, S. Mutashar.; validation, S. Gharghan.; H. Abduljaleel and S. Mutashar.; investigation, S. Gharghan.; resources, S. Mutashar.; data curation, S. Gharghan; writing—original draft preparation, H. Abduljaleel; writing—review and editing, S. Gharghan.; H. Abduljaleel and S. Mutashar.; supervision, S. Gharghan. and S. Mutashar; project administration, S. Gharghan.All authors have read and agreed to the published version of the manuscript.

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Data availability statement

The data that support the findings of this study are available on request from the corresponding author.

Conflicts of interest

The authors declare that there is no conflict of interest.

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