**Abstract:** With ever-increasing concerns on health and environmental safety, there is a fast-growing interest in new technologies for medical devices and applications. Particularly, wireless power transfer (WPT) technology provides reliable and convenient power charging for implant medical devices without additional surgery. For those WPT medical systems, the width of the human body restricts the charging distance, while the specific absorption rate (SAR) standard limits the intensity of the electromagnetic field. In order to develop a high-efficient charging strategy for medical implants, the key factors of transmission distance, coil structure, resonant frequency, etc. are paid special attention. In this paper, a comprehensive overview of near-field WPT technologies in medical devices is presented and discussed. Also, future development is discussed for the prediction of different devices when embedded in various locations of the human body. Moreover, the key issues including power transfer efficiency and output power are addressed and analyzed. All concerning characteristics of WPT links for medical usage are elaborated and discussed. Thus, this review provides an in-depth investigation and the whole map for WPT technologies applied in medical applications.

**Keywords:** Medical devices; implantable devices; magnetic resonance; wireless power transfer; nerve stimulation; pacemaker; endoscopy; drug delivery; wireless charging

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**1. Introduction**

With the increasing consciousness in medical diagnostic and monitoring systems, research in the medical engineering field has gradually focused on healthcare systems. In this area, implantable medical sensors and devices play a vital role in ensuring continuous monitoring online for different diseases, which can also significantly reduce cure cost [1,2]. A healthcare system overview [3] also predicted that the range of implantable biomedical devices, such as neural stimulators, drug delivery systems, detectors, monitors, etc., will have a strong correlation with the improvement of microsystem technologies. However, the challenges of developing healthcare system are obvious. First, the lead linkage of an on-board battery is hard to predict or detect after implanting. Moreover, the regular replacement of implanted batteries is adopted to extend the duration of the system, which increases the danger to patients. These two scenarios require replacement surgeries, which could lead to infection or direct damage to nearby human organs. In order to improve system reliability and reduce the danger of surgery, some research work is attempting to use longer-lasting batteries with certified safety standards [4], which increases the size and weight of implanted devices. However, miniaturization is another focus for the future healthcare system that restricts the application of large batteries. In order to solve the abovementioned problems, wireless power transfer (WPT) technology has been proposed by different researchers over the last 10 years, which guarantees the rechargeable characteristics of implantable devices with reasonable sizes and without surgeries.
WPT is a technology that transmits power through the electromagnetic field. Regarding the transmission distance, this technology is categorized into far-field and near-field WPT [5]. The difference among these methods is normal based on the transmission distance and frequency. Particularly, implantable biomedical devices mainly focus on near-field coupling power transmission, which can guarantee the acceptable efficiency and power level. In fact, the closeness of the two coils ensures a high power transfer efficiency with an immense usage of the excited magnetic flux [6]. Normally, due to the increase of electromagnetic energy absorption of the human body, WPT for medical devices operates at a recommended range between kHz to MHz [7]. The frequency band could be further narrowed with concerns that a possible impedance of functions of coils. As the frequency is fixed, the distance between the energy transfer coils needs to be relatively small. Therefore, near-field WPT is widely adopted to support implanted biomedical devices, while inductive coupling is the most popular option with an overall high transmission efficiency [8].

Aligned with these aforementioned interests, some papers analyze previous development paths and future possibilities of WPT technologies in implantable devices. For electrical and electronic applications [6,8,9], the majority of studies have classified appliances based on the differences among the categories of the WPT adopted. Although this method is reasonable, it is not suitable for medical devices. In fact, there is no standard or orthodox means to determine the methods of WPT technologies used for specific implanted devices. Moreover, the rapid development of this field requires a regular update for current and future practical markets.

In order to clarify various applications of WPT in biomedical implants and to further evaluate their future potentials, this review paper investigates the key performances such as power transfer capability, power level, efficiency, safety requirements, etc. In fact, this study mainly reviews the characteristics, features, operation principles, control strategies, advantages, and disadvantages of WPT for medical applications, the classification of which can be aligned with the categories of biomedical devices to achieve a comprehensive analysis of previous works. The current thresholds and future opportunities are also summarized at the end of each sub-section. Thus, the challenges and foresight for WPT in medical devices are revealed.

This paper is organized into seven parts. Sections 2–4 mainly discuss WPT technologies in different medical applications. Section 5 summarizes safety issues and regulations, while Section 6 recapitulates different perspectives on WPT in various biomedical devices. Finally, a conclusion is drawn.

2. Nerve Stimulation

Nerve stimulation is an effective method to alleviate the negative impacts of neurodegeneration. According to nerve categories, nerve stimulation can be divided into three types—brain stimulation, spinal spur, and peripheral stimulation [10]. All three therapies take similar physical principles to achieve stimulus, while WPT and wireless telecommunication are adopted to maintain a relatively long duration and precise control. The details of WPT for nerve stimulation are investigated in-depth in the following subsections.

2.1. Deep Brain Stimulation

Deep brain stimulation is a therapy that aims to ameliorate the impacts of neurodegeneration, which can cause neuron death and a series of diseases, such as Alzheimer’s, Parkinson’s, and amyotrophic lateral sclerosis [11]. Also, focal brain stimulation, which is categorized under brain–machine interface (BMI), is renowned as one form of neuro-stimulation. This is achieved by providing the spur with an electrical signal [12]. Currently, there are two methods of stimulation—conventional electrical stimulation and optogenetic stimulation [13]. Although their biochemical processes are different, both systems require an initial electricity input and follow electrical waves or light illumination. Based on shared demand, it can be concluded that the development of BMI requires a method to extend the lifetime of the system while maintaining the sensitivity of the brain.
One study [14] developed an inductive coupling WPT system for brain implants with minimized mismatch loss, which had a relatively small size of the receiving coil. This design provided a rechargeable implant that could increase the frequency of wireless telemetry with power guarantee. The primary coil obtained a circular feature with an inner diameter (D) of 12 mm and a trace width (w) of 3 mm. There was also a differential feed gap (G) of 3.175 mm for laminating on a two-layer FR4 substrate. For the secondary coil, the shape was similar to a cube with a length (L) of 1 mm, with an inner gap (V) of 0.8 mm and a differential feeding gap (GL) of 0.9 mm, as shown in Figure 1. The whole system had an operating power gain of 0.1% with a transmitting distance of 15 mm (Ad). The in vitro test was held in a developed cerebral spinal fluid (CSF) phantom with an optimal frequency of 400 MHz, which belonged to the Medical Device Radiocommunications (MedRadio) Service core band for general diagnostic and therapeutic applications [15].

![Diagram of deep brain stimulator](image)

**Figure 1.** Design of deep brain stimulator. (a) Transmitter; (b) general view; (c) receiver.

In addition, other related designs of recent deep brain implants are given in Table 1, which summarizes the current development trends and opportunities. For the wireless power link itself, the crucial parameters are the power transfer efficiency (PTE) and output power. These two criteria illustrate the sufficiency of power supply, which is the cornerstone of the feasibility of the system. When applying WPT in implants, safety issues including radiation and heating effects are also essential. However, not all studies of evaluations include these two parameters, especially heating impact. Here, only specific absorption rate (SAR) is summarized as the factor for safety assessment. In addition, it should be noted that the other parameters, such as size, weight, etc., might be the key parameters to evaluate the system in practical situations. However, this paper mainly uses PTE and output power as the key parameters due to easy evaluation. Only a few studies have discussed SAR impacts due to the lack of in vivo testing and human body simulation. There is therefore no detailed summary of safety issues in this section. In addition, the encapsulation material and media simulation are also discussed here to demonstrate the current safety concerns and mitigation strategies. According to Table 1, the PTE for deep implants is relatively low but the output power sufficiently supplements the implanted battery, which is generally designed with a relative size [16,17].

Actually, there are two studies that show a high PTE, and their testing scenarios include the air as the transmitting media. In fact, the power loss caused by electronics devices in small implants is quite significant, which greatly leads to the large difference for different transfer systems. Moreover, the WPT design, the distance involved, the coils used, the material of the implant, the size, the power level, the transfer frequency, etc. all greatly affect the PTE of the prototype. Additionally, the power requested in a regular period of stimulus is lower than the power input. Furthermore, in vivo or in vitro tests are adopted by most of the researchers to enhance the reliability of safety evaluation.
Based on the abovementioned parameters, WPT in deep brain therapy obtains great research potential.

(1) Convenience enhancement: Analyzing deep brain activities can be easier than before. With two transmission strategies, namely a wearable energy supplier or a recharge cage, it can allow the object to move freely during the charging period.

(2) Potential for full implantation: Fully embedded devices have become the mainstream of the brain research field. Due to the difficulties in plug-in charging and minimization, WPT offers an efficient alternative for practical applications.

(3) Possibility for stretchable implants: Flexibility of coil structure ensures the possibility of applying the stretchable optogenetics system [18].

However, there are also several limitations of WPT applications in deep brain therapy.

(1) Minimization difficulty: In order to protect other parts of the brain, the overall size of the implants needs to be small enough. Aside from difficulties in minimizing the functional parts of devices, materials and techniques for downsizing powering sections are still under development.

(2) Hazardous impacts: Exposure under megahertz may lead to unpredictable healthy issues, such as impacts on genes, which would lead to a decrease in fertility [19–22].

(3) Complicated transmitting gap characteristics: It is difficult to align the results of simulations and in vitro and in vivo tests, since the characteristics of biological tissues are hard to determine, especially when these characteristics vary between individuals.

The future development is also draw as below:

(1) Development of new materials and energy storage techniques will accelerate WPT application in implants: Nano super-capacitors and GaN-based capacitors can be used to store energy from a harvesting source, such as piezoelectric nanowires [23].

(2) Combination of wireless communication and power transfer: Brain implants generally aim for activity analysis, and the combination of two main functions could downsize these devices.

(3) Some new topologies of WPT can be adopted in deep brain therapy, which could significantly improve the PTE and the output power, as well as the therapy effect.

2.2. Peripheral Stimulation

The principle of peripheral stimulation is similar to the cortical spur, while it focuses on recovering the dysfunctional limbs and sensors. This section classifies the peripheral stimulation into three parts, based on the nerve category.

2.2.1. Vagus Nerve Stimulation

General peripheral nerve implants refer to the devices that rebuild the dysfunctional motor and sensory functions in the limbs with patterned stimulation [6]. This kind of application requires precise data collection and long-term operation in order to avoid further damage to nerves and muscles [24]. In order to meet these requirements, the implanted system adopts the analogous characteristics of deep brain stimulators, although the locations are different. Similarly, the chargers are designed to be wearable and tolerant of misalignment.
Table 1. Comparison of wireless power transfer (WPT) in deep brain stimulation.

| Category | Frequency | Input 1 | Output | Efficiency 2 | Transmitting Distance | Transmitting Coil | Receiving Coil | Material/Encapsulation 4 | Media |
|----------|-----------|---------|--------|--------------|-----------------------|-------------------|----------------|------------------------|--------|
| Cortical [25] | 907.5 MHz | 20.1 mW | 15.6 µW | -0.04% | 16 mm = 11breath+ 5air | Annulus: D_{trans} = 41.3 mm D_{int} = 10 mm | Cubic: L = 1 mm | PDMS | Head equivalent phantom |
| Cortical [25] | ISM band | 26.8 mW | 15.6 µW | -0.05% | 20breath+5air+2skirt+5air | Annulus: D_{trans} = 41.3 mm D_{int} = 10 mm | Cubic: L = 1 mm | PDMS | Porcine model |
| Cortical [25] | 907.5 MHz | ~29 mW | 118 µW | -0.4% | 16 mm = 11breath+5air | Annulus: D_{trans} = 41.3 mm D_{int} = 10 mm | Planar square: L = 6.5 mm | PDMS | Parylene C & medical grade silicone & C | Porcine tissue & Mouse back |
| Cortical [15] | 400 MHz | 19 mW (SAR1g), 82 mW (SAR10g) | 100 mW | 0.1% | 15 mm = 12breath+3air | Annulus: D_{trans} = 12 mm | Approx. cubic L = 1 mm | medical grade silicone & C | Porcine tissue & Mouse back |
| Cortical [26] | 13.56 MHz | N/A | N/A | 0.56% | Movable 6 | Cuboid: 40 cm x 40 cm x 20 cm | Planar square: L = 10 mm | N/A | Movable 6 |
| Optogenetic [27] | 7.2 MHz | N/A | N/A | 3.16% | 15 mm = 12breath+3air | Planar square: L = 10.5 mm | Planar square: L = 10.5 mm | medical grade silicone & C | Porcine tissue & Mouse back |
| Cortical [14] | 400 MHz | 19 mW (SAR1g), 82 mW (SAR10g) | 100 mW | 0.1% | 15 mm = 12breath+3air | Annulus: D_{trans} = 12 mm | Approx. cubic L = 1 mm | medical grade silicone & C | Porcine tissue & Mouse back |
| Cortical [28] | 400 MHz | 18 mW | 57 µW | 0.32% | 10 mm | Annulus: D_{trans} = 18 mm D_{int} = 12 mm | Approx. cubic L = 1 mm | medical grade silicone & C | Porcine tissue & Mouse back |
| Deep brain [7] | 403 MHz | ~99 mW | 10.52 mW | 10.62% | 5 mm = 3breath+2skin | Annulus: D_{trans} = 19.86 mm | Annulus: D_{int} = 15.94 mm | medical grade silicone & C | Porcine tissue & Mouse back |
| Deep brain [26] | 13.56 MHz | 0.5 A | 8.5 mW | 0.56% | Movable 6 | Cuboid: 40 cm x 40 cm x 20 cm | Approx. cubic L = 1 cm | medical grade silicone & C | Porcine tissue & Mouse back |
| General [29] | 120 kHz | 16V/250 mW | N/A | N/A | Movable 6 | Annulus: D_{trans} = 44 mm D_{int} = 8 mm | Thickness = 1 mm | LTPS TFTs | N/A |
| General [30] | 160 MHz | N/A | 60 mW | 60% | Movable 6 | Annulus: D_{trans} = 35 mm D_{int} = 34.4 mm | Annulus: D_{trans} = 10 mm D_{int} = 9.8 mm | 1X PBS | N/A |
| General [31] | 13.56 MHz | 12 W | 50 mW/mm² | N/A | Up to 30 cm | Two Planar squares: L = 30cm Distance = 9 cm | N/A | Uniform bilayer of parylene/PDMS | In vivo test (mouse) |

1 Input unit is changed due to the information provided from related research works. 2 Output power is defined as the activation power of micropump or metal melting. 3 Due to the ambiguous input and output power, the power transfer efficiency of most links cannot be calculated. 4 Coating refers to the material used to separate tissue and implanted devices, while encapsulation refers to the material used to envelop the circuits. 5 This design obtains an intermediate antenna for the high power-transfer efficiency. 6 “Movable” refers to a patient or research target that can move freely with receiving coils during charging. 7 This design uses a stretchable wire for receiving coil. 8 The efficiency is for wireless links but not for the overall systems. The high values are equal to 1.
Figure 2 demonstrates the research work [24] that developed a system containing a bio-electronic interface that links the injured muscle/nerve and the peripheral muscle in order to restore dexterous function. This process is achieved by connecting the proximal end of the injured part to the distal end of the peripheral nerve with an electromyography (EMG) signal. The telemetry is maintained by a wireless system with power offered from another coil. For WPT, the primary and secondary coils have a square design with a side length (L) of 16 mm. The link can provide 5 V to power the implanted amplifier with a frequency of 13.56 MHz. This design can control stimulation with data feedback. Other mature systems, such as ReStore, provide the pulse with pre-designed programs [32–34].

![Figure 2. Design of vagus nerve stimulation in arms.](image)

2.2.2. Retinal Nerve Stimulation

Retinal prostheses are widely adopted and promising therapies for eye-handicapped patients with common outer retinal degenerative issues, including age-related macular degeneration (AMD) and retinitis pigmentosa (RP) [29,35,36]. These problems are the dominant causes of most retinal blindness, especially in developed countries with an aging population [37]. For patients, visual function is maintained through an electrical stimulus on intact neural cells. Incitement is provided by implanted electrodes that deliver the surrounding images captured by a miniature digital camera mounted on eyeglass frames [37–39]. Due to the structure and characteristics of subretinal and epiretinal areas and nerve systems, both power and data are transmitted wirelessly to minimize the size of implanted devices and hazardous impacts. Generally, retinal recovery is categorized into two methods based on the implanted location; subretinal access ensures a better alignment, while the epiretinal approach is easier to be implanted [36]. The major consequence of the different locations of the wireless system is the change of media material, which could affect the coil coefficient and further dwindle the PTE. However, this section discusses the WPT by coalescing these two methods.

Wong et al. [31] proposed an innovative retinal implanted device. It replaced a traditional primary power supply with a photo-sensing circuit that adopted a phototransistor and a ring oscillator to provide stable power for wireless telemetry and power transfer. The design of the WPT system, which is shown in Figure 3, is different from previous studies on antenna design. The primary antenna is a circular Litz coil with a diameter of 3.0 cm (D1) and four windings, while the secondary coil is a thin aluminum film consisting of 100 windings with a total length of 2.8 m. The power is transmitted through magnetic resonance coupling with a frequency of 160 MHz. According to the study’s outcomes, the maximum power transfer efficiency was around 25% based on the energy requirement as 320 µW
of a photo-sensing circuit that is integrated with receiver antenna and other receiving-side elements in
the eyeball. The system was also tested via an in vitro experiment in phosphate-buffered saline (PBS).
It turned out that the artificial units could provide the same stimulus as a living retina.

Figure 3. Planning design of retinal neurostimulator.

In addition to improvements in energy resources, the overall system has been updated in recent
years, especially for the circuit design. To further illustrate current research emphasis and future
development directions, a summary of WPT for retinal devices is presented in Table 2, below.

From Table 2, in order to follow the trend of transmission efficiency and safety improvement,
several research groups [36,39–41] enlarged the receiver coil by relocating the secondary antenna
from the back to the anterior. Meanwhile, hermetic packaging was used to coat for the encapsulation
with biocompatible materials. Due to this relocation, the coil design changed to fulfill minimization
requirements, as the implanted coil was wound as a spiral for the curvature of the eyeball. Additionally,
current research also seeks to enhance the efficiency and safety aspects of misalignment tolerance [29]
and converter improvements [28]. Another study focuses on recovering the sense of dark and bright
(instead of regaining vision) with the wireless system to power up the artificial retina [42].

From the above analysis, WPT shows a great potential in retinal prosthesis from the following
three aspects:

(1) Minimization can be maintained with the flexible design of coils, while the wireless system
obtains the functions of both power transfer and data telemetry;
(2) The duration can be extended with the current photo-sensing circuit by the consistent power input;
(3) Safety is guaranteed with the hermetic design and biocompatible materials.

Also, due to the special conditions of retinal use, the application of WPT has the following
difficulties:

(1) Difficult implantation surgery: The non-planar structures of the target organs indicate a possible
bending of receiving antenna, which would lead to relatively low power transfer efficiency.
(2) Misalignment concerns: For devices designed to recover vision, it is hard to ensure alignment
due to the 360° rotation ability of eyeballs, as the primary coil is fixed on wearable glasses while
the receiving antenna is implanted inside the body. However, symmetry can be maintained with
a secondary coil located with a certain degree and spot, while the regained vision is limited.
(3) Change in transmission gap: Subretinal implantation is adopted for the majority of devices, as this
ab externo surgery obtains a minimally invasive influence. However, the structure and composite
of the retina could change with external impacts and age, which would lead to unexpected
changes in electromagnetic permittivity and permeability.
Table 2. Comparison of wireless power transfer in retinal prosthesis.

| Category     | Frequency | Input       | Output          | Efficiency | Transmitting Distance | Transmitting Coil | Receiving Coil | Material/Encapsulation | Media                |
|--------------|-----------|-------------|-----------------|------------|------------------------|-------------------|-----------------|------------------------|----------------------|
| Subretinal   | 125 kHz   | ~100 mW     | 0.76 mC/cm²     |            | 20 mm                  | Annulus: D_outer = 41.3 mm D_inner = 33.7 mm | Annulus: D_outer = 10.3 mm D_inner = 8.3 mm | Flexible polyimide substrate & Silicon | Air & phosphate buffered saline solution & minipig |
| Subretinal   | 125 kHz   | ±2.5 V      | N/A             |            | 30 mm                  | Annulus: D_mean = 19 mm | Spherical D = 19 mm T = 2 mm | Polydimethylsiloxane (PDMS) & Titanium | Air & phosphate buffered saline solution & minipig |
| Subretinal   | 160 MHz   | N/A         | 320 µW          |            | 1 cm                   | Annulus: D_mean = 60 mm 4 windings | Tot. Length = 100 m 100 windings | LTPS TFTs | IX PBS |
| General [28] | 1 MHz     | 9 V         | 16 V/250 mW     |            | 7 mm                   | Annulus: D_outer = 44 mm D_mean = 8 mm Thickness = 1 mm | Annulus: D_outer = 22 mm D_mean = 18 mm Thickness = 0.5 mm | N/A | Air |
| General [29] | N/A       | N/A         | 60 mW           | 60%        | 7 mm                   | Annulus: D_outer = 35 mm | D_mean = 34.4 mm Thickness = 4 mm | N/A | Air |
| General [43] | 10 MHz    | N/A         | 25 mW           | 36%        | 21 (outside)+4 (inside) mm | Annulus: D_outer = 42 mm D_mean = 34.4 mm | Annulus: D_outer = 20 mm D_mean = 14.8 mm | N/A | Air & saline |

1 Input unit is changed due to the information provided from related research works. 2 Output power is defined as the activation power of the micropump or metal melting. 3 Due to the ambiguous input and output power, the power transfer efficiency of most of the links cannot be calculated. 4 Coating refers to the material used to separate the tissue and implanted devices, while encapsulation refers to the material used to envelope the circuits. 5 This design obtains an intermediate antenna for high power-transfer efficiency.
Based on the previous analysis, the future of WPT in retinal implants can be indicated as follows:

1. **Alignment improvement:** Techniques with a high tolerance for misalignment and rotation will be developed. A combination of different coils to replace a singular receiver might be adopted. The strategies are also requested to maintain the low hazardousness and high convenience.

2. **Self-powered design:** Reuse the received light to power up the system.

3. **Minimization of implanted electrodes:** To increase the preciseness of stimulation and data analysis, the number of electrodes would increase under the constraints of mass and size.

### 2.2.3. Cochlear Nerve Stimulation

As a mature implant technology [44–48], the cochlear implant has experienced a long development path [49] as the first commercial wireless powered implantable device [50]. Generally, cochlear implants require both wireless communication for auditory information transmission and wireless power transfer to extend operation duration [50]. In order to meet all these requirements, implanted coils can realize all functions within a small space. For information transmission, the outside components, which are usually attached behind the ear, can capture the sound with microphones and transfer the information to electrical signals with a signal processor. Then, the signals can be transmitted to inner implants through wireless telemetry. After that, the implanted coil generates the electricity that stimulates the existing auditory nerve fibers to deliver the information to brain [51,52]. The implanted stimulator unit, which interprets the sound to electrical signals, is powered by a battery located outside the ear. This battery can be designed with rechargeable characteristics to continuously supply energy via WPT [53]. According to the commercial design, the coil distance is several millimeters, as they are merely separated by skin [54]. With this relatively small transmitting distance and simple gap texture, the coil link can maintain a high power-transfer efficiency with near-field magnetic resonance or inductive coupling [49]. As the power requirement ranges from 20mW to 40mW [49,55,56], the relatively small coils are sufficient for the power input. The general design for the WPT link for cochlear implants is presented in Figure 4.

![Figure 4. Wireless power transfer for a cochlear implant system.](image)
as the structure of support devices is highly restricted. For the health impacts, long-term exposure to radiation can lead to unpredictable hazardous effects due to changes in DNA or gene scale [58]. The potential ramifications are the degradation of the neuroendocrine system [59], fertility decrease, and muscle stiffness with loss of protein [22].

2.3. Spinal Stimulation

Similar to nerve stimulation, spinal nerve stimulation aims to assuage pain or dysfunction due to spinal cord injury. To be more specific, epidural spinal cord stimulation (ESCS) focuses on locomotor recovery for patients with incomplete spinal cord injury and multiple sclerosis [60]. Although the biochemical principles are different from previous therapies, the wireless link is designed with high similarity. To avoid redundancy, spinal stimulation is investigated without a table summary, due to there being relatively fewer applications than the other nerve therapies.

One high-efficient magnetic resonant WPT for spinal cord stimulation [61] utilized the unconventional multi-sine (MS) transmission waveform, which achieves an overall system efficiency of 50.7% with 6.78 MHz. The wireless link is designed with a four-coil system, with two planar circular coils at each side. For the transmitter, the inner length and width of the driven coil are 18 mm (D1) and 6 mm (w2), respectively, while the primary coil with 21 mm (D1) inner diameter and 27 mm (w1). For the receiver side, the parameters of the load coil are 18 mm inner and 20 mm outer, while the secondary coil is 8 mm inner and 20 mm outer. The transmitting distance is set as 18 mm, which is simulated as fat and skin in an in vitro test. Generally, the wireless link is located with a pulse generator placed in a pelvic or thoracic cavity [62]. This research work is designed to charge right after the spine, which is shown in Figure 5.

![Figure 5. Design of spinal stimulator. (a) Transmitter; (b) receiver; (c) general view.](image)

Compared to other stimulations, the spinal cord spur illustrates a flexible WPT location, especially compared to deep brain stimulation. The spinal cord, which is across the whole back, has plenty of options to locate the implant, while the pulse generators can be set near to the skin to achieve high energy-transfer efficiency. However, the healing targets of this stimulus are duplicated by brain implants, while the latter can maintain more functions. The current research of this kind of spur is still limited, but it will likely flourish in the future.

For WPT application, various location choices assure efficiency enhancement. The unconventional MS transmission [61] waveform could be utilized to enhance efficiency without considering a high increase in specific heat absorption. Duplex amplitude-shift keying (ASK) [63] and load-shift keying (LSK) [60] communication can also be applied to improve the power and data transmission efficiency, respectively. Additionally, the soft and stretchable coils are also attractive for muscle movement concerns [18].
3. Assistant Devices

Assistant devices refer to the implanted chargers that assist the biomedical implants to extend their operation duration. The most distinguishing feature of these devices is that the charging functional section is separated from the biomedical appliances. As the medical devices are moving during the working period, the charging coil is fixed in a nearby area with a relative efficiency. Additionally, the moving area of appliances should be restricted to ensure that a short lead/wire can successfully transmit the energy without safety issues led by an induced effect [64]. Based on these characteristics, only cardiac implants can be counted as a kind of assistant devices.

Cardiac pacemakers are implanted below the collarbone to maintain an adequate heart rate by delivering electrical pulses [65]. This function is mainly realized through two components: a pulse generator and a lead system. As the pulse is transmitted to the heart by employing leads, a case that contains the pulse generator, battery, and related circuits is implanted. As the case can be located near the skin, near field magnetic resonant WPT is widely adopted for this device. Additionally, the coil design is relatively flexible, as the space constraint is mitigated with lateral implantation.

With these understandings, an effective WPT charging system for pacemakers was designed [66]. The study displayed an efficient power link with a Helmholtz transmitter as shown in Figure 6. Two transmitting coils are wound with Litz wires with $165 \times 0.1$ mm to reduce skin and proximity effects. The outer diameter is $280$ mm ($D_{\text{outer}}$), while the inner diameter is $190$ mm ($D_{\text{inner}}$) for both coils. The two coils are separated by $190$ mm, which ensures that a normal human can stand or lie between them. Similarly, the receiving part is composed of two coils, which are wound with $100 \times 0.07$ mm Litz wire. Two symmetrical coils are designed to enhance the power transfer efficiency, while 16 ferrite cores are added to each transmitting coil. However, the induced effect by the current transmission should be taken into consideration [64]. According to the in vitro test, the power link achieved a high transfer efficiency of $88\%$, while the misalignment concerns, including rotation ($0$–$90^\circ$), x-direction shift, and height change, were discussed to ensure a sufficient power input in most of possible situations.

![Figure 6. Wireless power transfer for cardiac implants. (a) Transmitter; (b) system configuration.](image)

The studies of WPT in cardiac pacemakers illustrates the possibility of separating the wireless power link and the pacer. As shown in Table 3, the combination of functional and powering parts could result in a relatively low efficiency, as the magnetic flux can reach the receiving coil easily with a shorter transmitting distance. Additionally, the output power values are various among research works. These values indicate the duration of the charging period, but a fast process might lead to a perilous effect. For pacemakers, titanium is adopted as the encapsulation material on large scale due to its great metal shell ductility. However, the eddy current effect is aroused due to metallic shielding. This effect will lead to a decrease in PTE and an increase in SAR with heat generation. As the receiving coil is assembled inside the cardiac case, there is no additional coating analysis for the implanted coil. All listed studies fulfill the safety regulations on SAR based on different standards. In this table, the parameter is also included, as most research works thoroughly discussed the SAR or thermal...
effects. The emphasis on the safety issues indicate that the research has arrived at the optimization and commercial stage.

As a highly developed application, the coil structures are comparable, but the number of coils can increase to enhance the power transfer efficiency (PTE).

With these understandings, the potential of applying WPT in pacemakers can be summarized as follows:

(1) Effective energy supply: The fixed location and neglectable movement of implanted coil curtail the necessity of misalignment mitigation strategies. This would also escalate the PTE.

(2) Large demand: Pacemakers consume considerable energy comparing to other implanted devices. As a consequence, invasive energy supply techniques are demanding.

(3) High flexibility in coil design: As the coils can be detached from the major system, the space for receiving coil increases and can be applied for complex coil design or combination.

Also, there exist several difficulties in implementation.

(1) Varied transmitting distances among individuals: The thickness of chests for different people can affect the promotion of implanted WPT.

(2) Safety concerns led by additional wire: As extra wires might be used to transmit the current to the pacers, the further induced effects [57] should be taken into consideration.

(3) Health impacts due to long-term exposure to radiation: Radiation can influence the human body directly on DNA, genes, and protein. For pacemakers, the effect of heart attack due to the degradation of protein [67] should be considered.

In addition, the future development directions are given as

(1) Innovation in compensation topologies, such as LCC-C, LCL [68];

(2) Adding ferrite cores to wireless links to intensify the electromagnetic field for high PTE;

(3) Accurate numerical evaluation of SAR and thermal effects through simulation human body models [69].
Table 3. Comparison of wireless power transfer for pacemakers.

| Category          | Frequency | Output | Efficiency | Transmitting Distance | Transmitting Coils | Receiving Coils | Encapsulation   | Media                      | Safety Parameters          |
|-------------------|-----------|--------|------------|-----------------------|--------------------|----------------|----------------|---------------------------|-----------------------------|
| Separated [66]    | 160 kHz   | 5 W    | 88%        | 0-10 cm               | 2 planar spiral coils: n = 25, Δd = 190 mm, D_{outer} = 280 mm, D_{inner} = 190 mm, 16 ferrite bars per coil | 2 planar spiral coils: n = 13, Δd = 5 mm, D_{outer} = 50 mm | Titanium case | Male mannequin | 2 W/kg SAR (IEEE) |
| Combined [70]     | 20 kHz    | 500 mW | 77%        | 8 mm = 3 + 2 + 3     | A planar spiral coil: n = 29, 18 AWG, D_{outer} = 70 mm Planar helical coil with ferrite film, n = 13, 58x48 mm², Copper strand (160 leads, D = 0.1 mm) Δd_{film} = 0.29 mm | A planar spiral coil: n = 39, 24 AWG, D_{outer} = 46 mm Planar helical coil with ferrite film: n = 9, 44.5 x 30.5 mm², Δd_{film} = 0.29 mm Flexible printed circuit | Titanium case | Unipolar lead | Pork (3 mm air + 3 mm skin + 2 mm fat) | 2.7 V/m (ICNIRP) |
| General [68,71]   | 300 kHz   | 3.07 W | 78%        | 8 mm = 1 + 2 + 5     | A planar spiral coil: n = 13, 18 AWG, D_{outer} = 70 mm Planar helical coil with ferrite film: n = 9, 44.5 x 30.5 mm², Δd_{film} = 0.29 mm Flexible printed circuit | 2 planar spiral coils: n = 29, 18 AWG, D_{outer} = 70 mm Planar helical coil with ferrite film, n = 13, 58x48 mm², Copper strand (160 leads, D = 0.1 mm) Δd_{film} = 0.29 mm Flexible printed circuit | Titanium alloy material TC4 | Pork (1 mm pig skin + 2 mm pig fat + 5 mm pig lean) | 2.89 W/kg psSAR (IEEE) 4.22 °C increase |
| Separated [64,69] | 300 kHz   | 1 W    | 80%        | 10 mm                 | A planar spiral coil: n = 10, D_{outer} = 36.6 mm Planar helical coil with ferrite film: n = 13, 58x48 mm², Copper strand (160 leads, D = 0.1 mm) Δd_{film} = 0.29 mm Flexible printed circuit | A planar spiral coil: n = 22, 18 AWG, D_{outer} = 36.6 mm Planar helical coil with ferrite film: n = 13, 58x48 mm², Copper strand (160 leads, D = 0.1 mm) Δd_{film} = 0.29 mm Flexible printed circuit | Titanium case | Air, saline solution | 2 W/kg SAR (IEEE) |
| Separated [64,69] | 13.56 MHz | 1 W    | 80%        | 10 mm                 | A planar spiral coil: n = 4 (SP), D_{outer} = 35.4 mm | A planar spiral coil: n = 22, 18 AWG, D_{outer} = 36.6 mm Planar helical coil with ferrite film: n = 13, 58x48 mm², Copper strand (160 leads, D = 0.1 mm) Δd_{film} = 0.29 mm Flexible printed circuit | Titanium case | Air, saline solution | 2 W/kg SAR (IEEE) |
| Combined [72]     | 13.56 MHz | 200 µW | 6.0%       | 20-30 mm *            | A planar spiral coil: n = 18, 23 AWG, D_{outer} = 40 mm, D_{inner} = 10 mm | A planar spiral coil: n = 4, 30 AWG, D_{outer} = 5 mm, D_{inner} = 1 mm | Titanium case | Euthanized pig | 2 W/kg SAR (IEEE) |
| Combined [73]     | 198 MHz   | 300 mW | NA         | 200-300 mm *          | 2 octagonal coils fabricated on one PCB board: n = 4, n = 3, D_{outer} = 160 mm | A planar square coil: n = 6, 22 AWG, L = 4.9 mm | Titanium case | Ex vivo Langendorff heart models | NA |

1 The distances with * indicates that the values are estimated through coil diameters. 2 Some design obtains several coil-sets for the power transmitting side, but all transmitting coils are outside the body. 3 Ferrite cores may exist to enhance the electromagnetic intensity for the receiving side, and all receiving coils are inside the body. 4 Safety parameters are summarized in this table, as most of pacemaker research included SAR/thermal analysis.
4. Movable Implants

Multi-functioning robotic capsules (MRCs) refer to devices that are able to travel to the areas where it is difficult to apply therapy through an open wound [74]. This is a promising technology for the direct diagnosis and treatment in the abdominal cavity without interrupting the patient’s daily activities [75]. Traditionally, wireless capsule endoscopes (WCEs), as one of the MRCs, are used in the gastrointestinal (GI) or digestive system, where a natural pipeline system is provided. As the only technique that can allow the direct visualization of the full length of the GI tract including the small bowel for many pathologies, there are several relevant academic studies and commercial products [76–78]. In this decade, the utilization of MRCs has broadened with the development of miniaturized active control systems, actuations, and locomotion mechanisms. These multi-functioning robotic capsules can be used in drug delivery systems, semi-surgeries, etc. [75]. With this increase in functions, the overall power required by one capsule has escalated. However, the movable and minimal nature of WCE restricts the possibility of taking batteries to provide sufficient energy. Thus, WPT is crucial to facilitate the future development of WCEs by extending the overall operating duration significantly. In order to clarify the differences between a robotic wireless capsule and capsule endoscopy, this section is divided into two parts to discuss the WPT characteristics in various conditions. The case for a multi-functioning capsule is a drug delivery system.

4.1. Gastrointestinal Endoscopy

Gastrointestinal disorders [79] influence many people due to detrimental diseases such as gastric cancer [80,81], tumors [82,83] and bleeding [84]. Aside from effective therapy, early detection is also important for prevention. Wireless capsule endoscopy provides a possibility of painless, quick implantation, while the operation duration can be relatively long. The current increasing demand for real-time healthcare system indicates a high request of WCE in the future, as it offers an opportunity for easy detection and treatment without additional wounds. For this application, the difficulties of applying WPT fall into alignment, power transfer efficiency, and safety concerns.

In order to evaluate the overall power transfer efficiency at different locations inside the body, one study [76] conducted an in vivo test in swine. The design adopted a high frequency of 1.2 GHz to ensure the mid-field antennas could sufficiently transfer the requested power. The wireless link was designed with two small square loop antennas with a length of 6.8 mm. The transmission distances varied with the implanted locations. For the picked swine, the transmitting gaps were 30 cm, 75 cm, and 25 cm for the esophagus, stomach, and colon, respectively. The results were positive, as the output power is 37.5 µW, 123 µW and 173 µW for implants at these three locations. Moreover, the mean transmission efficiencies were calculated as 0.076%, 0.025%, and 0.035% for the referred locations in the in vivo test. To simulate the composition of the transmission gap, the study used a mixture of entrails and other chopped tissue. For safety concerns, simulations were performed to determine the maximized power input according to the 10 g SAR Institute of Electrical and Electronics Engineers (IEEE) standard. This study is special as it held an in vivo test, which is shown in Figure 7, while most of the other works only include simulations or in vitro tests. Although both methods demonstrate the feasibility of the design, this method illustrated the significant differences in efficiency and output power stability between simulations and real experiments. These misalignments could be a result of unanticipated transmitting distance and coil direction changes. With this understanding, in vivo tests are recommended for movable minimal robotics, and studies on improving the tolerance of load variation are advocated and discussed. In regard of these two directions, Table 4 summarizes the studies in the last 10 years.

Table 4 lists the recent studies on wireless endoscopies. Power transfer efficiency remains relatively low no matter how the frequency changes. This illustrates the high difficulty of enhancing the power to the load. In order to transfer sufficient energy for continuous operation, the uniformity of the excited electromagnetic field was taken into consideration in most studies. Generally, this is realized by designing the transmitting coils through two approaches—solenoid and Helmholtz [75]. Solenoid
transmitting coils can induce a relatively high magnetic flux density, while Helmholtz coils provide the advantage of high uniformity [85]. Additionally, receivers were designed with a ferrite core to increase the magnetic field absorption capacity. Aside from efficiency enhancement, safety is also a crucial concern. All studies used the IEEE high tier standard for 10 g tissue and evaluations were held within vivo or in vitro tests.

Figure 7. WPT for wireless capsule endoscope (WCE). (a) Configuration of transmitter/receiver; (b) proposed testing locations.

Based on the aforementioned analysis, the potential of applying WPT in wireless endoscopies are given below.

1. The high design flexibility of power receiver: As the robotics may obtain different functions, the outlook and inside mechanical structures are various. WPT provides the opportunities for changing the receiving coil with relatively low ramifications on efficiency when compared to other power supply techniques.

2. Guarantee of sufficient power load: The power input can vary within the restriction of IEEE SAR standards, so the power output can be ensured even with unanticipated load variation. Additionally, a ferrite core can be added to intensify effective magnetic flux.

3. Potential on combined function: As the coils can transfer both energy and information, the implanted robotics capsules can deliver its location for positioning. This function would be helpful for fast charging and device location [86].

Also, there exist several difficulties in implementation.

1. Low power transfer efficiency: As the robotic capsules obtain mechanical missions, the misalignment and rotation of receiving coils can be expected. Under this situation, the electromagnetic field leakage of the primary coil is relatively high.

2. Hard to be popularized: Transmitting conditions can vary with individual conditions, such as gender, nationality, age, obesity rate, etc. Although in vivo tests can illustrate a sufficient power output, real performances cannot be guaranteed.

3. Safety concerns led by information leakage: Due to immature encryption technologies, the leakage of operation frequency could result in a safety issue.

In addition, regarding the current development of WPT and multifunctional endoscopy, the future developments of WPT in these biomedical devices can be summarized as

1. Redesigning wireless link to obtain the high tolerance of misalignment;
2. Current research emphasis is on the numerical or in vitro analysis and improvement, while these limited studies have demonstrated differences among results from these analyses and in vivo test [87];
3. Adding ferrite core with the high permeability to increase power received.
Table 4. Comparison of wireless power transfer in multifunctioning robotics.

| Category       | Frequency | Output | Efficiency | Transmitting Distance 1 | Transmitting Coils 2 | Receiving Coils 3 | Media 4 |
|----------------|-----------|--------|------------|-------------------------|----------------------|-------------------|---------|
| Helmholts [88] | 400 kHz   | 790 mW | 3.00%      | 310 mm *                | Two octagonal coils: n = 13, D = 330 mm, Louter = 690 mm, Linner = 620 mm | Dmean = 10 mm, ferrite core: 6.6 mm × 6.6 mm × 5.5 mm | Air     |
| Charging system [77] | 13.56 MHz | 24 mW  | 3.04%      | 1000 mm                 |                      |                   |         |
| Solenoid [89]  | 164.7 MHz | 26 mW  | 0.02%      | 70 mm                   | Transmitting coil: Dmean = 80 mm, n = 1 | Receiving coil: Dmean = 9 mm | Air     |
| Helmholtz [90] | 218 kHz   | 600 mW | 2.00%      | 250 mm *                |                      |                   | Pork chops |
| Solenoids [91] | 220 kHz   | 750 mW | 3.55%      | 345 mm *                |                      |                   | Pig intestine |
| Helmholts [92] | 246 kHz   | 534 mW | 4.90%      | 200 mm *                |                      |                   | Air     |
| Helmholtz [78] | 250 kHz   | 570 mW | 5.40%      | 175 mm *                |                      |                   | Air     |
| Helmholts [79] | 250 kHz   | 758 mW | 8.21%      | 175 mm *                |                      |                   | Air     |
| Solenoids [93] | 218 kHz   | 500 mW | 4.08%      | 200 mm *                |                      |                   | Air     |

1 The distances with * indicate that the values are estimated through coil diameters. 2 Some designs use several coil-sets for the power transmitting side, but all transmitting coils are outside the body. 3 Ferrite cores may enhance the electromagnetic intensity for receiving side, and all receiving coils are inside the body. 4 Only transmitting media is included in this table, as no studies discussed the impacts of encapsulation.
4.2. Drug Delivery System

There are two forms of drug release systems—passive devices that mainly control the rate of diffusion and active devices that require external triggers to control the release time [94]. Although the principles are different, both can be equipped with WPT systems for long endurance and high precision. Actually, the current advances in 3D printing, micromachining, and flexible electronics have promoted the development of pharmacologic delivery to ensure the preciseness and accuracy of drug therapy. By adopting wirelessly controllable actuators, key factors including the location, area, time, and amount of drug can be controlled for the long-term treatment of chronic disease [95], including cancer, cortical dysfunction, and organ lesions. Generally, wireless drug delivery systems are ultra-miniaturized and consist of the drug containers, control circuitry, actuators, and a battery. To balance endurance and size, the research on wireless powered actuators to replace the battery has become extensive.

Another typical delivering system relies on a micropump, which administers the demanded liquid drug flow. According to a related study [96], drug release is achieved by an external pressure increase that is realized by the electric current activation of hydrogen and oxygen.

4.2.1. Active Devices

Active devices, which focus on time control, require the power input to activate the lip control system or simply melt the sealing [94]. The purpose of applying WPT is to release drugs, which results in a relatively low power consumption and a low required preciseness of degree of opening, but the time length for unblocking is restricted. With these characteristics, the majority of researchers adopt a specific metal as a sealing material and use the power to melt it for release. Although convenience is ensured with this design, the metal may affect the electromagnetic field activated by the transmitting antenna.

One study [97] developed an integrated ionic polymer metal composite (IPMC) cantilever valve that can be activated by the voltage difference to deliver the drug. The implanted IPMC actuator is activated with a voltage applied by integrating the planar receiver coil through near field magnetic resonance at the field frequency of approximately 25 MHz with 0.6 W input power. An in vitro experiment was carried out to demonstrate the effectiveness, and the performance was illustrated with a coil distance of 4.5 cm (Δd) and an output voltage of 1 Vdc while the two loops are fully aligned. The receiver was designed as a planar square coil with a length of 1 cm (L). The structure of the delivery system is shown in Figure 8. The study also discussed the impact of transmitting gap and misalignment degree, but there was no possible mitigation.

Figure 8. An example of WPT in a drug delivery system: active devices with metal sealing.
4.2.2. Passive Devices

Different from active devices, passive drug delivery systems guarantee the flow rate of drug release, which implies the control of the opening level of a valve or the pump inside [94]. This design illustrates that passive devices are chosen to transfer electromagnetic energy to the physic force that drives chips. The difficulty of applying this system is that the preciseness of the control system is required that the characteristics of fluid in the surroundings will impact the flow rate of drugs. Moreover, a feedback system is also required to control the pump regarding location once the implanted devices move around inside the human body [98].

Another typical delivering system relies on a micropump to administer the demanded liquid drug flow. According to a study [96] displayed in Figure 9, the drug release is achieved by an external pressure increase that is realized by electric current activation of hydrogen and oxygen. The electric current, 0.33 mA, is driven by a WPT system with an input power of 9 V and 0.3 W. The system could obtain a successful dose delivery of 6 fixed micropumps for one week with 2 cm (Δd) transmitting distance and accurate alignment, which ensures two weeks of consistent administration of an anti-cancer drug. For coils, the receiver is circular with a diameter of 17 mm (D₁), while the transmitter is similar to a rectangle with a 310 × 140 mm² sectional area. An improvement in minimization and duration is achieved when compared to the similar work in 2014 [99], and an in vivo test was performed for the chronic application.

![Figure 9. WPT configuration in passive devices with micropumps.](image)

According to Table 5, the efficiency of WPT is not a vital benchmark to evaluate the performance of wireless links. Instead, the effectiveness of WPT in drug delivery systems is only illustrated by the output energy. Once the output energy can activate the wireless actuators, the wireless link is efficient, as there may not be any energy storage in the implants. Aside from the difference in evaluation standards, the capsulation materials for drug delivery are similar to those previously mentioned for nerve stimulation, as both applications mainly focus on biocompatibility and safety. However, the media of transmission gaps are more varied than in other situations. This is due to the relative frequent moves during operation. For examples, water or saline solutions are used to replace the complex in vivo conditions for expediency.
Table 5. Comparison of WPT in drug delivery devices.

| Application       | Category | Frequency | Input 1 | Output 2 | Efficiency 3 | Transmitting Distance | Transmitting Coil | Receiving Coil | Coating/Encapsulation 4 | Media                      |
|-------------------|----------|-----------|---------|----------|--------------|-----------------------|-------------------|----------------|-------------------------|---------------------------|
| Cervical tumor cells [96] | Passive  | 25 MHz    | 0.6 W   | 1 Vdc    | N/A          | 4.5 cm                | N/A               | ~ 10 x 10 mm² | PDMS                    | 0.2% agarose gel & DI water |
| Cancer cells [99]  | Active   | 2 MHz     | 9 V     | 0.33 mA  | N/A          | 3 cm                  | Spiral 310 x 140 mm² 8 turns | Spiral D = 17 mm 6 turns | medical grade silicone & Parylene C | 1X PBS                     |
| Cancer cells [95]  | Active   | 2 MHz     | 9 V/3 W | 0.33 mA  | N/A          | 3 cm                  | Spiral 310 x 140 mm² 8 turns | Spiral D = 17 mm 6 turns | medical grade silicone & Parylene C | 1X PBS & Mice               |
| Cerebral cortex [100] | Passive  | 13.56 MHz | 5 V/0.5 W| 1.34 µW  | 1.22-0.06%  | 0.5 cm              | Annulus: Douter = 28 mm Dinner = 17 mm | Annulus: Douter = 24 mm Dinner = 14 mm | N/A                      | 1X PBS & Mice               |
| General [101]      | Passive  | 6.7 MHz   | 10 V    | N/A      | N/A          | 0.15 cm              | Annulus            |                  | Nation & PDMS            | Blue 38 & Air & DI water    |
| General [102,103]  | Passive  | 13.64 MHz | N/A     | 1.89 Vdc 0.65 W | 1-5 cm      | N/A                  | N/A               | ~ 10 x 10 mm² | PDMS                    | Water                      |
| General [104]      | Active   | 185 MHz   | 1.1 W   | N/A      | N/A          | 1-5 cm              | N/A               | ~ 9.2 x 2.7 mm² | Nitinol & Parylene C    | Water                      |

1 Input unit is changed due to the information provided from related researches.
2 Output power is defined as the activation power of micropump or metal melting.
3 Due to the ambiguous output power, most power transfer efficiency of the link cannot be calculated.
4 Coating refers to the material used to separate the tissue and implanted devices, while encapsulation refers to the material used to envelope the circuits.
Based on Table 5, the potential of applying WPT in drug delivery system can be summarized as

1. Controlled timely administration: The WPT can precisely control the dissolution of gold sealing to release the drug in an optimal time range.

2. Efficient treatment: For central nervous system disorders, conventional treatments, such as oral ingestion and intravenous injection, are not efficient. This is due to a highly selective permeability of the blood–brain barrier, which complicates transporting processes. Implanted drug containers can release the drug directly to the lesion area without traditional penetration. This localized treatment could also minimize the side effects of drug with a low dosage and limited additional absorption [105].

3. Stable energy supply: For drug delivery systems with a relatively low power requested among implants, vertebrates’ daily activities could offer sufficient energy through body heating or movement [106,107]. Although the prospect of self-fulfillment is attractive, the power sources are less stable than that of WPT. The system would also require additional supporting devices.

The difficulties of WPT implementation are also listed as

1. Low power transfer efficiency: The non-planar structures of targeted organs indicate a possible bending of receiving antenna, which would lead to a relatively low power transfer efficiency.

2. Unnecessary operation duration: The function of WPT mainly focuses on the convenience and safety in the drug delivery system, while the duration is mainly controlled by the amount of implanted drug. Once the required dose is small enough that the administration could be activated by the power generated inside body [108], the necessity of WPT is doubted due to size concern.

3. Larger energy consumption than expected: The wireless actuator movement might be hindered by interstitial fluid. Therefore, a larger force than expected is required [96].

Based on the aforementioned summary, the future development of WPT in drug delivery system can be predicted as follows:

1. Adopting biocompatible materials to ensure the low hazardousness and high convenience;

2. Flexible implanted system to achieve the high accuracy and long duration;

3. In vivo/vitro tests to evaluate the result accuracy of numerical analysis and simulation;

4. Wearable transmitters rather than cages are vital for real human applications, which requires a high convenience standard.

5. Safety Standards and Regulations

The compliance of the proposed WPT link with safety regulations has also been investigated. Generally, adverse health impacts caused by implanted WPT are categorized into two parts—non-thermal and thermal effects. Non-thermal effects are initiated due to exposure to an electromagnetic field under 10 MHz. The induced electric field inside body would cause the stimulation of muscles, nerves and sensory organs. For thermal effects, they are restricted precisely, which are generated due to heat absorption by tissues [71]. Broadly, the current regulations [109,110] limit the temperature increase of active implantable medical devices to lower than 1 °C compared to the normal body temperature. At the frequency of interest, the guidelines of the Institute of Electrical and Electronics Engineers (IEEE) and the International Commission on Non-Ionizing Radiation Protection (ICNIRP) provide basic restrictions for electromagnetic fields in terms of the specific absorption rate (SAR). SAR measures the rate at which energy is absorbed by the human body when it is exposed to a radio frequency (RF) electromagnetic field [111]. It is defined as the power absorbed per mass unit of tissue. To narrow down the complexity and contradiction of this electromagnetic field (EMF) guidelines, those generally adopted standards are listed, but they may not meet some specific requirements in specific countries [111,112].
This section summarizes the existing safety standards and lists other specific hazardous impacts of WPT.

5.1. Current Safety Guidelines

According to this review, the regulations for radiofrequency, 100 kHz to 300 GHz [110,113], are considered, since the frequency for most implanted WPT falls into this range. Currently, the standards for exposure limits are categorized based on the charging environment. This is determined by whether the patients are conscious of being exposed in the electromagnetic field [114]. The uncontrolled environment, which is also known as the general public, is defined with exposure expectations. This exposure scenario is set with a lower hazard tolerance comparing to the controlled environment with the opposite definition. Although individuals with implanted devices understand their exposure, the limits for the uncontrolled environment are discussed with the concern of wearable charging in a public area.

To simplify and clarify the guidelines, only the up-to-date regulations are counted and listed in Table 5. The power limits are calculated through the voltage and current root-mean-square value [110,113]. Due to different sensitivities of the human body, limbs, and head to electromagnetic exposure, the limits for these three parts are discussed separately. For ICNIRP 2010, this analysis focuses on the frequency range lower than 100 kHz [110]. For the latest version from ICNIRP, the exposure scenario is defined with frequency from 100 kHz to 300 GHz which includes all aforementioned medical applications. However, the ICNIRP 1998 [113] is discussed here, and the ICNIRP 2018 is still under revision and obtains similar results. Moreover, the implantable devices are generally designed with a frequency lower than 1 GHz, and only the range lower than 6 GHz is summarized here. According to the ICNIRP reports [111,113], the power restrictions for head, body and limbs in the uncontrollable environment are 2 W/kg, 0.08 W/kg, and 4 W/kg, respectively.

For the IEEE 2005 guidelines [115], the frequency limits range from 100 kHz to 3 GHz with a more stringent standard for the head in uncontrollable exposure with 1.6 W/kg per 1 g. Table 6 summarizes the prominent standards for the WPT in the frequency range of 100 kHz to 300 GHz.

| Guideline | Basic Restriction | Whole Body Avg. (W/kg) | Head/Trunk (W/kg) | Limbs (W/kg) |
|-----------|-------------------|------------------------|------------------|--------------|
| IEEE 2005 | Controllable      | 0.4                    | 10               | 20           |
|           | Uncontrollable    | 0.08                   | 1.6 (per 1 g)    | 4            |
| ICNIRP 2010 | Controllable | 0.4                    | 10               | 20           |
|           | Uncontrollable    | 0.08                   | 2                | 4            |
| ICNIRP 1998 | Controllable | 0.4                    | 10               | 20           |
|           | Uncontrollable    | 0.08                   | 2                | 4            |

The standards are all restricted with 10 g cubic mass and 6-min interval. IEEE localized standards are peak spatial average SAR.

5.2. Special Requirements of Implantable Devices

(1) Required specialized standards: Different from general exposure in the electromagnetic field, metallic implants could produce a potentially strong field. As the safety standards from IEEE and ICNIRP only include general exposure, the potential field generated by implants is currently defined as an untested condition [116]. To provide adequate protection for implant users, special standards are required with evaluations of currents, E-filed, or SAR in the human body [111]. Moreover, a combination of measurements with numerical simulations and experimental tests are encouraged to obtain a high level of detail and accuracy [117].

(2) Difficulties of precise assessment: For medical implants, with frequencies from several hundred KHz to MHz, both electric and magnetic fields contribute to hazardous exposure. For near field applications, the electrical fields in tissue can be evaluated by SAR based on eddy current,
which is the major contributor [118]. For the high frequency range, a significant enhancement of local absorption in the skin can be measured with standing wave effects, which is due to the reflections at the boundaries of tissues with strong dielectric contrast [119–121]. Furthermore, the drawbacks of inaccurate tissue modeling could exacerbate the difficulty of correct hazard assessment. The complexity of the human body indicates various in vitro model designs, in both structure and material aspects. For tissue, which is considered non-magnetic [122], the properties of conductivity and permittivity should be focused [123–126]. Different properties will affect tissue absorption and electric field distribution and further impact SAR [127,128].

(3) Encapsulation: For implantable devices, the packaging material is a crucial topic to ensure biosafety. For WPT, the preferred metals could lead to hazardous impacts once they are implanted. In order to avoid these negative ramifications, encapsulating the whole system with biocompatible material is necessary [129,130]. Generally, the adopted encapsulation materials aligned with the original design of implanted biomedical devices, such as polydimethylsiloxane (PDMS), NuSil, etc. [8]. However, different encapsulations would also affect PTE. For the material concerns, metamaterials could intensify the electromagnetic field while ensuring that the mass augment is under control [131]. In contrast, encapsulation materials could also attenuate magnetic flux, which should be circumvented. Moreover, metallic shells would generate eddy currents, which would result in a temperature increase on the surfaces of the implants [70]. With these potential influences, the impact of encapsulation architecture should be taken into consideration.

(4) Energy encryption: In addition to physical safety, information protection is crucial to avoid the privacy violations or data interface. Energy encryption [132], which is proposed in WPT for electric vehicles, aims to eliminate the possibility of receiving energy from unauthorized suppliers or transferring energy to unauthorized receivers. This cyber security is achieved by adopting chaotic encryption [133], which can avoid code-breaking, appropriate resonant converters, and control strategies to fulfill the power level and operational requirements. This system is designed for simultaneous multi-charging. For WPT in implantable devices, this process should be scaled down to meet the minimization requirement in order to guarantee charging safety.

6. Foresights and Development Trends

Although fruitful achievements have been made on the development of coil design, the control methods, the efficiency enhancement strategies, and the encapsulation materials of WPT systems in implantable devices still provide many opportunities for further study:

(1) To develop control strategies or coils structures to enhance the tolerance of unanticipated misalignment and rotation;
(2) To reduce the electromagnetic flux leakage to maintain a high transmission efficiency;
(3) To apply metamaterials to wireless links while limiting a large mass increase to intensify electromagnetic field;
(4) To integrate WPT and wireless information transmission into the same channel to form a hybrid system, hence manipulating power and control simultaneously;
(5) To devise techniques for determining the transmission materials, hence maintaining an accurate simulation analysis;
(6) To minimize the auxiliary circuits, compensation networks, and coils to ensure a minimal precarious influence;
(7) To design the encryption methods for WPT to ensure no information leakage;
(8) Other advanced design methodologies, control strategies, and new ideas to enhance the system performance of the implant devices.
7. Conclusions

In this paper, a review of the application of WPT in implantable biomedical devices has been investigated and discussed, with emphasis on the impact factors for WPT output. The factors include the implanted location, transfer media, and safety warrants. Their characteristics and impact potential are evaluated based on WPT system features, such as input power, efficiency and transfer distance, current thresholds, etc. For devices such as cochlear implants and pacemakers, the feasible application of WPT requires the evaluations of safety concerns. For developing devices, especially endoscopies and drug delivery systems, the optimization of coil design to enhance the tolerance of load variation is necessary. It is anticipated that a boost in research combining WPT and wireless telecommunication, especially for nerve-related therapy and healthcare system, will be a fertile area for development in the coming years. The key summation is listed below.

(1) For brain stimulations emphasizing the deciphering of brain functions, the current strategies all regard MHz operating frequency, which may affect the information transmission quality. As a result, the methods of combining wireless power transfer and wireless telecommunication are expected in the near future.

(2) For peripheral stimulation focusing on long-term therapies, the current studies achieve a sufficient energy supply, but the safety issues are still waiting for exploration.

(3) For the spinal stimulations with huge future potential, the wireless link designs are expected to have higher power levels and transfer efficiency.

(4) For the pacemakers and cochlear implants, which are highly developed, their safety and reliability are highly required.

(5) For the movable implants which are under developing, different coil combinations are proposed to enhance misalignment tolerance, while the overall power transfer efficiency is still quite low. Positioning strategy may be requested for further power transfer efficiency enhancement.

(6) For endoscopy with short operating duration, the transfer of power to the receivers is feasible, while the potential of operating without a battery is expected.

(7) For drug delivery systems with requested activation, wireless power transfer can be used for the temporary relocation of the whole device to enhance the preciseness of the system in the near future.

(8) For safety standards currently focused on SAR, new regulations are expected to involve more details related to time, location, information linkage, and reliable simulation methods.

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