Received August 30, 2017, accepted September 13, 2017, date of publication September 27, 2017, date of current version October 25, 2017.

Digital Object Identifier 10.1109/ACCESS.2017.2757267

INVITED PAPER

Review of Near-Field Wireless Power and Communication for Biomedical Applications

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This work was supported by the research fund of Signal Intelligence Research Center supervised by the Defense Acquisition Program Administration and the Agency for Defense Development of Korea.

ABSTRACT Near-field magnetic wireless systems have distinct advantages over their conventional far-field counterparts in water-rich environments, such as underwater, underground, and in biological tissues, due to lower power absorption. This paper presents a comprehensive review of near-field magnetic wireless power transfer (WPT) and communication technologies in a variety of applications from general free-space systems, to implantable biomedical devices we find of particular interest. To implement a fully wirelessly-powered implantable system, both high-efficiency power transfer and high-rate data communication are essential. This paper first presents the history and the fundamentals of near-field WPT and communication in free-space systems, followed by technical details for their specific use in implantable biomedical devices. Finally, this paper reviews recent advances in simultaneous wireless information and power transfer and highlights their applications in implantable biomedical systems. The knowledge reviewed in the paper could provide intuition in the design of various wireless and mobile systems such as wireless body area networks, small-cell 5G cellular, as well as in-body biomedical applications, especially for efficient power and data management and higher security.

INDEX TERMS Near-field wireless power, near-field wireless communication, biomedical applications, implantable device.

I. INTRODUCTION

Electromagnetic (EM) fields from an emitting antenna are generally divided into two regions, the near-field and the far-field. Although the boundaries between these regions are not sharply defined, they usually have distinctive characteristics [1]. While the far-field EM radiation dominates at greater distances, the non-radiative near-field behaviors of EM fields dominate when close to antennas or scattering objects. The world of mobile electronics has exploded recently, and there has been a dramatic increase in the level of interest in near-field wireless power and information transfer technologies, such as near field communication (NFC), radio frequency identification (RFID), and wireless power transfer (WPT) based on inductive coupling and magnetic resonance. These near-field technologies exploit the near-field magnetic field via inductive coupling between coils. RFID is usually used in access control and healthcare applications [2], while NFC is used for data exchange, fintech applications, and various smart devices [3]. WPT systems are typically applied in various mobile devices, home appliances, and electric cars [4].

Near-field magnetic systems have distinct advantages in lossy dielectric media such as water, underground, and biological tissues. In such environments, the conventional far-field EM technologies experience significant path loss as a result of high energy absorption in the medium [5], [6]. The near-field magnetic systems are more efficient than the conventional far-field counterparts for power and information transmission since they experience much less energy absorption in a lossy dielectric medium [7].

In the human body, high energy absorption also leads to more substantial heating in surrounding tissues, similar to the operating mechanism of a microwave oven [8], [9]. For this reason, conventional far-field EM technologies and near-field magnetic systems need to be operated under proper guidelines and regulations which limit exposure to radiated power, to ensure human safety [10]. Because of their lower path loss, the near-field magnetic systems can transfer power and information using less transmission power compared to conventional far-field EM technologies, while satisfying specific absorption ratio (SAR) constraints. Since near-field magnetic systems ensure better safety, they are more suitable for implantable biomedical devices.

Furthermore, the near-field magnetic systems are highly reliable in lossy dielectric medium [11]. Conventional far-field EM technologies exhibit variable channel conditions and propagation delays due to the inhomogeneous permittivity of the materials in transmission environments such as soil and the human body [12]. In contrast, near-field magnetic systems experience negligible channel variations even in an inhomogeneous lossy medium, because these materials have similar permeability [13].

Despite the above advantages, there have been significantly fewer works on near-field magnetic transmission technologies than those pertaining to conventional far-field EM technologies, partly because it is more difficult to conduct analysis and simulation without the far-field plane wave simplification. To facilitate advances in near-field wireless technologies, substantial research is necessary to develop theories, design methodologies, and measurement techniques.

To serve as a comprehensive reference and to inspire future research, this paper presents an overview of the near-field magnetic wireless power and communication technologies. Among other applications, this paper focuses on implantable biomedical applications which require simultaneous wireless information and power transfer (SWIPT). According to studies in [14], $8 \sim 10\%$ of the population in America and $5 \sim 6\%$ in industrialized countries have experienced benefits from implantable biomedical devices. Since the late 1950s and early 1960s, research related to implantable biomedical applications has been active and steadily growing [15]. Compared to wireless technologies in free-space, implantable biomedical devices have additional issues to deal with, including miniaturization to allow minimally invasive operation, biocompatibility, packaging, and hermeticity, in addition to the power absorption limits of the human body and other safety issues [14]–[17].

It is worth noting that near-field magnetic techniques are not the only methods by which implantable biomedical applications can send power and information. Besides far-field EM technologies, there are other applicable approaches, such as ultrasonic and optical methods [18], capacitive coupling [19], [20], intra-body communication (IBC) [21], and so on. However, these methods have their own limitations, which can be briefly summarized as follows.

1) The ultrasonic based communication methods can deliver data to a deep implantation site [18]. However, they have limited bandwidth below the 10 MHz carrier frequency and can only send data over a distance of a few cm, due to large absorption loss at high frequency [22]. The power reception also shows significant vulnerability to misalignment between transmitter (Tx) and receiver (Rx) [23].
2) Optical links cannot work deep inside of the body due to the high absorption factor [20], and the efficiency of power delivery is not sufficient to ensure reliable power for implantable biomedical devices.
3) Capacitive coupling is a good telemetry method for chip-to-chip and implantable biomedical applications, but it has an extremely short operating range (a few mm), which is a significant disadvantage [24].
4) IBC is suitable for transmitting data from implantable biomedical devices, however it is not appropriate for power transmission [25].

Among the various methods, the near-field magnetic systems are the most suitable for wireless power and data transmission for implantable biomedical devices, since they can efficiently deliver power and information without causing harmful effects to the human body. Nonetheless, the near-field magnetic systems also have a few limitations, such as low data rates, short transmission range, and sensitivity with respect to the relative position of Tx and Rx. These issues may not be critical in existing applications which only require low data-rate communications at short range. Furthermore, depending on applications, the short transmission range of the magnetic system may not be a disadvantage. Some researchers have reported the short range characteristic of the magnetic communication system to be a benefit in terms of achieving security and frequency reuse [26], [27]. However, those issues need to be solved to apply near-field magnetic technologies to various future applications which may require higher data rates, longer transmission distance, and mobility support. To overcome these issues, researchers have proposed a number of promising solutions, which are presented in the following sections.

To facilitate the design of implantable biomedical devices with both WPT and communication, this paper aims to present a comprehensive review of both near-field WPT and communication technologies, from fundamental principles to state-of-the-art results, for both free-space (Sections II and IV) and implantable biomedical applications (Sections III and V), respectively. Furthermore, in Section VI this paper discusses SWIPT in near-field magnetic systems for implantable biomedical devices. Fig. 1 provides an outline of the paper. Table 1 summarizes the review results of near-field magnetic communication and/or WPT for free-space and/or implantable biomedical applications. This paper

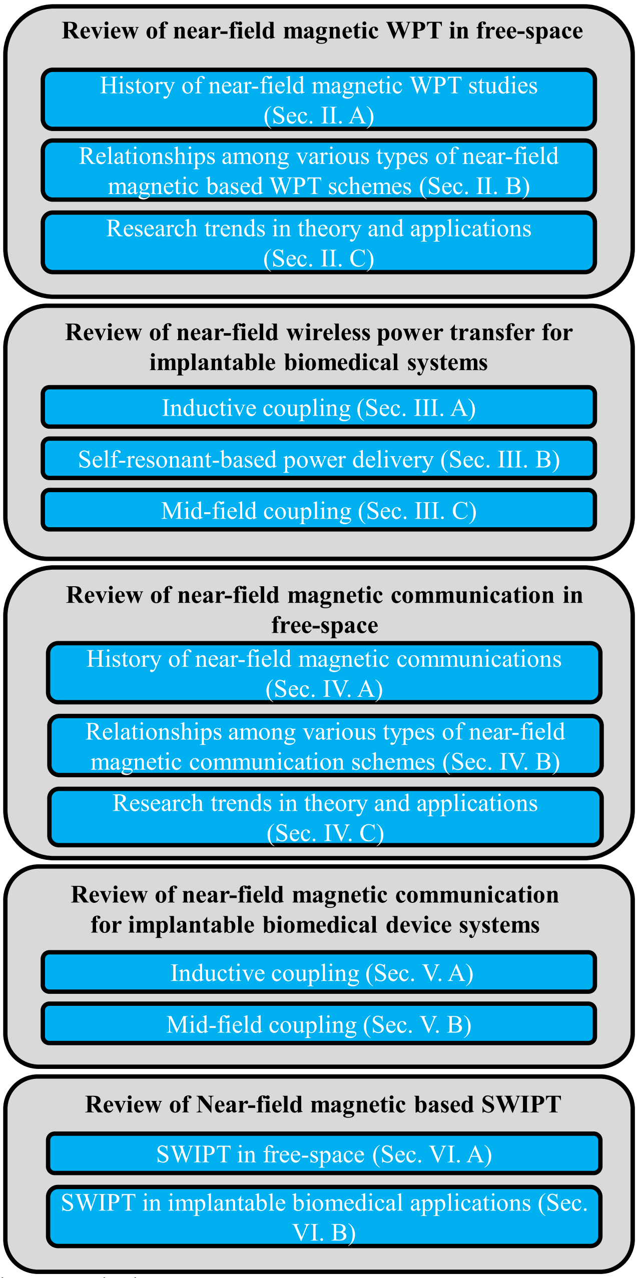

FIGURE 1. Organization of the paper.

differs from previous survey/review papers in the literature on near-field WPT and/or communications in two main respects:

1) Firstly, it presents a comprehensive overview of near-field WPT and communication technologies including their history, fundamental principles, recent trends, as well as specific discussions pertaining to both free-space and implantable biomedical applications.
2) Secondly, it reviews SWIPT by focusing on its applications in near-field magnetic systems for implantable biomedical devices.

II. REVIEW OF NEAR-FIELD MAGNETIC WPT IN FREE-SPACE

A. HISTORY OF NEAR-FIELD MAGNETIC WPT STUDIES

In 1865, Maxwell introduced what became known as Maxwell's equations, in which Coulomb's law, Gauss's law, Ampere's law, and Faraday's law were integrated into simultaneous equations to predict the existence of an electromagnetic wave [32]. In 1877, Hertz experimentally verified the existence of an electromagnetic wave [33]. High voltage was applied to a Tx dipole antenna with a spark gap. The electromagnetic energy arising from the Tx antenna was then propagated to a Rx antenna, which emitted sparks between the spark gap of the Rx antenna. It can be said that this was the first ever WPT experiment. In 1884, Poynting derived Poynting's theorem based on Maxwell's equation regarding the energy conservation of an electromagnetic field [34]. The Poynting vector indicates the energy density and direction of a propagating electromagnetic field. Energy flow in the near-field WPT is explained by the Poynting vector [35], [36].

Between the late 1890s and 1910, Tesla developed a high-voltage high-frequency generator to demonstrate the potential of WPT [37]. In 1926, the Yagi-Uda antennas were introduced [38]. Yagi-Uda antennas are now widely used as directional antennas, consisting of a fed radiator and non-fed directors and reflectors. The directors and reflectors are excited by the near-field of the radiator, and act as an array antenna. It can be said that power excitation from the radiator to the directors and reflectors in the Yagi-Uda antenna is a kind of near-field WPT. WPT systems using magnetic induction for devices implanted in the human body have also been investigated since the 1960s [39], [40]; details of these implant devices will be addressed in Section III. In 1969, Richard et al. published a book which discussed the relationship between circuit theory and field theory from the viewpoint of electromagnetic energy [41]. In 1978, a WPT system for moving vehicles with a powered roadway using magnetic induction was proposed [42]. In the next year, the system was experimentally tested [43]. Since then, WPT systems for moving electric vehicle have been studied continuously [44], [45]. Since the 1980s, magnetic induction WPT systems have been broadly introduced in home appliances, such as electric toothbrushes, shavers, and cordless phones. Starting in the late 1990s, SONY developed the wireless IC card "FeliCa", which employed magnetic induction as a power supply for IC chips and for data transfer purposes [46]. In 2007, the concept of "Self-resonant WPT" was proposed by an MIT group involving the fields of power electronics, magnetics, microwaves, and antennas [47], [48]. A photograph of the experiment, in which power was transferred through the members of the laboratory, caused a stir in the scientific and industrial world. It is remarkable that even in the traditional form of induction WPT before MIT, resonance capacitors were widely used for power factor compensation due to leakage flux [49]. From the viewpoint of

TABLE 1. Summary of existing survey in related areas.

Reference	Scope	Main contribution
[2]	RFID	i) A detailed overview of RFID, ii) research trends with related publications, iii) classification of RFID applications, and iv) potential and future RFID research directions.
[3]	NFC	i) NFC technology with various approaches including standards, ecosystems, business issues, applications and security issues and ii) future NFC research directions.
[13]	Magnetic communication with non-conventional media	i) A brief background of magnetic communication and its motivations, ii) various promising applications of magnetic communication, and iii) a first attempt to present the future direction of magnetic communication.
[4]	Wireless charging and network applications	i) The history of wireless charging, ii) various fundamental of wireless charging methods, iii) standards and implementations of wireless charging, and iv) practical strategies for network applications.
[14]	Implantable medical devices	i) Technologies of implantable medical devices from an engineering perspective including biocompatibility, package and hermeticity, structure design and delivery system, power management and wireless communication.
[15]	Implantable medical devices	i) The early history of implantable medical devices including basic circuits, ii) requirements, characteristics, and design approaches of implant medical devices, iii) challenging issues of implantable devices, and iv) research trends.
[17]	Implantable medical devices	i) Current situation and highlighted challenges of detail requirements of implantable medical systems including power supply, wireless communication, remote monitoring, bio compatibility, and infections for future advancements.
[29]	Modulation techniques for implantable devices	i) Various modulation and demodulation techniques of the existing wireless implantable medical devices, ii) current challenges and problems of the modulation technologies, and iii) suggestions and discussions for the progress of implantable medical devices.
[18]	Wireless communication for implantable medical devices	i) Various wireless data link methods, ii) consideration of simultaneous wireless power and data link including inductive coupling and RF link, and iii) example of related systems.
[30]	WPT for implantable medical devices	i) Wireless power telemetry with an emphasis on the inductive link, ii) corresponding power management circuits, and iii) recent related research trends such as energy harvesting.
[31]	WPT for implantable medical devices	i) Various WPT methods for implantable medical devices including inductive coupling at low-frequency, MHz range, and midfield powering at sub GHz, ii) practical applications, and iii) technical challenging issues.
[12]	RFID technologies for implantable devices	i) Sensing and communication requirements of implantable medical devices, ii) introduction of RFID based implantable devices, iii) miniaturization of RFID based implantable devices, iv) wireless powering using passive RFID-implant devices, and v) biomedical/therapeutic applications.
[16]	Wireless powering and data link for implantable medical devices	i) Basic design consideration for implanting wireless powering and data link including basic circuit models and ii) typical applications including detail block diagram of system.

impedance matching, the MIT scheme was equivalent to the commonly used transformer coupling model [50], [51]. In ''self-resonant based WPT'', however, by using self-resonant antenna instead of an LC resonator, a high quality-factor (Q-factor) was achieved. Q-factor is a dimensionless parameter which is defined as stored energy over energy loss in a period. A high Q-factor can greatly increase the power transfer efficiency (PTE) of a system. Recently, two WPT organizations have proposed WPT standards. The 'Wireless Power Consortium' (WPC) proposed a 'Qi' standard which operates at 100 $\sim$ 205 kHz [52]. Qi is the most popular WPT standard and has been adopted by various companies such as Nokia, LG electronics, HTC, SONY, and IKEA. Another standard organization 'AirFuel Alliance' was established to combine the 'Power Matters Alliance' (PMA) and 'Alliance for Wireless Power' (A4WP) [53]. The PMA standard operates at 277 $\sim$ 357 kHz which is a frequency range similar to Qi. A4WP standardized 'Rezence' at 6.78 MHz with the support of Witricity, Intel, Samsung, Qualcomm, AT&T, and Duracell. Qi and PMA use inductive coupling technologies, while Rezence technology is based on the self-resonant inductor.

B. RELATIONSHIPS AMONG VARIOUS TYPES OF NEAR-FIELD MAGNETIC BASED WPT SCHEMES

There are a few types of near-field magnetic WPT schemes and their relationships are depicted in Fig. 2. The most traditional near-field magnetic WPT scheme works by means of magnetic induction, shown as ''Type 1'' in Fig. 2. In this scheme, a resonant capacitor can be used for power factor compensation due to leakage flux. However, the load impedance is determined by the power demand of the load. As a result, when the transmission distance increases, the transmission efficiency decreases in accordance with the magnetic coupling coefficient k_m.

Another type of the near-field magnetic resonant WPTs is magnetic induction with complex conjugate matching, shown as ''Type 2'' in Fig. 2. To maximize PTE, an impedance matching network is introduced between the Rx coil and the load. Since coils are inductive in nature, resonance can be obtained by inserting a capacitor to form an LC tank, as shown in Fig. 3. In practical terms, the separation between Type 1 and Type 2 is not clear. For example, consider the near-field magnetic WPT for a moving vehicle [54]. In such schemes, the Rx device on the vehicle may have a resonant

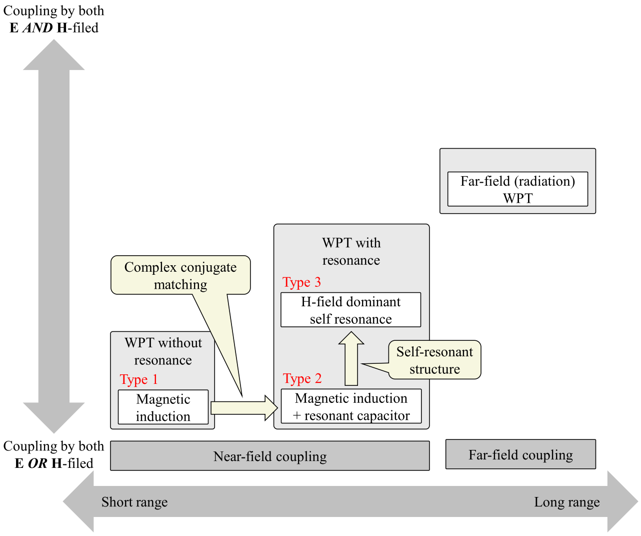

FIGURE 2. **Relationship among various kinds of WPT scheme.**

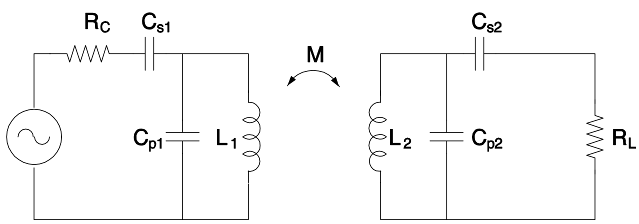

FIGURE 3. **Lumped circuit model for inductive coupling. The coils can be shunt tuned by inserting capacitors C_{p1} and C_{p2} in parallel of series tuned with C_{s1} and C_{s2}.**

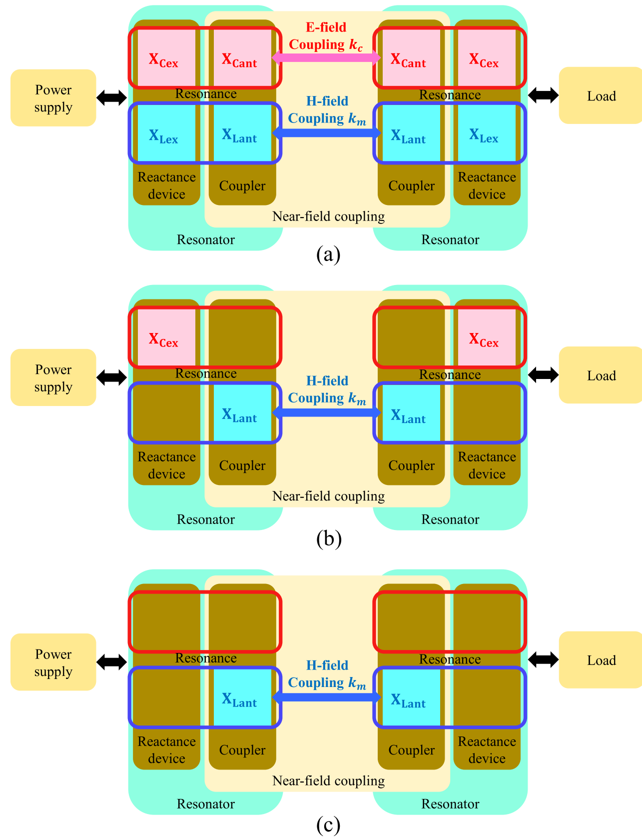

FIGURE 4. **Unified resonant-WPT model: (a) unified model, (b) magnetic induction with capacitor (Type 2 in Fig. 2), (c) self-resonant based WPT (Type 3 in Fig. 2).**

mechanism. However, the Tx devices under the road may not have a resonant mechanism because they commonly exist as just a pair of conducting transmission lines.

Another near-field magnetic resonant WPT is the self-resonant based WPT [55], which is "Type 3" in Fig. 2. As explained in the previous subsection, achieving a high Q-factor by using the self-resonant coil can increase PTE. This technique typically uses four coils, namely, driver, primary, secondary, and load coils, and its operation was initially explained by the coupled-mode theory [56] in [47] and [48]. In a self-resonant based power delivery system, extra coils on the driver and load side provide a way of tuning the input and output coupling, and therefore the input impedance and the output impedance, respectively. In other words, the extra coils play the role of an impedance matching network [57]. The PTE can be greatly enhanced because the Q-factor of the coils and the impedance matching network can be kept high in order to compensate for small mutual coupling k_m between the source and the load coils. When extra coils are used for impedance matching, self-resonant based power delivery can be understood as inductive coupling, since each pair of coils are coupled through the magnetic field. Therefore, the self-resonant based system can also be analyzed and optimized using self-inductance and mutual inductance in circuit theory [58], [59], which is generally more familiar to electrical engineers than coupled-mode theory.

Resonant WPT systems are based on electric-field (E-field) coupling and magnetic-field (H-field) coupling, and in order to explain them as an integrated system, a unified WPT model was proposed [60]. Fig. 4 (a) shows the unified model. The power supply feeds electric power to the Tx resonator, and power is extracted from the Rx resonator. Both resonators have a coupler and reactance device. The coupler is a device that generates a near-electric and/or magnetic field. The coupler includes a coupling coil and a self-resonant antenna. The reactance device is a device that is not intended to generate an electric and/or magnetic field. The reactance device includes a resonant capacitor and a matching inductor.

The coupler has inductive and capacitive reactance components which are expressed as $X_{C_{ant}}$ and $X_{L_{ant}}$, respectively. The reactance device has inductive or capacitive reactance component of $X_{C_{ex}}$ and $X_{L_{ex}}$, respectively. With regard to the resonance, the resonant frequency of the system is determined so that the reactive impedance of the resonator (i.e., the Reactance device and Coupler) becomes zero: $X_{C_{ant}} + X_{C_{ex}} = X_{L_{ant}} + X_{L_{ex}}$. In terms of coupling, E-field coupling occurs between the capacitive reactance of the Tx and Rx couplers with the electric coupling coefficient k_c, and H-field coupling occurs between the inductive reactance of the Tx and Rx couplers with the magnetic coupling coefficient k_m.

In a WPT system which uses a coupling coil and a resonant capacitor, as shown in Fig. 4 (b), the coupler mainly has inductive reactance, while the capacitive reactance of the coupler itself is quite small. Therefore, an external form of reactance is necessary to realize resonance. During the exchange of energy between the external capacitive reactance and inductive reactance, energy loss occurs, which reduces the Q-factor of the system, and hence degrades the PTE.

In a WPT system using a self-resonant antenna, as shown in Fig. 4 (c), the antenna itself has both inductive and capacitive reactance capabilities. Therefore, without using a resonant capacitor, resonance can be achieved. By employing a self-resonant scheme, a high Q-factor is achieved because the loss in the resonant capacitor is avoided. In the magnetic-coupling self-resonant scheme, the electric field is not negligible because of its resonant nature, as shown in Fig. 4 (c). Suppressing the electric field coupling by modifying the geometry of the self-resonant antenna can improve the Q-factor and extend transmission distance [61]. On the other hand, by utilizing electric field coupling, the transmission distance can be extended by using a repeater consisting of a single-wire conductor [62].

C. RESEARCH TRENDS IN THEORY AND APPLICATIONS

Achieving a maximum PTE is a priority goal in WPT systems where impedance matching and coupling coefficient are the key parameters of high PTE. Most early studies of WPT systems tried to achieve high PTE by assuming that the Tx coil and Rx coil needed to be closely located and perfectly aligned. Reference [63] introduced the relationship between port impedance and coupling coefficient, and the change in image impedance patterns, to provide in-depth insight into system performance. Reference [64] showed the maximum transmission efficiency of a two-port network, which represents a pair of couplers. A theoretical analysis based on an antenna and a matching circuit approach [65], an equivalent circuit approach [66], and field theory [35] were also conducted.

Besides the studies on high PTE for a given limited transmission range, research groups also attempted to extend range and support mobility with maximum PTE. Although the transmission range of inductive coupling based magnetic WPT systems are usually smaller than the Tx coil dimension, [67] introduced self-resonant based WPT systems for distances longer than the Tx coil dimension using a high Q-factor. Reference [68] presented a resonant WPT system using an intermediate perpendicular resonant coil between the Tx coil and Rx coil. Reference [69] developed a dipole coil resonance system (DCRS) with a ferrite core for a stronger and wider magnetic field, enabling 209W at 5 m with a 20 kHz carrier frequency.

To support mobility, unpredictable loads and changes in the Rx position have to be addressed. References [70] and [71] proposed an analytical model of a near-field inductive coupling magnetic system and magnetic resonance system to compensate the effect of misalignment. To make near-field magnetic WPT systems robust and reliable, two approaches have generally been employed: robust structure and electrical tuning.

The first approach utilizes robust structures, which can prevent or compensate the efficiency degradation caused by an unpredictable position change, by exploiting 3-dimensional (3D) coils, antenna arrays, and metamaterials. The 3D coil structure can prevent a reduction in PTE caused by rotation misalignment due to omnidirectional magnetic fields [72], and the Tx coil array structure produces a broader magnetic field, which can provide energy to the Rx coil even with lateral misalignment on the array structure [73], [74]. Asymmetrical Tx and Rx coils can also increase the transmission coverage [75]. Array structures are popular and can produce a uniform magnetic field in an appointed area using antiparallel resonant loops [76], multi-loops having the same axis [77], and overlapped arrays [78]. Reference [79] considered move-and-charge schemes which were combined with a Tx array structure and omnidirectional Rx coil to prevent cases of both lateral and rotation misalignment. Recently, some research groups have adopted metamaterials to extend the operating range of near-field magnetic systems [80] and to compensate for the losses that occur with misalignment [81].

The second approach is based on electrical tuning to adjust the resonant characteristics. Efficiency can be degraded by mismatched conditions, such as variable mutual inductance depending on relative position, and changing loads. To compensate, its critical to dynamically adapt operating conditions and power transfer ranges to these changes. For example, methods to dynamically adjust operating frequency, the input and output impedance of the coils, and the load impedances of the rectifier are discussed in detail in [82] and [83]. Reference [84] compensated the decreased power efficiency via displacement using frequency shift with an electrically tuned Class-E inverter. They achieved a nearly constant PTE within a certain coupling coefficient range. Tuning can be also applied using a matching network at a given operating frequency. Reference [85] solved PTE degradation by frequency splitting with critical coupled systems by adjusting the capacitance of the resonator. They verified the proposed method via testbed, by applying offline tuning. Reference [86] presented an automated impedance matching system using switching capacitor circuitry although it still had unsolved issues, such as the power consumption of the feedback loop and high system complexity.

Magnetic beamforming can simultaneously achieve the goals of both long transmission range and high PTE with mobility. To the best of the authors' knowledge, the studies about magnetic beamforming techniques are very recent trends in the communication and WPT area. In the first study on magnetic beamforming in 2014, [87] proposed a Tx coil array which could efficiently transmit power to a Rx about 50 cm away from the Tx device, without alignment by phase control. In 2015, a power transfer method for powering up to 6 different devices was proposed by the same research group [88]. Reference [89] conducted a coupling and circuit analysis of the Tx array components to prevent power leakage. They presented a theoretical analysis and well matched measurement results with various coil arrangements. Reference [90] proposed a beamforming algorithm for maximum PTE with an orthogonal 3D coil based Tx coil array. The aforementioned studies considered a single Tx device and multiple Rx devices. Reference [91] presented a multi-input single-output (MISO) WPT using signal processing and optimization methods with perfect parameter information between the Tx coils and Rx coil. They showed a near-optimal solution using the proposed non-convex system mode. This work was later extended to the general setups with distributed transmitters with optimized locations and general magnetic multiple-input multiple-output (MIMO) WPT systems with arbitrary number of transmitters and receivers in [92] and [93], respectively.

[94] introduced a magnetic beamforming system from a different point of view, which considered the security of the WPT system to prevent unwanted power transfer to unintended Rx. Issues still remain in the magnetic beamforming systems, such as power leakage between the Tx array components, the highly complex computations needed to determine the optimal current of each transmit coil, and estimate the system parameters. However, there are clues to solve these issues. Reference [89] reported enhanced performance with reduced leakage, and [90], [91] mentioned the linear relationship between the optimal current of each Tx coil and the mutual inductance between each Tx coil and Rx coil in particular cases. Reference [95] presented a non-coupling coil pattern to mitigate the power leakage and computation complexity issues. A non-coupling coil can reduce the power leakage in the Tx array, since the coil pattern reduces coupling between the Tx array components. Reducing power leakage can lead to better PTE and longer transmission distance than can be obtained with conventional coupling coil pattern systems. Additionally, reducing coupling can avoid the complicated calculations needed to determine the optimal current of each Tx coil, by using the linear relationship with mutual inductance between each Tx coil and Rx coil.

WPT technologies are promising for various applications including mobile devices, home appliances, sensor networks [96]–[98], chip-to-chip applications [99], vehicles [100] and medical applications. To apply WPT technologies to these applications, challenging issues still need to be worked out, such as EMC/EMI issues to meet safety regulations [101], [102], RF-to-DC conversion efficiency, in addition to the issues mentioned in this paper.

III. REVIEW OF NEAR-FIELD WIRELSS POWER TRANSFER FOR IMPLANTABLE BIOMEDICAL SYSTEMS

Although methods of powering implantable biomedical applications can basically employ the same technologies employed in free-space systems, they also need to consider the following issues. a) To be minimally invasive, the size of the implant needs to be highly restricted. b) Human tissue, through which the electromagnetic wave propagates, is a lossy medium. This leads to additional power losses, as heat in the tissue, and degrades the overall PTE. c) Safety regulations that address tissue heating by external electromagnetic sources place a certain limit on the potential amount of power that can be delivered [16]. Considering these factors, various design parameters such as operating frequency should be revisited for applications with biomedical purposes.

We first review the mechanisms of inductive coupling and magnetic resonance for powering implantable devices. These methods are as widely used for the wireless powering of implantable devices as they are in free-space. To make the survey more comprehensive, this section also introduces another method, mid-field coupling. When the Rx device size is much smaller than the separation between the Tx and Rx device, researchers have proposed operating at higher frequencies, in the mid-field range, where both reactive and radiative energy can be utilized [103].

A. INDUCTIVE COUPLING

The most popular technique for delivering power wirelessly to biomedical implants is inductive coupling, which was first used to power an artificial heart [104]. Since then, there have been several comprehensive analyses of inductive links over tissues [105]–[108]. Table 2 lists the parameters of the inductively coupled systems described in the literature [109] although this list is not exhaustive.

TABLE 2. Frequency, Range, and coil size of select implants.

Implant type	Carrier (MHz)	Range (cm)	Source coil (cm)	Receiver coil (cm)
Generic [111]	1-2	0.5	1	0.47
Generic [112]	4	0.8	-	-
Generic [113]	10	1.5	3.5	2.7
Cortical visual [114]	5, 10.33	1	-	-
Retinal [115]	1-10	-	-	-
Retinal [116]	2.5	0.575	1.34	1.34
Muscular [117], [118]	2	-	9-20	0.12
Muscular [119]	6.78	-	-	-
Muscular [120]	2	-	-	-
Bio-sensor [121]	27	9	-	5

To maximize PTE, impedance matching networks and coupling are as important in implantable biomedical applications as they are in free-space applications. However, the limitations of implantable devices introduce significant obstacles to realizing a proper impedance matching network. To compensate the improper Rx coil inductance resulting from its limited size, inductance tapping and voltage doubler circuits have also been considered for impedance matching [106]. In addition, ferrite cores [39] and efficient rectifying circuits [110] have been employed to enhance coupling between the Tx coil and Rx coil. Table 3 lists the different tuning configurations utilized to achieve high PTE for each application.

Implantable biomedical applications also have misalignment issues similar to those encountered in free-space, due to the highly curved anatomical surfaces and relative movement of organs during daily life [131]. Many techniques have also been studied to improve the robustness of the link, and these

TABLE 3. Tuning configurations of select studies.

Reference	Transmit tuning	Receive tuning	Resonance
[122]	-	shunt	unloaded
[124]	series	shunt	loaded
[125]	shunt	shunt	unloaded
[107], [126]	stagger	stagger	-
[127]-[129]	series	series	-
[130]	series-shunt	shunt	-

are summarized in [109]. In [123], the dependency of the efficiency on the coupling coefficient k_m was reduced by operating at the optimal voltage transfer ratio. A similar idea was proposed in [124], where k_m maximized trans impedance to desensitize its variation with k_m. In [107] and [125], the voltage transfer ratio was desensitized to variation in k_m by staggered tuning circuits. Using coils with free-running oscillation, it was shown that the efficiency is independent of coupling if k_m is greater than the inverse of the loaded Q-factor of the Rx coil [126]–[129]. In [130], a circuit approach was described to improve the tolerance to coupling variation, using coils at self-oscillation. In [132], the design parameters for planar spiral coils were optimized to maximize PTE. [133] implemented soft and flexible coils using a liquid metal alloy for more comfortable implants, with comparable system-level performance as metal based counterparts. In most of these techniques, the idea of using resonant LC tanks on both coils to enhance the efficiency is widely embraced [47], [83], [134].

B. SELF-RESONANT BASED POWER DELIVERY

Resonant based power delivery has been applied in implantable and wearable devices [71], [135], [136]. Reference [71] used multiple inductor coils of a few centimeter size for coupling and matching purposes. To achieve high PTE in a deeply implanted location, a high Q-factor is required in implantable biomedical devices. Since the Q-factor of those coils (a few hundreds) is much higher than that of discrete inductors (a few tens) at corresponding frequencies, the efficiency reported in [71] can be much higher than that of a system using matching networks made of discrete LC components.

Another benefit of using a multi-coil solution is that it provides designers with more degrees of freedom to optimize the inductive link [58], [136]. Spacing between the primary and secondary coil can be adjusted to control the input impedance.

Exploiting this feature, a three-coil link was also proposed in [136], to achieve both high efficiency and power delivery. However, having extra coils generally requires larger sized implants and more complex designs, and introduces several design constraints on the inductor geometry [137]. Moreover, it is very difficult to fabricate coils with high Q-factors within a scale of a few millimeters. As a consequence, self-resonant based coupling has not been adopted for powering millimeter-sized miniature implants. To operate implants whose sizes are much smaller than the distance from the Tx, it is advantageous to use mid-field coupling, which exploits a higher frequency range [103], [138]–[140].

C. MID-FIELD COUPLING

Microwave operating frequencies can be used to maximize PTE. As frequency increases, electromagnetic waves attenuate faster in tissue because the skin depth is reduced. The antenna efficiency of an electrically small Rx, however, generally increases with frequency. One can therefore expect an optimal frequency at which PTE is maximized for a given Rx structure. Specifically, when the Rx was limited to millimeter-size and subjected to the constraints of a realizable range of Q-factor, it was shown that the optimal frequency lies in the low-GHz range [88]. At low-GHz frequency, a power link with a typical source and Rx separation in an implantable system corresponds to mid-field operation, where both reactive and radiative modes of fields can be utilized to power up the Rx [141].

When the powering system operates at mid-field, the source structure can be tuned to maximize efficiency, similar to an antenna array concept operating with a far-field antenna. Unlike a far-field operation, however, the purpose is not simply to control the direction of plane wave propagation. Instead, the purpose is to produce a field distribution in which the electric field causing the heating of tissue is minimal per given energy delivery to Rx. The source structure generating the minimum dissipated power in tissue was analytically found in [142] and [143].

The starting point of such an analysis is to model the human body as a multilayered structure (Fig. 5 (a)), where Green's function is analytically known. The thickness and the dielectric property of each layer follows the geometry of the human body and the Debye model [144]. The source is modeled as a current sheet above the tissue to represent an arbitrary source, invoking the equivalence principle in electromagnetic theory. Fig. 5 (b)-(e) shows the solution for multilayers approximating the chest wall – skin, fat, muscle, bone, and cardiac tissue – at a maximum operating point (2.6 GHz) compared to a coil source. The optimal current density, shown in Fig. 5 (c), consists of alternating current paths spaced approximately every half wavelength. The current paths propagate inwards, generating fields that exhibit a focusing effect, as shown in Fig. 5 (e). In contrast, the field generated by a coil source shows divergent power flow lines, as shown in Fig. 5 (d).

The feasibility of operating millimeter-sized cardiac implants was verified by experiment in [140]. A source resembling the optimal current distribution was fabricated with metal slots. Four microwave ports were used to excite the structure. A microelectronic stimulator, about 2 mm in diameter and 3 mm in length, weighting 70 mg, was fabricated with an Rx coil of about 2 mm diameter attached. The device reports the power received based on the pulse rate of a light emitting diode, which can be calibrated precisely to the power transferred to the coil [143]. With 500 mW of

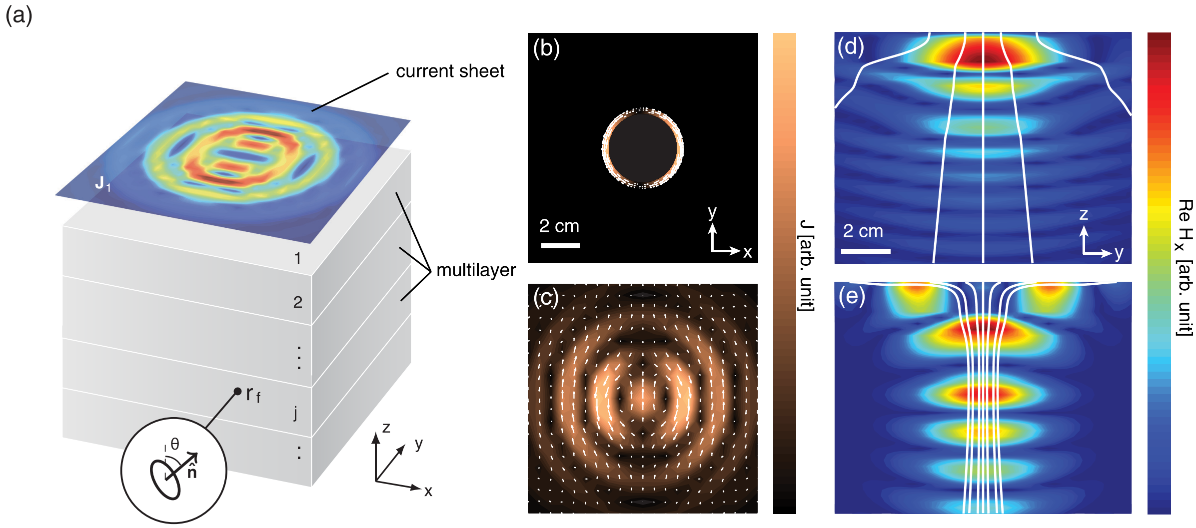

FIGURE 5. **(a) Multilayer tissue model and a current sheet above it to represent arbitrary source. (b-c) Current distribution of the coil source (top) and the optimal source (bottom) at 2.6 GHz. (d-e) The magnetic field component aligned with the receiver dipole moment $\hat{x}$ and the Poynting vector (white) generated by the coil source (top) and the optimal source (bottom) at 2.6 GHz.**

output power from the source, 200 μW of power could be transferred to the device. Considering that modern pacemakers require just 8 μW to operate [144], this is more than sufficient to perform various advanced electronic functions. The resultant PTE also approached the value derived from theory.

IV. REVIEW OF NEAR-FIELD MAGNETIC COMMUNICATION IN FREE-SPACE

As previously noted, near-field magnetic communication technologies, which have the same principles as near-field WPT technologies, are types of short-range communication technologies based on magnetic fields, such as RFID, NFC, and payment systems. The main concern of near-field magnetic WPT technologies is the PTE. However, the primary goals of modern near-field communication technologies are throughput and reliability. In this section, we present the history of typical near-field magnetic communication schemes, and introduce the principles of near-field magnetic communication and various related technologies, followed by descriptions of technological issues and recent trends.

A. HISTORY OF NEAR-FIELD MAGNETIC COMMUNICATIONS

RFID is the oldest standard application to employ near-field magnetic communication technologies. In 1948, Harry Stockman published the landmark paper ‘‘Communication by means of reflected power’’ enabling communications using reflected power [145]. In the 1960s, RFID commercial activities began with a few established companies, and an electronic article surveillance system (EAS), the first wide-spread commercial application of RFID, was presented to the world to prevent theft. This system, termed a one-bit tag, can only detect the presence or absence of the tag.

In the 1970s, more work on RFID was conducted by many companies, academic institutions, and government laboratories. In 1975, Alfred Koelle, Steven Depp, and Robert Freyman presented an important result in a paper entitled ‘‘Short-range radio-telemetry for electronics identification using modulated backscatter’’ [146], which is the starting point of a completely passive tag with a transmission range of tens of meters. After this paper, General Electric, Westinghouse, Philips, and Glenayre tested this passive identification tag system at the Port Authority of New York and New Jersey. Based on an RFID system consisting of a microwave system and an inductive system, various applications became widespread, including animal tracking, vehicle tracking, and factory automation. Further size reductions and functional improvements were achieved by utilizing low-voltage and low-power complementary metal-oxide semiconductor (CMOS) logic circuits in the 1970s.

In the 1980s, RFID applications were rapidly expanded with the development of the personal computer (PC), since the PC can provide a convenient and economical means of collecting and managing data from RFID systems. At the same time, various nations became interested in different applications of RFID. Transportation and personnel access applications drew more interest in the United States (US), while Europe was more interested in short-range systems for animals, industry, business, and electronic toll collection applications.

In the 1990s, the US installed more than three million RFID tags for electronic toll collection purposes in North America. Europe applied RFID tags for electronic toll collection, access control, and a wide variety of other commercial applications using both microwave and inductive technologies. These electronic tolling applications were installed in Australia, China, Hong Kong, Philippines, Argentina, Brazil, Mexico, Canada, Japan, Malaysia, Singapore, Thailand, South Korea, South Africa, and in other countries. These types of electronic toll collection systems were the first commercially successful application of RFID technology [147].

Currently, RFID is used in various commercial fields including the airline industry, railways, manufacturing, agriculture, hospitals, livestock, supply chains, food, payment applications, tracking applications, and identification applications [2]. RFID is also now becoming more popular for short-range interactions in the Internet of Things (IoT) [148], [149].

However, RFID cannot support peer-to-peer communication, which requires initialization of communication at either end by both parties. NFC is one of the alternative near-field magnetic communication methods to RFID, including peer-to-peer communication. NFC technologies were developed by Philips and Sony jointly in 2002. These were based on RFID and contactless smartcard technology for contactless communications. In 2004, ISO/IEC 18092 "Near field communication – interface and protocol (NFCIP-1)" came into effect [151]. In 2007, Nokia 6131 NFC-enabled handsets and NFC tags were supplied to 500 selected subscribers by Innovation Research & Technology in the United Kingdom (UK). This was the first consumer trial of NFC technology. In 2010, Samsung provided the first Android phone, Nexus S, which included a NFC function. In 2011, Google demonstrated NFC for gaming and sharing a contact, URLs, apps and video, and Research in Motion presented the first device to include the PayPass service, the payment service of MasterCard which can use NFC. In 2012, the first NFC-enabled 'smartposter' was used in a campaign by the UK restaurant chain EAT and Everything Everywhere. In 2013, Samsung and VISA announced a partnership to develop mobile payments using NFC. In 2014, AT&T, Verizon and T-Mobile released the NFC payment service Softcard with Smartphone, and Apple applied their NFC payment service, Apple Pay, to their devices [151]. In 2015, Samsung released Samsung Pay, which can support both magnetic card and NFC payment technology [152].

RuBee (IEEE 1902.1) is another subset of RFID. RuBee shares the same operating principle of the other near-field magnetic communication systems. However, RuBee is more similar to WiFi or Zigbee than RFID and NFC in terms of protocol. The RuBee protocol was designed for real-time asset visibility networks in harsh environments by Visible Assets. A RuBee work group (IEEE 1902.1) was formed in late 2006 with 17 corporate members, and the Food and Drug Administration (FDA) classified RuBee as a non-significant risk (NSR) class 1 device in medical visibility applications in the same year. In 2009, a final specification of RuBee was issued as an IEEE standard [153].

Magnetic field area network (MFAN) is the latest scheme employing near-field magnetic communication, using a 128 kHz carrier frequency for underground applications, such as underground state monitoring and earthquake management, underground facility management, building and bridge monitoring, and environmental pollution surveillance [154]. The MFAN standard ISO/IEC 15149 includes an Air interface, in-band control protocol for WPT, relay protocol for extended range, and security protocol for authentication.

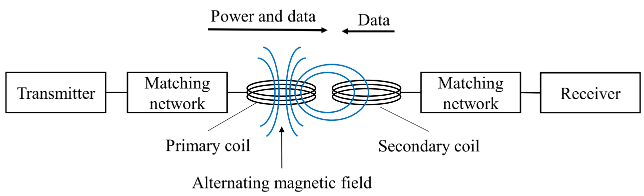

FIGURE 6. Simplified concept of inductive coupling.

B. RELATIONSHIPS AMONG VARIOUS TYPES OF NEAR-FIELD MAGNETIC COMMUNICATION SCHEMES

Before explaining the differences of the above-mentioned near-field magnetic communication schemes, we need to talk about the basic principle of near-field magnetic communication technologies. This principle is commonly applied to all of the types of near-field magnetic systems described in Fig. 2. Near-field magnetic communication systems are operated in the near-field region. A primary coil generates a magnetic alternating field. When a secondary coil is placed in the near-field of the primary coil, these coils are coupled by inductive coupling, and energy from the alternating field can be induced in a secondary coil according to Faraday's law. This is the basic principle of magnetic systems, as shown in Fig. 6. Additionally, if both the primary coil and secondary coil have identical resonance frequencies, the resonance circuit of the secondary coil produces sympathetic resonance. Current in the secondary coil then flows against the external magnetic alternating field, affecting the external magnetic alternating field [155]. Using this phenomenon, a passive inductive coupling based RFID tag can obtain power and can transmit data streams to an inductive coupling based RFID reader. The following schemes all employ the same inductive coupling based RFID, although there are a few differences in other components, such as network topologies and protocols.

NFC is a short-range half-duplex communication protocol which uses inductive coupling and an operating frequency of 13.56 MHz. It is commonly used with NFC mobile devices, NFC tags, and NFC readers, and the operating modes of NFC include the reader/writer mode, card-emulator mode, and peer-to-peer mode, as shown in Fig. 7 [3]. In the reader/writer operating mode, an active NFC mobile can read and write the data stored in the NFC tag. This mode follows ISO/IEC 14443 Type A, Type B, and FeliCa, and data exchanges are performed according to the near field data exchange format (NDEF) from the NFC Forum. In the card-emulation mode, NFC readers and NFC mobiles act as traditional RFID readers and as a contactless, read-only RFID smart card based on ISO/IEC 14442 Type A, Type B, and FeliCa. The NFC technology can be integrated with existing types of RFID infrastructure using the card-emulator mode [3], [151].

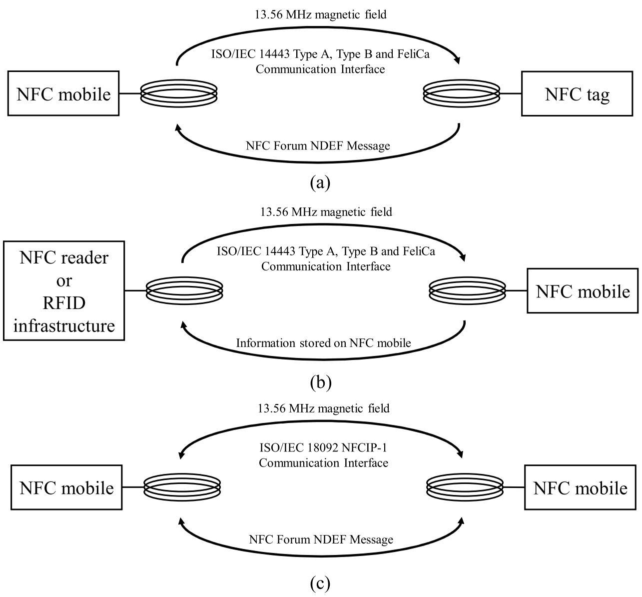

FIGURE 7. Operation mode of NFC: (a) read/write mode, (b) card emulator mode, (c) peer to peer mode.

The peer-to-peer mode is unique to NFC and unlike RFID, since RFID can only read data from a card or a tag. In the peer-to-peer mode, two NFC mobiles can exchange information in both directions according to ISO/IEC 18092 as NFCIP-1. The NFCIP-1 protocol basically operates a "request-response model" and includes an error handling function using the acknowledge (ACK) character frame.

RuBee operates at 1 to 50 feet (0.3 to 15 m) with a long wavelength (131 kHz) for harsh-environment and high-security asset-visibility applications. RuBee does not use a reflected signal for backscattering, but actually transmits a data signal on demand, similar to WiFi and Zigbee, with a slow data rate (1,200 baud) and a small packet size. Therefore, RuBee cannot provide a high data rate, however, by using a magnetic field with an electrically small size and low power operation, RuBee can guarantee high security communication to eliminate the risk of eavesdropping [157. According to a peer-reviewed study of the Mayo Clinic, RuBee does not affect implantable biomedical devices such as pacemakers or interplantation cardioverta defibrillator (ICD)s. Furthermore, RuBee does not create EMI or EMC problems in the operating room [158]. Therefore, RuBee is suitable for near-field short range magnetic communications that require long battery lifetimes, safety, and high security with an omnidirectional magnetic field.

MFAN is unlike the other near-field magnetic communication approaches in terms of network topology. Other near-field magnetic communication systems support a broadcast request- response or peer-to-peer method. MFAN can establish a centralized network which is similar to Zigbee, consisting of one MFAN-coordinator (MFAN-C), multiple MFAN-nodes (MFAN-N), and an MFAN-repeater (MFAN-R). The role of MFAN-C is access control and time resource management as the center of the network. MFAN-R can compose a small network with MFAN-Ns which are located outside the range of MFAN-C, to expand the network. Every node can operate as an MFAN-C, MFAN-R, and MFAN-N [154].

C. RESEARCH TRENDS IN THEORY AND APPLICATIONS

The primary challenges of RFID and NFC are usually high cost, privacy, security, and recognition issues [3], [147], [148], [159]. However, this paper focuses more on the fundamental issues of the near-field magnetic systems, such as limited channel capacity in free-space use, rather than these standard-specific issues.

Before introducing specific high data rate schemes, we briefly discuss the theoretical basis of path loss, link budget, and channel. Akyildiz provided the path loss of a magnetic induction (MI) waveguide for underground applications [160]–[162], and Agbinya provided a link budget model [163]–[165] power equation and the channel capacity of magnetic communication systems in various magnetic relay configurations [166]. Reference [7] provided clear theoretical demonstrations of the benefits of magnetic communication systems in the near-field with lossy mediums which have high permittivity and conductivity, such as the human body.

To enhance channel capacity or throughput, there have been numerous studies involving many aspects, e.g., tests of different modulation schemes or the use of advanced structures such as MIMO and relays. The general modulation schemes used in conventional EM communication systems, such as amplitude, frequency and phase modulation, ranging from binary to higher-order encoding modulation, can be also applied to near-field magnetic communication systems [167]. Amplitude shift keying (ASK) or amplitude modulation (AM) schemes are the most common, and rely on varying the amplitude of the signal. ASK can emulate the zero/one logic of digital communication, as the modulation depth of ASK can be controlled up to 100% from 0%. When the modulation depth is 100%, it is called on-off keying, which is a special case of ASK. FSK or frequency modulation (FM) modulates the data by altering the frequency of the carrier signal. Phase shift key (PSK) or phase modulation methods send the data by changing the phase of the carrier signal. ASK and FSK are relatively simple schemes for data transmission. However, some applications of near-field magnetic communication systems require low power or powerless systems with simple architecture. Load shift keying (LSK) or load modulation (LM) schemes identify the reflected impedance in the Tx coil via the changing load impedance of Rx, and can thus reduce power consumption. This back telemetry link can easily deliver simple information, like RFID. LSK schemes have advantages in terms of power consumption and hardware complexity. However, these schemes cannot support high data rates. Several research groups have applied carrier-less pulse modulation (PM) based methods, which have been adopted from the impulse radio-ultra wideband (IR-UWB) schemes of conventional EM communication systems, to near-field magnetic communication systems to obtain wide bandwidth, low hardware complexity, and low power consumption [168], [169]. Reference [170] presented a review of the PM based near-field magnetic communication systems to compare with carrier based EM communication schemes, and capacitive coupling methods. In addition, [170] reported useful information about the PM based near-field magnetic system, such as the relationship between distance and data rate, crosstalk, and constant magnetic field scaling. In PM based systems, data is transmitted via sub-nanosecond pulses with wide bandwidth, and the PM based methods are usually applied to chip-to-chip communication and implantable devices with proximity distance. The bandwidth of the Tx coil and Rx coil should be two times higher than the signal frequency to suppress inter-symbol-interference (ISI), which is caused by a ringing signal from the Rx. Reference [169] presented PM based multi-channel near-field magnetic systems to achieve high data rates in a 4-stacked system in a package. The system inevitably had crosstalk issues between array components. To reduce the crosstalk, the system used the time division multiple access (TDMA) technique between adjacent array components. Reference [171] used PM methods with the H-bridge architecture to achieve high data rates for mobile applications. However, the H-bridge requires high power consumption. To reduce the H-bridge power consumption, the system applied transmission time control. The proposed system also introduced reliable communication by using an additional amplifier and differential detection method, which can make a decision regardless of signal strength when the system receives a weak signal due to large distance and misalignment. However, to achieve wide bandwidth the proposed methods require a low Q-factor, which decreases transmission range and SNR. Therefore, these methods may not be suitable for implantable biomedical applications which require a high Q-factor to receive enough power for reliable operation [172]. In Section V, we introduce implantable biomedical devices including PM based near-field magnetic communications with high Q-factor.

Other research groups have tried to achieve a high data rate by using special coil structures such as relay coils and coil arrays. Akyildiz, Agbinya, and Masihpour reported magnetic relay based approaches, which are promising solutions to obtain extended range and enhanced channel capacity, respectively. Reference [11] proposed a spread resonance (SR) strategy which can achieve increased channel capacity with an MI waveguide, which was proposed by Solymar [173]–[176]. In this strategy, a resonant frequency of each relay coil is deviated from the original central frequency. This enhances bandwidth and reduces path loss, whereas the SR strategy leads to increased system complexity. They realized a magnetic communication system, which included the MI waveguide system and underground testbed, using software defined radio (SDR) [177]. Furthermore, they included measurement of the 3D coil to prevent rotational

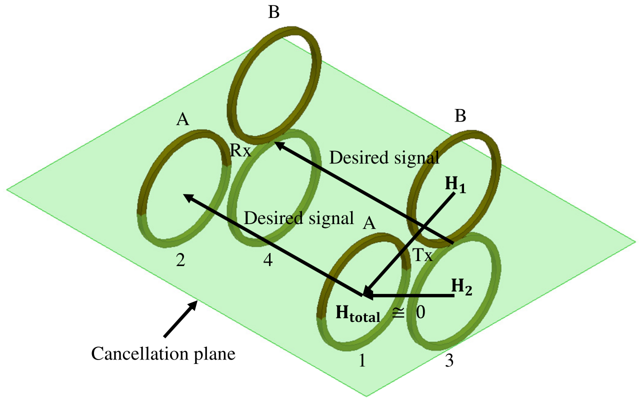

FIGURE 8. **The heterogeneous multi-pole loop antenna array MIMO is composed of circular loop antenna and quadrupole loop antenna [184].**

misalignment, and applied this omnidirectional coil to underwater communication systems [178]. Reference [179] showed the investigation results of various coil relay configurations, which were derived using a detailed theoretical analysis. The study showed variable received power with various magnetic relay methods, which were dependent upon excitation methods, to find a better range extension method. Reference [180] presented a multi-hop relay method for range extension and enhanced channel capacity in non-line of sight conditions.

The magnetic relay methods are promising solutions for achieving high data-rates and range extension in acceptable situations that allow additional relay coils. In peer-to-peer applications, on the other hand, MIMO is a promising method for increasing channel capacity. Reference [181] presented a link budget of various antenna configurations, such as single-input single-output (SISO), single-input multi-output (SIMO), multi-input single-output (MISO), and MIMO, without considering crosstalk between array components, and they proposed another multi antenna method using a frequency splitting phenomenon [182]. They showed the potential of frequency splitting MISO or SIMO with a linear binary chirp modulation in the strongly coupled region via hardware, using a field-programmable gate array (FPGA). Reference [183] also demonstrated the potential of a 3×3 magnetic MIMO configuration with multiple coils, which were arranged orthogonally to each other. Reference [184] proposed another magnetic MIMO configuration using a heterogeneous multi-pole loop antenna array with a crosstalk cancellation effect, as shown in Fig. 8. This configuration obtained parallel multiple streams without complicated signal processing methods, which are required in the conventional RF MIMO system. With quadrupole loops, antenna B can generate a cancellation plane, which is composed of cancellation points of the opposite magnetic fields H_1 and H_2. When circular loops, antenna A is located in the cancellation plane, and there is no crosstalk between the quadrupole loops and circular loops, as shown in Fig. 8. They also presented a testbed with the magnetic MIMO using SDR [185].

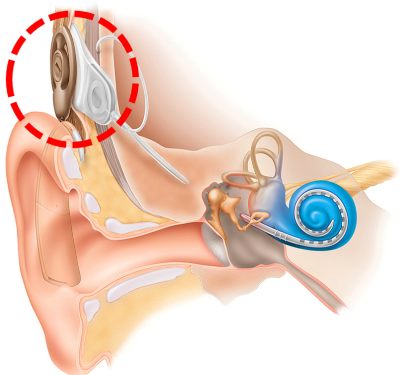

FIGURE 9. Cochlear implant: the part shown in dotted line is coil part for wireless power and communication.

V. REVIEW OF NEAR-FIELD MAGNETIC COMMUNICATION FOR IMPLANTABLE BIOMEDICAL SYSTEMS

Various wireless communication technologies have been applied in implantable biomedical devices, as mentioned in the introduction. Among these technologies, near-field magnetic communication systems are promising because they provide the advantages of near-field magnetic technologies. Fig. 9 shows a cochlear implant which is one of the most successfully commercialized implantable devices using the near-field magnetic technologies. As we mentioned in Section III, the near-field magnetic technologies used in free-space can be applied to implantable biomedical applications, as lo ng as several considerations of implantable biomedical devices are taken into account, such as size reduction for minimal invasive operation, biocompatibility, packaging, and hermeticity. Additionally, low data-rate issues are also significant in implantable biomedical applications, because of the high data-rate demands of neural prosthesis applications, which aim to achieve performance similar to human organs, such as cochlear, retinal, and various other neural systems. Cochlear implants and visual prostheses require a large volume of data transmission between external sensors and the implantable biomedical devices, and an invasive brain-computer interface (iBCI), such as prosthetic limbs need to transmit a large amount of data from the nervous system and a sensory system to external devices [186]. For example, to obtain resolution similar to the human eye, a visual prostheses requires about 40 Mbps to support 600∼1000 electrodes [123]. However, a conventional visual prosthesis can only support 60∼100 electrodes. To satisfy the data rate requirement of these promising applications, we can apply the capacity enhancing methods mentioned in Section IV. This section reviews the trends in near-field biomedical communication technologies in terms of inductive coupling and mid-field coupling.

A. INDUCTIVE COUPLING

Fig. 10 shows a general schematic overview of the near-field magnetic technologies which use inductive coupling between the external component and the implantable biomedical devices [16], [189]–[191]. Among the various modulations schemes discussed in the previous chapter, ASK and FSK are more attractive in terms of the minimal size of the implantable biomedical device, due to simple circuitry [190], [194], [195], although they experience SNR degradation and low data rates.

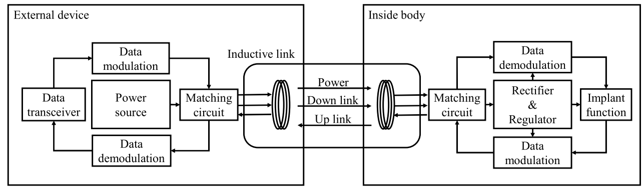

FIGURE 10. The general schematic overview of the near-field communication and power transfer telemetry link using an inductive coupling.

LSK schemes may be an alternative option to satisfy strict size and power consumption requirements, such as battery-less operation and simple circuitry. Accordingly, they usually play an important role in monitoring and status feedback processes, such as impedance around the electrode, chemical states, neural responses, and battery states [16], [106], [196], [197]. However, similar to ASK and FSK, LSK cannot support high data rates.

PM based methods can be considered for high data rate communication schemes with low power consumption, as we mentioned in Section IV. However, the presented schemes may not be appropriate for implantable biomedical applications that require a larger distance than the on-chip coil applications, due to low Q-factor needed to obtain wide bandwidth [196].

Ghovanloo's group proposed improved PM methods for implantable devices. Reference [196] proposed pulse harmonic modulation (PHM) which uses multiple pulses more than twice to suppress ringing signal from the Rx, with specific time delay and amplitude. By using PHM, the system did not require a higher resonant frequency to avoid ISI caused by the ringing signal, or additional resistance to obtain an excessively wide bandwidth. To verify the PHM scheme, they fabricated a fully integrated transceiver which achieved 10.2 Mbps [186], and presented an enhanced version, which could transmit data up to 20 Mbps, including automatic gain control in the Rx to reduce power consumption and ISI [123].

B. MID-FIELD COUPLING

Poon *et al.* [103] proposed a different optimal operating frequency in the low-GHz range for better efficiency in biological media, as explained in Section III. A detailed overview of mid-field coupling is provided in Section III in terms of power transfer. This subsection briefly introduces mid-field coupling communication systems for implantable biomedical applications. The original purpose of the mid-field coupling was to sustain high PTE while reducing the size of the implantable devices. Moreover, it can provide additional inherent advantages such as high data rate, less sensitivity in the link gain to changes in antenna orientation, and good separation decoupling between the power transfer and communication circuitry [118], [119], [124]–[126].

Reference [197] attempted to overcome the sensitivity issues. They proposed a power efficient backscattering method for the uplink from the implantable devices to an external device. Because the separation between external devices and the implantable device was much larger than the size of the implantable device, the backscattering method inherently suffers a high bit error rate (BER). Even worse, tissue composition and the separation can be time-varying depending on the motion of patients. This uncertainty in an operating environment can degrade the quality of the backscattering method severely. The proposed method was able to achieve robust operation in variable conditions. By using reconfigurable load modulations, it maximized the distinguishability between two loads in the backscattered signals for a given environment. Other parameters, such as pulse width and data rate, were also reconfigurable to accommodate uncertainties in the environment. For example, the pulse width of a clock can be adjusted to trade-off between the signal quality and the amount of energy harvesting. Lastly, employing asynchronous amplitude modulation with data encoded in the pulse-width can eliminate the synchronization circuitry and bulky external clock, minimizing the power consumption as well as the implantable device size.

VI. REVIEW OF NEAR-FIELD MAGNETIC BASED SWIPT

Previous sections have addressed the near-field magnetic technologies for WPT and communication, respectively. These two lines of work aim to maximize power transmission efficiency and chancel capacity, respectively, while in general a trade-off exists in achieving these two goals in an integrated system. The SWIPT systems were first introduced for near-field applications in [198] and [199], although the term SWIPT was first coined in [200] for far-field applications. Nonetheless, the concept of SWIPT was already being used in implantable biomedical applications to achieve high-rate communication and high- power transfer with small from factors [197]. Section V described implantable biomedical applications requiring high data rates, such as cochlear and visual prosthesis. To achieve a high data rate, the systems require a sufficient energy supply as well. Therefore, the implantable biomedical devices need to optimize both the functions of communication and WPT.

In this section, we first briefly introduce the SWIPT technologies by following the relevant literature in wireless communication, and then focus our discussion on recent studies of near-field magnetic based SWIPT systems for both the free-space and implantable biomedical applications.

A. SWIPT IN FREE-SPACE APPLICATIONS

1) FAR-FIELD EM BASED SWIPT

Recently, there has been a lot of interest in SWIPT systems in wireless communications (see, [201], [202] and references therein). Generally speaking, SWIPT systems are able to achieve substantially improved energy and spectrum efficiencies, as compared to separate WPT and wireless communication systems that operate independently. Typical SWIPT receivers need to be designed to be able to separate received signals for information decoding and energy harvesting, in either time, power, or antenna domains, as shown in Fig. 11. When the system adopts time switching (TS), the Rx is switched in time between information decoding and power rectification, as shown in Fig. 11 (a). The TS technique has the advantage of simple hardware implementation at the Rx. However, it requires accurate time synchronization and scheduling. The power splitting (PS) technique splits the received signal into two streams of different power levels, as shown in Fig. 11 (b). The PS technique can achieve close to the optimal rate-energy performance trade-off theoretical bound [201]. However, this technique requires higher Rx complexity than the TS technique. Fig. 11 (c) shows an antenna switching (AS) technique which switches each antenna element at the Rx between decoding and rectifying to achieve SWIPT.

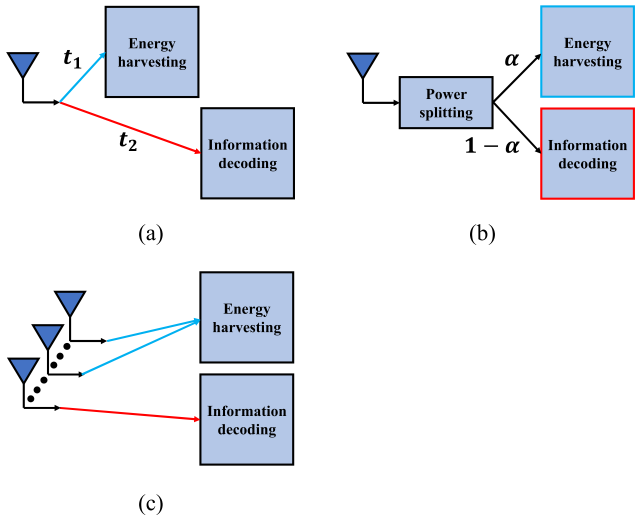

FIGURE 11. SWIPT receivers in different domains: (a) time, (b) power, and (c) antenna.

2) NEAR-FIELD MAGNETIC FIELD BASED SWIPT

Some research groups have attempted near-field magnetic based SWIPT systems. Even before the far-field EM based SWIPT was widely studied, near-field magnetic systems already supported simultaneous power and data transmission, such as passive RFID, NFC, and various telemetry systems including implantable biomedical devices. In this subsection, we present a brief overview of the SWIPT schemes using near-field magnetic based technologies in the free-space. Reference [204] took an approach that was similar to conventional far-field EM based SWIPT studies, and applied it to a near-field magnetic system. They proposed an optimal power allocation (OPA) strategy for an orthogonal frequency-division multiplexing (OFDM) system to maximize the transfer power efficiency under the required channel capacity. They presented the conditions for the existence and boundaries of the OPA, a low-complexity algorithm for the OPA, and compared their performance via simulation. Reference [205] proposed a new SWIPT system which was composed of one data stream and multiple parallel power streams for free-space applications such as the NFC based access point for multiple devices. This was based on their previous work, about magnetic beamforming using an orthogonal 3-D coil structure for the WPT [90]. They proposed two optimization solutions: maximization of the sum of the received power of all the power Rx coils, and maximization of the minimum received power among all power Rx coils under the constraint of a data stream, via simulation and theoretical analysis. References [204] and [205] reported related results in terms of a theoretical analysis. Reference [206] introduced a practical circuit model and measurement results for an FD based SWIPT using a single inductive link. The carrier frequency for power and data were 22.4 kHz, and 1.67 MHz, respectively. To minimize the effect of the data-carrier on power efficiency, the data-carrier had to be set at least an order of magnitude higher than the power-carrier. This scheme achieved reliable communication and efficient wireless powering at low-carrier frequency. Reference [207] presented a specific near-field magnetic based SWIPT system for electric vehicle charging. This system supported wireless powering and an uplink telemetry scheme from the vehicle to the charger, which included battery status, vehicle identification code, and emergency messages, using the same inductive link.

B. SWIPT IN IMPLANTABLE BIOMEDICAL APPLICATIONS

SWIPT is an essential system for future implantable biomedical devices. To achieve the requirements of both high data-rate and efficient power transmission, many research groups have applied various telemetry methods. In this subsection, we review the SWIPT systems employed for implantable biomedical applications. Cancelling or reducing crosstalk is important to achieve simultaneous high data rate and high PTE in near-field magnetic based SWIPT systems for implantable biomedical devices. To solve the crosstalk issue, some research groups have proposed multi-carrier systems supporting multiple links, while the other groups have tried single carrier approach. This subsection mainly discusses near-field magnetic based SWIPT methods to solve the crosstalk issues, and finally describes mid-field coupling based SWIPT.

1) NEAR-FIELD MAGNETIC BASED SWIPT

To apply the SWIPT technologies to implantable devices, it is possible to use different carrier frequencies for different purposes, for uplink, downlink, and power transmission. Fig. 12 shows various multi-carrier SWIPT systems for implantable biomedical devices. Fig. 12 (a) is composed of a multi-carrier system to obtain multiple streams for separated data and power transmission with two inductive links. However, these

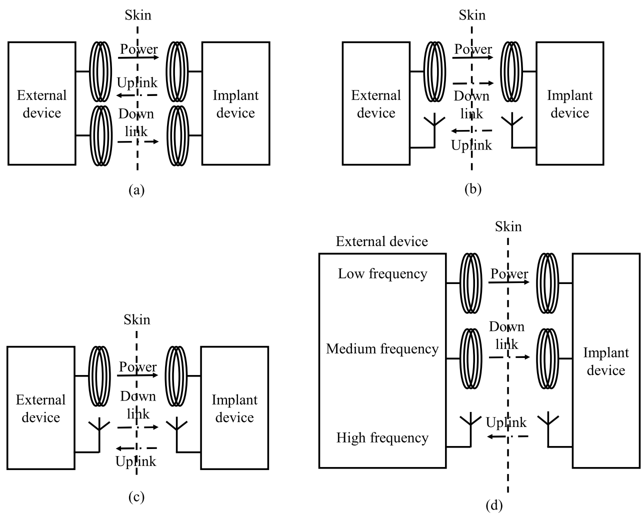

FIGURE 12. **Multi-carrier based SWIPT schemes: (a) multiple inductive coupling (power-downlink, uplink) [208], [209], (b) inductive coupling (power, uplink), high frequency (downlink) [112], [210], (c) inductive coupling (power), high frequency (uplink, downlink) [211], [212], (d) three carrier signal: inductive coupling – low frequency (power), inductive coupling - medium frequency (downlink), high frequency (uplink) [213].**

schemes have the most significant crosstalk issues among the multiple inductive links for power and data signals. Even when the power and downlink signals use different carrier frequencies, crosstalk between the power and data signals cannot be avoided, because of the significantly different power levels between these two signals [208]. Usually, this interference can be cancelled by a high-order filter which requires a reasonable spectral separation between the power and data signals. However, [208] introduced a crosstalk cancellation scheme without a high-order filter using non-coherent DPSK with a high data rate. Reference [209] also applied a multi-carrier system to support high-rate full duplex data communication with efficient power transmission using two inductive links. They presented a comparison between an orthogonal coil structure and a proposed coplanar coil structure to reduce the crosstalk effect; the coplanar coil was more robust to lateral misalignment than the orthogonal type.

Fig. 12 (b), (c), and (d) present far-field EM links and inductive links used to obtain bidirectional data transmission and power transfer. In Fig. 12 (b), they transmit simple command data and power from an external device to implanted devices, and the implanted device sends sensor data to the external devices using an RF link [113], [210]. In Fig. 12 (c), the inductive coupling transmits only power, and data is passed through the RF link [211], [212]. Ghovanloo reported the design of a wireless link for a high-performance implantable biomedical device using three carrier signals at three different frequencies, as shown in Fig. 12 (d). However, these schemes bring high complexity to the circuit and antenna structure [213]. The main challenge of this method is cancelling the interference between different carrier signals. To ensure robustness against interference, they applied several methods as follows: 1) using a coil structure with an orthogonal geometry, which can reduce the crosstalk [213]; 2) using FSK modulation, which is more robust against noise and interference than ASK; 3) applying a higher data carrier frequency five to ten times greater than the power carrier frequency to guarantee enough space between the carrier harmonic components in the frequency domain [196]. Furthermore, this research group recently presented a fully-integrated near-field wireless transceiver which can support SWIPT using a pulse delay modulation (PDM) scheme. PDM is a low-power carrier-less modulation scheme which offers a broad bandwidth with robustness against strong interference from the power carrier signal [184].

Reference [214] proposed a single-carrier near-field magnetic system which can support SWIPT in high data-rate implantable biomedical devices. LSK is usually a popular modulation in single-carrier near-field magnetic SWIPT systems [195], [215], [216]. However, LSK may not be capable of achieving both high efficiency WPT and a high data-rate, since there is a trade-off between transfer efficiency and data-rate. To achieve high PTE, implantable systems require a high Q-factor. However, the data rate of LSK is inversely proportional to the Q-factor. Furthermore, an extra load for LSK leads to impedance mismatching and additional power consumption from the real load [217], and the rectifier cannot receive power from the coil during data transmission. Reference [218] described a solution involving passive phase shift keying (PPSK) which can achieve simultaneous power and data transmission via a single inductive link without the extra load. The PPSK scheme can reverse the current phase of the secondary coil using half-period switching, where the reversed current leads to increased current in the primary coil, and the primary coil can receive data due to this phenomenon. However, the proposed scheme has additional power losses due to the disruption of resonance. To overcome these problems, [214] proposed a new backscattering data modulation technique, cyclic on-off keying (COOK), which can simultaneously support high efficiency power transfer and high-rate communication with a small loss. Although COOK shows behavior similar to conventional LSK and PPSK, to synchronize the received signal it closes the switch only during a single cycle when the voltage of the secondary coil is zero and the current is at its peak. This synchronized switching does not disturb resonance. Unlike the multi-carrier methods, the single-carrier schemes take into account the additional power loss resulting from the discontinuous power transfer and disruption of resonance. However, the issues related to the trade-off between power efficiency and data-rate still remain.

2) MID-FIELD COUPLING BASED SWIPT

The mid-field coupling based SWIPT concept has also been considered for implantable biomedical applications. Reference [197] described a wireless power and data transceiver with sub GHz band for low-power implantable devices. In this system, the downlink trade-off between power efficiency and data recovery depends on the modulation depth of ASK. For implantable biomedical applications with power constraints, the modulation depth should be minimized to ensure a stable power supply. However, the minimized modulation depth makes it difficult to recover the data and clock, and in the backscatter link, adjusting pulse width [219] and optimal modulating load [220] affect the data rate and energy consumption, respectively.

3) CHALLENGING ISSUES OF NEAR-FIELD MAGNETIC BASED SWIPT

Many research groups have achieved sufficient power efficiency and high data rates in downlink telemetry. However, the SWIPT technologies still have challenging issues, especially at the uplink due to an asymmetry of size and power consumption between external devices and implanted devices [214]. Furthermore, to implement near-field magnetic based SWIPT systems, it is necessary to consider various parameters and constraints in addition to carrier frequency and link structure, to both achieve performance goals and address safety issues. To design a high-efficiency and high-rate near-field magnetic SWIPT system, we need to take into account tissue effect, coil size, and the geometric position between the coupled coils to achieve high coupling efficiency and high power conversion efficiency. However, this causes large design complexity because of interactions between the parameters. For example, high frequency can be used to obtain size reduction and a high Q-factor in the coils. However, high frequency leads to parasitic value sensitivity, increased source driving requirements, and difficult rectifier implementation. In addition, the tissue effect varies depending on the carrier frequency, thus, establishing the optimal frequency involves a trade-off between tissue effect and transmission efficiency [221]. To satisfy the performance and safety issues, engineers should have a comprehensive understanding of the system in implantable biomedical devices.

VII. CONCLUSION

Implantable biomedical devices are strong candidate applications of near-field wireless systems, due to the advantages of using near-field magnetic technologies inside the body. This paper presents a comprehensive review of the near-field wireless powering and communication schemes for both free-space and implantable biomedical devices, from historic development to recent trends. Among the various issues in near-field magnetic systems, we focused on three aspects: transmission range, misalignment, and limited channel capacity. Section II provided a detailed history of the WPT from Maxwell to the present day, and compared different WPT schemes in terms of coupling performance. Recent trends were also introduced, concerning transmission range and misalignment issues with the maximum PTE in free space. To extend the transmission range, self-resonant coupling, relay, and DCRS can be applied, and misalignment error can be compensated by robust structures, using arrays or metamaterials, while electrical parameter tuning can be utilized for impedance matching. These methods can also be applied to implantable biomedical devices to satisfy their particular requirements, including minimal size, low-power consumption, low-complexity, and safety, as discussed in Section III. Mid-field coupling based wireless powering was also introduced for smaller sized implants, to operate in the sub-GHz range. Sections IV and V dealt with near-field communication methods which share the fundamental principle of near-field wireless powering. Section IV provided an overview of typical near-field magnetic communication systems such as RFID, NFC, RuBee, and MFAN, and presented both the limitations and promising methods to achieve high data-rate communication in free space, such as relay, MIMO and PM based modulation. Section V discussed near-field communication methods mainly used to achieve high data rates in implantable biomedical applications. The PM based methods and special structures discussed are also applicable to implantable devices, much the same as their use in free-space, while mid-field coupling was also shown to be effective since it can achieve wide bandwidth because of the high carrier frequency. In Section VI, we briefly introduced the conventional far-field EM based SWIPT system, and then discussed recent studies on near-field magnetic based SWIPT systems in free-space and implantable biomedical applications. Future applications require high data rate communication and highly efficient power transmission at the same time, and so the trade-off between them has to be considered, where the Q-factor of the coils and their cross coupling are the primary factors that need to be taken into account for system performance optimization.

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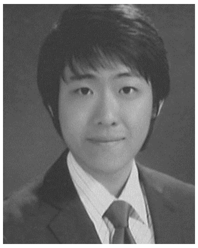

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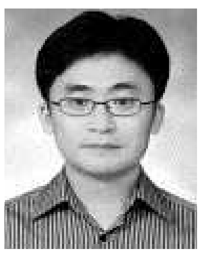

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