

A review of wireless charging systems for cardiac pacemakers

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Abstract— Some of the advantages of wireless power transfer technology are reliability, safety and convenience of use, features suitable for biomedical applications. In recent years the implementation of these systems has increased due to improvements in technology and efficiency. This paper reviews the state of the art in wireless power transfer for cardiac pacemaker charging using inductive power technology with resonant converters. Resonance frequency, magnetic coupling or typical topologies are the most studied concepts by researchers in wireless charging systems for pacemakers. The compilation of authors and knowledge in this field provides an overview of the current status and allows further progress to be made in making this technology viable.

Keywords—Wireless power transfer (WPT), inductive power transfer, active implantable medical devices (AIMD), cardiac pacemaker, resonance coils, review.

I. INTRODUCTION

Active implantable medical devices (AIMD) are surgically inserted under the skin and stimulate, treat or monitor dysfunctional organs. Examples of AIMDs are cochlear implants, cardiac pacemakers, implantable cardioverter defibrillators, implantable neurostimulators, brachytherapy implants, glucose monitors, left ventricular assist devices, retinal implants, phrenic nerve stimulation, deep brain stimulation, gastric stimulators or implantable infusion pumps [1]–[5]. Progress in these technologies is constant, and this includes the method of powering the devices, which remains a challenge. Currently, most commercial solutions use disposable lithium-ion batteries, which have a limited lifetime of 5 to 15 years [6]–[8].

There are alternative methods to supply AIMDs under investigation [7], such as self-recharging with harvesting methods [9]–[11]. However, the most widespread solution in AIMDs for powering the devices or charging the batteries is to implement wireless power transfer (WPT). The main

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advantage is the extended lifetime, which avoids surgical operations that put patients at risk [12], [13].

Wireless charging is a convenient, reliable and safe technology for powering and charging electrical devices. The advantages over other power methods are that they are efficient, cost-effective and safer by eliminating the use of physical connectors [14]. Depending on the transfer distance, WPT systems can be either near field or radiative far field. The first ones can be capacitive and inductive transfer and the second ones are microwave, laser, optic, radiofrequency or ultrasound [15]–[18]. Some of these techniques applied in AIMDS are studied in [3].

Inductive power transfer (IPT) is the oldest and most widely used WPT technology. The advantages of magnetic inductive coupling also include ease of application, high efficiency over a short distance and guaranteed safety [19], [20]. Most papers are focused on IPT systems, but some others use ultrasound [17] or microwaves [21] techniques. The typical mode of operation of IPT converters is resonance and they are called magnetic coupling resonant converters (MCR), which is the one covered in [8].

One of the most used AIMDs are cardiac pacemakers. The main elements of a pacemaker are the generator consisting of the battery and the heartbeat control and the leads that connect the heart to the generator and carry the electrical messages to the heart. There are two main classifications of pacemakers: according to the number of leads they can be single, dual and bi-ventricular; and according to the mode of operation they can be with fixed-rate, 'on demand' and rate responsive [22], [23].

Numerous studies of IPT resonance converters have been performed in pacemakers and even implemented in mice [24]. The research papers focus on the design particularities of IPT systems for AIMDs. The main factors considered are operating frequency, coil size, coupling factor, weight or human exposure [7], [25]. In this paper, we have compiled the different analyses of these factors together with the different topological solutions that have been provided by the research community.

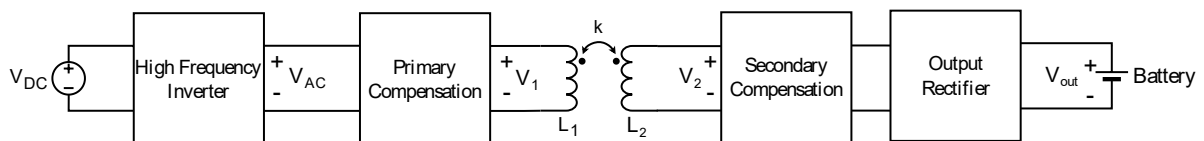


Fig. 1. Fundamental blocks of a inductive power transfer system

II. APPROACHES IN PACEMAKER WIRELESS CHARGING

A typical IPT system consists of four basic components (Fig. 1): an inverter stage that generates a periodic current; a transmitter coil that is a part of a resonance tank (primary compensation); a receiver coil that includes a resonance tank (secondary compensation); and a rectifier with high-frequency filtering [26]. From the diagram, the main parameters that define these systems are the resonant frequency, the magnetic coupling between the coils, the communication transfer, and the battery. This section includes the features of each of these parts in a pacemaker charging system.

A. Magnetic coupling

The functional element of the system is the coupled coils. Depending on the quality of the coupling, they can be tightly coupled ($k > 0.2$) or loosely coupled ($k < 0.2$). The power transfer efficiency depends on the coupling quality. The higher the coupling, the higher the energy transfer, so the objective in IPT systems is to increase the k [15], [18].

This factor is a function of the geometry of the coils, their mutual position and the materials between them that influence the magnetic field distribution [18], [26]. Coupling k is also related to the self-inductances L_1 and L_2 and the mutual inductance M through (1). The influence of each parameter has been studied considering the particularities of AIMDS.

$$k = \frac{M}{\sqrt{L_1 L_2}} \quad (1)$$

The pacemaker case is implanted under the collarbone at a distance of 15 mm at most [27]. The secondary coil is implemented outside the pacemaker case [12], [28] being a better option than embedded or inside case [29]. The distance between coils is variable due to the different breathing and tissue thickness of each patient, but typically between 8 and 12 mm [30]. On the other hand, there are misalignment issues between the primary and secondary. Distance and misalignment affect coupling which leads to changes in operating point and efficiency [19], [31].

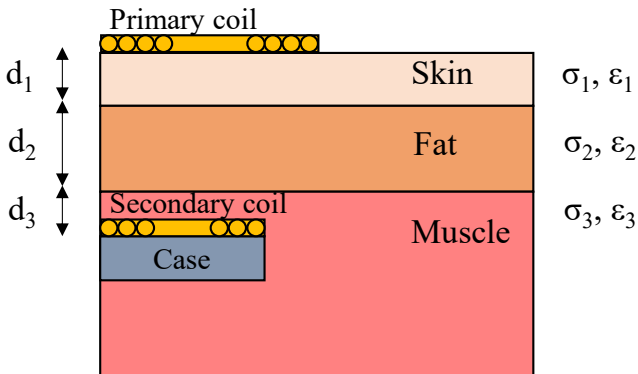


Fig. 2. Tissue simplification model with its main parameters

Another factor influencing the coupling is the separation medium between the coils, which in this case are body tissues. Physical tissue models, or "phantoms", are used for experimental validation of transcutaneous systems. Phantoms must represent, at the frequency of interest, the conductivity (σ), dielectric permittivity (ϵ) and losses, as well as imitate the structure and shape of the tissue [32]. The tissue simplification model normally adopted by researchers is a three layer of skin, fat and muscle as shown in Fig. 2 [33], [34]. While primary

coil is in air, secondary is inside tissue. To simulate this medium, some researchers use animal tissue such as 1 mm pig skin, 2 mm pig fat and 5 mm pork lean [34], [35]. Another method is a liquid phantom which represents tissue as homogeneous, such as a saline solution of sodium chloride and water between the two coils [30].

In the secondary side, the proximity with the pacemaker titanium case induces eddy currents [31]. To solve it, the case surface in contact with the coil can be covered by a thin layer of ferrite of 1-mm thickness [28], [30]. Ferrite and aluminium plates on the primary side is also beneficial for directing field lines and increasing coupling [27]. However, it is not recommended by some researchers to use ferrite due to biocompatibility [31], [36].

Comparisons of coil geometries have been made and the one with the lowest resistance and best coupling is the planar spiral type [31]. Within spiral coils, circular (Fig. 3) and square geometries have been implemented in literature [37]. As it appears in the comparison studied in [31] of spiral coils, the best results are obtained in circular geometries, being the preferred one in the state of the art.

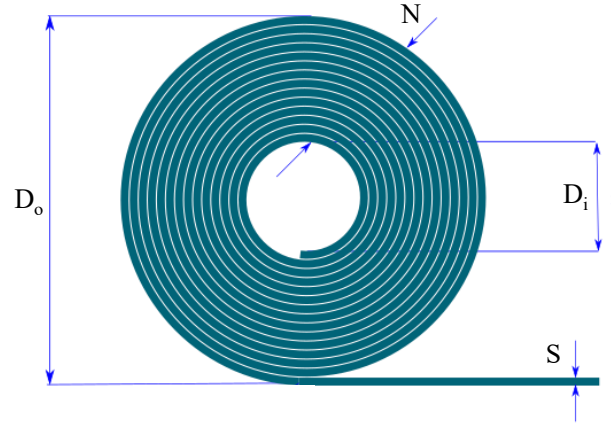


Fig. 3. Design parameters of the circular spiral geometry

TABLE I. REVIEW OF SPIRAL COILS IN PACEMAKER IPT SYSTEMS

Paper features	d (mm)	Max k	Coil	Geometry	N	D ₀ (mm)	D _i (mm)	S (mm ²)
Secondary with magnetic shield [12]	10	0.22	L ₁	Circular	8	44	36	0.8
			L ₂	Square	5	30	16	0.5
Ferrite film layers and PCB secondary coil [35]	8	0.35	L ₁	Elliptical	13	58x48	0	2.5
			L ₂	Elliptical	9	44.5x30.5	0	1.20 x 0.072
Ferrite film in both coils [30]	8	0.33	L ₁	Circular	16	80	17	2.5
			L ₂	Circular	9	45	15	1.5 x 0.1
Spider-web coils geometry [38]	10	-	L ₁	Circular	14	85	60	0.36
			L ₂	Circular	4	60	28	0.36
Comparison of various couplings and shielding [28]	10	0.28	L ₁	Circular	10	50	36.6	0.33
			L ₂	Circular	4	35.3	30.1	0.33

The primary side is unrestricted, while the secondary side must be dimensioned to fit the case. TABLE I. shows the dimensions of coils designed for pacemakers. The outer diameter of the secondary coil is fixed to fit the dimensions of a generic pacemaker, while the inner diameter of the primary is slightly larger than the outer diameter of the secondary to make the system more robust in case of misalignment [28].

It is worth mentioning other coupling strategies in the field of biomedical devices that improve energy transfer such as four-coil systems [39], sandwich topologies with ferrite bars [40] or multi-bundle secondary coil [41].

B. Frequency

The inverter is another main part of the IPT system since it generates AC voltage with a frequency at which the resonance occurs. Within the frequency range, low frequencies from 20 kHz to 200 kHz and medium-high frequencies from 300 kHz to 13.56 MHz are used.

The electric field is proportional to the frequency, and the conductivity of the human body increases with frequency. In biomedical applications, lower frequencies are characterised by better penetration of the magnetic field inside the body, but in terms of efficiency and transferred power the performance is lower [12], [18], [28]. However, compliance with Electric and Magnetic Fields (EMF) standards is facilitated by generating less eddy currents at low frequencies [27].

Higher transmission frequencies improve the transmission efficiency of the system but could lead to tissue heating, stray current losses and electromagnetic compatibility (EMC) problems in AIMDs [28]. Another consideration is the limitation of the penetration of the field into the human body [25]. Despite the disadvantages, multi-megahertz (MHz) systems are being considered for small low-power systems [42].

Unlicensed frequencies of interest for WPT applications for AIMDs are: 133-150 kHz, 6.78 MHz, 13.56 MHz, 27.1 MHz, 40.7 MHz, 868-915 MHz, 2.45 GHz, 5.2 GHz and 5.8 GHz. The frequencies 6.78 MHz and 13.56 MHz are common as they are industrial scientific medical (ISM) bands, reserved internationally for medical purposes [25], [28].

The compilation TABLE II. of pacemaker topologies includes frequencies used in the prototypes that researchers have developed.

C. Battery and power

Currently, traditional pacemaker batteries are non-rechargeable primary batteries. This type of technology can store higher energy density, but once the charge is finished, they must be replaced. The battery, of lithium-ion technology, is typically 2.75-3.0 V and around 1 Ah capacity, depending on the manufacturer [23]. Examples of values of primary batteries in pacemakers are 0.92 Ah batteries for single-chamber pacemakers that could last 10.4 years, and 1.3 Ah lithium batteries for dual-chamber pacemakers that could last 11.3 years, both with voltages of 2.8V [43].

Secondary or rechargeable batteries have lower energy densities, but rechargeable batteries have a longer lifetime. In addition, the main advantage of rechargeable batteries in AIMDs is that surgical interventions are avoided [44]. However, the lower energy density means that the battery capacity is much lower for the same size. From Fig. 4, the size

of the battery is half of the titanium case of the traditional pacemaker.

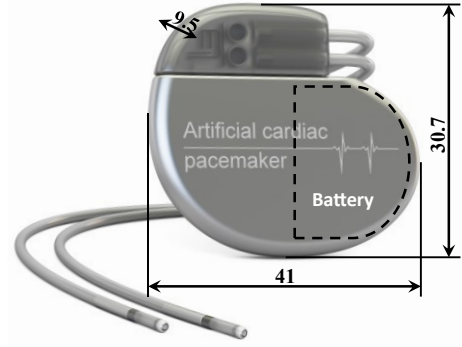


Fig. 4. Typical size of a pacemaker and battery position

The power consumption in a pacemaker is mainly composed of two loads: a continuous load of 10-20 μ A of the device, and a communication load for 30 min every 6 months of 200-400 μ A in new generation pacemakers [45]. For this consumption and a 3.6 V - 100 mAh lithium battery, the approximate duration without charging would be one year at the most according to (2). Taking into account the waiting time of the patients, the charging process is usually sized at 2C (30 minutes waiting time) [43].

$$t_{\text{discharge}} = Q_{\text{batt}} / I_{\text{discharge}} \quad (2)$$

As a battery charging system, the load changes depending on the state of the battery because it depends on the voltage and current supplied (3). During battery charging, the power varies according to the charging modes, which in lithium batteries are constant current and constant voltage modes. This is a factor to consider when designing the system. In section III, topologies robust to load variations are proposed.

$$R_L = \frac{V_{\text{batt}}}{I_{\text{DC}}} \quad (3)$$

D. Data transmission

Data communication solutions in WPT systems can take two forms: through magnetic coupling of the power line and using a dedicated transceiver on both sides.

Communication from the primary to the secondary called forward telemetry or downlink can be done directly by modulating the input source in amplitude or frequency. However, communication from the secondary to the primary or back telemetry (uplink) is done by perturbing the signal transmitted by the coupling. This perturbation can be by load modulation called load shift keying (LSK) or by frequency shift keying (FSK) [46], [47].

Research in power system control data communication with LSK has been carried out on different biomedical implants due its very low-power consumption, simple implementation, and robustness. Some examples are gastroesophageal monitoring [48], or biomedical devices in general [49], [50], the two last with an explanation of the developed communication protocol and modulation and demodulation process.

The other communication technique is by a dedicated device. The use of a dedicated transceiver isolates the power

and data transmission channels, allowing these two links to be designed independently, as in [51] with 2.4 GHz modules. However, it can lead to additional interference problems when working in other frequency bands and the size of the secondary can increase considerably.

III. TYPICAL TOPOLOGIES

This section lists the most common topologies for wireless pacemaker charging in the state of the art. The choice of the correct IPT topology must consider the specifications for AIMDs which are mainly low power and frequencies in the ISM (high frequencies). In addition, the variable parameters in the wireless charging system of a pacemaker that affect the operation are coupling and load conditions and the system must be designed to be robust to these variations.

A. DC/AC inverters

Operating at high resonant frequencies allows the size of the converter and coils to be reduced. The main DC/AC for high frequency inverters topologies are class E and class D power amplifiers, [14]. Fig. 5 shows the two inverters: the class E inverter consisting of a choke coil (L_{ch}), a shunt capacitor (C_{sh}) in resonance with each other and a single transistor (Q_1), and the class D inverter comprising two transistors (Q_1 and Q_2).

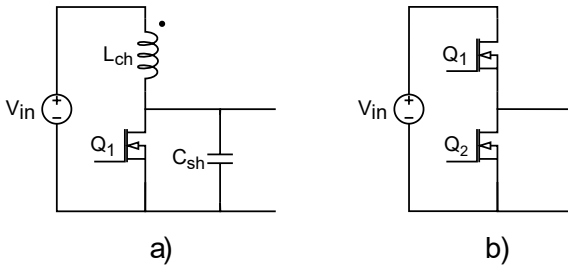


Fig. 5. Power amplifiers a) Class E and b) Class D

Class E amplifiers provide a very compact and efficient solution for high-frequency WPT systems. The reason is the Zero Voltage switching (ZVS) operation naturally at certain frequencies. However, the dependence of the resonant operating point on load and coupling variations complicates the design of this amplifier. The Class-D amplifier is more robust since it does not need resonance for inverter part operation, but when operating at very high frequencies the switching losses can become unacceptable [42], [52].

The Class-E and Class-D amplifier topologies working at MHz frequencies have been suggested in biomedical literature. In [40] a class E is proposed for a 5 W pacemaker charging system and 88% transmission efficiency is achieved. There are also battery - less power systems with class E [53]. In [38], a class D power amplifier board is implemented with a ZVS to deliver 4 W with 95% transmission efficiency.

B. Compensation tank

Within the basic resonators, the most efficient topologies are series-series (SS) and series-parallel (SP). Fig. 6 shows the equivalent SS and SP circuits with the parasitic resistances of the coils (R_1 and R_2) and the equivalent output resistance (R_{AC}). From this circuits capacitance at which resonance occurs (4) can also be obtained [19].

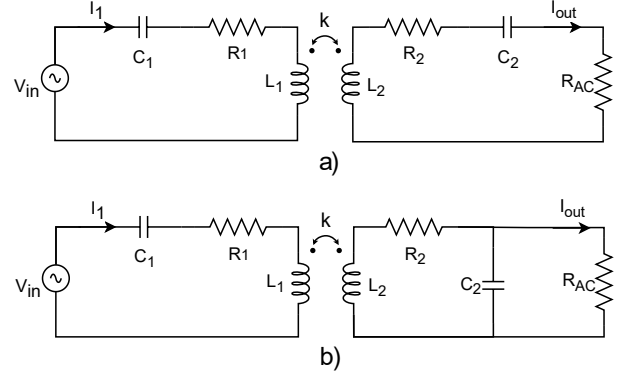


Fig. 6. Compensation tanks a) Series – Series and b) Series – Parallel

$$C_1 = \frac{1}{\omega^2 L_1} \quad C_2 = \frac{1}{\omega^2 L_2} \quad (4)$$

The series-parallel topology is suitable for lower power ratings. In addition, it is robust to load variations. [15]. The SP topology requires less secondary inductance than SS. Especially useful in applications to minimise the size of the receiver coil or where a constant voltage transfer ratio is desired without the need for feedback control, e.g. in biomedical applications [29].

Hybrid topologies improve the efficiency of the magnetic link over basic ones, but complicate the design and increase volume, price and weight [14]. In [35], primary side is LC tank with a series capacitor and secondary side is also series capacitor due to space constraint (Fig. 7). An LC-S topology

TABLE II. COMPILATION OF TOPOLOGIES FROM PAPERS PROTOTYPES OF IPT CHARGING SYSTEMS FOR PACEMAKERS

Paper references	Description	Power	Maximum Efficiency	Resonant tanks	Inverter topology	Rectifier topology	Frequency	Distance	Coupling coefficient
[35]	Impedance matching method to achieve stable resonance.	3 W	78.4% (WPT)	LCC-C	-	Full bridge	300 kHz	8 mm	0.35
[30]	LC-S compensation topology operating with zero phase angle (ZPA)	1.9 W	82.39 % (Coils)	LC-S	-	Full bridge	300 kHz	8 mm	0.33
[12]	Simulation of magnetic shield to improve link efficiency	1.5 W	-	SP	-	Full bridge	300 kHz	10 mm	-
[38]	Commercial Class D and spider web coil	4 W	97.91 % (Coils)	SP	ZVS class D	Full bridge	1.78 MHz	10 mm (10-100)	-
[28]	Method to identify topology and operational frequency	-	-	SP less f	Full bridge	-	300 kHz	10–60 mm	0.282
[28]	Method to identify topology and operational frequency	-	-	SS more f	Full bridge	-	13.56 MHz	10–60 mm	0.2484
[40]	Sandwiched topology with a controllable working distance	5 W	88% (Coils)	SS	Class E	Full bridge	160 kHz	50 mm	0.0824

has fewer component number and better output current characteristic than LCC according to [30].

In TABLE II the different topologies power and efficiencies are presented.

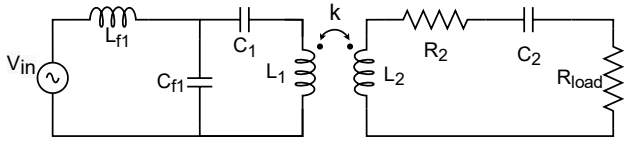


Fig. 7. LCC – C topology

IV. HUMAN EXPOSURE CONSIDERATIONS

The design of the device must comply with the associated safety regulations to protect patients from harm caused by electromagnetic radiation. For any AIMDs, the safe levels of power absorbed by the tissue must be considered, as tissue heating damage may occur. The factor used to measure the maximum permissible exposure (MPE) in electromagnetic field environments is the Specific Absorption Rate (SAR) [3].

The IEEE Base C95.1 standard and ICNIRP express the SAR limitation. The standards specify the mass in grams on which to calculate the average SAR and then the SAR is represented per unit mass. According to IEEE 1992, the maximum SAR value should be less than 1.6 W/kg for 1 g and less than 2 W/kg for 10 g [3], [30], [32].

The current standard uses a homogeneous equivalent model to calculate the SAR. Equation (5) for estimating the value is composed of the electric field strength (E), the density (ρ) and the conductivity (σ). It can also be estimated in terms of the temperature rise (T) over a given exposure time (t) for specific tissue points with specific heat capacity (c). The quantity can be integrated into a volume or mass of tissue, the details of which are defined in each SAR standard [32], [33].

$$SAR = \frac{\sigma \cdot |E|^2}{\rho} = c \frac{\Delta T}{\Delta t} \quad (5)$$

Some of the researchers of pacemaker systems do safety assessment-oriented SAR calculation. Analysis of 3D finite element simulations [35] in COMSOL [12], [29] is common. The influence of induced currents on the titanium case that alters the efficiency of the IPT system has also been analysed in [12].

V. CONCLUSIONS

In this paper, a study of wireless charging systems in cardiac pacemakers has been developed, from the particularities of coupling to the typical topologies used by researchers. Two summary tables have been made with the main papers in this technology. Finally, electromagnetic exposure considerations have been summarised with limit values applied in pacemakers.

From the analysis, it can be concluded that the most suitable topology for pacemaker contactless charging is the SP tank due to its performances in low power, and the Class D tank because it is more robust to variable pacemaker conditions. Although clinical trials and studies are needed to evaluate the safety and efficacy of these systems, advances in wireless technology are improving the scenario of wireless pacemaker chargers to reach feasibility.

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