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Wireless Power Technology for Biomedical Implants

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1. Introduction

Biomedical Implants require a clean and medically safe source of energy to perform their operations. Early implants such as pacemakers sourced their power from small lithium ion batteries. While this solution allows for the operation of the implantable device without a wire connecting the internal and external circuitry, limited battery life causes the impracticality, health risks and expense of operating on patients for the mere purpose of replacing the battery.

A significant development in the supply of power to implantable devices is the use of inductive coupling to charge an implantable rechargeable battery. The concept may be understood by considering two windings of a weakly coupled transformer, where the core is in fact a large air gap as shown in Fig. 1. The primary winding is the transmission coil, and the secondary winding is the receiving coil. Numerous applications have been developed along this wireless battery charging idea, varying in size, frequency and coil structure (Li & Bashirullah, 2007; Lim et al., 2005).

The concept of wireless battery charging has been extended to the idea of supplying power only wirelessly, where implanted batteries are not used at all. Real-time powering systems save a considerable amount of implant space; however they require the constant supply of wireless energy to the implant.

Wireless power transfer has generally been implemented in the kHz to MHz range, and the transfer of energy becomes less efficient as the frequency of transmission increases (Vaillancourt et al. 1997). In addition to making circuit elements smaller, this naturally makes the design of highly efficient transmission circuits a point of interest.

The use of wireless power technology in implantable devices led to a more flexible range of applications which were not otherwise possible; retinal prosthesis being one such example.
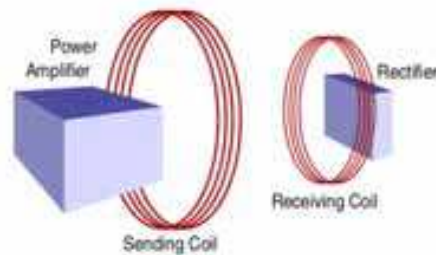
Fig. 1. A general block diagram showing the flow of power to an implantable device.

Implanted devices generally consist of power and data receiving-coils. The power signal is generally rectified and used to store energy or directly supply the implant's electronic blocks. The data signal is demodulated and supplied to an implanted processor, which turns the information into a form understood by an actuator, which stimulates the body as required.

Medical prosthesis generally involves sending a signal to a transducer in order to conduct a particular function, which the body cannot perform. Fig. 2 shows a general block diagram of a prosthetic device, where information is processed and sent to the implant wirelessly. It is further processed and sent to the organ being stimulated. An example of prosthesis is muscular stimulation, where a signal is sent from an electronic device directly to muscle fibres to stimulate muscles in cases where the nervous system fails to do so. Another example of prosthesis is the Bionic Eye.

Telemetry involves using an implant used to monitor a particular parameter in the body, such as temperature, blood pressure, or other parameters which electronic transducers have been designed for. Implanted telemetry units generally comprise blocks similar to those used in prosthesis units. Fig. 3 shows a block diagram of a telemetry device, where information is sensed, digitised, modulated and wirelessly transmitted to an external device.

The supply of power to both prosthetic and telemetric implants involves similar principles. An oscillating unit produces the desired transmission frequency, a power amplifier, power transmission coils and rectifiers.

2. Switched Power Amplifiers

The supply of wireless power to biomedical implants begins with power amplifiers, which supply energy to an antenna or coil at a particular frequency. As the size of implants decreases so does the available space for receiving antennas. This leads to shorter wavelength antennas and lower size constraints for coils, which increases the frequency of power transmission. Higher transmission frequencies place a particular focus on utilising power amplifiers, which operate efficiently at these levels. Switched power amplifiers have been a popular choice to drive inductive power links for implantable electronics due to their ability to minimise losses at higher frequencies. Three types of such amplifier configurations will be discussed in this section.

Fig. 2. A block diagram of a prosthesis system, where the external element is on the left and the internal element on the right.
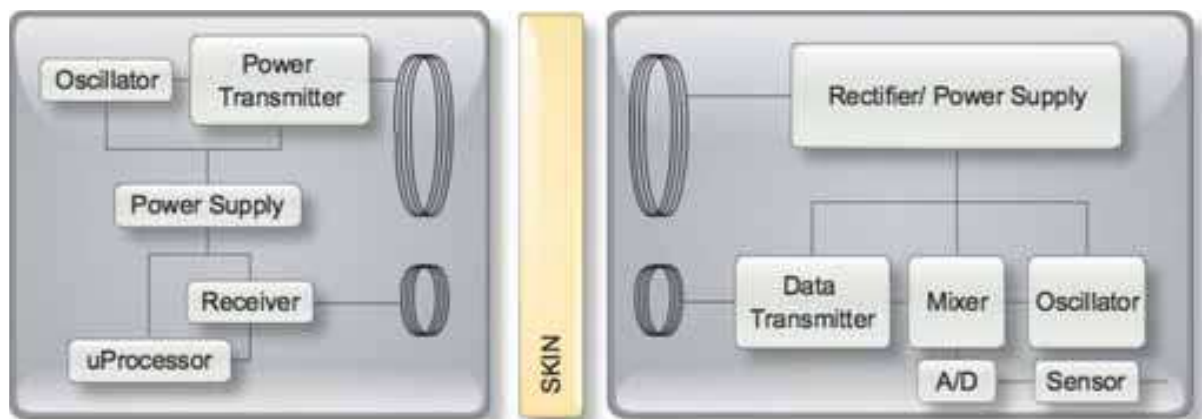

Fig. 3. A block diagram of a telemetry system, where the external element is on the left and the internal element on the right.

2.1 Class-D Amplifier

The Class-D amplifier consists of an inverter, which switches two transistors on and off alternatively to generate a theoretically square wave. The output of the inverter is connected to a series RLC network as shown in Fig. 4, which is resonant at the fundamental frequency of the square-wave, producing a sinusoidal signal at this frequency. The gain of the amplifier is given by (1) as a function of R, L, C elements shown in Fig. 4. The theoretical efficiency of the Class-D amplifier is 100%, which assumes that all circuit elements are ideal.
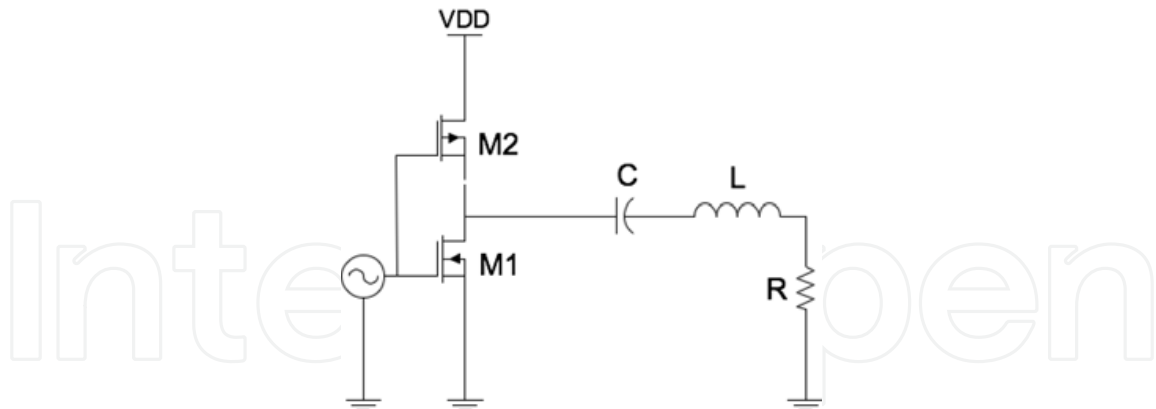
Fig. 4. Class-D power amplifier

$$\frac{v_{out}}{v_{in}} = \frac{s\dfrac{R}{L}}{s^2 + s\dfrac{R}{L} + \dfrac{1}{LC}} \tag{1}$$

In reality, circuit elements are not ideal and several losses have been analysed with a focus on parasitic drain-source capacitance in each of the transistors, which becomes significant in higher frequency RF designs (El-Hamamsy, 1994).

The drain-source capacitance, C_{ds} actually introduces a capacitor where an otherwise open circuit should ideally exist and at high frequencies, typical capacitor values are in the order of pico Farads. This means that parasitic capacitance C_{ds} becomes a significant circuit element, which allows the dissipation of energy during switching cycles thus decreasing the amplifier's efficiency.

2.2 Class-E Amplifier

The Class-E amplifier, shown in Fig. 5 is a circuit designed for high frequency applications, making its application suitable as a power transmitter for biomedical implants (Troyk, 1992). It comprises a capacitor C_1 across the transistor terminals, which absorbs the transistor's parasitic capacitance and forms a key component of the circuit's high efficiency operation at high frequencies by shaping the voltage across its terminals in a particular way.

The principle of the class-E amplifier's high efficiency operation lies in the shape of the voltage across C_1, which is shown in Fig. 6. Parameters for the amplifier are chosen such that the voltage at this point is zero when the transistor is switched on such that no stored energy is dissipated from the capacitor. The voltage is shaped such that the rate of change of voltage (dv_{c1}/dt) across this point is also zero (Sokal, N.O. & Sokal A., 1975; Kazimierczuk, M., 1986; Sokal, N.O., 2000). This feature enables robustness to phase or frequency irregularities in practice.
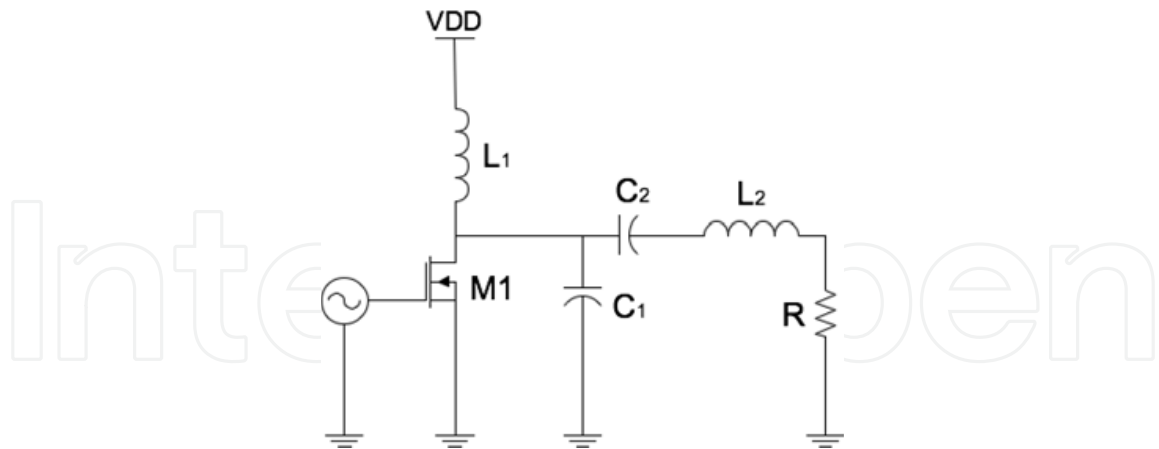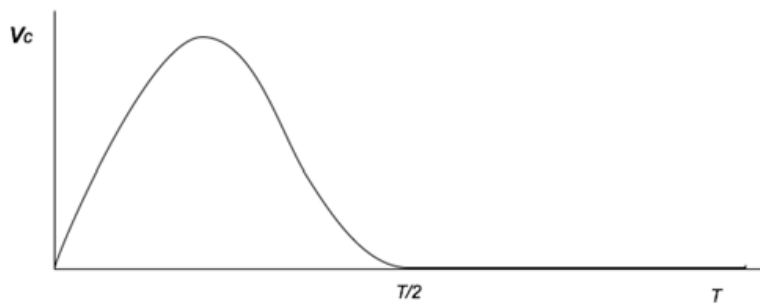
Fig. 5. Class-E power amplifier

Fig. 6. Voltage across C_1 for one period.

Analysing the amplifier in the frequency domain is an elegant way to model the performance of the circuit. The small signal model corresponding to the Class-E amplifier is shown in Fig. 7. It is possible to interpret the input signal supplied to the class-E amplifier as a series of alternating step inputs, as shown in Fig. 8. The voltage signal supplied to the gate of the transistor (base if it is a BJT) is translated to a step increase in the current flowing through the drain. This means that parameters such as resonant frequency ω and damping factor ζ may be used to optimise the response of the class-E amplifier upon a step increase or decrease in the current. The impedance seen by the current source of Fig. 7 is expressed by (2).

$$Z_L = sL_c \parallel r_o \parallel \frac{1}{s(C_1 + C_{ds})} \parallel \left(\frac{1}{sC_2} + sL_2 + R \right) \tag{2}$$

$$Z_L = \frac{s^2 L_2 C_2 + sRC_2 + 1}{s^3 L_2 (C_1 + C_{ds}) C_2 + s^2 \left[\frac{L_2 C_2}{r_o} + R(C_1 + C_{ds})C_2 \right] + s \left[\frac{RC_2}{r_o} + (C_1 + C_{ds}) + \frac{L_2 C_2}{L_1} + C_2 \right] + \left[\frac{1}{r_o} + \frac{RC_2}{L_1} \right] + \frac{1}{sL_1}} \tag{3}$$
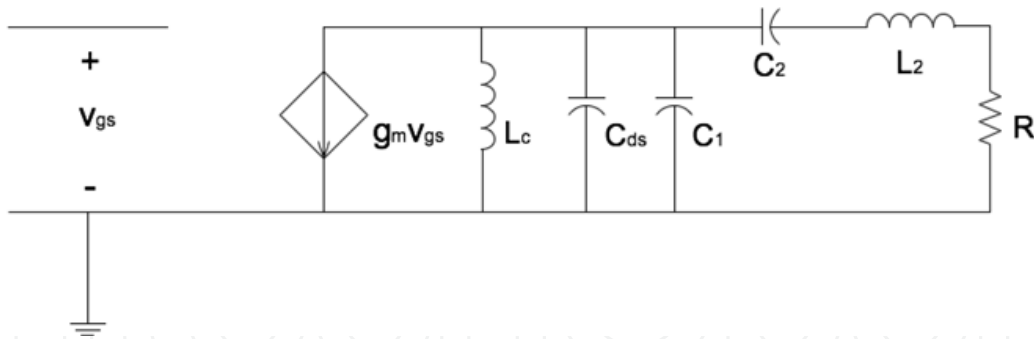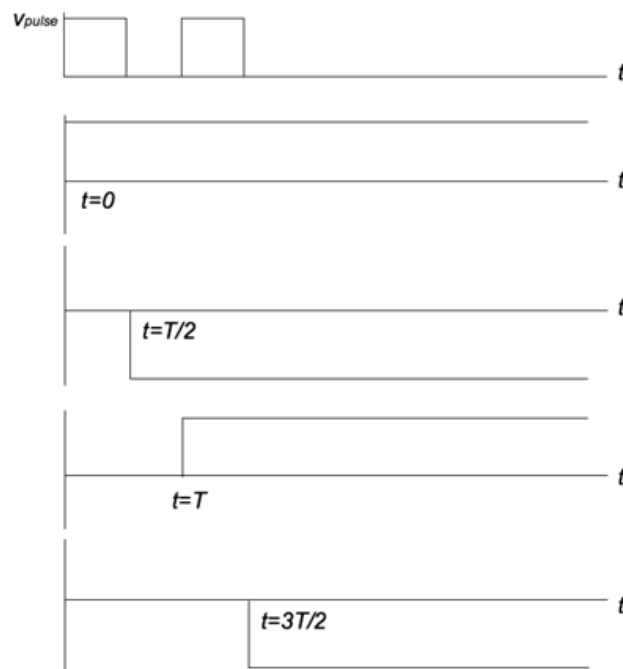
Fig. 7. Small signal model of a class-E power amplifier

Fig. 8. A clock signal split up into several time-displaced step inputs

The expression for the voltage across C_1, called $v_{in,}$ is determined by multiplying the impedance Z_L by a step input of $1/s$, as shown in (4).

$$v_{in} = \frac{1}{s} Z_L \tag{4}$$

The output voltage of the circuit is considered to be the voltage across the resistance R, and obtaining an expression for the output voltage is as simple as applying a voltage divider as shown in (5).

$$v_{out} = \frac{R}{R + sL_2 + \dfrac{1}{sC_2}} v_{in} \tag{5}$$

Substituting the expression for Z_L obtained from (3) into (4) and then (5) allows the determination of the transfer function (6).

$$\frac{v_{out}}{v_{in}} = \frac{s^2 RC_2 g_m}{s^4 L_2 (C_1 + C_{ds}) C_2 + s^3 \left[\frac{L_2 C_2}{r_o} + R(C_1 + C_{ds}) C_2 \right] + s^2 \left[\frac{RC_2}{r_o} + (C_1 + C_{ds}) + \frac{L_2 C_2}{L_1} + C_2 \right] + s \left[\frac{1}{r_o} + \frac{RC_2}{L_1} \right] + \frac{1}{L_1}} \tag{6}$$

The conditions that allow for the maximum transfer of power occur when the denominator of (6) is minimal, that is when the *j*-component is zero. Using the relationship $s=j\omega$ the expression (7) shows the resonant frequency with respect to circuit parameters.

$$\omega_n = \sqrt{\frac{\dfrac{1}{r_o} + \dfrac{RC_2}{L_1}}{\dfrac{L_2 C_2}{r_o} + R(C_1 + C_{ds})C_2}} \tag{7}$$

If the parasitic elements of r_o and C_{ds} are approximated to very large and very small values respectively, the transfer function in (6) will become:

$$\frac{v_{out}}{v_{in}} = \frac{RC_2 g_m}{s^2 L_2 C_1 C_2 + s R C_1 C_2 + (C_1 + C_2)} \tag{8}$$

Since (8) is a second order transfer function, the resonant frequency ω_n and damping factor ζ are determined to be:

$$\omega_n = \sqrt{\frac{1}{L_2 C_1 \| C_2}} \tag{9}$$

$$\zeta = \frac{R}{2} \sqrt{\frac{C_1 \| C_2}{L_2}} \tag{10}$$

These two parameters are useful tools to analyse the points at which zero switching occurs. Parameters are chosen such that the resonant frequency ω_n matches that of the input signal, and the damping factor ζ is used to control the point at which zero switching occurs.
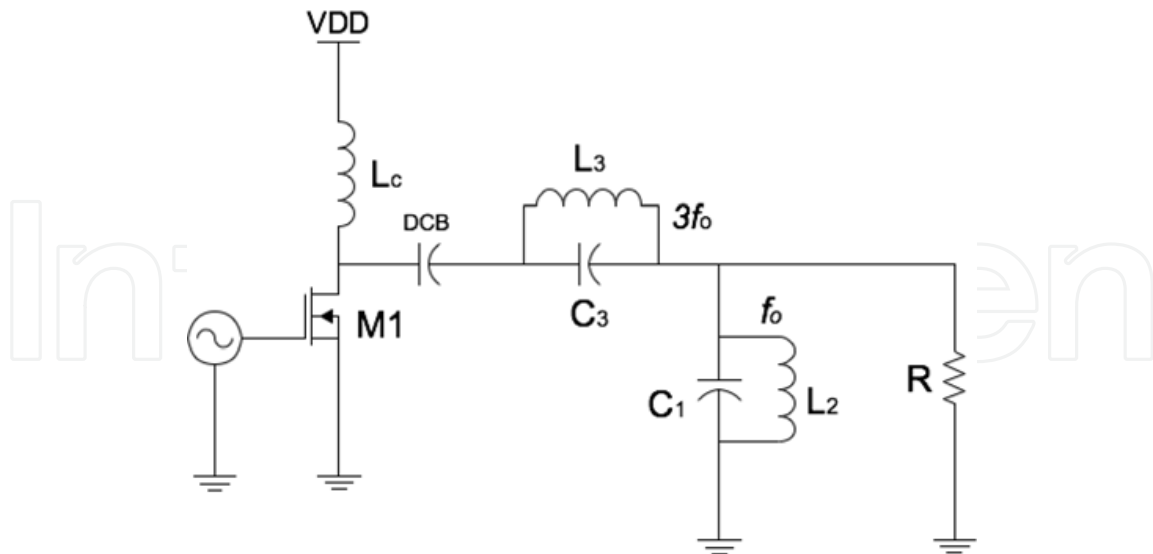
Fig. 9. The class-F power amplifier

2.3 Class-F Amplifier

The class-F amplifier, like its class-E counterpart, includes a transistor and a choke inductor. It differs from the class-E amplifier by its load network, which includes multiple resonating circuits that resonate at frequencies in the harmonic spectrum of the driving signal, which is supplied to the amplifier's input. Like its class-D counterpart, the amplifier does not account for the parasitic capacitance across the transistor's terminals, making it less commonly used as a wireless power transmitter compared to the class-E.

3. Coils and Energy Transfer

Wireless power is most commonly transferred with inductive links, which comprise primary and secondary coils that operate together as a weakly coupled transformer (Yang et al., 2007). The structure of each coil determines several factors in the coil's ability to inductively transfer power. It is important to consider the way in which different coil structures produce different field patterns. Aspects which influence the transfer of power with inductive coils involve coil size, separation, shape and performance at different orientations.

It is more efficient to transmit wireless power at lower frequencies (Vaillancourt et al. 1997), and as the complexity of implants is increasing, data rates are required to keep up with the increased sophistication. Wang et al. (2006) proposed the advantages of biomedical implants operating in dual frequency bands for sending power and data, and the dual-band concept has been extended upon in subsequent research. The optimal orientation of power and data coils were determined to be orthogonal, based on interference and spatial restrictions (Silay et al., 2008).

3.1 Spiral Coils

Biomedical implants mostly employ traditional wire-wound cylindrical inductors for the power transmitting and receiving coils. Zeirhofer & Hochmair (1996) investigated the enhancement of magnetic coupling between coils using a geometric approach. It is concluded that coupling is enhanced when turns of the coil are distributed across the radii rather than concentrating them at the outer radius of the inductors. Printed spiral coils are therefore worthwhile investigating for use in biomedical implants.
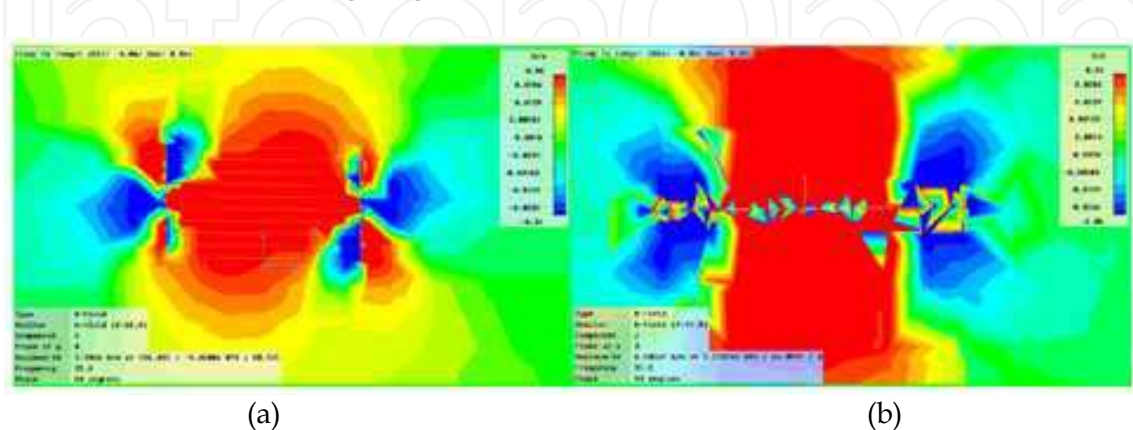

| (a) | (b) |

Fig. 10. A comparison of flux patterns for two similarly sized coils simulated in CST. (a)The coil on the left is a cylindrical helix coil, while the one on the right (b) is an Archimedean planar spiral.

Spiral coils are not always suitable for biomedical implants. Implementing a spiral coil for retinal prosthesis for example (Wang et al. 2006) may not be desirable from an aesthetic or psychological perspective. Similarly, if a large number of turns are required, the equivalent spiral coil would be much larger.

3.2 Spiral Coil Arrays

Research in this field has focused mainly on optimising certain orientations of primary and secondary coils, from finding optimal coupling factors to space-efficient packaging. In situations such as pre-clinical monitoring, the issues related to wireless power transfer to implanted devices become more difficult to manage, mainly due to random and variable movement by the subject of the experiments.

A common pre-clinical scenario involves an enclosure in which the subject is free to move. Zimmerman et al. (2006) investigated the optimisation of wireless power transfer in such a situation, monitoring the overall transfer efficiency by varying transmission frequency and the number of turns on the secondary coil, which was 1cm from the primary coil. The system produced 3V at 1.3mA in the implant itself, accounting for a tilting angle of 60^0. The primary coil was a cylindrical wire-wound coil, wrapped around the circumference of the base of the enclosure.

Fig. 10 shows the difference in field patterns between two similarly sized coils in the order of 100mm x 100mm, one a cylindrically wire-wound coil and the other, a planar Archimedean spiral as simulated by advanced electromagnetic simulation software from

Computer Simulation Technologies (CST™). The green areas in the colour plot show zero magnetic field intensity. The spiral coil produces a stronger field for a similar size, which makes it a clear point of interest to investigate for a power transfer scenario in an enclosure. The planar profile of the coil is also less intrusive. If a single power transmission coil is implemented on the base of an enclosure as a spiral, it is interesting to consider the fact that the magnetic field produced by the coil will not be uniform. In fact, it varies greatly within the spatial limits of the spiral, as shown in Fig. 12 (a) and (b). Note that there are permanent "blind spots" within which the secondary coil will not be able to receive any energy.
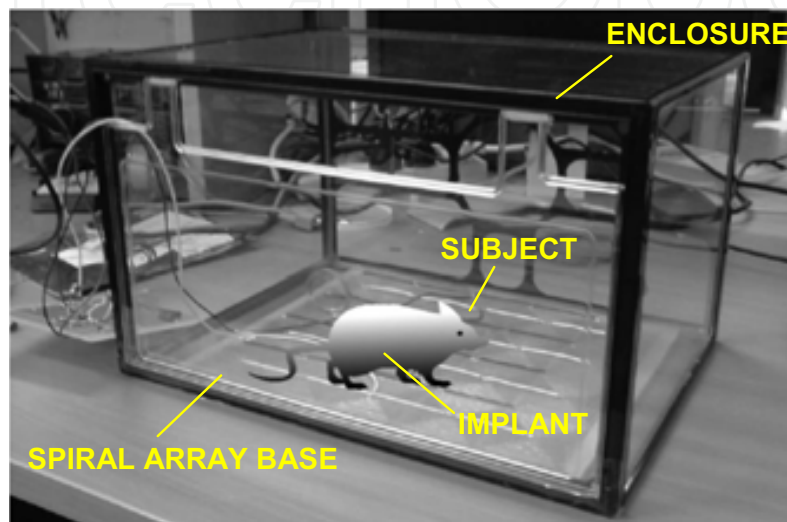

Fig. 11. A sample pre-clinical testing scenario showing an enclosure, n array of spirals connected in series and parallel and a subject.

This leads to the idea of producing an array of several small coils, connected in series or parallel networks, depending on the desired array's impedance. Fig. 12(b) shows a simulation conducted in an identical scenario and scale to that of Fig. 12(a) . It is evident that there are smaller and less powerful neutral areas, appearing green, while having more distributed and more intense zones. This idea was extended to a simulation of 20 coils connected in an array. Fig. 13 shows that the z-component of the magnetic field level on the surface of the base had very few neutral green zones.

The behaviour of the spiral array should be investigated further, by experimenting with materials of different dielectric constants, eventually introducing non-live tissue such as raw meat, with the aim to resemble its performance in an implanted environment. Fig. 13 shows the system concept of collecting data from an implant in a mouse roaming on the spiral array base. The orientation of the mouse varies and simulations are required to understand the coupling between the spiral array base and the implanted coil.
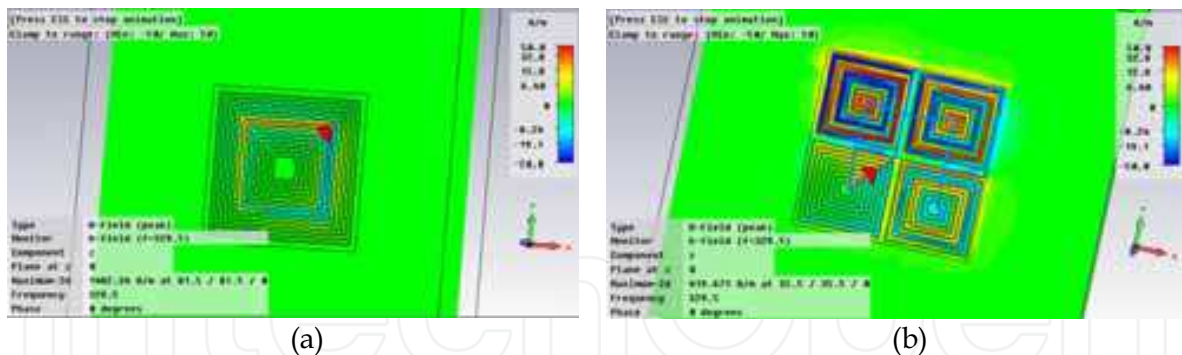
Fig. 12. A comparison between two scenarios simulated in CST. (a) The magnetic field pattern of the z-vector on the plane of the spiral. (b) Four spirals covering the same area as (a).
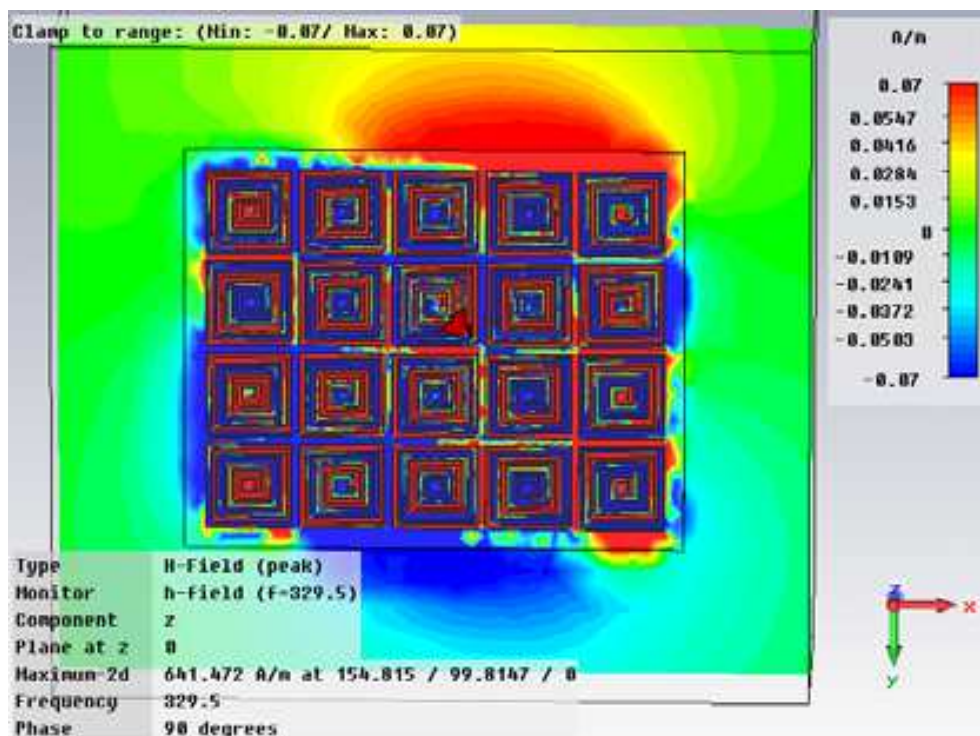

Fig. 13. A simulation conducted on CST to determine the magnetic field patterns of an array of 20 spiral coils.

4. Rectification

Rectification is an important element of wireless power transfer, in that it is vital to capture and utilise the power received at the secondary coils of the implant. In higher power non-biomedical applications, rectification is a straightforward task. Diode bridges and voltage regulating units such as the ones shown in Fig. 14(a)-(b) are typical examples. However, implementing this at higher frequencies with lower power levels is difficult. The first obstacle is the diode, which is the basic building block of any rectifier. It has an immediate voltage drop ranging from 0.1 to 0.7V, depending on the type of diode, meaning that it is favourable to use less number of diodes in most medical inductive power transfer

applications (Li & Bashirullah, 2007; Chaimanonart & Young, 2006; Zimmerman et al., 2006). The highest frequencies used to transfer power inductively are in the order of MHz, and most designs simply send more power to compensate for the forward voltage drop of the rectification diodes (Sauer et al., 2005).

(a)   (b)   (c)
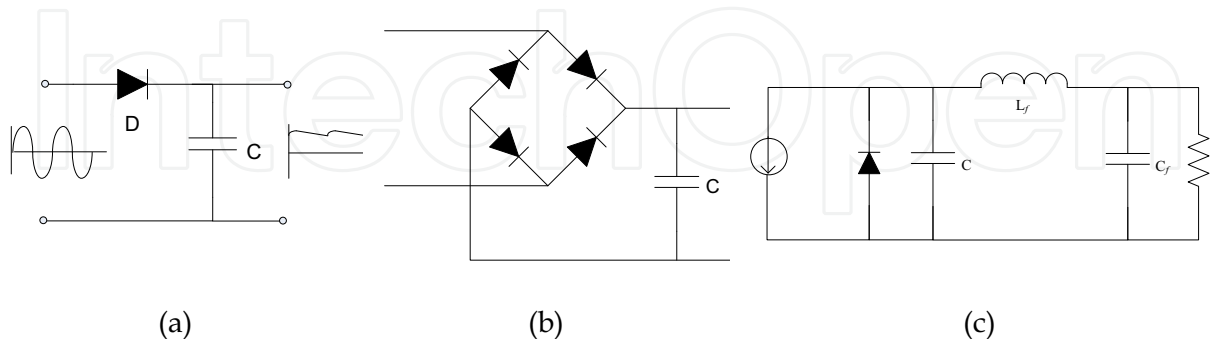
Fig. 14. (a) Half-wave diode rectifier. (b) Full-wave bridge rectifier. (c) Class-E rectifier (Kazimierczuk, 1989).
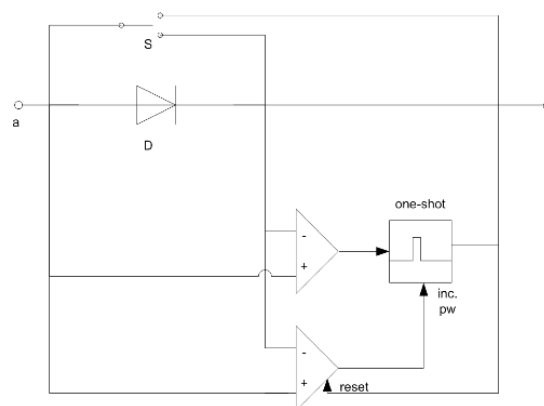

Fig. 15 An active diode presented by Lehmann & Moghe (2005)

The Class-E rectifier, shown in Fig. 14(c), has the potential to work well at high frequencies in implantable devices. It is designed for high efficiency operation at high frequencies due to its zero switching properties. It is mostly used in DC/DC converters (Kazimierczuk & Jozwik, 1989; Reatti et al., 1993), and despite not having been implemented in biomedical applications, it is an attractive circuit for use in implantable electronics.

A concept introduced by Lehmann & Moghe (2005) aims to remove the voltage drop on a rectifier all together by using comparators to sense the initiation of the diode's forward-bias, then sending a one-shot pulse to a switch, allowing the energy to bypass the diode and supply the load, as shown in Fig. 15. The concept has been tested with an input voltage at 5 MHz and the target implant supply voltage at 3V with variable loads of 2kΩ and 10kΩ. The results from these investigations show up to a 70% increase in received power compared with an on-chip passive rectifier.

5. Conclusion

This chapter has presented an overview of current techniques in the field of wireless power transfer technology for biomedical implants, including the types and theory behind switch mode power amplifiers, techniques in transmitting between two coils, and the types of rectification employed in implantable devices. System optimization can be achieved in all three sections. Class-E power amplifiers provide a robust and highly efficient means to transmit power to implantable devices due to their zero voltage and voltage derivative switching. The magnetic coupling between the primary and secondary coils can be analysed using full wave simulators. This enables one to achieve optimized coil designs after taking tissue properties and device orientation into account. The arraying technique of primary coils has shown promising results in eliminating blind spots. Several interesting solutions are developing for the rectification of received power to implants, including creative ways of avoiding voltage drops across diodes, which are a fundamental element of rectification.

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Biomedical Engineering can be seen as a mix of Medicine, Engineering and Science. In fact, this is a natural connection, as the most complicated engineering masterpiece is the human body. And it is exactly to help our "body machine" that Biomedical Engineering has its niche. This book brings the state-of-the-art of some of the most important current research related to Biomedical Engineering. I am very honored to be editing such a valuable book, which has contributions of a selected group of researchers describing the best of their work. Through its 36 chapters, the reader will have access to works related to ECG, image processing, sensors, artificial intelligence, and several other exciting fields.

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