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High Efficiency Inductive Link of Wireless Power Transfer for 0.18 um CMOS Implantable Medical Device in Telemedicine Embedded System

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Abstract: *The Implantable Medical Device IMD is a computer system that is Embedded into the human body. IMDs have extended the ability of physicians to diagnose and treat diseases, making great contributions to health and improve the quality of life of patients and, play an important role in keeping them alive. The new generations of IMDs are increasingly incorporating more computing and communication capabilities. The solutions to optimize the battery life time challange are RF energy harvesting, wireless power transfer and provide lower power transmitters. In this paper, methodologies to transfer and harvest energy in implantable medical devices are introduced and discussed to highlight the uses and significances of various potential power sources followed by in-depth discussions of the most popular method, namely inductive linking of wireless power transfer. The circuits have been designed in UMC 0.18 um CMOS technology. The circuits simulations were done using Cadence Tools to optimize the design and get the required specifications for the proposed system. In general, the choice of power amplifier's operating class is based on requirements regarding linearity and power efficiency. For applications in which linearity is a critical issue whereas good efficiency is desired, linear-mode amplifier e.g. Class AB is suitable with a simple topology, leading to optimal costs of production and system operation. So, In this paper, a 13.56 MHz Class AB CMOS PA with CG driver stage followed by CS power stage was designed using the proposed method for WBAN applications. The simulation results showed that the proposed PA achieved maximum PAE of 34.74% has an average gain of 21.6dB over the frequency of interest in Industrial, Scientific and Medical (ISM) bands.*

Index: *CMOS analog circuit design , Integrated Circuits , Implantable Medical Device IMD , Energy Harvesting , Inductive Link , Wireless Power Transfer WPT, Wireless Body Sensor Network WBSN, Medical Wireless Technology, remote powering , power management, Telemedicine.*

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I. Introduction

Recent development in Radio Frequency (RF) CMOS technology creates new trends in medical industry, which improves the quality of services in diagnosis and medical operations. Wireless body sensor network (WBSN) is a new technology that allows collecting vital information from implanted body sensing devices acting as nodes of the WBSN [1]. Implantable Medical Devices (IMD) are spreading widely in telemedicine system. The medical developments in terms of implantable devices have brought a robust change in the life of people. Medical devices have extended the ability of physicians to diagnose and treat diseases, making great contributions to health and quality of life. The approach to quality of devices has depended largely on regulation. The critical nature of medical devices has caused them to come under stringent regulations. Clearance to market devices is granted only after the Food and Drug Administration (FDA) has determined through its classification and review procedure that there is reasonable assurance of the safety and effectiveness of the device. Such regulatory requirements are necessary and appropriate [2]. A rigorous but responsive and responsible regulatory process helps to ensure that new medical technologies represent the state of the art, have the real potential to do good as demonstrated in scientifically grounded studies and reach patients promptly. It also damages global competitiveness and increases healthcare costs directly and indirectly. The IMDs technical capacities are expanded, making it possible to establish Internet connection in case of necessity and emergency situation for the patient. While the web connectivity of implantable devices was advanced, Wireless Sensor Network (WSN) based systems can realize long-distance signal transmission so as to make it possible for doctors to follow up patients' health conditions remotely. By exploiting capabilities of wireless medical devices physicians are able to conduct real-time monitoring without causing any inconvenience for their patients. Implantable devices are studied and analyzed based on systematic called nano-medicine as shown in Figure.1 [1]. In this figure two main subcategories: "Implantable Sensors" and "Sensory Aids" are shown. Usually, using batteries is undesirable because of weight, size and life time issues. Therefore, energy harvesting or remote powering solutions are necessary. However, harvesting the power inside the living body is insufficient to power multi-sensor systems with short range two-way communication. Remote powering provides a reliable solution to continuously monitor the patient.

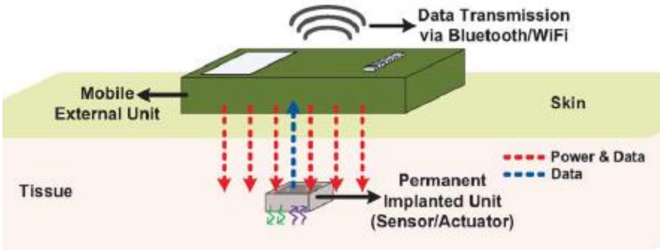

Figure 1: Implantable Medical devices in systematic nano-medicine view

Figure 1.illustrates a stand-alone implanted system. A permanent implanted unit is placed under the skin inside the body and consists of sensors and/or actuators. To avoid using the batteries, the implanted system is activated and charged by a mobile external unit. The operation of the implanted system is initiated by relocating the implant with the mobile external unit. The measured data is transmitted from the implanted unit to the external unit. For the comfort of the patient, the external unit is removed after finishing the operation. The external unit also configures the implant by down-link communication. The mobile external unit communicates with a control unit such as a smartphone or a base-station via bluetooth and WiFi. Power added efficiency is an important metric in the PA design as it measure how effectively the DC power is converted to RF output power. In addition, group delay is very important as it is measure of phase nonlinearity [3].

In this paper, a proposed design for maximizing the power added efficiency is presented. A class AB CMOS PA with CG driver stage followed by CS power stage was designed using the proposed method for WBAN applications. The simulation results showed that the proposed PA has an average gain of 21.6 dB over the frequency of interest 13.56 MHz. In addition, The PA achieved maximum PAE of 34.5% over the frequency of interest in Industrial, Scientific and Medical (ISM) bands.

II. Implantable Medical Devices: Roles and Future Expectations

Definitions of self-care, e-care, mobile-health and Internet of Things (IoTs) have emerged, changing the traditional roles of doctors and patients. People are capable of monitoring their health every day with innovative devices while large scale medical data could be assessed and diagnosed continuously by caregivers located thousands miles away. In many cases, the devices need to be inserted to stay safely and securely in the body for a period of time; thus a new expectation is raised for IMDs that the implant needs to be able to communicate with external units for real-time tracking and sensing, diagnosis and treatment. In order to achieve that, a sufficient power source becomes mandatory, no matter whether the device is active or passive. In this upcoming scenario, power is needed not only for running the implant but also for feeding the high-consuming wireless communication, and this would concurrently drain out the battery of the implant and consequently reduce the lifetime of the device. A conceptual view of a future IMD is sketched in Figure 2. With these expectations, an IMD (1) can be powered from an external unit, such as hat (2) if the IMD is in the brain/head location. If there is a subcutaneous IMD in the arm/hand location, it can be powered by wrist device, (3) For wireless communications or wearable belt (4) if the IMD is located in the abdomen, IMDs can directly send data to a smartphone or via external units. Algorithms may be applied to filter, interpret and sort the data for diagnoses, storage or real-time examinations by distanced care-givers. Big data processes and mobile cloud systems will be needed to facilitate the infrastructure. This will be the model of healthcare in the future which can significantly save time, and cost, providing efficiency and efficacy for our society. Obviously, reliable micro-sensors and communication, durable power and systems as well as smart algorithms and designs, are essential. Other issues such as safety, usefulness, fashion and comfort also need to be improved in order to bring cutting-edge not only to patients but also general public.

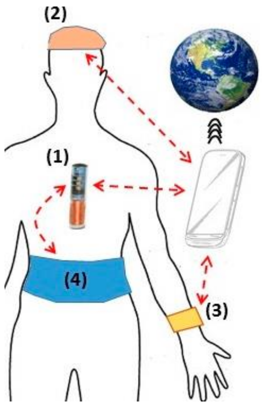

Figure 2 : Overview of IMDs and other components. (1) Implant ; (2 - 4) External devices

III. Embedded Systems and Implantable Medical Devices IMD

Embedded systems are regarded as an innovative and lightweight technical solutions. Main benefits of embedded systems are: connectivity, productive performance and cheap cost of components. These advantages make embedded systems widely applicable in the industry of implantable devices [3]. The framework of Nano-medicine has taken advantage of computing devices integration. A variety of electronic medical tools are being employed to address the needs of healthcare system. For example, continuous and automatic management of health conditions would not be possible without involvement of implantable medical devices (IMDs) which are type of medical devices embedded inside the human body for medical intervention. Although implantable devices so far have met wide approval of the professionals, several limitations in the design and structure of IMDs still impede further expansion of the implant market. Hence, development of completely new technique is strongly required to prevent any undesired external impact on IMDs.

III.1 Mapping IMDs to Embedded Systems :

Embedded device and systems need many considerations leading to secure system design and development maturity. Generally, the internal structure of IMDs involves sensors, radios, actuators, batteries, CPUs etc. We identified a number of aspects which play the key role in mapping the implants to embedded systems. One of the recent developments of modern nano-technology are “wearable” or “implantable” sensors which monitor health conditions and obtain medical data of high accuracy. Enhancement of micro-electro-mechanical systems (MEMS), progress in developing digital electronics and new opportunities of wireless communication cumulatively allow for substantial expansion of the range of implantable and indwelling medical sensor-based tools. Assuming that, the devices appropriate for implantation in the human body hereby have become the key focus area for many scientists [4].

III.2 Designing Challenges of IMD :

The main objectives of IMD designers are to improve reliability of device and enhance its safety features, meaning while reducing its power consumption and usage costs. In the early years of implantation practice the manufacturers primarily focused their efforts on enhancing battery power of the implants, wireless communication or testing new materials that could be used in implant production. Considering that IMDs have immediate contact with the patient’s organism, all IMDs are subject to safety and effectiveness checks to assure strict compliance with standards and requirements accepted health care. However, this general approach needs to be reconsidered in response to new risks. Among the existing deficiencies of IMDs, weak control system is the most critical factor in destabilizations of device’s functioning. A single threat model cannot be adopted for all IMDs: each and every implant is designed to serve a different purpose, hence, sensibility and deficiencies of the devices largely vary.

III. 3. Communication Challenges for IMD:

Figure 3 overviews the FCC standards and technologies for short and long range devices and applicable US regulations. FCC classified WSN-based medical systems into two groups: short-range devices and long-range devices. During the long-range transaction data is received by remote monitoring station, whereas short-range transactions are intended for locally based receivers. Recently, FCC revised its standards for medical devices to keep up with the technological progress. Thus, wireless medical products are released to US market only having been authorized by FDA and certified in FCC. With regard to the fact that wireless systems incorporate radio technology, all such devices fall under the scope of FCC’s regulation that are listed in Figure 3. As known, both Zigbee and Bluetooth protocols demonstrate best efficiency in transmitting data at short ranges and with low expenditures of power. In contrast to Bluetooth, which is usually used for facilitating connection between receiver and base, Zigbee protocols are mainly devised for managing chain of interconnected system. Accordingly, groups of implants may as well be controlled with the aid of Zigbee[5].

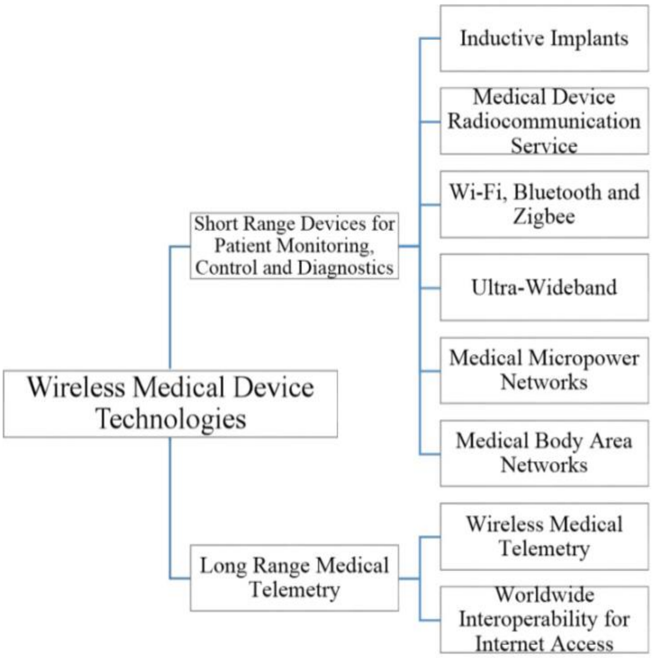

Figure 3: Wireless Medical Device Technologies' category

This work will use the **inductive implants** of short range devices for patient monitering, The following bands are designated for ISM applications of radio communications services in accordance with RRS5.150 of the Radio Regulations [6] :
13.553 – 13.567 MHz (centre frequency **13.560 MHz)**
26.957 – 27.283 MHz (centre frequency 27.120 MHz)
40.66 - 40.70 MHz (centre frequency 40.68 MHz)
433.05 - 434.79 MHz (centre frequency 433.92 MHz)
2 400 - 2 500 MHz (centre frequency 2 450 MHz)
5 725 - 5 875 MHz (centre frequency 5 800 MHz), and
24 - 24.25 GHz (centre frequency 24.125 GHz)

IV. Methods of Powering IMDs:

In this paper, methodologies to transfer and harvest energy in implantable medical devices are introduced and discussed to highlight the uses and significances of various potential power sources followed by in-depth discussions of the most popular method, namely inductive linking. We introduce methods to power active IMDs, approaches are categorized into two main groups: (1) IMDs that work independently with or without one-time battery, or sustainably; and (2) IMDs which could be either battery-based or batteryless, obtaining power transferred from an external unit. Figure 4 presents these two groups [7].

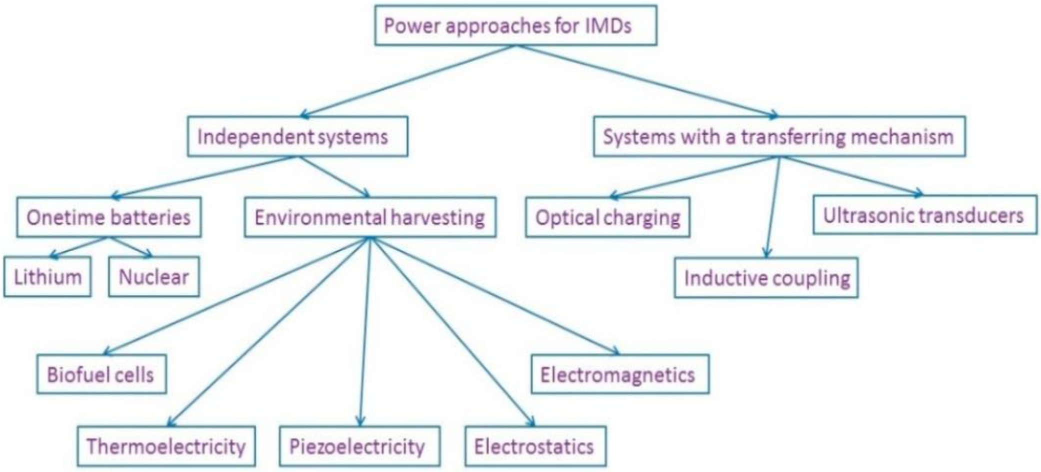

Figure 4 : Approaches and Methods of powering IMDs

4.1. Independent Systems :

In general, batteries store energy in the forms of chemical substances which can produce electricity. Batteries contain anodes, cathodes and electrolytes to allow ions to move thus forming currents. There are three categories of power capabilities used for battery performance (low rate, medium rate and high rate). Although the definition is not yet clear, low-rate batteries should be able to provide a constant current of 100 μA while high-rate ones can supply a pulse power of at least 5 W for 10 s. Thus, the gap could be filled with medium-rate batteries. Among IMDs, pacemakers use low currents, neuro-stimulators and drug pumps need medium-rate ones and implantable defibrillators/cardioverters (ICDs) require extremely high power as well as additional longevity [8].

4.2. Dependent Systems (Impantable Systems with an External Unit) :

While independent systems can provide solutions which bring comfort and avoid complexity, they showing their weaknesses in the areas of reliability, low output power (MFCs, thermoelectricity) as well as potential to cause toxicity and failing to deliver biocompatibility. Systems with external units to transfer energy continuously are therefore of interest with the increasing need of communication between IMDs and smart devices. The power can be sent through the body tissues optically, mechanically or electromagnetically.

4.2.1. Inductive Link Powering :

Recently, inductive coupling has been widely investigated to power up IMDs. The principle behind is based on a mutual inductance between two coils in which one is located outside the body while the other is integrated with the implanted device. As the external antenna transmits a varying electromagnetic signal, a voltage would be induced in the receiver coil Figure 5. The wireless power efficiency depends on the resonance frequency (or operating frequency), distance, alignment, and coupling matching between the transmitter and the receiver coils [1,2,5,9]. In addition, the inductive technology can be exploited to transmitted data from outside to inside the body and vice-versa without using a radio-frequency (RF) transmitter or receiver with data rates up to few hundreds kilobits per second [1,2,10,11].

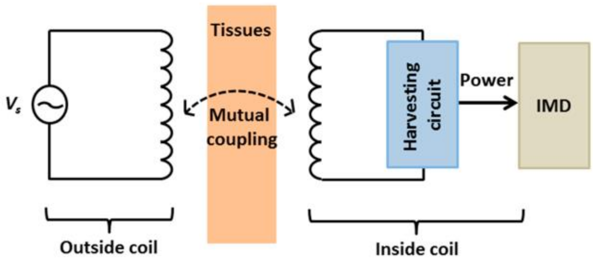

Figure 5. Inductive link Powering

Parramon et al. developed an inductive coupling power source which could generate 19mW at a carrier frequency of 10 MHz. This was used to power a microsystem for electromyography (EMG) recording implanted in rabbit muscle. The diameter of the inside coil, the diameter of the outside coil, and the distance between the two coil were 10 mm, 20 mm and 15 mm, respectively [12]. Further, in 2007, Ghovanloo and Najafi demonstrated a system-on-chip (SOC) by combining application-specific integrated circuit (ASIC) design with off-chip components (LC tank; filters) delivering 50 mW over a 5 mm distance at 5/10 MHz frequency [13]. There are several essential factors in inductive coupling approach, such as misalignment, size of the implant antenna, unknown side effects and the limited carrier frequency due to tissue absorptions, thus low-megahertz ranges (0.3–30 MHz) have been widely used [5].

4.2.2 Inductive Linking : Possibilities and Challenges

Wireless powering via an inductive link has been extensively investigated for a wide range of applications in the recent years owing to the capability to provide sufficient power, the reliability and the possibility to integrate it with other electronic components. Further, the inductive link could also be exploited for data transmission [2,4,13]. In the domain of medical devices, numerous approaches were proposed with innovations in design, materials and circuits in order to target specific applications. For IMDs, the operating frequencies are usually in the low MHz or KHz range to minimize the power absorbed by the tissue which may cause tissue heating and side effects [6,13]. A number of studies have been carried out to improve the power transfer efficiency **(PTE)** of regular 2-coil systems [14,15], in which coil geometry, coil dimensions, number of turns, and coil losses were examined for optimization. Recently, MEMS technology and advanced materials have enabled the fabrication of miniaturized coils on either flexible or hard substrates, which could be integrated monolithically with other electronic components. While the problem of a smaller coil resulting in less coupled energy remains unsolved, another challenge that the thin fabricated metal film of the coil (usually <1 μm) would cause an extremely low quality factor (Q), and consequently low PTE.

In practical scenarios, most anatomical surfaces are highly curved and most organs are relatively mobile during daily activities, requiring IMDs to be flexible and to remain functional. However, the inductance of the flexible coil antenna would change with respect to mechanical environmental cues, thus varying the resonance frequency of the LC tank in the IMD. If the changes are significant, the PTE would be reduced drastically. This calls for the implementation of an adaptive mechanism to ensure the resonant frequencies are matched on both sides [15,16]. However, this may add components to the room-limited IMDs. Another practical issue is **misalignment** which has been widely investigated [4,16,17]. For specific IMDs attached to moving organs such as the stomach, positional and angular misalignment becomes critical. Recent investigations have shown the superiority of spiral structures for transmitter antennas to produce a large-cover beam size, minimizing misalignment issues [18]. Nonetheless, misalignment is case-dependent and unavoidable, requiring thorough calibration and investigation in a simulated environment before actual use.

Instead of using a conventional 2-antenna system, multi-input and multi-output (MIMO) systems have been investigated [19]. Obviously, as the field is continuously active, multiple IMDs can be used simultaneously, however it would lead to a complicated case as it affects the mutual inductance between any two antennas and consequently, the PTE. Further, it is also difficult to place a 'repeater' to improve the PTE and distance. The most possible case that could help is using multiple transmitters with one receiver in an IMD[20][21]. For example, two or more transmitters can be located around the torso to increase the power sent to IMDs. Nonetheless, it would be hard to obtain constructive superposition and field optimization in dynamic cases with the inevitable daily activities of users. It would be improper to not mention the biological effects caused in tissues by the electromagnetic field [21]. Although it has been studied, attention was paid to the acute effects generated by heating, while long-term health issues are concerns preventing patients and the public from accepting and using IMDs with inductive coupling.

V. Inductive link powering for Implantable Medical Devices :

The solutions to optimize the battery life time challange are RF energy harvesting, wireless power transfer and provide lower power transmitter. In this work we will use Inductive linking for wireless power transfer. The power consumption of typical implanted system with advanced sensing and processing blocks is estimated to be 2–3 mW. Accordingly, we have used magnetic coupling for remote powering, because for short distance (less than 10 cm) it is the most effective and the penetration in the body is better compared to the other methods[22]. On the other hand, the size constraints limit the area of implanted coil to 12 mm. In order to design optimal coils, number of turns and conductor width and spacing are critical elements. In addition, it is necessary to find the optimal frequency and transmitting coil size. The frequency choice aims to maximize the induced voltage which is given by

$$V_{ind} = \mu_0 \,.\, \mu_r \,.A.\, N \,.\, \omega \,.\, H_{eff} \qquad (1)$$

where μ_0 , μ_r , A, N , ω *and* H_{eff} are the magnetic permeability constant of vacuum, the relative permeability of implanted medium, the loop area of the implant coil, its number of turns, the angular frequency of the effective magnetic coupling, and the effective magnetic field strength, respectively. The induced voltage is proportional to the frequency. However, AC resistance of a coil also increases as the conductive losses due to skin and proximity effects start to dominate after a certain frequency [15]. The frequency choice is also crucial to deliver the power efficiently in the body. The dielectric properties of the body tissues change with the frequency. Accordingly, the penetration depth of a signal inside the body decreases with frequency [16]. The optimal power transmission frequency depends on the kind of the body tissues, the distance between the transmitter and receiver and a possible air gap. In order to determine a value suitable for a generic use, assuming a tissue (fat) with thickness of 2 cm, a 3-D simulation determines an optimal frequency[17]. Figure 6 shows the tissue effect on the optimal power transmission frequency and also the power transmission efficiency. The optimal frequency shifts from 31.3 MHz to 17.1 MHz considering the tissue effect. Accordingly, the power transmission frequency is moved to the closest ISM band of 13.56 MHz which is authorized for the inductive remote powering applications [18]. The power transmission efficiency of the remote powering link is maximized when the coils are tuned at the same operation frequency.

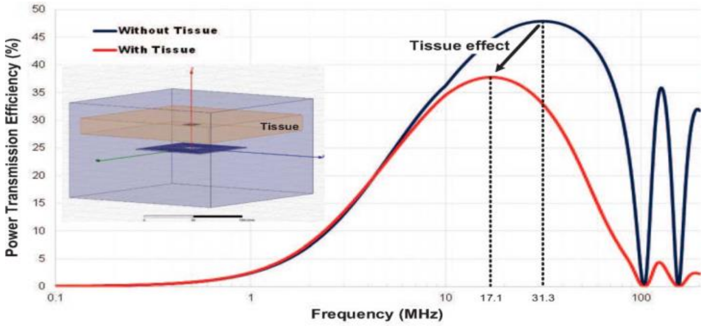

Figure 6. Tissue effect on optimal power transmission frequency[17]

Assuming that the coils are tuned to the same frequency, the maximum power transmission **efficiency** for a certain load can be expressed by [19]

$$\eta_{\max} = \frac{1}{1+\dfrac{1}{k_{12}^2\, Q_{PC}\, Q_{IC}}} \qquad (2)$$

where k_{12}^2 , Q_{PC} and Q_{IC} are the coupling coefficient, the quality factors of the unloaded Powering Coil and the unloaded Implant Coil, respectively as shown in Figure 7. The maximum power efficiency can be obtained by maximizing not only the coupling factor, but also the quality factors of the coils [20], [21]. A geometric optimization of the remote powering link is required to obtain an efficient remote powering link. However, dependency of remote powering efficiency to many geometric parameters of the coils, e.g., number of turns, width of the conductors and spacing between them, outer and inner diameters and also the coupling between the coils, makes the optimization extremely complicated. The geometric parameters of coils are swept to obtain the optimal coil pair. The lumped model parameters of the optimized coils which are shown in Figure 7 are extracted by using analytical equations and field solver software[22]. The power transfer efficiency PTE of the remote powering link is calculated by using the equation(2). The geometric parameters of the coils are characterized in a 3-D electro-magnetic field simulation software[17]. Finally, the coils are produced on printed circuit board (PCB) for reliability and reproducibility and characterized for operation frequency. Figure 7, also shows the external link driver. Its output power depends on the amplitude of the supply voltage which is adjusted to deliver the required power level to implant. A power feedback control, controls to continuously supply sufficient power to the implant despite coupling variations.

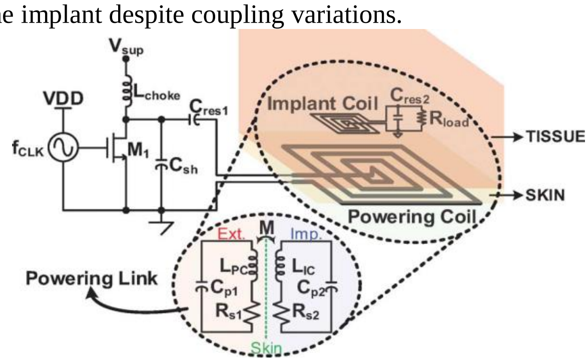

Figure 7. Inductive link powering and external link driver

The remotely powered systems consist of two parts. The external part generates the magnetic field for the implanted system and receives the data transmitted by the implanted system. The internal part creates a supply voltage for the implanted system from the induced current and communicates with the external part. The external unit is composed of a link driver, a receiver, a supply controller, and a powering coil. **The external link driver is chosen as Class-AB power amplifier** to drive the powering coil due to the **high drain efficiency of Class-AB**. Figure 7 shows the inductive link powering and external link driver of the remote powering link. The link is presented by L_{PC}, L_{IC} and *M*, are the powering and implant coils and mutual inductance, respectively. The C_{res1} , C_{res2} capacitors are used for tuning the coils for operation frequency. In addition, R_{Load} represents the load of the remote powering link which is the input resistance of the implanted rectifier. The M1, L_{Choke} and C_{sh} present the switching transistor of the amplifier, the choke inductor and the shunt capacitor, respectively. L_{Choke}, C_{sh} and values are chosen in order to maximize the power efficiency of the power amplifier [22]. The operation frequency of the amplifier is fixed to 13.56 MHz by an external oscillator circuit. The output power (Pout) delivered to the load R_L , seen by PA, is proportional to:

$$P_{out} \ \alpha \ \frac{V_{\sup}^2}{R_L} \tag{3}$$

The Class AB amplifier output stage combines the advantages of Class A amplifier and Class B amplifier while minimising the problems of low efficiency and distortion associated with them. The Class AB Amplifier is a combination of Class A and Class B in that for small power outputs the amplifier operates as a class A amplifier but changes to a class B amplifier for larger current outputs. This action is achieved by pre-biasing the two transistors in the amplifiers output stage. Then each transistor will conduct between 180 and 360 degrees of the time depending on the amount of current output and pre-biasing. Thus the amplifier output stage operates as a Class AB amplifier. A Class-AB power amplifier plays the role of amplifying the modulated RF signal and driving the antenna through an integrated matching network.

VI. Propoed Embedded Electronics for Inductive Link Powering for IMD :

This work design a Class AB structure which is a cross between Class A and Class B structure. The idea here is to use a push-pull structure, where each device is biased slightly above threshold. Circuit implementations of a Class AB PA are similar to those of the Class B architecture; what differs is the biasing and output waveforms. This architecture is like Class A in that each device does carry current under nominal bias, but it is like Class B in that neither device is on for the entire cycle. By allowing the two devices to conduct current for a short period, the output voltage waveform during the crossover period can be smoothed out and thus can reduce the distortion at the output. Thus this structure can provide linearity close to Class A structure with **efficiency** close to class B structure.

Depending on whether linearity or efficiency is the dominant metric, the bias point can be chosen to be close to the threshold (the Class B bias point), in which case both efficiency and linearity would approach the Class B levels, or it can be chosen such that the device remains on for most of the input cycle (closer to the Class A bias point), in which case the efficiency and linearity would start to approach the values of a Class A PA. Several Class AB PA's have been reported in the literature, with efficiencies ranging from anywhere between 30% and 60% [18].

The critical selection is the chosen architecture and bias point. For output power levels of 20 dBm, a linearly amplifying PA biased in Class AB with little distortion losses is a good choice for one-stage PA design. The next step is to look at the transistor width. The goal of around 25 to 35 % drain efficiency signifies a DC current of around 200 mA. This in turn leads to a transistor width of 1 mm for a gate bias between 1.1 V and 1.2 V.

6.1 Design Optimization of CMOS Class AB PA as a link driver for IMD :

The proposed design will be given in this section. Figure 8 shows the schematic of the power amplifier followed by 2-section LC matching network to transform the antenna load of 50 Ω to the load of the PA. The first section is constructed by a 10.5 nH series inductor (L1) and a 0.11 pF shunt capacitor (C1). The second section is constructed by inductor (L_{bond}) of about 2 nH inductance value. The notation C_{pad} is used to model the parasitic capacitance of the pad and the ESD protection. Post-simulation shows that $C_{pad}$is 0.538 pF.L_{bond} together with C_{pad} form the second section of the matching network. The simulated characteristics (S_{11} and S_{21}) of the matching network is shown in Figure 9. For operating frequency of 13.56 MHz, the pulse width is T = 1/f = 73.746 ns. The half of pulse width is 36.87 ns. We choose the shunt capacitor to the drain terminal of the PMOS as large as possible such value is 10pF or 20pF, Figure 16. The transistor sizes of the NMOS and PMOS at the switching stage were 1.6mm and 4mm respectively. V_o = 4.153 V , $P_o = (4.153)^2 / 2x50 =$ 173 mW = 22.4 dBm.

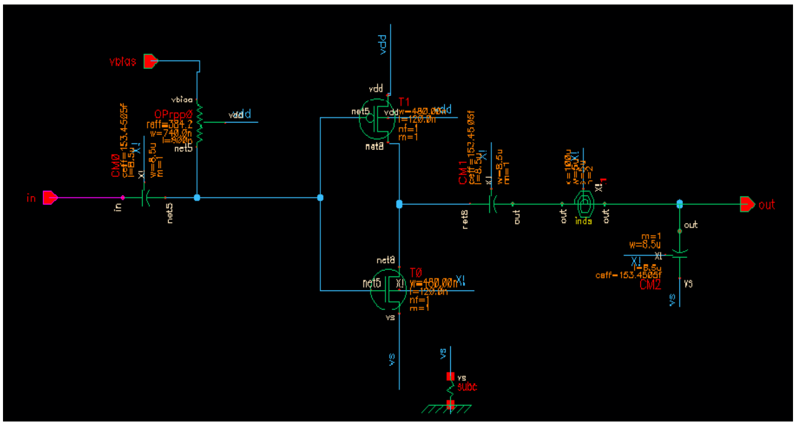

Figure 8. Schematic View of the proposed CMOS Class-AB power amplifier

The DC current from Vdd=5V is 99.75 mA, then the dissipated power is 0.498 mW (~0.5mW).

Then drain efficiency is $\eta_D = \frac{P_o}{P_{dc}} = \frac{0.173}{0.498} = 34.74\%$, The input power is $P_i = 5X\ 1.222X\ 10^{-4} = 2.05mW$

Then the Power Added Efficiency $PAE = \frac{P_o}{P_{dc} + P_i} = 34.74\ \%$

Performance summary of CMOS Class-AB PA , Frequency 13.56 MHz, V_o =1 V, Vo= 4.153V, P_o =173mW = 22.4 dBm, PAE= 34.74 % , P_D =0.498 mW

Simulated results scattering parameters S11 and S21 of Class-AB power amplifier.

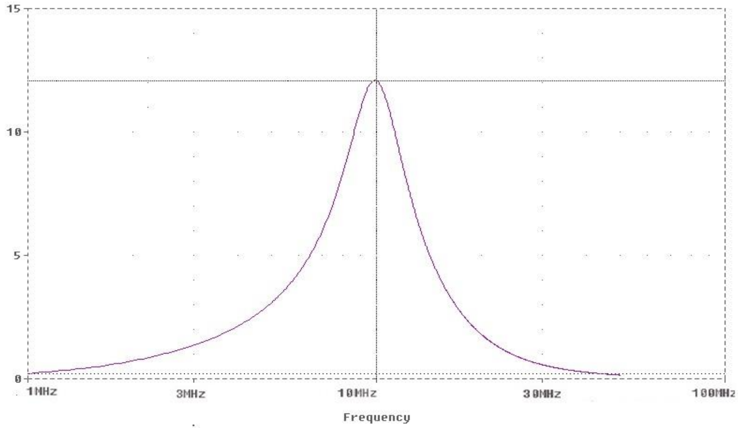

Figure 9. Voltage Gain S_{21} = 21.6 dB

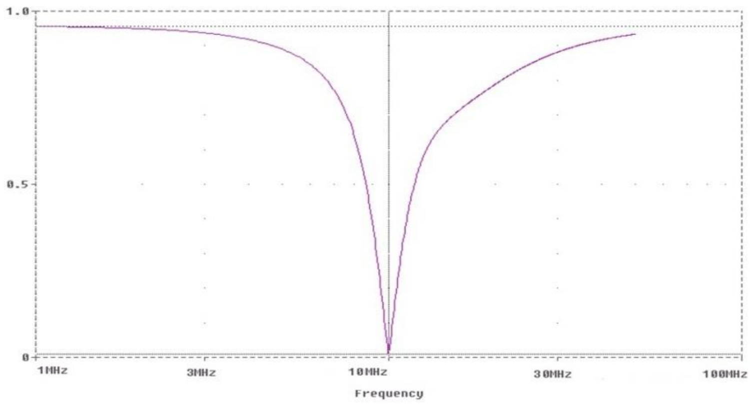

Figure 10 Input Reflection Coefficient S_{11} = -38.6dB

VII. Conclusion and Future work:

The paper presents a remote powering for the implantable medical devices to be used along with different sensor/actuators networks. A permanent implanted device is placed under the skin and stores the measured data. A mobile external unit charges the implanted device and receives the data from the implant.The choice of power amplifier's operating class is based on requirements regarding linearity and power efficiency. For applications in which high efficiency is desired, linear-mode amplifier e.g. Class AB is suitable with a simple topology, leading to optimal costs of production and system operation. A Class-AB Power Amplifier is proposed to improve the efficiency for the efficient power transfer. The power required by the implantable electronics including the one for the wireless communication is 0.5 mW. In this paper, methodologies to transfer and harvest energy in implantable medical devices are introduced and discussed to highlight the uses and significances of various potential power sources followed by in-depth discussions of the most popular method, namely **inductive linking**. The circuits have been designed in **UMC 0.18 um CMOS** technology. The circuits simulations were done using **Cadence Tools** to optimize the design and get the required specifications for the proposed system. So, In this paper, a 13.56 MHz Class AB CMOS PA with CG driver stage followed by CS power stage was designed using the proposed method for WBAN applications. The simulation results showed that the proposed PA achieved maximum PAE of 34.74% has an average gain of S21 of 21.6 dB and, and S11 of -38.6dB over the frequency of interest in Industrial, Scientific and Medical (ISM) bands.

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