A Remotely Powered Implantable Biomedical System With Location Detector

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Abstract—An universal remote powering and communication system is presented for the implantable medical devices. The system be interfaced with different sensors or actuators. A mobile external unit controls the operation of the implantable chip and reads the sensor's data. A locator system is proposed to align the mobile unit with the implant unit for the efficient magnetic power transfer. The location of the implant is detected with 6 mm resolution from the rectified voltage level at the implanted side. The rectified voltage level is feedback to the mobile unit to adjust the magnetic field strength and maximize the efficiency of the remote powering system. The sensor's data are transmitted by using a free running oscillator modulated with on-off key scheme. To tolerate large data carrier drifts, a custom designed receiver is implemented for the mobile unit. The circuits have been fabricated in 0.18 μm CMOS technology. The remote powering link is optimized to deliver power at 13.56 MHz. On chip voltage regulator creates 1.8 V from a 0.9 V reference voltage to supply the sensor/actuator blocks. The implantable chip dissipates 595 μW and requires 1.48 V for start up.

Index Terms—CMOS analog circuit design, data communication, implantable biomedical system, implantable medical device, inductive link, integrated circuits, location detection, locator, power management, remote powering, wireless power transfer.

I. INTRODUCTION

FUTURE trend in patient monitoring is to use implanted devices capable of measuring relevant vital parameters and transmit them to an external data processing station [1]–[4]. This offers multiple advantages, the small and light weight implants are more comfortable for the patient. Moreover, the possibility of a low power local data processing with data storage allows to achieve continuous long-term monitoring with a stand-alone implanted device. In addition, a mobile external station is removed for the comfort of the patient after the implanted device is charged and the data is received. Animal research is another application of implantable electronics [5], [6]. As an example, the effect of the new drugs on animals can be verified [7].

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Fig. 1. Conceptual design of stand-alone implantable system.

The key requirements for an implanted system are the size, the autonomy and the data communication capability. The system size normally should be less than 1 cm^3 [8]–[10]. Autonomy is important for operations which requires long monitoring periods. Usually, using batteries is undesirable because of weight, size and life time issues. Therefore, power 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.

Fig. 1 illustrates a stand-alone implanted system. A permanent implanted unit is placed under the skin inside the body and consists of sensors 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.

This paper presents an universal platform for implantable multi-sensor systems. Basic blocks for a micro-integrated system, with possibility of being assembled as a System In Package, SIP, are experimentally verified. In addition a locator system is proposed to find the location of the fixed platform in the body to power and communicate efficiently. The paper is organized as follows: Section II explains the details of the optimized remote powering link. The external and the internal electronics of the remote powering is discussed in Section III. Section IV overviews of the proposed locator system. Section V describes the electronics of data communication. The simulation and measurement results are presented in Section VI. Finally, Section VII concludes the paper.

Fig. 2. Tissue effect on optimal power transmission frequency.

II. OPTIMIZED REMOTE POWERING LINK FOR IMPLANTABLE SYSTEM

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 [11], [12]. On the other hand, the size constraints limit the area of implanted coil to 12 by 12 mm². In order to design optimal coils, number of turns and conductor width and spacing are critical elements [13], [14]. 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

$$V_{\text{ind}} = \mu_0 \cdot \mu_r \cdot A \cdot N \cdot \omega \cdot H_{\text{eff}} \quad (1)$$

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 [17] determines an optimal frequency. Fig. 2 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].

Fig. 3. Magnetically coupled remote powering link and external link driver.

The power transmission efficiency of the remote powering link is maximized when the coils are tuned at the same operation frequency. Assuming that the coils are tuned to the same frequency ($\omega_{\text{ext}} = \omega_{\text{imp}}$), the maximum power transmission efficiency for a certain load can be expressed by [19]

$$\eta_{\text{max}} = \frac{1}{1 + \frac{1}{k_{12}^2 Q_{PC} Q_{IC}}} \quad (2)$$

where k_{12} , Q_{PC} and Q_{IC} are the coupling coefficient, the quality factors of the unloaded powering coil and the unloaded implant coil, respectively. 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 the coils are swept to obtain the optimal coil pair. The lumped model parameters of the optimized coils which are shown in Fig. 3 are extracted by using analytical equations and a field solver software [22], [23]. The power transfer efficiency of the remote powering link is calculated by using the equation in [19]. The geometric parameters of the coils are characterized in a 3-D electromagnetic field simulation software [17]. Finally, the coils are produced on printed circuit board (PCB) for reliability and reproducibility and characterized for operation frequency.

Based on the aforementioned optimization method, we designed the optimal remote powering link. Table I represents the design parameters of the optimized coil geometries and the measurement results of the optimized coils.

Fig. 3 also shows the external link driver. Its output power depends on the amplitude of the supply voltage (V_{sup}) which is adjusted to deliver the required power level to implant. A power feedback control described with more details in Section III, controls V_{sup} to continuously supply sufficient power to the implant despite coupling variations.

TABLE I
DESIGN AND OPTIMAL COILS PARAMETERS

Parameter	Unit	Powering	Implant
Operation frequency (f_0)	MHz	13.56	
Distance between coils (d_{12})	mm	30	
Outer dimension	mm	80	12
Number of turns	—	9	8
Conductor width	mm	2	0.2
Conductor spacing	mm	1.6	0.2
Conductor height	μm	30	20
Inductance (L_{PC}, L_{IC})	μH	4.32	0.78
Series resistance (R_{s1}, R_{s2})	Ω	3.81	0.89
Parasitic capacitance (C_{p1}, C_{p2})	pF	2.44	3.46
Reson. capacitance (C_{res1}, C_{res2})	pF	29.4	181
Coupling factor (k_{12})	—		0.02

III. ELECTRONICS OF REMOTE POWERING

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 a class-E type of amplifier to drive the powering coil due to the high achievable drain efficiency. Fig. 3 shows the class-E power amplifier together with the remote powering link. The link is presented by L_{PC} , L_{IC} , and M the powering and implant coils and mutual inductance, respectively. The C_{res1} and 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 M_1 , L_{choke} , and C_{sh} present the switching transistor of the amplifier, the choke inductor and the shunt capacitor, respectively. L_{choke} , and C_{sh} values are chosen in order to maximize the power efficiency of the power amplifier [24]. The operation frequency of the amplifier is fixed to 13.56 MHz by an external oscillator circuit.

The output power (P_{out}) delivered to the load seen by the power amplifier (R_L) is proportional to [25]

$$P_{out} \propto \frac{V_{sup}^2}{R_L}. \quad (3)$$

Due to the weak coupling between the powering and the implant coils, the reflected impedance of the implanted side to the external side is low [20]. Therefore, R_L is dominated by the series resistance of the powering coil at the operation frequency ($R_L \approx R_{s1}$).

A low supply voltage (V_{sup}) is necessary for the power amplifier to deliver few mWs to the implanted system due to low R_L value. V_{sup} is controlled by the supply controller according to the power feedback data in order to change the transferred power level. Fig. 4 illustrates the power feedback control loop. The supply controller is composed of a DC-DC converter which assures high power efficiency [26], [27]. A NMOS switch (M_{DC-DC}) which is controlled by the power feedback data, defines V_{sup} voltage according to the duration of the pulse. When the power level in the implanted system is

Fig. 4. Power feedback control loop.

Fig. 5. Simulated power efficiency of amplifier vs. load resistance of implanted system for different coupling coefficients.

insufficient, the power feedback data will be “0”. The data is inverted and M_{DC-DC} starts to charge the capacitor. Consequently, V_{sup} voltage increases until the power feedback data becomes “1”. When the data is “1”, M_{DC-DC} is turned off and V_{sup} voltage starts to decrease.

The variation of the load resistance or the coupling coefficient does not significantly change performance of the power feedback loop and the efficiency of the power amplifier. Fig. 5 shows the simulated power efficiency of the amplifier versus the load resistance of the implanted system.

The delay from detecting the rectifier output voltage level to updating the supply voltage of the power amplifier creates additional voltage ripples on the output voltage of the rectifier that could result in stability issues. The amplitude of the ripples mainly depends on the power feedback update rate and the DC-DC converter frequency response [26]. The update rate is limited by the communication data rate. In addition, the values of the inductor L_{DC-DC} and the capacitor C_{DC-DC} have been chosen experimentally to minimize the ripples on the rectifier output [27].

The internal unit consists of an implant coil, a passive full-wave rectifier, a voltage regulator, and a voltage level detection circuit [7]. In addition, a reference generation, a power on reset, a multiplexer, and a low-power transmitter are integrated on the same platform.

The active rectifiers are highly efficient for high-power loads. However, they are not suitable for low-power applications which demand less than 3 mW [28], [29]. The passive rectifiers are more effective for the low-power applications [30], [31]. On the other hand, the passive rectifiers suffer from the high voltage drop across a diode-connected transistor

which reduces the power conversion efficiency [32]. In order to decrease the voltage drop, the high-threshold transistor can be replaced by a low or zero threshold transistor. However, the backward leakage current is increased due to the non-disabled transistor. Accordingly, the power efficiency of the rectifier decreases. In this study, the induced signal in the implant coil is rectified and V_{rect} voltage is charged on the storage capacitor by a passive full-wave rectifier. The rectifier uses the charge-storage technique which decreases the forward voltage drop on the diode connected high-threshold transistor as presented in [7]. This technique reduces the gate-source voltage (V_{GS}) of the transistor and allows to achieve high power conversion efficiency at low-power loads.

A low-dropout (LDO) voltage regulator suppresses the ripples and generates a clean and stable supply voltage (V_{reg}) for the implanted system. In order to obtain high Power Supply Rejection Ratio (PSRR) on V_{reg} voltage, the reference voltage of the voltage regulator ($V_{\text{ref_LDO}}$) should also have an acceptable PSRR value. Moreover, $V_{\text{ref_LDO}}$ voltage should be constant and stable for wide range of supply variation where V_{rect} could change by more than 1 V. The reference generation circuit is a CMOS self-biased reference circuit [33]. Fig. 6 shows the proposed reference generation circuit together with start-up. When V_{rect} voltage increases, the current on the both M_3 and M_4 transistors are increased. Therefore, the V_{GS10} voltage also increases. This voltage increase is compensated by the decrease on V_{GS7} voltage. Accordingly, $V_{\text{ref_LDO}}$ voltage stays constant for wide range of V_{rect} voltage. In addition, a self-biased cascode current mirror is proposed to enhance the PSRR. The output resistance of the cascode current mirror is defined as

$$r_o = g_{m_{M6}} r_{o_{M6}} r_{o_{M4}} \quad (4)$$

where $r_o, r_{o_{M6}}, r_{o_{M4}}$ is the output resistance of the cascode stage, the output resistance of M_6 transistor, and the output resistance of M_4 transistor, respectively. In addition, $g_{m_{M6}}$ represents the transconductance of the M_6 transistor. The output resistance of the cascode current mirror is increased by the gain of the M_6 transistor compared to the single stage current mirror which introduces more fluctuation on the current of each branch and also $V_{\text{ref_LDO}}$ voltage. Fig. 6 shows the comparison of PSRR between the single and cascode current mirror. PSRR is enhanced by more than 25 dB at low frequencies and 5.5 dB at 27.12 MHz frequency. In addition, the cascode transistors are biased by the self-biased technique which decreases the voltage drop across the current mirror and allows to operate at low supply voltage with cascaded transistors.

The level detection circuit compares the output voltage of the rectifier (V_{rect}) with a reference voltage as illustrated in Fig. 4. When V_{rect} is under the reference voltage, the supply sensor sends the power feedback data as logic “0”. This increases the delivered power level to the implanted unit and results in a higher rectifier voltage. On the other hand, the power feedback data will be “1”, when the rectifier voltage is higher than the reference voltage. The voltage level detection circuit is implemented by a comparator and the voltage levels are adjusted by the voltage dividers. The sensor system collects the data from

Fig. 6. Proposed reference generation circuit with start-up.

Fig. 7. Comparison of PSRR between single and cascode current mirror.

different types of electrochemical, pH, and temperature sensors. The data source is switched between the sensors and the supply sensor by a multiplexer and the data is delivered to the transmitter.

In order to prevent malfunctioning and failures in the circuits especially in digital circuits due to the insufficient supply voltage, a Power on Reset (PoR) circuit is implemented. The level of V_{reg} is tracked by the PoR circuit [34]. The PoR circuit sends “power-up” signal and enables all circuits when the supply voltage is high enough. In addition, the PoR disables the all chip and sends “power-down” signal when V_{reg} voltage level decreases under a certain level where the circuits perform improperly. Moreover, the PoR circuit cancels all the operation if the supply voltage drops under a certain level or the power transmission is interrupted suddenly. All the operations are restarted by the PoR circuit when the supply voltage reaches a required level.

Fig. 8 shows the circuit schematic of Power on Reset. When V_{reg} supply voltage increases, the current and eventually V_{GS1}

Fig. 8. Circuit schematic of Power on Reset.

voltage of M_1 transistor increases. The increase on V_{GS1} voltage is delayed by charging the C_1 capacitance. When V_{GS2} voltage reaches to threshold voltage of M_2 transistor, M_2 transistor turns on and N times current of M_1 transistor starts to flow over M_2 transistor. Accordingly, the drain voltage V_{DS2} of M_2 transistor decreases and is tracked by M_8 transistor. Transistor M_8 produces a positive feedback which increases V_{GS2} voltage which also increases the drain current of M_2 . Accordingly, the gate voltage V_{GS3} of M_3 decreases under threshold voltage of M_3 and finally M_3 turns off at a certain V_{reg} voltage level. Therefore, the input of the inverter converges to V_{reg} voltage and “power-up” signal is supplied to enable all circuits. The inverter is used to obtain rail-to-rail signal. On the other hand, “power-down” signal is created to disable the circuits by the same mechanism which is performed reversely while V_{reg} voltage decreases.

IV. PROPOSED LOCATOR SYSTEM

During the magnetic power transmission, the coupling between the coils changes according to the positions of the coils with respect to each other. Therefore, there is a strong correlation between the misalignment and the coupling of the coils. The transmitted power needs to be increased to obtain a constant received power when the implant coil moves from the center to the edges of the powering coil. Accordingly, the supply voltage of the powering coil driver is increased to compensate the misalignment. Fig. 9 shows the effect of misalignment on the coupling coefficient and the required supply voltage of the driver which are normalized with respect to the concentric coil alignment values.

The permanent implanted device needs to be recharged and communicate with the mobile external device as shown in Fig. 1. However, the implanted device could move inside the body and it would be necessary to relocate it. The supply voltage of coil driver is tracked for constant received power by the power feedback loop. Also, this supply voltage is used to relocate the implanted device. Fig. 10 shows the proposed locator which can be used to find the location of the implantable device in the body. The locator tracks the supply voltage of the external coil driver and the LED ladder indicates the amount of the misalignment between the coils. The supply voltage of the driver

Fig. 9. Misalignment effect on coupling coefficient and supply voltage.

Fig. 10. Aligning mobile unit with implantable device using a locator circuit.

decreases non-linearly when the mobile unit moves towards to the center of implanted device as shown in Fig. 9. Therefore, a resistor ladder with non-linear voltage levels is implemented so each LED turns on for a fixed amount of misalignment. Using comparators, the supply voltage of the power amplifier is compared with the resistive ladder voltage levels. When the mobile unit moves towards to the implanted device, the supply voltage of the coil driver decreases and the LEDs on the indicator turn on respectively. Finally, all the LEDs turn on when the perfect alignment is achieved as shown in Fig. 10.

V. ELECTRONICS OF DATA COMMUNICATION

The quality factors of the powering and the implant coils are crucial to improve the power transfer efficiency. However, the high quality factor limits the high data rate communication by reducing the bandwidth. Therefore, the data communication deteriorates the power transmission efficiency when the remote powering and the communication share the same channel. In order to obtain efficient power transmission and high data rate communication simultaneously, a different frequency channel (868 MHz) which is defined by the regulations is used for data communication [35].

The power consumption of the implanted system is generally dominated by active transmitter power dissipation. Generating

Fig. 14. Experimental setup of remote powering system and data communication.

Fig. 15. Micrograph of the fabricated (a) power management blocks and (b) free-running oscillator.

Fig. 16. Post-layout simulation result of LDO voltage regulator.

The measured power efficiency of the integrated passive full-wave rectifier is 80% at 2 mW load while generating 2.2 V at the output of the rectifier. The reference generation circuit creates 0.9 V reference voltage (V_{ref_LDO}) and has power supply rejection (PSRR) response of 78.6 dB at DC and 47.8 dB at 27.12 MHz. Fig. 16 shows the post-layout transient simulation result of the LDO voltage regulator. The regulated output voltage has maximum ripple of 50 mV (1.79–1.84 V) when the unregulated input of the regulator decreases from 2.8 V to 2.1 V and the load current decreases from 3 mA to 100 μ A. The ripple rejection of the high-speed LDO voltage regulator is more than 60 dB at 27.12 MHz. The PoR enables the circuits when V_{reg} voltage reaches to 1.48 V. The PoR disables the circuits if V_{reg}

Fig. 17. BER performance of communication link for different communication distances.

TABLE II
POWER CONSUMPTION OF IMPLANTABLE ELECTRONICS

Circuit	Power (μ W)	Dist. (%)
Voltage regulator	234	13.8
Reference generation	10.9	0.6
Power feedback comparator	11.2	0.7
Power on reset	14.4	0.9
Transmitter	324	19.1
Total	595	35.0
Rectifier load power	1700	100

voltage decrease under 1.45 V. Fig. 17 shows Bit Error Rate (BER) performance of the OOK transmitter and the custom designed receiver for different communication distances. In order to measure the BER, more than 90 million random bits are created by a FPGA and transmitted across the communication link in air. Table II summarizes the power consumption breakdown of the implantable electronics.

The values of L_{choke} inductor and the C_{sh} capacitor as shown in Fig. 3 are optimized in order to maximize the drain efficiency of the amplifier. The remote powering link is optimized for 1.6 k Ω load resistance which also emulates the input resistance of the rectifier. The maximum remote powering efficiency is achieved as 21.7% at 1.6 k Ω load by using 180 nH inductance and 320 pF capacitance values when the distance between the coils (d_{12}) is 3 cm.

The 2-D lateral misalignment effect on power transfer efficiency is verified by moving the powering coil according to the implant coil as shown in Fig. 14. The platform allows to simulate different coupling conditions during the measurement. The platform have two docks (1 and 2) where the powering and the implanted coils are placed respectively. In addition, the powering coil can move on the dock 1 to simulate the 2-D lateral misalignment.

Fig. 18 shows the overall power efficiency distribution on the powering coil which includes the power efficiencies of the power amplifier, the remote powering link, and the integrated rectifier in order to obtain 2.2 V output voltage from the rectifier for 1.7 mW load power. The overall efficiency changes drastically from 16.8% to less than 1% when the implanted unit moves from center to edges. This nonuniform power transmission gives rise to change in the received power at the implanted unit. As a

Fig. 18. Overall power efficiency distribution on powering coil ($d_{12} = 3$ cm).

Fig. 19. Dynamic power adaptation technique.

result, the implanted unit cannot continue its operation if the received power drops under a threshold level. On the other hand, the circuits can be damaged if the received power increases too much. Therefore, the transmitted power level must be adapted dynamically by the power feedback control loop as shown in Fig. 4. Fig. 19 shows the dynamic power adaptation technique. When the coupling between the coils changes, the power feedback loop adjusts the transmitted power level. Accordingly, the rectifier voltage is kept between 2.20 V and 2.32 V by the feedback loop.

The magnetic field strength decreases from the center to the edges of the powering as shown in Fig. 18. As a result the power amplifier supply voltage follows the inverse of magnetic field trend to keep the implanted rectifier output at 2.27 V. Fig. 20 illustrates the power amplifier supply voltage change due to the horizontal and vertical lateral misalignments. The PA supply voltage needs to be increased from 0.28 V to 0.45 V to obtain constant voltage at the output of the rectifier while the implanted unit moves from center to edge of the powering coil by 3 cm.

Fig. 21 shows the locator system to relocate the implanted device. The supply voltage of the power amplifier changes non-linearly. In addition, the power transmission efficiency on the pow-

Fig. 20. Power amplifier supply voltage vs. lateral misalignment.

Fig. 21. Locator system. (a) Misalignment case. (b) Perfect alignment case.

ering coil decreases extremely on the edges less than 1% and the power amplifier needs to transmit high power which damages the circuit. Accordingly, the locator system is designed to keep the misalignment below 3 cm. The locator system has five non-linear voltage levels and each voltage level is chosen experimentally to represent approximately 0.6 cm on the powering coil. When the implanted device and mobile unit is not aligned properly, the indicator shows the distance required for perfect alignment. All the LEDs are turned on when the perfect alignment is achieved and the power transmission efficiency is maximized as shown in Fig. 21.

As a case study, this implantable remote powering and data communication micro-system can be used for bio-sensors. Bio-sensors are helpful to monitor endogenous molecules (e.g., glucose, ATP) and drugs in the body. These bio-sensors require a circuit which needs 711 μ W of power to calibrate and readout [42]. Our micro-system can supply up to 1.7 mW of power with 16.8% of power efficiency and has 0.6 mW power consumption. Therefore, 1.1 mW of power budget is quite sufficient for operation of the bio-sensor readout and calibration circuit.

TABLE III
SUMMARY OF REMOTELY POWERED IMPLANTABLE MICRO-SYSTEM

Parameter	Unit	Value
Technology	—	UMC 0.18 μm
Implantable	—	Yes
Total power consumption	μW	595
Powering type	—	Remote powering
Powering frequency	MHz	13.56
Powering distance	mm	30
Powering efficiency	%	16.8* @ 1.7 mW load
Rectifier power efficiency	%	80 @ 2 mW load
Voltage regulator PSRR	dB	60 @ 27.12 MHz
Supply voltage	V	1.8
Reference voltage PSRR	dB	47.8 @ 27.12 MHz
PoR enable/disable voltage	V	1.48/1.45
Uplink communication	—	OOK
Uplink com. frequency	MHz	869
Uplink com. data rate	Mb/s	up to 1.5
Uplink com. distance	cm	45
Downlink communication	—	ASK
Downlink com. frequency	MHz	13.56
Locator system resolution	mm	6

* Including the power efficiencies of the power amplifier, powering link and rectifier

TABLE IV
PERFORMANCE COMPARISON OF REMOTE POWERING LINK

Reference	Diameter (Ex./In.)(mm)	Dist. (mm)	Freq. (MHz)	Eff.
[43]	60/20	30	0.7	36%
[44]	30/30	10	6.785	51%
[45]	69/20	10	1	41.2%
[46]	64/22	20	0.7	82%
[47]	43/10	10	8	54.8%
This Work	80/12	30	13.56	21.7%*

* Including power efficiencies of power amplifier, powering link

TABLE V
PERFORMANCE COMPARISON OF SHORT RANGE COMMUNICATION

Reference	Data Rate (Mb/s)	Dist. (cm)	Freq. (MHz)	Power (mW)*	BER
[39]	0.33	13	433	1.81	0.003
[48]	1	—	916.5	3.8	0.001
[49]	136	20	2460	3	0.0017
This Work	0.5	30	868	0.32	$3 \cdot 10^{-5}$

* Transmitter power consumption only

Table III summarizes the performance of the remotely powered implantable micro-system. Furthermore, Tables IV and V compare this work with the recently reported remotely powered systems and the short range communication systems in the literature, respectively.

VII. CONCLUSION

The paper presents an universal remote powering and communication systems for the implantable medical devices to be used along with different sensor/actuator interfaces. 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. A locator system is proposed to relocate the implant in the body with 6 mm resolution for the efficient power transfer. A significant feature is the detection of the location of the implant. Moreover, this custom

designed receiver tolerates large data carrier drifts of the transmitter. The fabricated implantable electronics works with a relatively low supply voltage and provides all the functions needed for the external link thanks to an optimized remote power supply capable to operate over 3 cm distance. The power required by the implantable electronics including the one for the wireless communication is just 0.6 mW.

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