Wireless powering electronics and spiral coils for implant microsystem toward nanomedicine diagnosis and therapy in free-behavior animal

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A B S T R A C T

In this paper, we present a wireless RF-powering electronics system approach for batteryless implantable biomedical microsystem with versatile sensors/actuators on laboratory animals toward diagnosis and therapy applications. Miniaturized spiral coils as a wireless power module with low-dropout (LDO) linear regulator circuit convert RF signal into DC voltage, provide a batteryless implantation for truly free-behavior monitoring without wire dragging. Presented design achieves low quiescent-current and Line/Load Regulation, high antenna/current efficiency with safety considerations including temperature and electromagnetic absorption issues to avoid damage to the implanted target volume of tissue. Related system performance measurements have been successfully completed to demonstrate the wireless powering capabilities in desired implantable microsystems.

1. Introduction

Neuroscience engineering together with in vivo real time biomedical information, such as neural activity, recorded by implant neural sensors and microsystems is ultimately crucial for nanomedicine and bio-logical researches to identify and develop treatments for brain disorders i.e. epilepsy. For such applications in animal model studies, microsystems that consist of neural sensors, amplifiers, data converters/operators and data transmission units are used to capture neural information from animals and transfer the data to a laptop for further analysis.

Generally, at least two types of wire connection are required for powering and data transmission. Use of wire connection between animal and instrumentation can cause serious problem even lead to the death of implanted animal when wire drag happens. Swivels are sometimes used to solve the dragging, but the price can be another problem, the channel numbers are limited and the implanted laboratory animal still cannot behave in their nature condition because the wires. Benefit to the rapid developments of wireless communication, many different kind of wireless solution, including Zigbee [1] and Bluetooth [2], have been commercialized and applied in animal studies. Batteries are common used to eliminate the problem of power wires. However, batteries dominate the size and the cost of the implant device and become less attractive when the implant devices get smaller and need to replace their batteries occasionally [3]. Therefore, the demand for developing continuous monitoring in long-term, real-time prosthetics in free-behavior implanted animal is restricted by the capacity of the battery. A biomedical system constructed by bio-potential sensors [4] and actuators with miniaturized powering device receiving RF-power from an external host is then highly desirable.

Currently, there are various methods for wireless transfer of power to biomedical microsystems, including ultrasonic energy transmission [5], capacitive coupling [6] and inductive coupling. Ultrasonic energy harvest systems transfer the mechanical energy into the implant device through skin, but suffer from low energy conversion capability [7]. Capacitive coupling declares clean and quiet telemetry by using metal plates and skin tissue as capacitors [6], however, it can never be applied in animals that need to be monitored under freely moving status. Comparing to other two method, inductive link achieves more possibility to transfer energy through an open air field with reasonable transmission efficiency.

In this paper, we present an inductive coupled wireless powering system which contains inductive coils and low-dropout regulator design for implant microsystem toward free-behavior animal applications. Section 2 describes the overall wireless powering structure in an implantable microsystem. The development of the inductive coupling coils and low dropout regulator is detailed in Sections 3 and 4, following with performance test of the proposed wireless powering system in Section 5. Section 6 summarizes this paper with a brief conclusion.
2. Wireless powering system structure

Fig. 1 displays the block diagram of microsystem with bio-sensors i.e. ExG, chemical or temperature sensors, and stimulation actuators i.e. drag delivery for nanomedicine diagnosis and therapy. Wireless transceiver in the microsystem transmits bio-info data to and receives commands from an external host, while proposed RF–DC wireless powering system receives RF power from an external host and provides system supplies to the implant microsystem. The RF power produced by a Class-E amplifier is coupled into the receiving coil via a tuned LC network followed by a low dropout (LDO) regulators to produce stable system supplies [8]. The proposed wireless powering system can serve as an additional small module to be applied to different implant microsystems. The schematic of wireless powering system can also be observed in Fig. 1. External component includes PA and RF power transmission coil, on-broad components involves receive coil, rectifier, smoothing capacitance and bias diode, and the on-chip component includes LDO regulator with band gap bias circuit and thermal protection circuitry.

In the inductive coupled power transmission design, the main contribution to the power loss is the electromagnetic wave reflection at the air-skin interface and heat dissipation effect inside the tissues [9]. Also, reliable transmission efficiency with enough transmission distance becomes one of the considerations. Low RF powering frequency provides longer transmission distance but also makes receiver coil bigger. Furthermore, electromagnetic wave shows its absorption and reflection property by the skin tissue when frequency range is approximately higher 20 MHz and lower than 1 MHz, respectively. In this paper, 13.56 MHz radio frequency is chosen because it is one of the ISM band and its low tissue absorbability to avoid damage on animal tissue.

3. Inductive coupling spiral coils theory and design

3.1. Inductive link theory and design procedure

The radio frequency power is transmitted through coupled coils, which follows the electromagnetic induction of Faraday’s Law. Various approaches have been presented for inductive link optimization in implantable electronics [10–12]. Fig. 2 illustrates a simplified inductive powering model [13,14]. \( L_1 \) is the primary coil driven by an ac source \( V_s \), which is often an efficient class-E power amplifier. \( L_2 \), is the secondary coil that is integrated with the implant electronics. \( M \) is the mutual inductance between two coils. Coil windings have parasitic resistance and capacitance associated with them, which are represented by lumped elements \( R_1 \) and \( R_2 \). Capacitors \( C_1 \) and \( C_2 \) are added to form a pair of resonant LC-tank with \( L_1 \) and \( L_2 \), respectively.

Magnetic flux lines are formed around the primary coil as a result of the flow of current though it and the voltage induced in the secondary coil is due to the magnetic flux passing though. The equivalent resistance \( R_e \) reflected back into the primary coil can be expressed as:

\[
R_e = \frac{(\omega M)^2}{R_2 R_c} = \frac{R_a K^2 Q_1 Q_2}{R_{ac} + Q_1^2 R_s} \times R_{11}
\]

where \( K, Q_1 \), and \( Q_2 \) are the coupling coefficient and unloaded quality factor of the two coils. The mutual inductance of two aligned coils that having diameters \( r_1 \) and \( r_2 \), turns \( N_1 \) and \( N_2 \) and separated by distance \( x \) is given by the following expression [15]:

\[
M = \frac{\mu_0 N_1 N_2 r_1^2 r_2^2 \pi}{2 \sqrt{(r_1^2 + x^2)^3}}
\]

where \( \mu_0 = 4\pi \times 10^{-7} \text{H/cm} \). And the optimal transmission efficiency is readily found in the literature [16]:

\[
\eta_{opt} = \frac{K^2 Q_1 Q_2}{\left[1 + (1 + K^2 Q_1 Q_2)^{1/2} \right]^2}
\]

The shared magnetic flux between coils results in a coupling coefficient \( K \), where \( 0 < K < 1 \), is given by:

\[
K = \frac{M}{\sqrt{L_1 L_2}}
\]
single layer spiral coil structure were utilized in this work. Also, to maximize the coupling coefficient $K$, optimal values of geometry of the coils are obtained according to the expressions in [16]. Calculate $R_{S1}$ and $R_{S2}$ from desired rectifier performance and powering specification. Therefore, $L_1$ and $L_2$ can be designed and obtained and realized by experimental measurements. Then, the resonance capacitance $C_{S1}$ and $C_2$ can be found according to desired transmission frequency.

3.2. Fabricated results and measurements

Fig. 3 shows the optical photographs of the fabricated receiving and external coils made by 24/16 AWG copper wire with 1.5 cm and 4 cm in diameter. The inductance, parasitic resistance, parasitic capacitance and resonate capacitance of the transmission coil are 15 $\mu$H, 11.5 $\Omega$, 2.73 pF, and 6.4 pF, while the receive coil exhibits 521 nH, 16.7 $\Omega$, 3.71 pF and 260.7 pF, respectively. The antenna efficiency is characterized with PDMS coating as bio-protection in implanted tissue. Measurement result under conditions in air/tissue (pork) and with/without PDMS encapsulation is illustrated in Fig. 4, which results no obvious influence from packaging.

4. Low dropout regulator design

4.1. Circuit design

Fig. 5 illustrates the block diagram of the proposed low-dropout linear regulator. The architecture is modified from a typical low-dropout regulator topology [18] with power MOSFET and thermal protection unit to enhance the driving current and avoid temperature damage on tissue. The error amplifier amplifies the voltage difference between the reference voltage and divided load voltage and switches the power MOSFET (PMOS). Output voltage is described as:

$$V_{OUT} = V_{REF} \left( \frac{R_1 + R_2}{R_2} \right)$$

(5)

Another design issue in the linear regulator is the stability problem. According to bode plot analysis, the output capacitance and its equivalent series resistance (ESR) decides the zero point to increase the phase margin. However, the ESR must be carefully designed in appropriate range to ensure the system stability.

Fig. 6 shows the circuit schematic of the error amplifier. $M_{22}$ is the power MOSFET illustrated in Fig. 5. Also, stability is one of the design key points. In this design, ceramic capacitance with low ESR is used as output capacitance, which can cause low phase margin and lead into un-stable condition. Therefore, extra RC-compensation is introduced to enhance the system stability. In Fig. 6, current limitation control by two Poly resistances which is inverse proportional to current. PIP capacitance $I_{48}$ and resistor $R_{31}$ are used for Miller compensation. PIP capacitance $M_{44}$ and $M_{39}$ are used as internal smoothing capacitances for spike rejection. MOS $M_{13}$ acts as output switch, which controlled by thermal protection circuit.

Fig. 7A shows the detailed schematic of a temperature-independent bandgap reference voltage generator (RVG) [19]. In Fig. 7A,
the PNP BJT Q_{16} and Q_{17} act as negative- and positive-TC voltage devices produce a temperature independent voltage difference to the operational amplifier. A resistor R_{18} is used to adjust the output VREF. PIP capacitance M_{21} and resistor R_{2} are used for Miller compensation. M_{9} and M_{10} act as a startup for quicker response. M_{14} is an extra gain stage for higher PSRR. M_{20} is used for extra ESD protection, while the PIP capacitance M_{2} used as a filter and bias. Fig. 7B illustrates the thermal protection circuit schematic. The over-temperature protection (OTP) design is used to avoid thermal damage on tissue due to the high temperature (>40°C) caused by the operating circuit. In Fig. 7B, R_{3} is a ROND resistor in tsmc 0.35 μm process, which has high temperature coefficient (PTC1 = 1.51e -3). When T raise, voltage drop on R_{3} increase, then turn-on M_{14}, increase current flow passing M_{14}, turn-on M_{15}, pull down OTP. Hysteresis functions when temperature drop down lower than 40°C, M_{14} will not off immediately due to the current provided by BJT Q_{32}. Until the temperature is lower enough, say <37°C, voltage drop on R_{3} decrease, current decrease, M_{14} turn-off, M_{15} off, OTP pull high.

4.2. Fabricated results and measurements

The LDO regulator chip is fabricated via TSMC 0.35um 2P4M process. The die size is 1.42 × 0.95 mm². Fig. 8A shows the optical microphotograph. Fig. 8B is the test result of the power supply rejection ratio. The measured PSRR is around 70.833 dB and 0.165 dB at 10 kHz and 13.56 MHz, respectively. Comparing to the simulation result of PSRR (~81.44 dB), the main difference is that simulation consists of only pure capacitance, but in practice, parasitic inductance does exist in the output ceramic capacitance, which decreases the high frequency performance in PSRR.

The stability test of the LDO regulator displays the ripple and noise on the output voltage level under different current loading. Fig. 9A output stability observation under I_{OUT} = 200 mA. Result shows that the maximal noise level is lower than 3–4 mV. Low ripple and noise performance is one of the great property of linear regulator. Fig. 9B present the output ripple caused by a current loading from 0 mA to 200 mA. Output transient behaves a 308–264 mV spike, say 17–14% variance in output voltage, under the condition V_{IN} from 3 V to 6 V, respectively. The output transient performance can be easily improved by increase the quiescent current but also increase the power consumption. Fig. 9C illustrates the output ripple observation under a sudden maximal current loading. The maximal output current varies under different V_{IN} condition. Result shows that the peak of the spike is less than 300 mV. Line transient is the output current ripple caused by the input voltage variance. Fig. 9D shows the spike observation under loading current 200 mA with V_{IN} variance from 2.6 V to 6 V.
overall Line Regulation performance is between 1.5 and 4.5 mV, cause 0.0833–0.25% variance in output current.

Table 1 shows the measured Load/Line Regulation Result. Note that when the $I_{\text{OUT}} = 200$ mA under $V_{\text{IN}}$ from 5 V to 6 V, the system is thermal shutdown due to the over thermal protection (OTP). Generally speaking, the Load Regulation is between 0.5 and 5 mV, which is around 0.0278–0.278% variance in output voltage. Table 2 summarizes the measured performance of the presented LDO regulator comparing to other works.

Presented chip shows low drop out voltage, wide input range with reasonable quiescent current. Additionally, thermal-protection (<40 °C) design is also included in the present linear regulator to avoid damage in implanted target for implant device applications. Some equations used for specification comparison are listed below:

$$V_{\text{Drop}} = V_{\text{IN}} - V_{\text{OUT}} \cdot I_{\text{OUT \ MAX}}$$  \hspace{1cm} (6)

$$I_{\text{eff}} = \frac{I_{\text{OUT \ MAX}}}{I_{\text{IN \ MAX}} + I_{\text{Q}}}$$  \hspace{1cm} (7)

5. Wireless powering system performance

After the fabrication of spiral coils and LDO regulator chip, PCB level integration is utilized for system performance characteriza-
tion. To generate the RF powering, a power amplifier module using commercial chips is used to amplify the 13.56 MHz signal from a signal generator. Fig. 10 shows the fabricated wireless powering system, the class-E power amplifier and receive module. Fig. 11 illustrates the temperature raise test under 37 °C with maximal output current for 1 hr. Result shows that less than 2 °C raise is observed thus meet the implantation requirement [23,24].

Table 1: Load/Line Regulation Test Result.

<table>
<thead>
<tr>
<th>VIN (V)</th>
<th>2.6</th>
<th>3.3</th>
<th>4.2</th>
<th>5.0</th>
<th>6.0</th>
<th>Line Regulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>I_{OUT} = 0 mA</td>
<td>V_{OUT} (V)</td>
<td>1.828</td>
<td>1.8285</td>
<td>1.829</td>
<td>1.8295</td>
<td>1.8285</td>
</tr>
<tr>
<td>I_{OUT} = 50 mA</td>
<td>1.828</td>
<td>1.829</td>
<td>1.83</td>
<td>1.8305</td>
<td>1.8305</td>
<td>2.5 mV</td>
</tr>
<tr>
<td>I_{OUT} = 100 mA</td>
<td>1.828</td>
<td>1.83</td>
<td>1.831</td>
<td>1.8325</td>
<td>1.8325</td>
<td>4.5 mV</td>
</tr>
<tr>
<td>I_{OUT} = 200 mA</td>
<td>1.8285</td>
<td>1.8315</td>
<td>1.834</td>
<td>OTP</td>
<td>OTP</td>
<td>NA</td>
</tr>
<tr>
<td>Load Regulation</td>
<td>0.5 mV</td>
<td>3 mV</td>
<td>5 mV</td>
<td>NA</td>
<td>NA</td>
<td></td>
</tr>
</tbody>
</table>

Fig. 10. Fabricated full system with power amplifier, transmission coil, receiving coil and LDO regulator.

Table 2: Summary of LDO regulator performance.

<table>
<thead>
<tr>
<th></th>
<th>[20]</th>
<th>[21]</th>
<th>[22]</th>
<th>This work</th>
</tr>
</thead>
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<tr>
<td>Year</td>
<td>2006</td>
<td>2008</td>
<td>2010</td>
<td>2011</td>
</tr>
<tr>
<td>Technology (μm)</td>
<td>0.13</td>
<td>0.18</td>
<td>0.35</td>
<td>0.35</td>
</tr>
<tr>
<td>Input voltage (V)</td>
<td>3–4.5</td>
<td>NA</td>
<td>2–4.5</td>
<td>2.6–6</td>
</tr>
<tr>
<td>Output voltage (V)</td>
<td>2.8</td>
<td>1.2</td>
<td>1.8</td>
<td>1.8</td>
</tr>
<tr>
<td>Dropout voltage (V)</td>
<td>0.199</td>
<td>0.2</td>
<td>0.2</td>
<td>0.125</td>
</tr>
<tr>
<td>Current efficiency (%)</td>
<td>NA</td>
<td>99.9</td>
<td>NA</td>
<td>99</td>
</tr>
<tr>
<td>Output current (mA)</td>
<td>150</td>
<td>200</td>
<td>200</td>
<td>0–200</td>
</tr>
<tr>
<td>Quiescent current (μA)</td>
<td>114</td>
<td>20</td>
<td>30</td>
<td>45</td>
</tr>
<tr>
<td>Current limit (mA)</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
<td>252</td>
</tr>
<tr>
<td>Line regulation (mV)</td>
<td>1.5</td>
<td>NA</td>
<td>6</td>
<td>4.5</td>
</tr>
<tr>
<td>Load regulation (mV)</td>
<td>17.4</td>
<td>NA</td>
<td>0.09</td>
<td>5</td>
</tr>
<tr>
<td>PSRR (dB)</td>
<td>67@20 K</td>
<td>45@20 K</td>
<td>NA</td>
<td>71@10 K</td>
</tr>
<tr>
<td>Thermal protection</td>
<td>No</td>
<td>No</td>
<td>No</td>
<td>Yes (&lt;40 °C)</td>
</tr>
</tbody>
</table>

Fig. 11. Temperature raise test under 37 °C with maximal output current for 1 hr.

and 200 mA, respectively. The maximal output ripple is around 50 mV due to the low PSRR performance at 13.56 MHz. However, the ripple can be reduced down to less than 20 mV is extra 0.1 μF ceramic capacitance is added at the LDO output. Fig. 12C shows the Load Regulation under current loading varies from 100 mA to 200 mA. The Transient voltage is about 400 mV in peak. Table 3 shows the result of measured Load Regulation of the full system. Finally, the calculated specific absorption rate (SAR) is 0.27 W/kg, which is in the basic restriction of IEEE standard C95.1 (0.4 W/kg).

In ideal case, the quiescent current must be as low as possible during zero-to-low loading, because it amounts to a significant portion of the total drain current from the supply. However, in the other hand, the quiescent current of the circuit also determines the response speed of the amplifier. Error amplifier compares the reference voltage between feedback voltage and operates the pass element for different loading condition. Low quiescent current can lead to slow response in pass element control. In this paper, the Load Regulation achieves 0.5 mV under the condition of 2.6 V input voltage, which is comparable to other systems. From the application point of view, the current load of an implant microsystem for neural signal recording is relative stable. Therefore, the quiescent current of our design can be reduced to small loading variation condition. Another possible method for improving the system is to apply additional buffer between error amplifier and the pass device. Under the condition of very low quiescent bias current amplifier design, the objective of the buffer is to drive the power pass device quickly with minimal power. The buffer must produce low output impedance because the large pass device presents large parasitic capacitance. Also, higher slew rate currents of the buffer is required to quickly charge and discharge the parasitic capacitance of pass device. Additionally, the buffer should exhibit high input impedance to decrease the loading effects in error amplifier, which may limit the loop gain and bandwidth of the regulator. Clearly, adding additional buffer may increase the complexity of the system. Related studies will be consider as our next steps.
regulator achieves 45 coupling coils provide wireless power transmission with 0.83 coupling, provide a batteryless implantation for truly free-behavior low-dropout linear regulator circuit convert RF signal into DC voltage. Miniaturized spiral coils as a wireless power module with approach for implantable biomedical applications on laboratory animals. To summarize, the spiral coils as a wireless power module is presented in power management for batteryless medical instrumentation applications. The coil and circuit design, fabrication and system implementation are exhaustively discussed in this section. Finally, practical measurement result provides the detailed performance characterization of the presented wireless power module.

6. Conclusion

In this paper, we present a wireless RF-powering electronics approach for implantable biomedical applications on laboratory animals. Miniaturized spiral coils as a wireless power module with low-dropout linear regulator circuit convert RF signal into DC voltage, provide a batteryless implantation for truly free-behavior monitoring without wire droop. Carefully designed inductive coupling coils provide wireless power transmission with 0.83 coupling coefficient and high antenna efficiency. The proposed LDO regulator achieves 45 μA quiescent-current, 4.5/5 mV Line/Load Regulation, 71 dB PSRR with thermal protection to avoid damage to the implanted tissue. The full system exhibits less than 20 mV ripple output and maximal 400 mV transient peak voltage under current load varies from 100 mA to 200 mA. Related system performance measurements have been successfully completed to demonstrate the wireless powering capabilities under safety issues considerations (temperature and SAR) in desired implantable microsystems.

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References


Table 3

<table>
<thead>
<tr>
<th>Current</th>
<th>V_out</th>
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<tbody>
<tr>
<td>I_{0mA}</td>
<td>1.828 V</td>
</tr>
<tr>
<td>I_{50mA}</td>
<td>1.827 V</td>
</tr>
<tr>
<td>I_{100mA}</td>
<td>1.826 V</td>
</tr>
<tr>
<td>I_{150mA}</td>
<td>1.823 V</td>
</tr>
<tr>
<td>I_{200mA}</td>
<td>1.820 V</td>
</tr>
<tr>
<td>Load Regulation</td>
<td>8 mV</td>
</tr>
</tbody>
</table>

Fig. 12. (A) Output ripple observation under V_{in} = 2.5 V, I_{0mA} = 0 mA (B) V_{in} = 6.8 V, I_{0mA} = 200 mA (C) Load Regulation under I_{0mA} = 100–200 mA.

