

# A Low-Power FM Transmitter for Use in Neural Recording Applications

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**Abstract**—We present a low power FM transmitter for use in neural recording telemetry. The transmitter consists of a low noise biopotential amplifier and a voltage controlled oscillator used to transmit the amplified neural signals at a frequency of 433 MHz. The circuit is powered through a transcutaneous, inductive link. The power consumption of the transmitter is measured to be 465  $\mu$ W. Using a 1/8-wavelength monopole antenna, a received power level was measured to be  $-54.5$  dBm at a distance of one meter.

**Keywords**—RF telemetry, transmitter, neural recording, low power circuits

## I. INTRODUCTION

With advances in integrated circuit technology and the development of new biocompatible materials, biomedical implants are becoming more common. Modern MEMS technology has led to the development of large-scale neural recording devices such as microelectrode arrays containing as many as 100 independent electrodes. These arrays monitor neural activity in the brain, and have many potential uses in both the scientific and clinical realms.

A current limitation of this technology is the way in which these recorded neural signals are transferred from the recording device, which is usually implanted in the body, to signal processing equipment used for scientific or neuroprosthetic applications. In most modern neuroscience laboratories, long cables are used to connect the implanted recording device to external amplification and signal processing equipment. There are two main problems with the use of transcutaneous cables: (1) the risk of infection, and (2) noise pickup due to the high impedance and weak signals associated with neural recording.

One solution to these problems is to replace the cables with a wireless transmitter powered through a magnetic transcutaneous link. Such a system would eliminate any cables protruding through the skin, reducing the chance of postoperative infections. A transmitter implanted near the electrode array would also shorten the distance that the neural signal must travel over a wire, thus reducing a substantial source of noise. Ideally, the transmitting system would first amplify the weak neural signals, further improving resistance to any noise introduced during transmission.

In this work, we present an implantable 433-MHz wireless FM transmitter consisting of a low-noise

biopotential amplifier and a voltage-controlled oscillator (VCO). A brief system overview is given in Section II, while Section III discusses the implantable transmitter in more detail. Experimental results are presented in Section IV, and we conclude in Section V.

## II. SYSTEM OVERVIEW

A block diagram of a fully implantable recording system is shown in Fig. 1. The system can be divided into two main parts: the implant and the external electronics. The implant consists of neural signal amplifiers, an implantable FM transmitter for data telemetry, voltage regulation circuitry, and clock and data recovery circuitry. The external circuitry consists of a magnetic transcutaneous power transmitter. Coils on either side of the skin form a transformer that permit the transfer of power, though the frequency of this waveform must be kept under 10 MHz to prevent absorption by tissue. Commands may be sent to the implant by modulating the amplitude of the power waveform, but data rates cannot exceed this relatively low carrier frequency.

Typically, the power transmitter consists of an efficient class E power amplifier driving a transmitting coil, feedback circuitry to ensure oscillation, and circuitry to modulate command data onto the power waveform [1-3]. All transcutaneous power links suffer from low-efficiency coupling from the driving coil to the pickup coil. To compensate, the power delivered to the pickup coil can be increased to a certain degree, though coil heating can become a problem at high current levels. Furthermore, the amount of power dissipated by a small implant should be limited to around 100 mW to avoid excessive heating of the surrounding tissue.

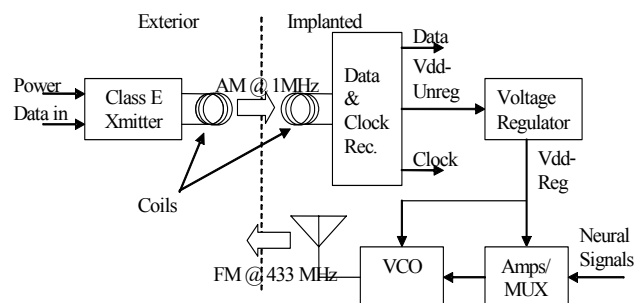


Fig. 1: Block diagram of a fully implantable recording system.

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Low-power operation is thus essential but increasingly difficult to achieve in multi-electrode systems requiring 100 or more amplifier circuits. A micropower wireless transmitter for telemetry of neural data represents a critical component in any practical implantable recording system.

### III. IMPLANTABLE FM TRANSMITTER

We developed a low-power FM transmitter that operates at 433 MHz. This frequency was chosen because it is the closest Industrial, Scientific, and Medical (ISM) band to the FCC-approved Medical Implant Communications System (MICS) band at 402-405 MHz. (The ISM band is suitable for testing and development purposes.) Moreover, using a high frequency makes small antenna designs possible.

For applications where the antenna is to be positioned directly underneath the skin, such as in this work, there will not be significant signal attenuation due to absorption into the tissue at this frequency. It has been shown that at 433 MHz, the depth of penetration through skin and tissue with high water content is 3.57 cm [4].

An integrated VCO was used as the FM transmitter; in practice, a separate power amplifier was not needed for short-range transmission. There are three main architectures for integrated VCOs in use today: the ring oscillator, the Colpitts oscillator, and the fully differential LC-tank oscillator. A Colpitts oscillator has been presented in [5], and a ring oscillator has been presented by [6]. For this project the fully differential LC-tank oscillator was chosen because this type of VCO provides a simple method for producing an FM output while having lower phase noise (frequency jitter) than ring oscillators.

A schematic of the VCO used in this work is shown in Fig. 2a. Transistors  $M_5$  and  $M_6$  act as variable capacitors each with a capacitance  $C$ .  $R_s$  is the parasitic series resistance present in the inductor  $L$  and limits the  $Q$  of the inductor. The  $Q$  factor is the ratio of imaginary impedance to real impedance, and is given by  $Q = (\omega L) / R_s$ , where  $\omega$  is the operating frequency given by  $\omega = 1 / \sqrt{2LC}$ .

It can be shown through small-signal analysis that the incremental input resistance seen when looking into nodes  $A$  and  $B$  (ignoring the inductor and capacitors), is given by

$$R_{in} = -\frac{1}{g_m}, \quad (1)$$

where  $g_m$  is the transconductance of the transistors  $M_1$ - $M_4$  and is given by

$$g_m = \sqrt{2\mu C_{ox}' SI_D} \quad (2)$$

where  $\mu$  is carrier mobility,  $C_{ox}'$  is gate oxide capacitance per unit area,  $S$  is the transistor width-to-length ratio, and  $I_D$  is dc drain current. Devices  $M_1$ - $M_4$  are sized such that their transconductances are equal.

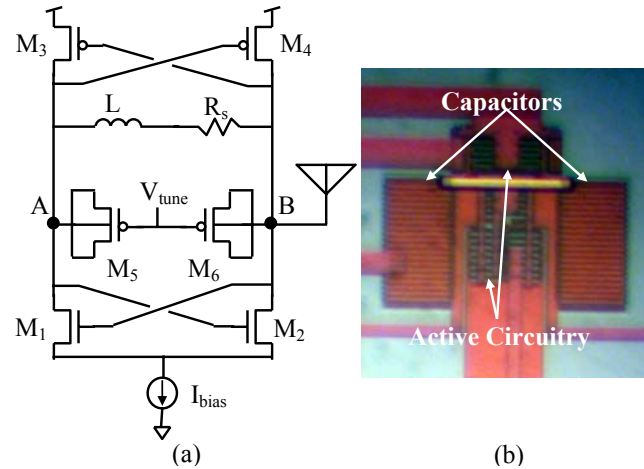


Fig. 2: (a) Schematic of the differential LC VCO used as a transmitter in this work. (b) Microphotograph of the fabricated VCO. Layout area measures  $117 \mu\text{m} \times 76.5 \mu\text{m}$ .

The circuit in Fig. 2a can now be modeled as an incremental resistance of  $-1/g_m$  in parallel with an LC tank. The inductor series resistance  $R_s$  can be transformed into an equivalent parallel resistance  $R_p = R_s(Q^2 + 1)$ . The parallel combination of the negative incremental input resistance of the active circuitry and the parasitic parallel resistance of the inductor,  $R_p$ , is termed  $R_{eq}$  and is given by

$$R_{eq} = \frac{-R_{in}R_p}{R_p - R_{in}}. \quad (3)$$

Equation (3) can now be used to establish the requirements for oscillation. From (3), if  $|R_{in}| > R_p$ , then  $R_{eq}$  is positive. A positive  $R_{eq}$  has the effect of loading the LC tank, thus causing the circuit to fall out of oscillation. Conversely, if  $|R_{in}| < R_p$ , then  $R_{eq}$  is negative. When  $R_{eq}$  is negative, it no longer loads the LC tank; it now supplies power to the LC tank. This condition of having  $|R_{in}| < R_p$  is what allows the VCO to maintain oscillations.

It is interesting to note that from (1) and (2), either the drain current ( $I_D$ ) or the width-to-length ratio ( $S$ ) of the transistors can be changed to achieve the required  $R_{in}$ . This means that making the transistors wider can lower the overall power consumption of the transmitter. The disadvantage to this approach is that the parasitic capacitances of the transistors also get larger. This has the effect of making the VCO less tunable.

There is yet another way in which the overall power consumption of the VCO can be lowered. Note that  $R_p$  is directly related to the  $Q$  factor of the inductor. This implies that for larger values of  $Q$ ,  $R_{in}$  can be larger and still sustain oscillations. Being able to tolerate a larger  $R_{in}$  results in a lower  $g_m$ , and for a fixed transistor size, a lower drain current can be used. This approach has the advantage of maintaining an acceptable level of tunability in the VCO.

To investigate this effect further, equation (2) and the condition for oscillation established by (3) can be used to

derive an expression relating the minimum bias current required to sustain oscillations to the  $Q$  factor of the inductor. This is given by

$$I_{b\min} = \frac{Q^2}{\mu C_{ox} S (\omega L)^2 (Q^2 + 1)}. \quad (4)$$

Fig. 3 shows equation (4) plotted on a logarithmic scale. Integrated (on-chip) inductors typically have  $Q$  values of five or less [7], while discrete (off-chip) inductors can have  $Q$  values of 20 or more at frequencies of 433 MHz. As seen in Fig. 3, the power dissipation of a VCO using a low- $Q$  integrated inductor can be an order of magnitude larger than that of a VCO using an off-chip inductor with a  $Q$  of 20-30.

With the criteria for oscillation known, transistor sizes can now be calculated. Table I gives the sizes of the transistors shown in Fig. 2a. The devices were made large enough to reduce the needed bias current while keeping the parasitic capacitances to a reasonable size. The sizes of transistors  $M_5$  and  $M_6$  are calculated from the following equation

$$C_{\max} = AC'_{ox} + 2WC_{ov} \quad (5)$$

where  $A$  is the transistor gate area,  $W$  is the transistor gate width, and  $C_{ov}$  is the overlap capacitance (per unit length) from the gate to the source/drain regions.  $C_{\max}$  is the maximum capacitance that transistors  $M_5$  and  $M_6$  can provide. For this work we chose  $C_{\max} = 1.28$  pF to achieve oscillations at 433 MHz with the inductor used: a small 47 nH surface-mount inductor with a tolerance of  $\pm 2\%$ . This device has a  $Q$  of approximately 20 at 433 MHz and measures 1.6 mm on its longest dimension. The self resonant frequency of this inductor is 1.3 GHz.

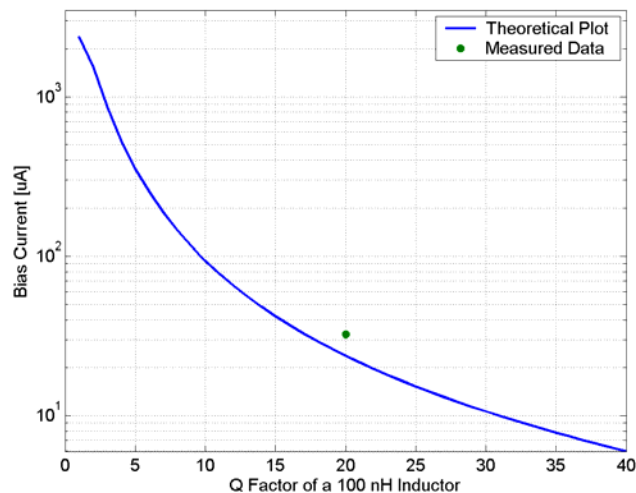


Fig. 3: Plot of VCO bias current  $I_{b\min}$  versus  $Q$  for a 47 nH inductor operating at 443 MHz. The star represents the measured minimum current required by a VCO using a 47 nH inductor with a  $Q$  of 20.

TABLE I  
VCO TRANSISTOR SIZES

Transistor	Width/Length ( $\mu\text{m}$ )
$M_1$	40.8 / 0.6
$M_2$	40.8 / 0.6
$M_3$	81 / 0.6
$M_4$	81 / 0.6
$M_5$	630 / 0.6
$M_6$	630 / 0.6

The inductor is the only off-chip component in the transmitter. An off-chip inductor was chosen for this project because of the high  $Q$  factor relative to integrated inductors. Alternatively, the  $Q$  value of integrated inductors can be raised using postprocessing steps such as etching away the substrate underneath the inductor [7]. By using an inductor with a  $Q$  of 20, the calculated minimum bias current required for oscillation is 20  $\mu\text{A}$ .

In order to modulate the 433 MHz carrier signal, the output of a neural signal amplifier was connected to the  $V_{\text{tune}}$  input of the VCO. Changing this voltage varies the capacitance of  $M_5$  and  $M_6$ , which modulates the VCO oscillating frequency. The neural signal amplifier used in this work was previously reported in [8]. It has a gain of 40 dB over a 7.5 kHz bandwidth and draws 16  $\mu\text{A}$  of current. The input-referred noise is 2.2  $\mu\text{V}_{\text{rms}}$ .

The overall chip area consumed by the VCO was 117  $\mu\text{m} \times 76.5 \mu\text{m}$ . The total area consumed by the VCO combined with one neural signal amplifier was 117  $\mu\text{m} \times 590 \mu\text{m}$ . The area consumed by the off-chip inductor was 1.28  $\text{mm}^2$ . Using a 3.3 V power supply the calculated minimum power dissipation of the VCO and neural signal amplifier was 119  $\mu\text{W}$ . The low power consumption of the VCO is largely due to the high  $Q$  factor of the inductor.

#### IV. TESTING AND RESULTS

The transmitter was fabricated in a commercially-available 0.5- $\mu\text{m}$  3-metal, 2-poly CMOS process (see Fig. 2b). A printed circuit board was fabricated to facilitate testing. For all measurements, the circuit was powered from a transcutaneous power link using a pickup coil consisting of 4 turns of 28 AWG wire with an outer diameter of 4 cm.

In order to validate equation (4), we measured the minimum bias current needed to sustain oscillations. We found that 32  $\mu\text{A}$  was the minimum current needed for our VCO using a 47 nH inductor with a  $Q$  of 20. This measurement is plotted in Fig. 3 as an asterisk. The measured point falls within 25% of the theoretical line.

The center frequency of the VCO was measured to be 420 MHz. The maximum variation achievable in the output frequency of the VCO was 22 MHz. The measured

minimum power dissipation for the VCO was  $159 \mu\text{W}$ , which is in close agreement with the calculated value.

The phase noise of the VCO was then measured using an Agilent E4402B 3-GHz spectrum analyzer. The phase noise at an offset of 600 kHz from the carrier was measured to be  $-104.4 \text{ dBc/Hz}$ .

The transmitter was then tested using a 1/8-wavelength monopole antenna measuring 8.6 cm in length. The bias current for the VCO was increased to  $125 \mu\text{A}$  in order to attain desired transmission range and adequate received power levels. The current consumed by the neural signal amplifier was  $16 \mu\text{A}$ , and the supply voltage was set to 3.3 V. This resulted in a measured power consumption of  $465 \mu\text{W}$ .

A 300 Hz,  $0.66 \text{ mV}_{\text{pp}}$  sine wave with a 1.65 V dc offset was applied to the input of the amplifier. A WinRadio 3150E FM receiver with a 30.5 cm monopole antenna was placed one meter away from the transmitter. Fig. 4a shows a plot of the recovered demodulated data. The frequency of the received waveform is 300 Hz, as expected. A photograph of the packaged transmitter chip detailing the off-chip inductor and antenna is pictured in Fig. 4b.

With the transmitting antenna parallel to the receiving antenna (which we define as position *A*) the received power was measured to be  $-54.5 \text{ dBm}$ . The orientation of the transmitting antenna was then changed with respect to the orientation of the receiving antenna. When the transmitting antenna was in the same plane as the receiving antenna but rotated by  $90^\circ$  (defined as position *B*), the received power dropped to  $-60.4 \text{ dBm}$ . When the transmitting antenna was pointed directly at the receiving antenna (position *C*), the received power continued to drop to  $-67.0 \text{ dBm}$ . (According to antenna theory, a monopole antenna should not radiate from the end of the antenna, but the nearby inductor and chip package act as parasitic radiating elements in different planes.)

The antenna was then removed and the above measurements were repeated with only the off-chip inductor

acting as an antenna. This is equivalent to using a  $0.002\lambda$  antenna. In position *A*, and at a distance of one meter, the received power was  $-61.4 \text{ dBm}$ . When the transmitting antenna was in position *B*, the received power dropped to  $-62.9 \text{ dBm}$ . As antenna theory would predict, the received power further dropped to  $-75.0 \text{ dBm}$  when the transmitting antenna was in position *C*. This was only slightly above the noise floor of the measurement equipment, which was  $-85 \text{ dBm}$ .

## V. CONCLUSIONS

We have presented an implantable FM transmitter that dissipates only  $465 \mu\text{W}$  of power. It was demonstrated that this transmitter can successfully transmit data distances of up to one meter using only a small off-chip inductor as an antenna. The results demonstrate the feasibility of using a micropower transmitter in a fully-implantable neural recording system. A relationship between the  $Q$  factor of inductors and the overall power dissipation of the VCO was also developed. We see that a relatively high- $Q$  inductor is necessary for low power operation. This implies that with some postprocessing steps, integrated inductors could eventually be used to realize a low power, fully integrated transmitter for biosignal recording applications.

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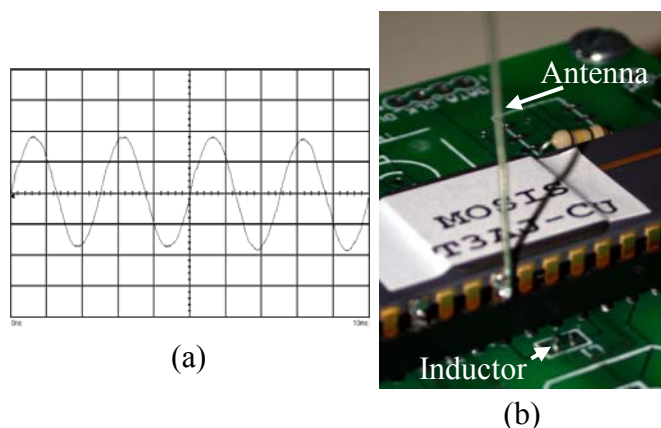


Fig. 4: (a) Recovered data showing a frequency of 300 Hz. (b) The transmitting antenna was a 1/8-wavelength monopole.