In-Vivo EEG Recording Using a Wireless Implantable Neural Transceiver

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Abstract- **We have recorded continuous *in-vivo* EEG and single-unit electrical activity from un-tethered rodents using an inductively powered and implantable wireless neural recording device. The device uses an integrated circuit to amplify modulate and transmit neural signals. The IC transmits neural signals (15 µV to 15 mV) at 3.2 GHz to a receiver located outside the environment of a behaving test animal with an input output correlation better than 90%. The design of the IC and the inductive powering are described.**

***Keywords* - Brain-machine interface, Telemetry, Epilepsy**

I. INTRODUCTION

The continuous monitoring of electrical brain activity with implanted depth electrodes is essential for understanding the neural substrates of many physiological and pathological brain functions [1]. Such recordings are extremely difficult in freely moving experimental animals, particularly in small rodents. The wires attached to the implanted electrodes can restrain and interfere with the natural behavior of the animal. Consequently, it has been impossible to correlate truly normal behavior with neuronal activity. In addition, it would be ideal if brain activity could be monitored while animals live in a natural and enriched environment. Animals reared in such complex “natural” environments, rather than in standard laboratory cages, have been shown to have altered brain development, such as a larger number of cells in certain brain areas and considerable changes in neuronal morphology [2]. For practical long-term experiments, this requires the use of existing inductive powering techniques [3], as well as the development of a novel integrated circuit (IC) to measure, amplify, modulate, and transmit electrophysiological activity to a receiver located far from the subject, where the data can be processed in real-time.

This paper describes the design and implementation of a neural transceiver to fulfill these requirements. An IC was designed and fabricated in a 0.35 µm CMOS analog IC foundry process to take advantage of the low-cost and mixed-signal suitability of this technology [4]. The IC amplifies, modulates, and transmits, neural signals from within the skull to a remote receiver located outside the environment of the animal. The chip together with the inductive powering circuitry, electrodes, and antenna, make up the complete transceiver. The device has been implanted *in-vivo* and used to record EEGs from the hippocampus of an un-tethered rat.

II. TRANSCEIVER DESIGN

The transceiver board includes pickup-coils, rectifiers, capacitors, and connectors in addition to the IC. The connectors interface the implant with the electrodes, and with a 2.5 cm flexible and implantable monopole antenna. The inductive power is coupled onto the coils located on the reverse of the package and rectified using a simple diode-bridge full-wave rectifier. The rectified charge is stored on 1 µF capacitors and drives the biasing circuitry on the IC. All of these off-chip components are integrated onto a small bio-compatible package with the chip. The bio-compatible transceiver implanted into the animal is shown in Figure 1.

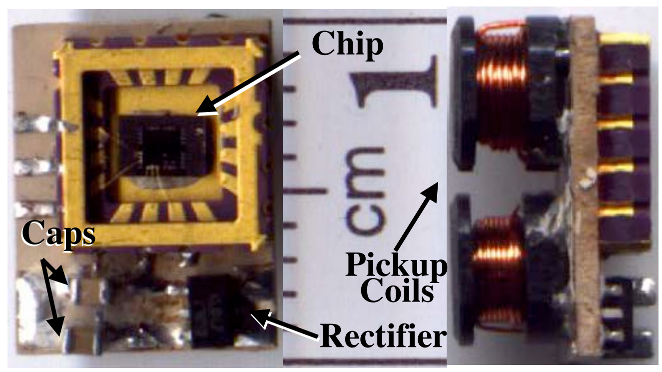

Fig. 1. Packaged transceiver top and side-view photographs.

The IC itself is further broken down into four distinct elements, the first of which is the biasing circuitry that receives power from the inductively coupled off-chip coils. The biasing circuitry contains a voltage-independent current source which generates a series of ground-referenced biasing potentials to drive the other three components of the chip. The first of these is the operational transconductance amplifier (OTA). The OTA is the input to the transceiver from the brain. It provides the first stage of amplification and is thus the most critical in terms of noise. The topology employed is fully differential This coupled with the use of bipolar electrodes eliminates the risk of amplifier saturation with dc drift, and drastically reduces the effect of common-mode noise. Compensation capacitors are added to the outputs of each differential stage to limit the bandwidth. The amplifier provides a gain of 34 dB over the 10 kHz bandwidth of interest with a sharp roll-off to deeply attenuate high-frequency noise from the inductive powering element. The amplified low-noise output from the OTA drives the third element of the IC: a 3 GHz voltage-controlled oscillator (VCO) with integrated passive components. By controlling the resonant frequency of the tank-circuit in the VCO, the OTA forces the VCO output to be a frequency modulated representation of the amplified signal. The output is then channeled to the fourth and final element of the IC, a class-C power amplifier (PA) attached to an implanted antenna, and then transmitted wirelessly to a receiver outside of the environment of the animal. The size of the packaged implant is less than $5 \times 5 \times 10$ mm^3 = 0.25 cm^3. The size of the four-channel IC shown in Figure 2 is less than 1 mm^2.

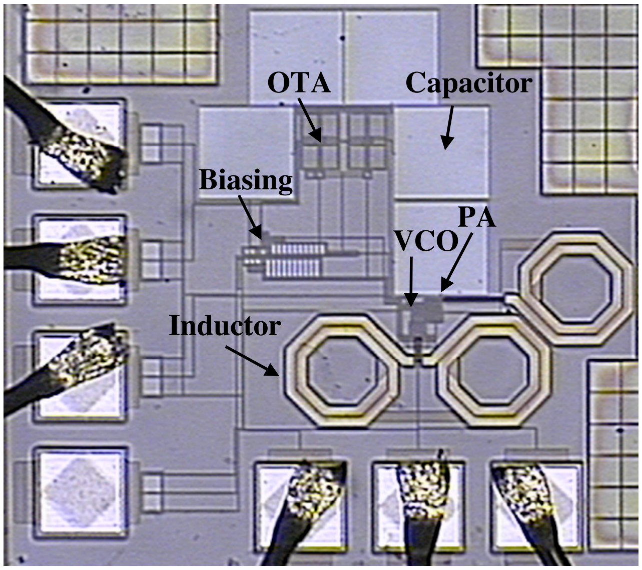

Fig. 2. IC micrograph.

The OTA must have a gain of 50 to place the 15 μV to 15 mV amplitude variations of the signal of interest within the optimal range of modulation of the voltage-controlled oscillator in the next stage. The bandwidth must be 10 kHz to accommodate the full range of desirable neural signal inputs while not amplifying any fraction of the 2 MHz inductive power signal that may have been coupled into the OTA input. A diagram of the specific topology chosen for the OTA is shown in Figure 3.

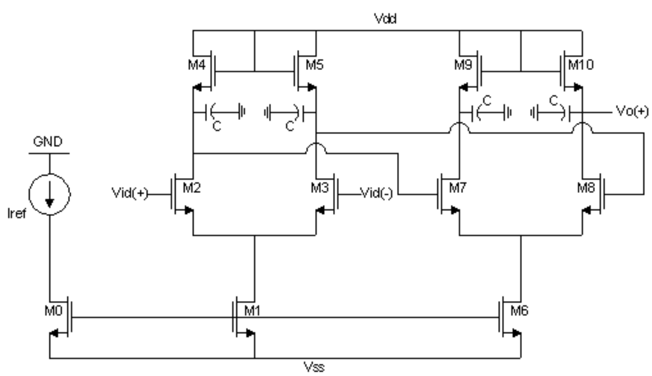

Fig. 3. Operational Transconductance Amplifier.

This topology consists of a two-stage amplifier employing the ratio of the gate area of the diode-connected transistors M4,5,9,10 to the differential transistors M2,3,7,8 to provide the desired gain. Transistors M1,6 mirror the current forced through M0 by the source circuitry and force it through the differential pairs located at their drains. The four capacitors attached to the outputs of the differential stages define the location of the poles in the frequency response of the amplifier. These are tuned to situate the poles at 10 kHz, which results in a gain of –20 dB at 2 MHz. The OTA output modulates the capacitance of the varactors in the cross-coupled oscillator shown in Figure 4. This topology was chosen for its simplicity, reliability, and low-power consumption.

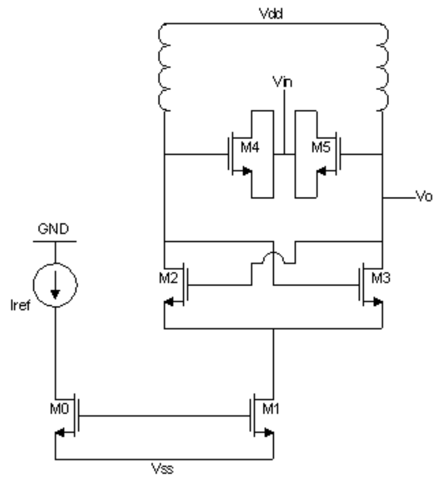

Fig. 3. Voltage Controlled Oscillator.

Transistors M0,1 mirror the current from the biasing circuitry. This current is then split between transistors M2,3 to form a positive feedback loop. The varactors are formed by M4,5, and their capacitance resonates in parallel with the inductors at 3.2GHz. The inductors themselves are integrated on chip with two turns per each of four layers for a total of eight turns. They were designed using existing modeling techniques [5]. At this frequency the tank circuit presents a high impedance to the drain of M2,3, and therefore achieve a gain > 1 which results in the oscillations. The low-frequency input voltage from the OTA modulates the depth of the depletion regions of M4,5 when their gates are held at a constant dc voltage, this in turn varies their capacitance. Variations in the input voltage and thus the capacitance, produce variations in the resonant frequency and thus the frequency of oscillations. In this manner a 3.2 GHz frequency-modulated (FM) signal is obtained from the low-frequency input of the VCO.

The Power Amplifier (PA) is perhaps the most difficult element to design in a low-power transceiver such as this one. PA efficiency is commonly defined as the ratio of the power delivered to the output to the power delivered to the amplifier. Given that there is a finite amount of power required just to bias the transistors into saturation, PAs driven just above this minimum must necessarily be very inefficient. High device aspect ratios help this situation by reducing the device dc power consumption, but at high frequencies of operation device parasitics introduce new inefficiencies directly proportional to the size of the output devices. Thus there is a complex set of tradeoffs which must be balanced to obtain a design which delivers the absolute minimum transmit power needed at the lowest possible total power consumption. Traditional Class A and B topologies are robust and power hungry. Although Class E-F topologies are theoretically more efficient, in practice they are exceedingly temperamental and unreliable requiring off-chip tuning devices and very large input signals for proper operation. The Class-C topology shown in Figure 5 is a good trade-off between the efficiency of the Class E-F topologies and the robustness of the Class-A-B topologies.

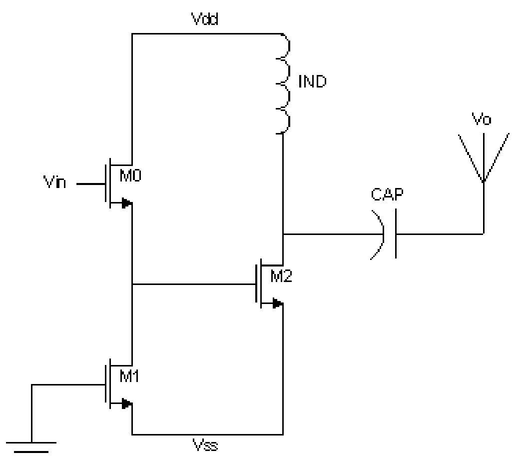

Fig. 3. Class-C Power Amplifier.

Transistors M0,1 in the PA are not a typical part of the Class C topology, they provide a low capacitance output for the VCO which minimizes the effect of the PA on the oscillating frequency. The PA is tuned for optimal performance at 3.2 GHz, reducing the power wasted in amplifying undesirable noise signals. The capacitor blocks the dc path from the power supply to the 50 Ω antenna. Transistor M1 sets the biasing voltage at the base of M2, and thus the current through that transistor and the overall gain of the PA. Althought the non-linearity of the PA results from the power-efficient operation of M2 on the edge of saturation, since the information in the signal is encoded in the frequency and not the amplitude, little or no information is lost.

III. Experimental Results

A. Lab Bench

The overall transceiver accomplishes a high correlation between the input signal and the received/demodulated signal as shown in Figure 6. The graph shows that for artificial signals > 10 μV a reproduction of > 70% of the original is attainable. For signals > 30 μV, the reproducibility is > 90%. The measured dc gain of the amplifier is 44, the roll-off is at 10 kHz, and the unity gain bandwidth is measured at 250 kHz allowing the gain to drop to almost –20 dB by the 2 MHz point, as intended. The sensitivity of the entire transceiver depends largely on the low-noise characteristics of the input stage. Given the gain of this stage, signal strength in subsequent stages will be much less susceptible to noise levels which are typically constant across a device. To quantify the noisyness of the amplifier we calculate what is called the input-referred noise. This is not a quantity which can be measured directly, rather it is the ratio of the noise measured at the output of the amplifier to the gain of the amplifier. For our OTA, the input referred noise is ~8 μV.

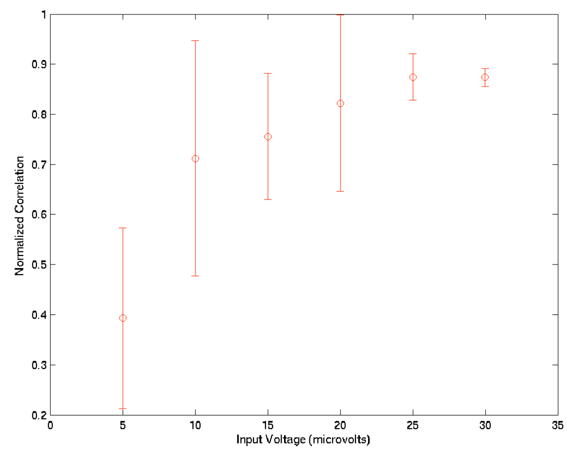

Fig. 6. Normalized input-output correlation as a function of input amplitude.

The quality of the oscillator is typically quantified by the phase noise. An ideal VCO would have a frequency power-spectrum consisting of a single impulse at the desired frequency. In practice the width of that impulse is increased by noise and inefficiencies in the circuit. The VCO in the transceiver has a measured 0.1% phase noise of –34 dBc. The magnitude of this value is due primarily to two sources. The first is the upconverted flicker noise from the output of the OTA and the current mirror M1 in the VCO [6]. This low-frequency noise gets modulated up to the carrier frequency and pollutes the purity of the modulated signal. The other major source of noise arises from the non-idealities present in the integrated tank circuit. In particular, the on-chip inductors while saving space, and allowing for a more economical and mass-producible design, in addition to higher resonant frequencies, have a higher series impedance than a bond-wire inductor would have. This impedance lowers the Q, or quality factor, of the circuit and increases the spectrum of frequencies for which the gain of the positive feedback loop is greater than one. Thus the VCO oscillates at a range of frequencies simultaneously and the power spectrum suffers accordingly. In our design slightly higher then possible phase noise figures are acceptable since the 300 MHz modulation range of the VCO is sufficient for a relatively large number of multiplexed analog channels to operate simultaneously with little loss in signal integrity.

As a result of variations in the foundry process and an excessive sensitivity of the power amplifier to the biasing voltages, the consumption of the entire transceiver ranges from 5 mW to 8 mW. Additionally the power transmitted is only –36 dBm. This means that we have to generate a stronger magnetic field in order to drive the power amplifier, and in addition we must place the receiving antenna closer to the animal in order to pick up the diminished signal.

B. In-Vivo

The essential test for the neural transceiver is whether in addition to performing well in the artificial lab-bench tests, it is able to measure real biological signals from living animals. A sampling of EEG waveforms recorded *in-vivo* from the rat hippocampus is shown in Figure 7 demonstrating our ability to achieve this goal.

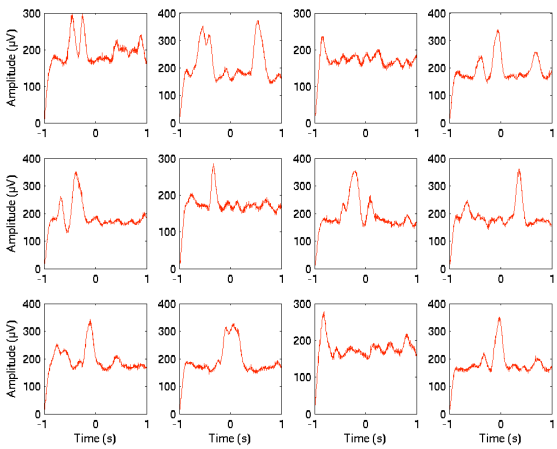

Fig. 7. Measured EEG Data.

The setup used to obtain the recordings consisted of an iCom 100 Watt Ham radio transmitter driving a large external coil in series with a passive impedance matching circuit. The rat was placed inside the coil with the transceiver mounted onto its skull. The transmitted signal was picked up and demodulated by an HP 8562A Spectrum analyzer. The output is sent to a computer where it is filtered using a second order Chebyshev digital filter with a cut-off frequency of 100 Hz. While the transceiver can record signals of up to 10 kHz with no loss in performance, in this particular implantation the electrodes were not within range of single-unit activity. The cutoff frequency chosen thus minimizes the noise while allowing us to see the EEG waveforms of interest in our studies on epilepsy.

V. Conclusions

The study of certain high-level neuroscience phenomena such as epilepsy is currently being hindered by the inability to examine the electrophysiological details of the phenomena while the subject is in a natural environment. This results from the constraints of the measuring methodology. The need for a miniature implantable wireless recording device to address these issues has been established. A design of an inductively powered single-channel neural recording device has been described. The transceiver uses an OTA to amplify neural signals, which are then frequency modulated by the VCO and transmitted by the PA. The recorded signal is a frequency-modulated (FM) version of the neuronal activity superimposed on a 3.2 GHz carrier wave. This FM wave is sent through a power amplifier that drives a 50 Ω antenna load and transmits the signal into free space. The signal is picked up by a similar antenna and demodulated using off-the-shelf equipment. The demodulated signal has a high degree of correlation with the original input signal for inputs as small as 15 µV and as great as 15 mV. A summary and comparison of designed and measured performance data for the implantable neural transceiver is shown in the table below.

SUMMARIZED DEVICE PERFORMANCE

Neural Transceiver Parameters	Designed	Measured
Gain	34 dB	32.9 dB
Phase Noise	NA	-34 dBc
Bandwidth	10 kHz	10 kHz
Power Dissipation	1.5 mW	5.8 mW
Transmit Power	15 µW	16 µW
Power Efficiency	1%	0.3%
Transistor Count	35 / channel	35 / channel
Passive Component Count	5 / channel	5 / channel
Transceiver Size	0.25 cm^3	0.25 cm^3
Cost	$50 / channel	$50 / channel

The device has a transmit range sufficient to allow for continuous recordings from animals in their natural environments. The power consumption of the neural transceiver makes inductive powering problematic but not impossible. In all other respects device behavior closely corresponds to design parameters both with artificial signals and with *in-vivo* recordings.

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