Wireless Neural Signal Acquisition with Single Low-Power Integrated Circuit

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***Abstract*—We present experimental results from an integrated circuit designed for wireless neural recording applications. The chip, which was fabricated in a 0.6-µm 2P3M BiCMOS process, contains 100 amplifiers and a 10-bit ADC and 902-928 MHz FSK transmitter. Neural signals from one amplifier are sampled by the ADC at 15.7 kSps and telemetered over the FSK wireless data link. Power, clock, and command signals are sent to the chip wirelessly over a 2.765-MHz inductive (coil-to-coil) link. The chip is capable of operating with only two off-chip components: a power receive coil and a 100-nF capacitor.**

I. Introduction

Progress in modern systems neuroscience relies on the ability to monitor and record electrical signals produced by neurons in the brain or peripheral nervous system. Despite advances in miniaturizing the electrodes used to detect neural activity, most modern electrophysiological studies require a wired connection to external amplifiers and recorders. The presence of a physical connection between electrodes and monitoring equipment limits the mobility of the subject under study and necessitates transcutaneous wires that present a risk of infection. As today's scientific instruments become tomorrow's medical devices to assist humans with disabilities (e.g., [1]), the need for completely wireless neural recording systems becomes more important.

Advances in circuit integration have led to the development of biopotential recording systems with wireless data telemetry (see [2] and references therein). Recent work at the University of Utah has produced an integrated circuit called the Integrated Neural Interface, or INI [3], that can be flip-chip bonded directly to a 100-channel MEMS Utah Electrode Array [4]. The goal of this project is to create a small (less than 1 cm), implantable neural recording system with low power dissipation (approximately 10 mW) so that surrounding tissues are not damaged by chronic heating. The earlier versions of the INI chip (designated INI1 and INI2) could not be used as stand-alone devices; they required a number of external components for operation and had to be powered from a battery. For long-term implantable use, batteries present a problem due to their size, mass, potentially toxic composition, and finite lifetime. Even rechargeable batteries would have to be replaced too often to be practical. (Modern pacemakers use non-rechargeable batteries that last for seven years before they are surgically replaced, but the power requirements for pacemakers is 2-3 orders of magnitude less than the ~10 mW needed for our 100-channel neural recording application.)

In this paper, we present results from a newly-fabricated neural recording chip, designated INI3. Unlike its predecessors, INI3 is capable of complete wireless operation: power and commands are sent to the chip via an inductive (coil-to-coil) wireless link, and data is transmitted from the chip via an RF telemetry link. The chip requires only two off-chip components: a coil to receive the power and command signals, and a single off-chip 100-nF capacitor to assist in power supply regulation. While similar in structure to the INI1/2 chips presented in [3], significant changes have been made to the RF transmitter circuits. These developments, along with experimental results from *in vivo* cortical recording sessions, are presented here.

II. INI3 System Design

Figure 1 shows a die photograph of the 5.4 × 4.7 mm^2 INI3 chip, which was fabricated in a 0.6-µm 2P3M BiCMOS process. The bulk of the layout area is consumed by a 10 × 10 array of neural signal amplifiers with bond pads that match the 400-µm pitch of a Utah Electrode Array, allowing non-toxic AuSn flip-chip assembly. Each amplifier has a gain of 60 dB with globally-programmable high- and low-frequency cutoffs that typically pass signals up to 5 kHz. One user-selected amplifier is digitized at 15.7 kSps by a 10-bit successive-approximation ADC. To limit the telemetry data rate to practical levels, 100 "spike detector" circuits use comparators to detect neural action potentials (spikes) that exceed a user-programmable threshold either in the positive or negative direction. The 6-bit DACs used to set individual thresholds for each channel are incorporated into the 100 neural amplifier blocks. The 100 latched comparators used for threshold crossing detection are grouped above the amplifier array to separate noisy digital circuits from sensitive analog amplifiers.

Additional on-chip circuits rectify the ac voltage on the power receiving coil and produce a regulated 3.3V dc supply for the chip (the chip can function anywhere between 3-4V). A 2.765-MHz inductive link supplies power to the chip; additional circuits recover this frequency and divide it by eight to produce a 345.6 kHz on-chip system clock with 50% duty cycle. Commands are sent to the chip by modulating the amplitude of the power signal. An on-chip command receiver detects amplitude changes in the unregulated voltage, waits for a specific 8-bit header signal, then reads in 852 control bits at a rate of 16 kbps. The control bits are used to set spike detection threshold levels, select a channel for the ADC, power down amplifiers that are not being used, set amplifier bandwidths, and configure the FSK-modulated RF transmitter.

Prior versions of the INI chip employed a power-optimized CMOS *LC* voltage-controlled oscillator (VCO) as a wireless transmitter [3]. This configuration required a high-resolution DAC to set the carrier frequency through the analog VCO control voltage. This approach was highly susceptible to frequency drift due to supply noise and temperature and power supply variations because the VCO was operated on a high-gain portion of its tuning curve.

To combat the problem of temperature and supply dependence, the transmitter core has been redesigned as a digitally-controlled-oscillator (DCO), shown in Fig. 2. The architecture of a DCO is identical to a VCO, but rather than relying on the output voltage of an 8-bit DAC to control the varactor capacitance, the capacitance in a DCO is directly controlled by an 8-bit digital control word. The measured tuning characteristic, shown in Fig. 3, is nearly linear and exhibits a temperature dependence of -96 ppm/°C from 35-39°C (four times better than previous INI VCOs).

Direct digital control of the capacitance is accomplished by taking advantage of the sigmoidal C-V curve of an accumulation-mode MOS varactor (see Fig 2 inset). The small, unit-sized varactors are divided into eight binary-weighted groups (only four bits are shown in Fig. 2 for simplicity), and the control terminal of each group is connected to one of the bits of the control word. When the i^{th} control bit is switched from 0 to 3V, the total tank capacitance is decreased by $2^i\Delta C$, corresponding to an increase in frequency. Though the relationship between frequency and capacitance is nonlinear, over the frequency range of interest the nonlinearity is mild and acceptable as illustrated in Fig. 3. This technique has no impact on the power dissipation of the oscillator. Furthermore, it is more compact than an approach employing switches and fixed capacitors, and it does not degrade the tank *Q*. This DCO design reduces temperature and supply dependence as compared to a traditional VCO because all of the varactors are operated at the extremes of the C-V characteristic where the capacitance is far less sensitive to small changes in the control voltage (see Fig. 2 inset).

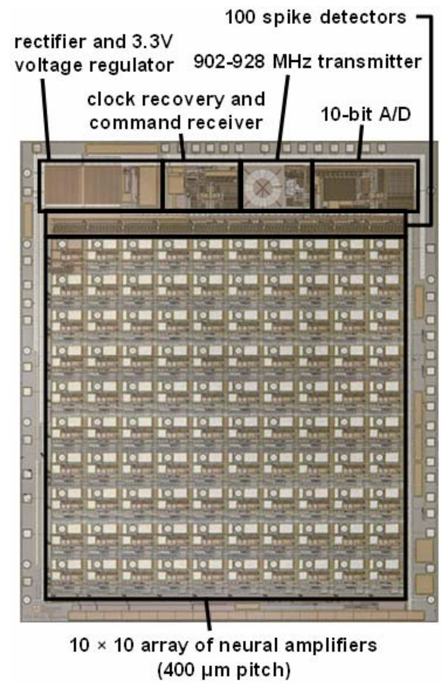

Figure 1. Microphotograph of 5.4 × 4.7 mm^2 INI3 wireless neural recording chip, fabricated in a commercial 0.6-µm 2P3M BiCMOS process.

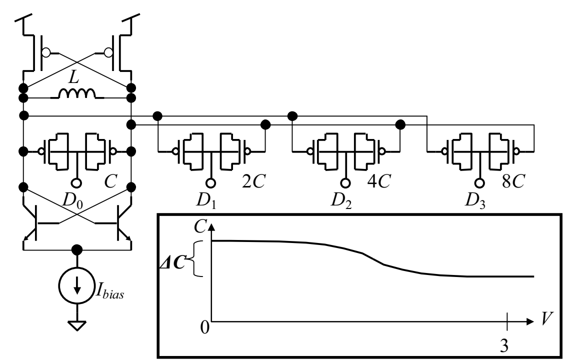

Figure 2. DCO schematic without FSK varactors (only four bits shown for simplicity) and accumulation-mode MOS varactor C-V characteristic (inset).

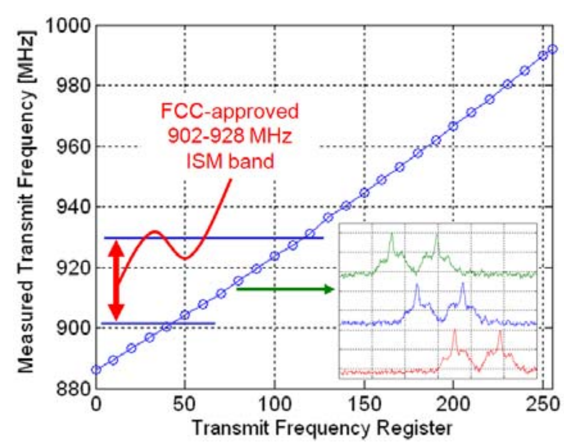

Figure 3. DCO tuning characteristic with measured output spectra for adjacent tuning codes 78, 79, and 80 (green, blue, red) from a range of 0-255.

This digital tuning technique is also used to implement FSK modulation (see Fig. 3 inset). The varactors used for modulation (not shown in Fig. 2) are approximately half the size of the unit varactor in the tuning array. With two additional control bits, the FSK frequency spacing is programmable from 165 kHz to 660 kHz.

The overall power dissipation of the RF transmitter is 500 µW, a 10× improvement over previous INI VCO designs. This large improvement is due to the shift in process technology from 0.5-µm CMOS to a 0.6-µm BiCMOS process with a thick top metal layer. Using this low-resistance metal, it was possible to design a 26-nH inductor with $Q = 11$ at 900 MHz, dramatically decreasing the power dissipation of the circuit. Additionally, the use of bipolar npn transistors in the $-g_m$ cell improves power dissipation by maximizing the design's g_m/I_{bias} ratio.

To complete the wireless interface to the INI3 chip, custom printed circuit boards were designed to send power and commands and receive telemetry data (see Fig. 4). The power board uses a class E power amplifier to create a ~60V_{pk} 2.765-MHz waveform from a 5V supply and drive a 5.8-cm diameter, 28-turn printed-circuit coil. The resulting ac magnetic field powers the INI3 chip. Techniques described in [5] were used to optimize the power link. A USB link to a laptop PC allows the user to send command strings that modulate the amplitude of the coil voltage; the INI3 chip can be completely reprogrammed in less than 100 ms.

The telemetry receiver board implements a 900-MHz FSK demodulator with programmable center frequency and USB interface. Demodulated data are streamed to the PC at the INI3 transmission rate of 345.6 kbps. PC software locates frame markers, checks parity bits, and decodes the data.

III. EXPERIMENTAL RESULTS

A. Battery-Powered Operation

Due to a design error in an on-chip bias generator, the neural amplifiers exhibited very poor power-supply rejection. When an isolated amplifier was powered from a benchtop supply, its input-referred noise was 4.8 μV_{rms}, near the design specification of 5 μV_{rms}. However, when the complete INI3 chip was powered from batteries using the on-chip voltage regulator, the input-referred noise increased to 20-30 μV_{rms}. With wireless inductive power, the input-referred noise increased to 30-40 μV_{rms}. For this reason, initial experiments were performed using battery power to reduce amplifier noise.

For these experiments, we recorded signals from a 100-channel microelectrode array (Cyberkinetics, Inc.) implanted in motor cortex of a cat. Recordings were performed approximately six months after implantation. One amplifier from an INI3 chip was connected to an electrode via a head-mounted connector. The cat was awake and resting comfortably during all recordings. The chip was powered from a 3V battery, but clock and command signals were sent wirelessly over an inductive link. A 5.8-cm receiving coil was connected to the INI3 chip and positioned 3 cm from the transmit coil shown in Fig. 4. The telemetry receiver shown in Fig. 4 was positioned approximately 5 cm from the INI3 chip to receive data.

Fig. 5 shows time-aligned spikes recorded from six different electrodes during a single recording session. Spikes from at least two distinct neurons can be observed in the top two traces.

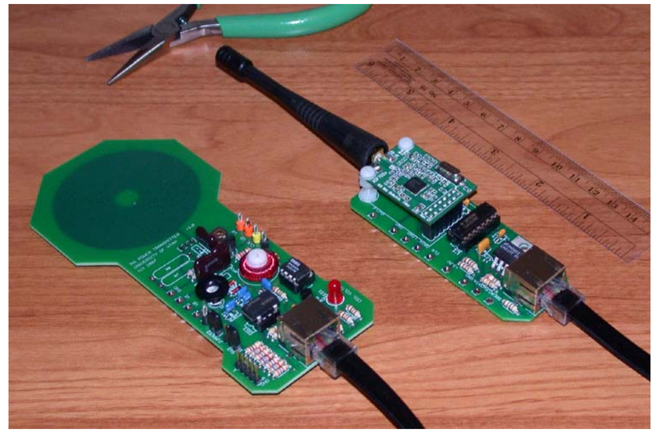

Figure 4. Custom circuit boards supporting wireless power and data link to INI3 chip. Left: 2.765-MHz inductive power/command unit with 5.8-cm printed-circuit coil. Right: 902-928 MHz RF telemetry receiver unit.

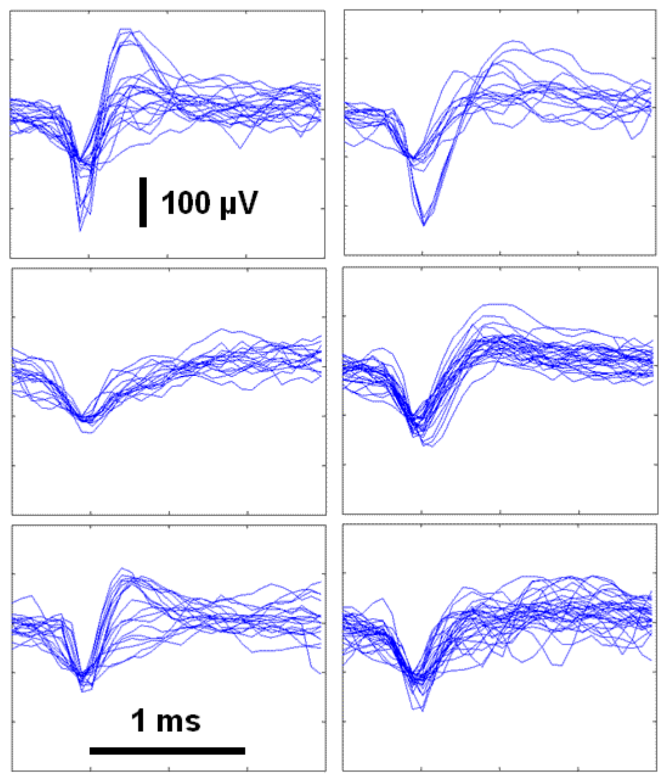

Figure 5. Peak-aligned neural spikes recorded in cat motor cortex from six different electrodes on the same day. The scale in all six graphs is identical. Data were recovered wirelessly using INI3 chip powered with a 3V battery; clock and command signals were sent wirelessly via an inductive link.

B. Complete Wireless Operation

We also obtained neural data using a completely wireless system with no battery. In this experiment, the chip received power, clock, and command signals from the inductive power link. As before, neural data telemetry was transmitted wirelessly approximately 5 cm to a receive antenna. The only off-chip components connected to the INI3 chip were a power receive coil and a 100-nF capacitor.

For this experiment, we monitored neural signals from a 100-channel microelectrode array (Cyberkinetics, Inc.) implanted in the premotor cortex of a rhesus monkey (Monkey D). Recordings were performed more than 22 months after implantation. As before, one channel of the INI3 chip was wired to an existing head-mounted connector. The chip was positioned near the head during recordings, and the boards in Fig. 4 were used for all power and communication.

Fig. 6 shows several time-aligned spikes recorded with this completely wireless system. Due to the low signal-to-noise ratio, we conducted a simple experiment to confirm that the signals we observed were in fact neural spikes. We recorded neural data in 5-second trials and counted the number of spikes that exceeded a fixed threshold in this time. In some trials, the monkey was resting. In others, the monkey was actively reaching for a target. Fig. 7 shows the number of spikes in ten such trials. Clearly, the neural signal is being modulated by reaching activity.

IV. CONCLUSIONS

We have demonstrated wireless, inductively-powered neural recording from a primate using a single-chip system with a minimal number of off-chip components. The INI3 chip is sufficiently small and low-power to permit implantation in future systems. The total power consumed by the INI3 chip is 8 mW. Of this, the 100 neural signal amplifiers consume 44% of the total power, the 100 spike detectors consume 22%, the voltage regulator consumes 13%, the RF transmitter consumes 9%, the ADC consumes 6%, and the command and clock recovery circuitry consumes 6%. The dramatic improvement in the RF transmitter power dissipation opens up the possibility of including a complete PLL frequency synthesizer in future INI chips to lock the transmission frequency to a multiple of the clock frequency.

Since the INI3 transmitter does not use an off-chip antenna, its transmission range is limited to a few centimeters. While this is sufficient for transcutaneous links in medical devices, there are many scientific applications where additional range would be useful. To this end, we included an on-chip differential class A output stage capable of driving an external antenna. Additional power may be allocated to this output stage, resulting in a transmission range of several meters, as described in [6].

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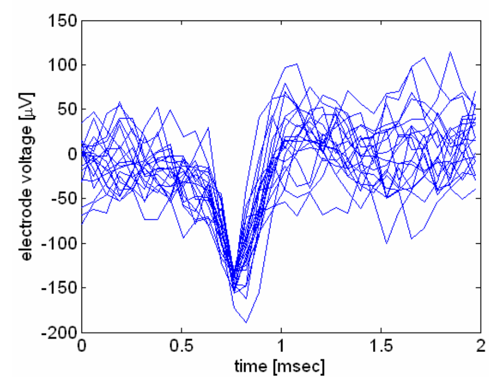

Figure 6. Spikes recorded from monkey premotor cortex during completely wireless operation: no battery was used; power, clock, and command signals were sent using a coil, and telemetry was received using an antenna.

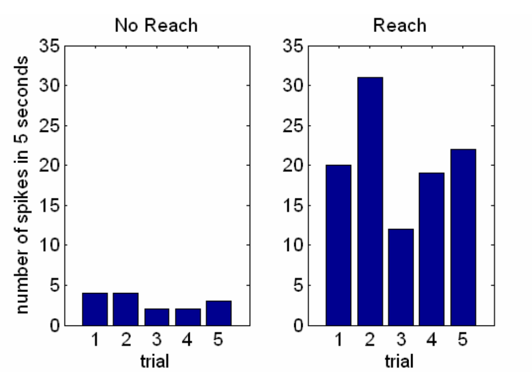

Figure 7. Spike counts recorded from monkey premotor cortex during completely wireless operation as in Fig. 6.