A Small, Light-Weight, Low-Power, Multichannel Wireless Neural Recording Microsystem

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Abstract—A small (<1cm^3), light-weight (<1g including batteries), low power (10mW, lasts 25 hrs), long range (22 ft.), 3-channel wireless neural recording system is designed, fabricated and characterized through in-vitro and in-vivo experiments. The transmitter comprises of an ASIC fabricated in 2-Poly, 2-Metal 1.5um AMIS process which can transmit data out wirelessly using on-chip VCO or low power commercial off-the-shelf (COTS) transmitters. The microsystem is employed in collecting neural signals from two different animal models: axons in cockroach leg and forelimb area of the motor cortex of a mature Long Evans rat.

I. INTRODUCTION

Multichannel neural recording/stimulation systems together with low-power and lightweight wireless communication systems are needed in many emerging applications, including remote sensing, sensor networks, and monitoring of biological signals in freely moving subjects. Many research groups in both academia and industry are involved in developing such systems and furthermore recent advances in Integrated Circuit (IC), process technology and Micro Electro Mechanical Systems (MEMS) provide means to implement systems meeting stringent constraints. Most of these microsystems are implemented at board-level that meet some needs, however design space highly depends on application and in some cases Application Specific Integrated Circuit (ASIC) design is the only solution. The microsystem presented here, is designed to study the neural signals of light-weight animal subjects while interacting in their natural habitat. The challenge of such a system lies in limited carrying capability of test targets (low weight), long range and long recording sessions needed to learn encoding of complex motor behaviors, auditory feedback and vocal learning in these species. To collect this information extra-cellular neural activity recorded by electrodes implanted in the host's forebrain should be processed and then transmitted out over a wireless link to a remote receiver.

In this work, a wireless battery-powered multichannel biotelemetry microsystem is designed and fabricated based on the ASIC's architecture presented in [1]. The test subjects for the in-vivo experiments were Long Evan Rats and Cockroaches which will be explained in detail in section III. The rest of the paper is organized as follows: Section II introduces the architecture of the system; section III presents the in-vivo/in-vitro experiments and analyzes the results. Section IV concludes the discussion.

II. SYSTEM ARCHITECTURE

The microsystem architecture is depicted in Figure 1 and the transmitter is shown in Figure 2. Implanted electrodes (impedance ~ 0.8 MΩ @ 1KHz) sense the action potentials with a typical magnitude of 50μV to 500μV in 100Hz to 5 kHz frequency range; the sensed signals are preamplified by low-noise, DC-Stabilized band-pass amplifiers with a low-pass cut-off frequency of 8 kHz, a high-pass cut-off frequency of 200 Hz and a gain of 40dB. The amplified spikes from the amplifier array are then time multiplexed along with a DC voltage so called Frame Marker which is used by the software at the receiver to separate the time multiplexed signals. Finally two transmitters are used for this design: an on-chip Colpitts oscillator with off-chip LC tank, and an off-chip low power transmitter, applying the non-linear Wide Band Frequency Modulation (WBFM) scheme to the multiplexed signal. To power the neural board transmitting by off-chip VCO, two Energizer-386 batteries are used, resulting in operation up to 25hrs. When using on-chip VCO, two Energizer-377 batteries are used which last for 5 hrs.

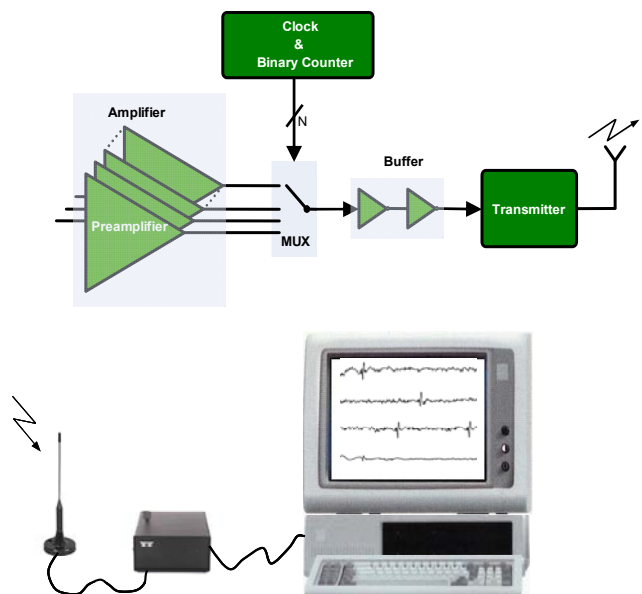Aiming at engineering a complete system (front-end,

Fig.1, The System's Architecture

signal conditioner, data modulator, wireless link, receiver), there are many unknown interrelated parameters only known by testing the entire system; a high-level system simulation can provide first-order information about the parameters. A Matlab-based model, simulating the entire system from the implanted electrode to the wirelessly received demodulated signal, is developed. Using this high-level simulation code, all the non-idealities of the entire signal path from the implanted electrode thermal noise to the limited receiver bandwidth can be calculated giving valuable insight in decision making at the initial design phase and therefore, in the first phase of design, a high-level Simulink model of the complete system was developed and employed in designing the presented system.

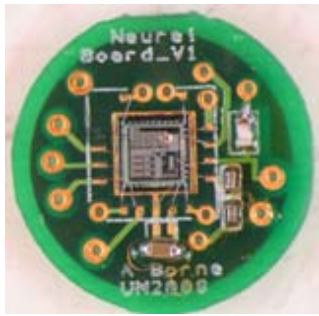The complete signal path is shown in Figure 3 and its various blocks are discussed below.

Fig. 2, The Assembled Microsystem:
Weight < 1g, Volume < 1cm^3

A. Preamplifier

The preamplifier consists of a High Pass Filter (HPF) with a unity gain, a two-stage compensated amplifier with a differential input stage and a common-source second stage employed in resistive closed loop configuration providing a gain of 40dB. An important issue needed to be addressed in designing of Neural Amplifiers, is the electrode-electrolyte interface. The recording electrodes are placed in Extracellular Fluid (ECF) and form an interface with the electrolyte. At the frequency of interest this interface can be modeled as a so called double layer capacitance (C_{dl}). Electrons are means of charge transfer in Electrode and subsequent circuitry, and in the ECF ions are responsible for charge transfer. Any potential sensed by electrodes is due to redistribution of charges (electrons on electrode-side and ions on the electrolyte-side) on C_{dl} in a non-faradic process [2]. Many designs use C_{dl} along with a large resistor to form a high pass filter, however this scheme has the drawback of high cut-off frequency variation in different in-vivo experiments as C_{dl} can varies between 150pF to 2nF from electrode to electrode. Another method uses capacitive coupling to DC stabilize the neural amplifier; however this approach too has its limitation due to C_{dl} variation. In this design we have used a PMOS source-follower (unity-gain buffer) driving an on-chip RC high pass filter with a high cut-off frequency of 200Hz to address the problem of

amplifier saturation due to electrode DC potential; the same circuit filters out the Local Field Potentials (LFP). Using this method the channel's performance (Signal to Noise Ratio) was constant over a 1V range of input offset (-500mVDC to 500mVDC) which is adequate to reject the electrode-electrolyte DC potential. The amplifier dissipates 80μW from a 3-V power supply and has an input referred noise of 12 μV$_{rms}$.
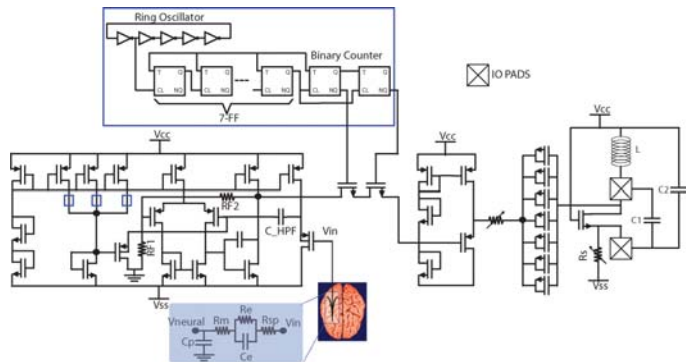

Fig. 3, The Complete Signal Path

B. Multiplexer and Buffer

The two-bit binary counter which commands the multiplexer has a five-stage ring oscillator with its output being down converted by a seven-stage T-Flip Flop (T-FF) providing a switching frequency of 16 kHz for each channel. For communication applications ring oscillators are not suitable due to their poor phase noise performance but in this application their phase noise performance does not pose a problem. One issue with this design is the variation of system clock with decreasing power supply voltage (224 Hz/V), which happens as the batteries discharge. To overcome these issues, techniques should be employed to desensitize the ring oscillator frequency to variation in power supply voltage. The current design compensates for clock variation in software by extracting the timing information every cycle through detecting the Frame Marker (FM). The Multiplexer gates are NMOS pass gates controlled by the binary counter and the buffer is a wide-band source follower which buffers the demultiplexed channels. The channels are mixed in time with a large DC voltage named Frame Marker (FM) which is used by the demultiplexing software at the receiver end. While this scheme (using Frame Marker) eases the design implementation it can lead to distortion at the receiver due to limited bandwidth of the receiver. Care must be taken to optimize the robustness of the demultiplexing process (depending on the relative amplitude of FM) and required bandwidth of the transmitter (affecting the fidelity of the demodulated signal).

C. Transmitter

Two transmitters are used: an on-chip single-transistor Colpitts VCO which employs three off-chip components (an inductor and two capacitors) as LC tank. It operates in 94 MHz to 98 MHz range and dissipates 1.48 mW from a 3-V power supply. To reduce power consumption a power amplifier is not used to drive the antenna and therefore through capacitive coupling to the antenna, the center frequency can vary. By using a receiver with wide bandwidth, high fidelity demodulation of signal can be guaranteed despite the center frequency variation; furthermore the introduced perturbations in signals can be filtered out by off-line software-based filters as the movement induced center frequency variation has its power in frequencies below 10Hz. An off-chip COTS transmitter, MAXIM2608 VCO, can be mounted on the back of the PCB (depicted in Fig. 2), which consumes 2.7mA from 3V supply. This VCO uses 4 off-chip components and has its center frequency set at 440MHz. The neural chip configured with on-chip VCO has been tested successfully on the bench but for in-vivo experiments we have employed the off-chip VCO.

D. Backend

A digital radio (Universal Software Radio Peripheral, USRP) from Ettus Research LLC, is used as the receiver. This device is programmable by a PC to which it is connected via USB2 port. The software is an open-source framework so called GNU radio which can be modified to meet individual needs. The main advantage of this platform is the wide bandwidth (up to 6MHz) which is needed in multichannel systems with high channel count. WinRadio AX-07B UHF/VHF antenna is used as the receiver's antenna.

There are interrelated parameters impacting the bandwidth required to demodulate the data and to gain the desired performance those parameters need to be optimized on the transmitter side with considering the constraints imposed by maximum band width of the receiver and selected center frequency. If one considers transmitting a single amplified channel of neural spikes, after some mathematic operations the required band width can be calculated as:

$$BW = f_0 \times \frac{\beta \times A \times (V_{pp})}{2 \times (C_{eq} + C_{VAR0} + C_{stray})} \tag{1}$$

BW: Band Width of the receiver [Hz]
f_0: the center frequency of the transmitter [Hz]
β: Varactor Capacitance Change per Volts [F/V]
A: Neural Amplifiers gain [V/V]
V_{pp}: Neural Signal Amplitude [V]
C_{eq}: Equivalent capacitance of LC tank [F]
C_{VAR0}: Varactor's capacitance with t voltage applied [F]
C_{stray}: parasitics capacitance [F]

The above formula assumes a low frequency input signal compared to center frequency. As (1) suggests, there is a tradeoff between amplifiers gain and modulation depth of the VCO with respect to fixed center frequency and BW of the receiver. For the system presented here, receiver's BW is set at 4MHZ, considering MAX2608 modulation depth at 440MHz center frequency, channel count and amplifiers intrinsic offsets.

An important performance metrics in wireless biotelemetry systems is the crosstalk present among channels. In communication schemes where data channels are sent using Time-Division-Multiplexing (TDM) a common source for crosstalk comes from channels leaking into adjacent channels as a result of limited receiver bandwidth. In the work presented here this problem is solved through using a wide BW digital radio. The architecture of the transmitter can induce crosstalk as well. For example in the presented microsystem the limited current capability of amplifiers together with RC constant of the multiplexer can lead to crosstalk among adjacent
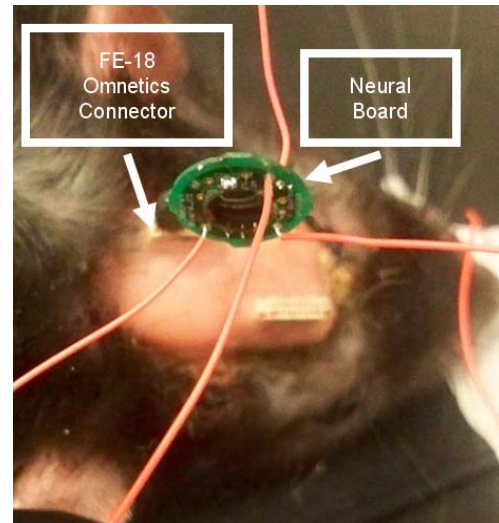

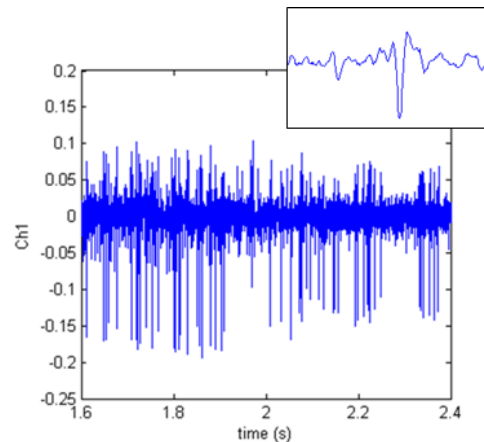Fig. 4, Recording from motor cortex of Long Evans Rat

Fig. 6, Recording from Femur of cockroach leg

channels but this problem can be avoided with proper sizing and design. To further reduce the crosstalk amplifiers' outputs are shielded with grounded lines and as a result, the cross talk is less than 2% among channels.

III. IN-VIVO RECORDING

Two test subjects were employed for in-vivo experiments: Cockroach and Long Evans rat; in recording from Long Evans rat (Figure 4) electrodes were inserted in the forelimb area of the motor cortex (coordinates AP: 2.5 mm, ML: 2.5 mm from bregma). Anesthesia was maintained through intraperitoneal injections of a mixture of 50 mg/ml ketamine, five mg/ml xylazine, and one mg/ml acepromazine at an injection volume of 0.125 ml/100 g body weight. The connector used on the rat was FE-18 Omnetics connector; to interface the FE-18 with our 3-channel neural board a custom made 3-channel connector was built which limits the functionality of the system to record from three adjacent sites at every recording. Figure 5 depicts the results. Before the wireless recording experiment, all the 16-channels were monitored through tethered recording and the results were used as a point of reference. The wired recording was in good agreement with wireless results. The second in-vivo experiment was on cockroaches. Cockroach leg contains sensory receptors and provides a stable platform to record neural activity from in initial prototyping of a design [3]. In this test the stainless steel pin were used as electrodes with the reference electrode inserted in Coxa and the recording electrode placed in Femur. The recorded action potentials are depicted in Figure 6.
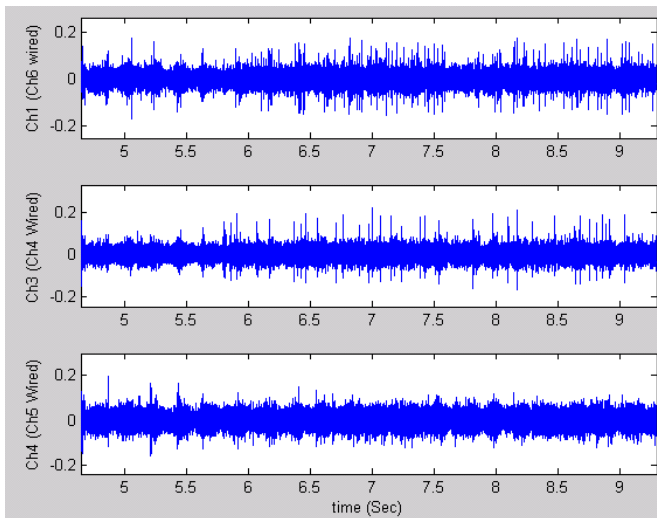

Fig. 5, Recording from motor cortex of Long Evans Rat

IV. CONCLUSION

The A small (less than 1cm^3 including the batteries) , light weight (less than 1g including the batteries), low power (10mW, lasts 25 hrs), long range (22 ft.), 3-channel wireless neural recording microsystem is designed, fabricated and its functionality is proved through in-vivo experiments on two animal models. Future work

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