24.1 A Miniaturized 64-Channel 225μW Wireless Electroocorticography Neural Sensor

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Substantial improvements in neural-implant longevity are needed to transition brain-machine interface (BMI) systems from research labs to clinical practice. While action potential (AP) recording through penetrating electrode arrays offers the highest spatial resolution, it comes at the price of tissue scarring, resulting in signal degradation over the course of several months [1]. Electroocorticography (EOg) is an electrophysiological technique where electrical potentials are recorded from the surface of the cerebral cortex, reducing cortical scarring. However, today’s clinical ECoG implants are large, have low spatial resolution (0.4 to 1 cm) and offer only wired operation.

To enable chronic and stable neural recording, we introduce a minimally invasive, wireless ECoG microsystem. Wireless powering and readout are combined with a microfabricated antenna and electrode grid that has >10× higher electrode density than clinical ECoG arrays, providing spatial resolution approaching today’s penetrating electrodes. Area- and power-reduction techniques in the baseband and wireless subsystem result in over 10× IC area reductions with a simultaneous 3× improvement in power efficiency over the state of the art (see Fig. 24.1.4), enabling a minimally invasive platform for 64-channel recording. The low power consumption of the IC, together with the antenna integration strategy enables remote powering at 3× below established safety limits [2], while the small size and flexibility of the implant minimizes the foreign body response. The improved implant safety and longevity gives wireless ECoG excellent prospects to become the technology of choice for clinically relevant BMIs in the foreseeable future [1].

Figure 24.1.1 illustrates the concept of the implantable ECoG microsystem and a block diagram of the IC, which includes circuitry for signal acquisition, a matching network, clock recovery, communication and power management. To mitigate the implantation of a large rigid structure, the IC is bonded directly to the human skull model [2]. A 1.5cm-diameter external antenna completes the link together with the electrodes and is used for both power and data telemetry. The IC was assembled together with the microfabricated ECoG electrodes and antenna on a PCB and implanted in an anesthetized Long-Evans rat over the left cortical hemisphere. All experiments were performed in compliance with the regulations of the Animal Care and Use Committee at UC Berkeley. Electrical recordings were made on all channels prior to and 15 minutes after the administration of Pentobarbital, a sedative. It is known that anesthesia causes increased Δ-band (1 to 4Hz) oscillations and depressed high-γ (65 to 125Hz) activity [8]. A representative channel is plotted in Fig. 24.1.6 showing that results are consistent with deepened anesthetic state.

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References:
Figure 24.1.1: Device concept for Electrocorticographic (ECoG) neural sensor, IC block diagram, and photograph of the high-density microfabricated electrodes and antenna.

Figure 24.1.2: Detailed front-end circuit diagrams.

Figure 24.1.3: RF-to-DC rectifier circuit and timing diagrams.

Figure 24.1.4: Front-end measurement results of full channel including ADC. State-of-the-art comparison is charted for fchop=8kHz.

Figure 24.1.5: Wireless subsystem measurement results: RF-to-DC rectifier output ripple for single and dual-mode rectification during data modulation, BER vs. antenna separation, and power/area breakdowns.

Figure 24.1.6: In-vivo system measurements: recordings from a single channel before and 15 min. after sedative administration. Spectral band power changes are plotted for all channels.
Figure 24.1.7: Chip microphotograph.