A Biosignal Analog Front-End Leveraging Frequency Translation

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***Abstract*—We propose a superheterodyne frequency mixing approach for extending the passband frequency response of existing biosensors such as neural recording devices. In this approach, we use a digital switching mixer to translate the frequency spectrum of a biological signal from its original passband to an intermediate frequency (IF). The IF signal is then amplified and digitized by an existing neural recording device, and the original signal passband is recovered by synchronous demodulation in post processing. We demonstrate this approach by translating a pre-recorded mouse electrocorticogram from a frequency range of 0.5 Hz to 100 Hz up to an IF of 407 Hz, enabling the use of an existing IC having a low frequency cutoff of 12 Hz. With an input impedance of 10 MΩ, the total root mean square error (RMSE) of the reconstructed signal is on the order of 41 μV.**

I. Introduction

This work focuses on the development of biosensors such as neural recording devices for electrocorticography (ECoG) [1], [2]. There are several challenges associated with ECoG measurements. ECoG signals typically occupy the frequency range of $0.5 - 100$ Hz, with amplitudes of $10 - 100$ μV. Large direct current (DC) offsets on the order of a few mV are often present due to electrochemical effects at the brain-electrode interface [2]–[4].

Implanted recording devices are generally low-power devices with a low supply voltage (often less than 2 V) and correspondingly limited voltage headroom. Given the high gain required (i.e. $>$ 40 dB), any DC offset needs to be mitigated in order to avoid amplifier clipping. Furthermore, ECoG electrodes have high impedances of $1-2$ MΩ or more, so the input impedance must be very high to avoid loading the electrode voltage, while still maintaining a high common mode rejection ratio (CMRR) of at least 80 dB [3].

The spectrum of the ECoG signal varies widely with brain activity. For example, in sleep monitoring, alpha ($7.5-14$ Hz), beta ($14-40$ Hz), gamma (>40 Hz), and delta ($0.5-4$ Hz) waves are of interest. On the other hand, integrated circuit (IC) designs are normally optimized for a single transfer function (e.g $12-1630$ Hz [5]) and fixed at the time of manufacturing.

Since it is impractical to redesign existing ICs for each spectral response of interest, a method for changing the passband response of existing neural recording devices is highly attractive. This work approaches this problem by utilizing superheterodyne frequency mixing of the input neural signal after alternating current (AC) coupling and amplification.

As depicted in Fig. 1, the up-converted signal is sampled by an existing infrastructure IC that was initially developed for neural recording in dragonflies [5]. The IC has a passband of $12-1630$ Hz and will hereby referred to as the DF chip (abbreviation for dragonfly). In this work, front-end frequency translation is used to extend the DF chip low frequency cutoff from 12 Hz down to 0.5 Hz. Fig. 2 depicts the measured frequency response of the sensor before and after the frequency translation. The low side -3 dB cutoff has been extended from 12 Hz down to 0.15 Hz, enabling new applications, such as ECoG sleep monitoring, which wouldn't otherwise be feasible with this IC.

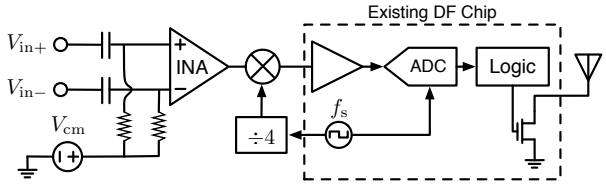

Fig. 1. Schematic of the proposed ECoG recording device.

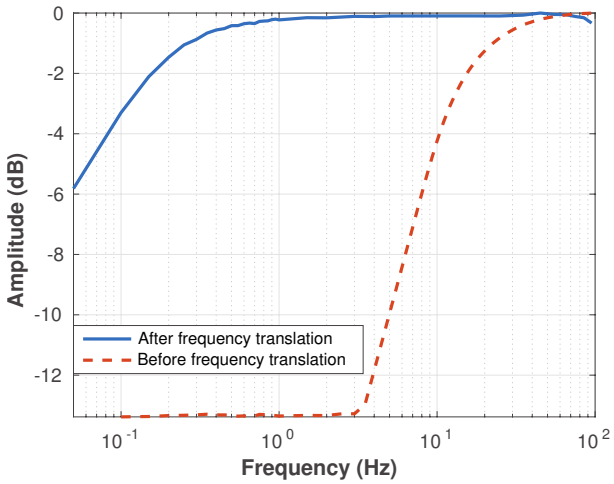

Fig. 2. Comparison of spectral response before and after frequency translation.

II. System Design

The existing DF chip [5] is equipped with 2 DC amplifiers and 14 differential biopotential amplifier inputs, as well as an 11-bit analog-to-digital converter (ADC) for signal digitization. It is clocked with a 20 MHz external quartz crystal. The chip is equipped with an RF power harvester that is implemented with a 4-stage Shottky diode voltage multiplier. The output of the power harvester is connected to

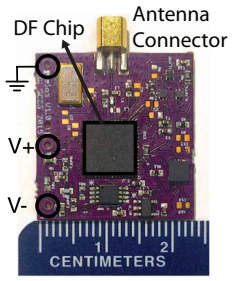

Fig. 3. The prototype biosensor board (25 mm×23 mm×3 mm)

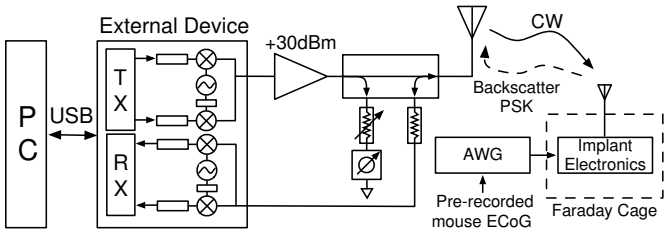

Fig. 4. Test setup for recording system.

a low dropout regulator (LDO) that generates a 1.23 V power supply. The digitized inputs are Hamming(16,11)-encoded and backscattered to the external device with a rate up to 5 Mbps using a binary shift keying (BPSK) modulation scheme.

A. Frequency Translating Front-End Design

To extend the low cutoff frequency of one channel of the DF chip, a frequency translating front-end was implemented using discrete components. Two 10 μF capacitors provide AC coupling in order to mitigate the DC offset from the electrodes, which can be up to a few hundred mV. Two 10 MΩ resistors provide a DC bias path for the amplifying stage, and are also connected to a μ-power 0.4 V voltage reference (LT6650) that provides a common mode reference. The input resistors set the input impedance to 10 MΩ and realize a high-pass filter (HPF) with the AC-coupling capacitors having a cut-off frequency of 0.15 Hz.

For the first amplifying stage, an Analog Devices AD8236 instrumentation amplifier (INA) was utilized. This amplifier was selected due to its very low quiescent current consumption (40 μA), low noise (5 μVpeak at 0.1−10 Hz), and low nominal operating voltage of 1.8 V. The INA is powered by a 1.23 V LDO output from the DF chip, and its reference input is driven by the 0.4 V voltage reference. The AD8236 is available in a tiny 1.57 × 2 mm wafer level chip scale package (WLCSP). The output of the amplifier V_{out} is given by:

$$V_{\text{INA}} = G_{\text{INA}}(V_+ - V_-) + V_{\text{ref}}, \tag{1}$$

where is the G_{INA} is the amplifier gain, which is set by a single resistor to 50 V/V, V_+ and V_- are the voltages at the positive and negative inputs respectively, and $V_{\text{ref}} = 0.4$ V is the voltage at the "reference" terminal.

The output of the INA is connected to a frequency mixer that is implemented using the Intersil ISL43L220 μ-power *dual* single-pole/double-throw (SPDT) switch. One input of the first SPDT is connected to ground and the other to the output of the INA. The control port of the switch is connected to a 407 Hz square wave signal. The latter is generated by a divide-by-4 clock divider circuit clocked by the 1628 Hz sampling clock (F_s) of the DF chip. The output of the first switch is amplified biosignal modulated with a square wave as

$$V_{\text{sw}+} = V_{\text{INA}} \cdot \Pi(F_{\text{s}}/4), \tag{2}$$

where V_{INA} is given by (1), and $\Pi(F_{\text{s}}/4)$ is the square wave with frequency $F_{\text{s}}/4 = 407$ Hz.

The second SPDT of the ISL43L220 is used to modulate the $V_{\text{ref}} = 0.4$ V signal that drives the reference terminal of the INA. Hence, the second output of the switch is given by:

$$V_{\text{sw}-} = V_{\text{ref}} \cdot \Pi(F_{\text{s}}/4). \tag{3}$$

The differential signal formed by $V_{\text{sw}+}$ and $V_{\text{sw}-}$ is connected to one of the input channels of the DF chip, so the common mode voltage output, V_{ref} of the INA is subtracted at the DF chip's built-in amplifier, resulting in a final output V_{DF} that is given by:

$$V_{\text{DF}} = G_{\text{DF}}\ (V_{\text{sw}+} - V_{\text{sw}-}) \tag{4}$$

$$= G_{\text{tot}}\ (V_+ - V_-)\ \Pi(F_{\text{s}}/4), \tag{5}$$

where $G_{\text{DF}} = 50$ V/V is the gain of the input amplifiers on the DF chip, and $G_{\text{tot}} = 2500$ V/V is the end-to-end gain of the system.

The sensor was fabricated on a 25 mm x 23 mm x 3 mm FR4 substrate as depicted in Fig. 3. Its measured power consumption is 1.06 mW including the frequency translation front-end. The front-end consumes only 56.6 μW, which represents only 5.3% of the overall system power consumption.

B. External System Design

An external system similar to the one presented in [6] was utilized, as shown in Fig 4. An external power amplifier feeds RF power from a National Instruments USRP B210 software defined radio to the external system antenna at up to +30 dBm. The output of the amplifier is connected to a self-jammer cancellation circuit [7]. The RF signals are downconverted to baseband using a National Instruments USRP B210, and Gnuradio and Matlab scripts that demodulate and deinterleave the packetized data [6].

The original waveform is recovered using coherent digital downconversion simply by multiplying the telemetered signal with a pulse train of 407 Hz. Because the mixer and the sampling clock are both derived from the same crystal oscillator, the harmonics that are produced from the modulation all fold to the same frequency bins, so there is no unwanted phase/amplitude drift. However, there is a residual LO feedthrough at $F_{\text{s}}/4$ that is caused by transient effects (charge injection at the SPDT switches) and modulation of the offset voltage during switching. This can be filtered out with finite impulse response (FIR) low-pass filtering (LPF). Fig. 5 depicts the spectrum of a 20 μV sinusoid at 25 Hz after downconversion and before filtering.

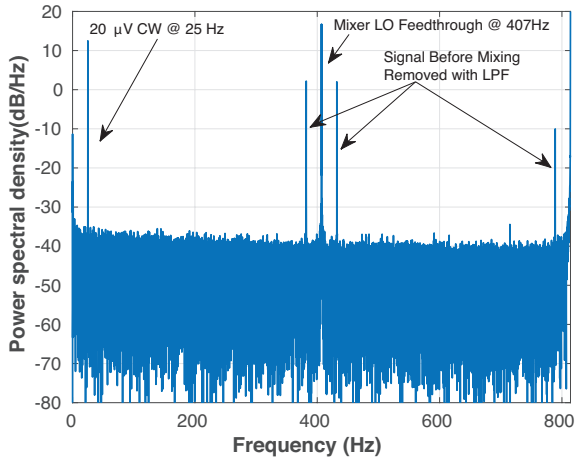

Fig. 5. The spectrum of a 20 μV sinusoid at 25 Hz after coherent digital downconversion but before digital filtering.

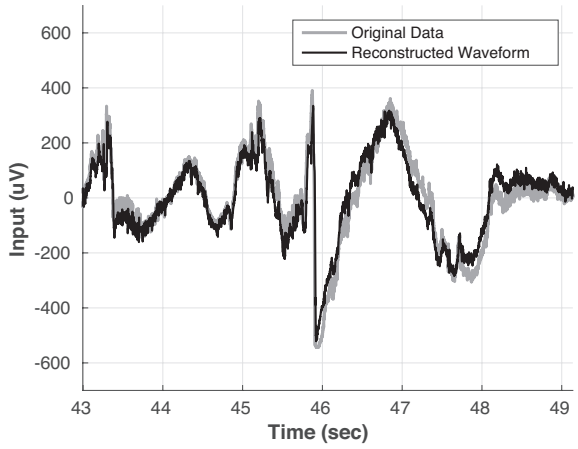

Fig. 6. Comparison of the pre-recorded mouse ECoG signal to the reconstructed waveform after frequency translation.

III. Experimental Results

The frequency response, linearity and noise were characterized in a Faraday cage to mitigate spurious coupling from unwanted sources (e.g. 60 Hz power line noise). An arbitrary waveform generator with its outputs connected to a resistor network that emulates the electrode impedance was utilized in order to feed the sensor with test signals.

Linearity measurements were conducted by feeding the recording device with a sinusoid with amplitude ranging from 1 μV to 400 μV and a frequency of $0.05 - 100$ Hz. The amplitude was estimated with periodogram techniques and measurements verified the HPF cutoff frequency of 0.15 Hz. The gain is flat to within 1 dB over the passband of $0.5 - 100$ Hz and over an amplitude of $1 - 300$ μV.

For the characterization of the system with a realistic biosignal waveform, the arbitrary waveform generator was loaded with 120 s of pre-recorded mouse ECoG data. A 6 s portion of the original vs. reconstructed waveform is depicted in Fig. 6. The total root mean square error (RMSE) is on the order of 41 μV which is essentially identical to the DF chip's original noise performance.

IV. Conclusion

This paper presents an approach for altering the frequency response of existing biosensor infrastructures in order to render them compatible for a broader range of applications. Particularly, an analog front-end was developed that integrates an amplifying stage and a digital switching mixer in order to upconvert a low-frequency bio-signal to a higher spectrum band that is within the passband of the IC of interest. After sampling of the upconverted signal the original signal can be reconstructed by utilizing synchronous demodulation in post-processing. To demonstrate the feasibility of the approach, a pre-recorded ECoG signal from sleeping mice with a frequency range of $0.4 - 100$ Hz was upconverted to an IF of 407 Hz and sampled using an existing IC that was originally developed for capturing neural signals from dragonflies. The original low frequency cutoff of 12 Hz was effectively reduced to 0.5 Hz and the signals were reconstructed with a minimal RMSE of 41 μV.

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References

[1] G. Schalk, "Can Electrocorticography (ECoG) Support Robust and Powerful Brain-Computer Interfaces?" *Frontiers in Neuroengineering*, vol. 3, pp. 9–10, Jan 2010.

[2] R. Muller, H.-P. Le, W. Li, P. Ledochowitsch, S. Gambini, T. Bjorninen, A. Koralek, J. M. Carmena, M. M. Maharbiz, E. Alon, and J. M. Rabaey, "A Minimally Invasive 64-Channel Wireless μECoG Implant," *IEEE J. Solid-State Circuits*, vol. 50, no. 1, pp. 344–359, 2015.

[3] F. Zhang, A. Mishra, A. G. Richardson, and B. Otis, "A Low-Power ECoG/EEG Processing IC With Integrated Multiband Energy Extractor," *IEEE Trans. Circuits Syst. I*, vol. 58, no. 9, pp. 2069–2082, 2011.

[4] A. Mahajan, A. K. Bidhendi, P. T. Wang, C. M. McCrimmon, C. Y. Liu, Z. Nenadic, A. H. Do, and P. Heydari, "A 64-channel ultra-low power bioelectric signal acquisition system for brain-computer interface," in *Biomedical Circuits and Systems Conference (BioCAS), 2015 IEEE*, Oct 2015, pp. 1–4.

[5] S. J. Thomas, R. R. Harrison, A. Leonardo, and M. S. Reynolds, "A Battery-Free Multichannel Digital Neural/EMG Telemetry System for Flying Insects," *IEEE Trans. Biomed. Circuits Syst.*, vol. 6, no. 5, pp. 424–436, Oct 2012.

[6] E. Kampianakis, A. Sharma, J. Arenas, and M. S. Reynolds, "A Dual-Band Wireless Power Transfer and Backscatter Communication Approach for Implantable Neuroprosthetic Devices," in *IEEE International Conference on RFID (RFID), 2017*, pp. 67–72.

[7] A. Boaventura, J. Santos, A. Oliveira, and N. B. Carvalho, "Perfect Isolation: Dealing with Self-Jamming in Passive RFID Systems," *IEEE Microwave Magazine*, vol. 17, no. 11, pp. 20–39, Nov 2016.