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# A Portable Platform for Low Power Electrochemical Impedance Measurements of Biosensors

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## Abstract

Electrochemical impedance spectroscopy is a measurement method that is currently gaining a lot of interest, e.g. as a low-power readout method for miniaturized biosensors. Many biosensor applications require miniaturization and portability of the sensor and the readout platform. For this reason, we present a low-power readout board (ø 40 mm) for impedimetric biosensors that covers high dynamic ranges of 500 Ω to 130 MΩ and can measure over frequencies ranging from 100mHz to 10kHz. The board is powered by a 610mAh Li-Ion coin cell, and the required sensor signal amplification is determined automatically in the software. We measured two dummy sensors covering the whole impedance range and compared the results to a laboratory potentiostat. The board shows a very high measurement accuracy of <0.5% for magnitude and phase of the impedance, high precision, and a battery lifetime of six months and can, therefore, enable many new portable applications.

**Keywords:** Electrochemical impedance spectroscopy, low power, portable potentiostat, biosensing, transimpedance amplifier

### Introduction and Motivation

Miniaturized electrochemical biosensors can enable many new point-of-care applications, such as environmental or health monitoring [1]. A common read-out method for these sensors is electrochemical impedance spectroscopy (EIS), where a small AC excitation voltage is applied to the sensor electrodes, and the resulting current is measured. This way, impedance changes of the sensing layer on top of the electrodes related to the target analyte concentration can be measured [1]. However, high impedances and wide dynamic ranges of these microsensor responses, as well as low power and portability requirements impose many challenges on the readout circuit for these applications [2]. In this work, we, therefore, present a small-sized potentiostatic readout board for impedimetric sensors employing an MSP430 low-energy microcontroller (MCU) and a transimpedance amplifier (TIA). Our printed circuit board (PCB) can measure high impedance and frequency ranges over several days with high accuracy and precision.

### The Measurement Platform

The miniaturized (ø 40 mm) impedimetric measurement platform is depicted in Figure 1 in a picture compared to the size of a 50-cent coin. The block diagram can be seen in Figure 2. The platform consists of a power management, a digital / mixed-signal part, and an analog part, which is the interface to the sensor. The power management, mixed-signal and digital part consists of the low-power MCU MSP430FR2355 by Texas Instruments. The MCU generates an ac excitation signal with an amplitude of 250 mV and automatically sweeps through frequencies from 10 kHz to 100 mHz, which covers the relevant dynamic range of many biosensors. The exact measurement frequencies can be set freely in the software. Additionally, the MCU reads back the excitation voltage and the sensor signal and applies post-processing algorithms to calculate

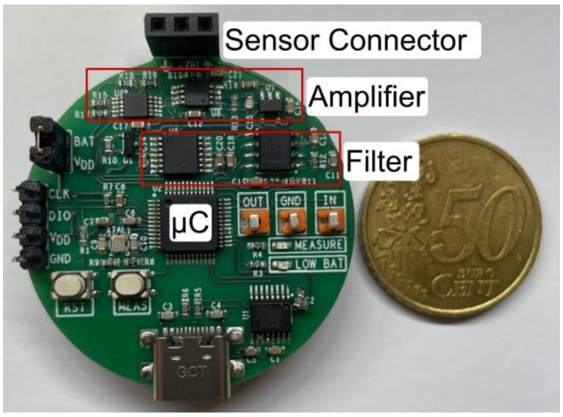

*Figure 1: Picture of the miniaturized PCB for impedance measurements.*

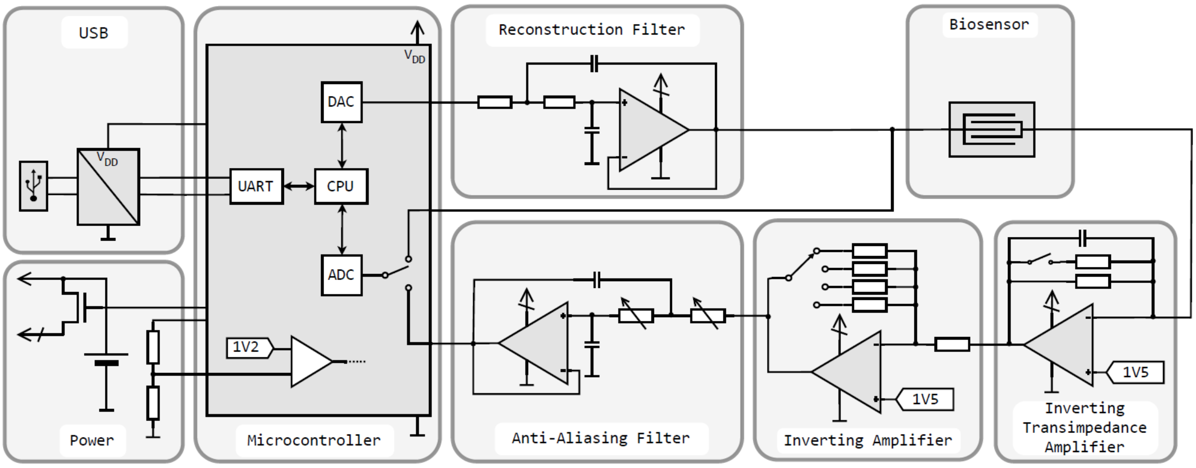

*Figure 2: Schematic block diagram of the circuit components, including power management with a MOSFET for disconnection of the power, when the board is not used and serial communication for data readout. A reconstruction filter smoothens the ac potential signal coming from the DAC of the MCU. An inverting TIA converts the current flowing through the sensor to a potential. Both, TIA and an inverting amplifier have various gains, which can be selected via software by a digital MUX. Before reading the signal back with a 12-bit DAC an antialiasing filter is implemented.*

the phase shift and amplitude of the sensor signal using the Goertzel algorithm. This digital signal processing technique is more efficient compared to Fast-Fourier-Transformation (FFT) when only a few spectral values are of interest. As we are always measuring at one fixed frequency, the Goertzel algorithm imposes less computing effort and, therefore, leads to power savings. Furthermore, the software automatically adjusts the amplifier gains during measurements for optimum accuracy depending on the sensor impedance. The analog part of the platform consists of a second-order Butterworth filter with a cutoff frequency of 20 kHz. The filter smooths out the steps in the excitation voltage, which result from the DAC of the MCU. After the filter, the sinusoidal voltage reaches the connected sensor. The resulting current response of the sensor is amplified and converted back to a voltage signal by the TIA. By using a 2:1 multiplexer, the resistance of the feedback loop can be switched, resulting in a 68dB or a 120dB amplification. To cover the high dynamic range of biosensors, an additional inverted amplifier with a 4:1 MUX is added after the TIA enabling a total of eight gain configurations (67 dB, 80 dB, 93 dB, 107 dB, 120 dB, 133 dB, 146 dB, 160 dB) and an impedance range from 500 Ω to 55 MΩ.

Before the internal ADC of the MSP430 converts the amplified signal it gets filtered by an anti-aliasing filter. A CR2450 coin battery provides the circuit with a nominal voltage of 3 V. The supply voltage is monitored using a voltage divider and the internal reference voltage of the MSP430. If it drops below 2.7 V, the MCU informs the user via a built-in LED. This ensures that the supply voltage doesn't drop below the 2.7 V limit of the analog components. To further minimize the current consumption, all analog components, i.e., OpAmps and filters, are switched off from the 3 V power supply in standby mode by the MCU using a PNP transistor, as depicted in Figure 2, leading to a minimum current flow of <20 µA. The coin cell has a capacity of 610 mAh. A full frequency measurement takes 70 seconds with an average current flow of 2.3 mA. 1 mA of this current results from the analog components, and the remaining current can be attributed to the MCU. Assuming one complete EIS spectral measurement every 15 min, our platform can measure for approximately six months without having to change the battery.

*Table 1: Amplifier gains and corresponding impedance ranges*

| Gain | 67 dB | 80 dB | 93 dB | 107 dB |
|---|---|---|---|---|
| \|Z\| (kΩ) | 0.5 – 2.5 | 2.5 – 11.6 | 11.6 – 57.9 | 57.9 – 263 |
| Gain | 120 dB | 133 dB | 146 dB | 160 dB |
| \|Z\| (MΩ) | 0.26 – 1.1 | 1.1 – 5.3 | 5.3 – 26.3 | 26.3 – 130 |

## Results and Discussion

The performance of the miniaturized EIS platform was characterized using two dummy sensors to cover the full impedance range with magnitudes |Z| from 500 Ω to 75 MΩ and phase angles φ from 0° to -85°. The equivalent circuits of these sensors are depicted in Figure 3a) for low and Figure 3b) for high impedance ranges with the corresponding capacitance and resistance values.

For the first validation of the platform, we measured over six decades at one frequency per decade, i.e., at 10 kHz, 1 kHz, 100 Hz, 10 Hz, 1 Hz, and 100 mHz. Before measurement, the platform was calibrated in order to eliminate the influence of in-accuracies in the analog components, parasitic capacitances, and signal processing on the measured phase shift. Therefore, instead of a sensor, eight different resistors were connected to the board in order to cover every amplifier gain. As a pure resistor should show a phase shift of 0°, the measured phase for each resistor is used as the calibration value and substracted during post-processing of the data. The calibration value strongly depends on the frequency and slightly on the amplifier gain. For example, at 100 mHz, the calibration values are 3.60° for 68 dB and 3.61° for 160 dB, whereas at 10 kHz, these values are 61.04 and 49.16°, respectively. The calibration step must only be done once for each measurement frequency and amplification after fabrication of the board and before the first measurements.

Figure 4a) shows the measurement results and standard deviations of our PCB (o) compared to the measurement results of a professional Gamry Interface 1010E potentiostat (x) and the theoretical curves of the dummy sensors (--). It can be seen that the measurements of our platform match very well with those of the Gamry potentiostat. Especially in the high impedance mode, the standard deviations of magnitude and phase are comparable to those of the Gamry Interface 1010E, and the differences are in the majority of the cases below 0.5%. Because the standard deviation is so low, the difference is almost not visible in the graphs. The deviations from the simulated values for the high-impedance sensor dummy occur for both the Gamry potentiostat and our PCB and stem from the tolerances of the analog resistors and capacitors and parasitic capacitances in the dummy cell.

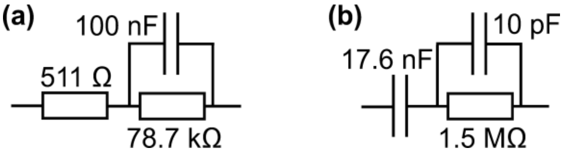

*Figure 3: Equivalent Circuits of the measured dummy sensors for a) low impedance ranges with R1= 511 Ω, R2=78.7 kΩ, and C1=100 nF and b) high impedance ranges with C2= 17.6 pF, C3=10 pF and R3= 1.5 MΩ.*

Figure 4b) shows the response of our sodium ($Na^+$) sensor to $Na^+$ concentrations of 1 mM, 25 mM, and 100 mM measured with the miniaturized readout board. The sensor impedance decreases with increasing $Na^+$ concentration and is for all measurements in the impedance range of our platform. Therefore, we can prove the applicability of our miniaturized, low-power readout circuit to low- and high-impedance biosensors with wide dynamic ranges. Table 2 compares the performance of this work to commercial solutions and other research articles. It should be mentioned that the AD5933 by Analog Devices is an integrated circuit that needs to be equipped with external components, e.g. for data readout and power supply. Thus, its size is not comparable to the other solutions. The table shows that our PCB outperforms all other

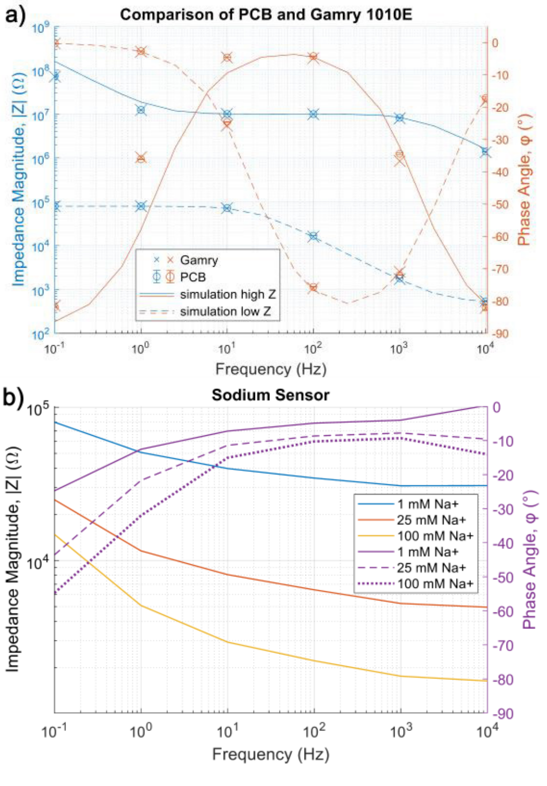

*Figure 4: Simulation and measurement results of a) a low impedance (- -) Randles Circuit dummy (R1=500 Ω, R2=80 kΩ, and C=100 µF) and a high impedance (–) sensor dummy that models our potassium sensor [3] (R3=10 MΩ, C2=10 µF, and C3=10 pF) using our measurement platform (o) and a Gamry Interface 1010E (x) potentiostat. (b) Measurement results of an impedimetric sodium biosensor.*

*Table 2: Comparison of this work to other miniaturized EIS readout circuits*

| | **$\|Z\|$ Range** | **$f$ Range** | **Accuracy** |
|---|---|---|---|
| This work | 500Ω – 130MΩ | 0.1Hz – 10kHz | <0.5%* |
| EmStat Go | 40Ω – 40MΩ | 16mHz – 200kHz | <1%** |
| AD5933 | 1kΩ – 10MΩ | 1kHz – 100khz | 0.5%*** |
| [4] | ~330kΩ | 10Hz – 100kHz | 0.6% |
| [5] | ~100kΩ – 1.1MΩ | 500kHz | - |
| [6] | ~25kΩ –160kΩ | 10Hz – 200kHz | ~1.5%(R)<br>~ 2%(C) |
| | **Power** | **Size** | **Battery Lifetime** |
| This work | 6.9mW | Ø40mm | 6 months |
| EmStat Go | ~30mW | 118x69x33 mm$^3$ | 6h cont. meas. |
| AD5933 | 33mW | 8x6x2mm$^3$ IC | 28 days |
| [4] | - | 21.6x20.3x 6mm$^3$ | USB powered |
| [5] | - | <80cm$^2$ | - |
| [6] | - | ~115x70mm$^2$ | - |

*(>1 MΩ / 10 kHz)<1.1%; **(>100 MΩ / 0.1 Hz) & (>10 MΩ / 100 Hz) <10%; ***lower with wide frequency sweeping [1]

work regarding the measurable impedance, which ranges from 500 Ω to 130 MΩ.
Only the EmStat Go from PalmSens can achieve similar impedance ranges, however, at slightly lower accuracy, bigger size, and higher power consumption and, therefore, lower battery lifetime.

## Conclusion and Outlook

In this work, we presented a small-sized impedimetric readout board for low-power EIS measurements of biosensors. The board can measure impedances from 500 Ω to 130 MΩ and thereby outperforms other miniaturized solutions. Impedance data can be recorded from 10 kHz to 100 mHz with high precision and very good accuracy of 0.5% and lower. The power consumption is 6.9 mW during measurements and below 60 µW in standby mode. The next steps will be to include a wireless data readout option using Bluetooth or NFC for power savings. However, with a battery lifetime of six months and by covering high impedance ranges and wide dynamic ranges, our portable board can already be used for most biosensors and enables new application fields.

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