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# A Stand-Alone Portable Potentiostat With Parallel Channels for Smart Electrochemical Analyses

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Marco Giannetto<sup>1</sup>, Maria Careri<sup>1</sup>, and Ilaria De Munari<sup>1</sup>, Senior Member, IEEE

**Abstract**—In recent years portable potentiostats have increased in popularity allowing to perform electrochemical analyses outside the laboratories, moving them to home or point of care (PoC) environments. In this context, the idea of performing multiple acquisitions at the same time is particularly attractive to deal with replicates or with simultaneous multiple quantitative assessments of different analytes, shortening the time required for the analyses and/or increasing data reliability. Multiple parallel channels on a compact and wireless device can maximize the overall system usability. In this article, a multichannel Wi-Fi-based portable potentiostat is described. The device features four independent channels, which can be conditioned with different voltages. The device was designed to minimize the component count and the power consumption, obtaining 23.5 mW per channel. The system was electrically characterized, obtaining an excellent linearity ( $R^2 = 0.9999$ ), a maximum channel-to-channel mismatch of 1% when the maximum current range is selected, and a negligible crosstalk among the channels. Moreover, the multichannel potentiostat was tested in the real case scenario of a semiquantitative evaluation of the anti-tissue transglutaminase target antibodies in celiac disease. Two replicates of IgG and IgA were simultaneously acquired, showing a good capability of separating the positive and the negative samples, with a reduction of the acquisition time of 76% with respect to a single channel solution.

**Index Terms**—Chemical sensors, electrochemical devices, multichannel, potentiostat, Wi-Fi sensors, wireless sensors.

## I. INTRODUCTION

THERE are several contexts in which portable and reliable devices for electrochemical analysis are exploited to provide easy-to-use, rapid, highly selective, and quantitative information on target analytes. Possible areas of application are the detection of biological contaminants in food [1], [2], [3], environmental monitoring [4], [5], [6], and clinical diagnostics [7], [8], [9], [10], [11]. Many of them are based on the use of a potentiostat, i.e., an electronic hardware elaborating the electrical signal generated from an electrochemical reaction.

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With reference to this, the main challenge tackled by the researchers is how to implement a compact portable device with cheap components and reduced power consumption without compromising the quality of the tests. It is worth noting that the overall portability of the device strongly depends also on the architecture of the system. In particular, for the development of a completely stand-alone instrument, it is preferable to avoid the use of additional devices for data elaboration and storage. Recently, several portable solutions have been commercialized [12], [13], [14]; however, for their operation, they rely on supplementary devices such as personal computers (PCs) or tablets/smartphones connected via USB and/or Bluetooth. Also, some interesting solutions are available in the literature [4], [15], [16], [17] with some advanced features like the connection to a cloud service, but all require the connection to an external device, limiting the portability and the overall usability.

Moreover, these solutions, as most of the available portable potentiostats, are designed for single analyte detection and are equipped with a single analog front end (AFE) to read and control the electrochemical sensor [4], [15], [16], [17], [18]. Many of these electrochemical acquisitions are time-expensive, with a single experiment easily lasting several minutes. Moreover, replicates are needed to increase the result precision to obtain statistically relevant results. Furthermore, sensor redundancy allows the implementation of advanced fault-tolerant algorithms and data-averaging techniques that can improve system robustness, efficiency, and accuracy [19]. Bioanalytical, pharmaceutical, and clinical applications require simultaneous electrochemical detection that cannot be obtained using a single-channel instrument. For example, parallel measurements of multiple cancer markers are useful for increasing the efficiency of cancer diagnostics and therapy monitoring [20], as well as for considering possible correlations between compounds [9]. Therefore, a potentiostat with multiple acquisition channels would significantly increase the efficiency of the experiments, implementing their parallel execution.

In this article, a multi-channel portable potentiostat with three electrodes for each channel suitable for both quantitative and semiquantitative electrochemical analysis, usable in a large variety of out-of-lab applications, is presented. The device features four completely independent and parallel channels, capable of simultaneously managing, with different voltages, four commercial screen-printed electrodes, each of which

83 composes a three-electrode electrochemical cell. Thanks to  
84 a six-channel analog-to-digital converter (ADC), the current  
85 measurement is carried out at the same time for all the  
86 channels. The device can be battery-powered and connected,  
87 through a Wi-Fi protocol, to a cloud analytics for data storage  
88 and elaboration. This allows a simple connection to wide-  
89 spread Wi-Fi networks for sharing data with different users  
90 without the need for external devices. Moreover, the cloud  
91 service can be interfaced with a web application for the remote  
92 control of the device. The aim of this article is the description  
93 and the characterization of the hardware device. A detailed  
94 description of a possible web application can be found in [18].

95 The proposed four-channel solution features a scalable  
96 architecture that can be easily expanded to meet user needs.

97 As a case study, an electrochemical immunosensor for  
98 the detection of anti-tissue transglutaminase antibodies  
99 (anti-tTG) was considered [21]. These antibodies represent  
100 reliable biomarkers for the diagnosis of celiac disease, for  
101 which both the IgA and IgM isotype of anti-tTG must be  
102 quantified. Indeed, it is common for patients affected by celiac  
103 disease to display a selective IgA deficiency which, in the  
104 absence of IgG quantification, would lead to a false negative  
105 outcome of the diagnosis. Therefore, the use of a multi-channel  
106 acquisition device is of paramount importance to allow both  
107 the simultaneous determination of the IgA and IgG isotypes  
108 of anti-tTG antibodies and the acquisition of at least two  
109 replicates during a single measurement cycle.

110 This article is organized as follows. In Section II, the related  
111 works are presented; in Section III, the architecture of the  
112 system designed is described; in Section IV, the experiments  
113 performed to characterize the device are reported and dis-  
114 cussed; and in Section V, conclusions are drawn.

## 115 II. RELATED WORKS

116 The key task of a potentiostat device is to measure the  
117 current flowing into an electrochemical cell due to a chemical  
118 reaction related to the target analyte. The electrochemical  
119 sensing systems require two [22] or three electrodes [7].  
120 A three-electrode electrochemical cell includes a working  
121 (WE), a reference (RE), and a counter electrode (CE), while  
122 in the two-electrode cell, CE and RE are unified. It is worth  
123 noting that, regardless of the number of electrodes in the cell,  
124 in order to carry out different types of analyses simultaneously  
125 in a multichannel system, it is important to keep independent  
126 the electrodes of the different cells under test. In this way,  
127 it will be possible to condition the different cells with different  
128 voltages according to the analysis to be carried out.

129 In [23], a multichannel potentiostat is presented. Thanks  
130 to four ADCs with single-ended inputs, the device is able to  
131 read four independent two-electrode cells simultaneously. The  
132 core of the system is a Raspberry PI Computer. Through I2C  
133 communication, up to eight potentiostats can be connected.  
134 It features a 12-bit DAC for generating the control voltages.  
135 Moreover, traditional potentiostats rely on a transimpedance  
136 amplifier (TIA) for current measurements. In this case, the  
137 authors exploited a shunt resistor, lowering the overall accu-  
138 racy of the device. The maximum current range is  $\pm 1.5 \mu\text{A}$ .

139 It is worth noting that, despite the high number of channels  
140 available in this solution considering the eight boards stacked  
141 together, the reading of the channels is not simultaneous, but  
142 it is serialized through the communication protocol chosen  
143 (e.g., I2C). Moreover, notwithstanding the two-electrode sys-  
144 tem results in a simplified circuitry [22], the three-electrode  
145 solution is more advisable since the current deriving from  
146 the chemical reaction flows through the CE instead of the  
147 reference one, avoiding the change in the potential of the WE  
148 [7]. For this reason, the three-electrode solution is the most  
149 widely adopted.

150 In [24] and [25], a compact architecture with 128 chan-  
151 nels is reported, where, to reduce the device dimensions,  
152 the single potentiostat channel contains only the essential  
153 components while the rest of the hardware is time-shared  
154 across channels and driven by a single  $\mu\text{C}$ . To increase the  
155 number of channels, a time division multiplexing technique is  
156 exploited. The measurement results are sent to a PC through  
157 a UART port, limiting the portability of the device. Moreover,  
158 the CE and RE are shared between the parallel channels  
159 limiting the possibility of conditioning the electrochemical  
160 cells with different voltages. The same limitation applies to  
161 the 8-channel potentiostat named “octopoti” presented in [24].  
162 This device cannot be considered a portable solution, as it  
163 requires a suitable external data acquisition system to read the  
164 outputs. Similarly, in [26], a 16-channel potentiostat requiring  
165 a multifunction PC with a data acquisition card to be powered  
166 and controlled is presented.

167 In [27], a six-channel wireless potentiostat is described.  
168 In this case, the solution turns out to be quite portable even if  
169 the choice of a low-range communication protocol such as  
170 Bluetooth does not eliminate the need to have an external  
171 device (e.g., PC, tablet, or smartphone) nearby, which acts  
172 as a gateway for the acquired data. Moreover, the data elab-  
173 oration and visualization require the LabVIEW data acquisition  
174 system running on an external device.

175 In [28], a three semi-parallel channels solution is reported.  
176 It requires a USB or Bluetooth connection to operate and  
177 relies on a smartphone or desktop APP for data processing and  
178 visualization. It features a 12-bit DAC and an ADC embedded  
179 into the ESP32 Microcontroller. The ADC is in common for  
180 the three channels. For this reason, although each of the three  
181 channels has dedicated electronics, the data sampling process  
182 is performed sequentially, alternating the reading of each ADC  
183 pin.

184 Also, some commercial solutions are available. MultiPalm-  
185 Sens4 and MultiEmStat4 from PalmSens [12] are benchtop  
186 instruments with up to 10 and 12 channels, respectively. The  
187 portability of these devices is very low, as they commu-  
188 nicate over a USB cable and are powered with an exter-  
189 nal 12V ac/dc adapter or the same USB port. A more  
190 portable device is the wireless dual-channel potentiostat  
191 Sensit-BT [12]. Another commercial solution is the STAT8000  
192 from Metrohm [15], a portable eight-channel potentiostat.  
193 Both are battery-powered and feature a Bluetooth inter-  
194 face. The STAT8000 requires a PC for data elaboration and  
195 visualization.



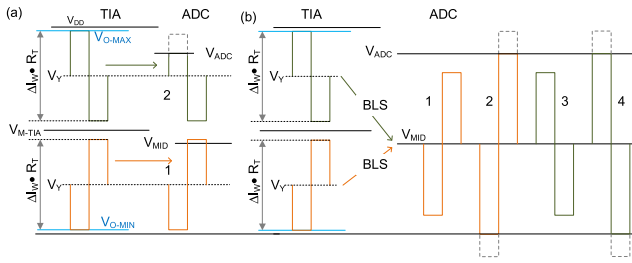


Fig. 3. Voltage waveforms at the TIA output and ADC input (zero-mean signal case). (a) Without BLS. Case 1:  $V_Y < V_{M-TIA}$ ; Case 2:  $V_Y > V_{M-TIA}$ . (b) With BLS between ADC and DAC. Cases 1 and 2:  $V_Y < V_{M-TIA}$ ; cases 3 and 4:  $V_Y > V_{M-TIA}$ . In cases 1 and 3 the reading range is TIA limited, and in 2 and 4 is ADC limited.

276 than 45

$$277 \quad \text{GBW}_{A2} > \frac{C_{RW} R_H}{2\pi (C_T R_T)^2}. \quad (3)$$

278 For the present design, the LTC6082 (quad opamp) was  
279 used. It exhibits a typical GBW of 3.5 MHz with a current  
280 consumption of 330  $\mu\text{A}$  and can be compensated over the  
281 range of feedback resistors.

282 In the design of the readout channel of the current flowing  
283 into the WE terminal ( $I_W$ ), the signal swing at the TIA output  
284 should be carefully considered. Indeed, the reading range of  
285 the potentiostat cannot exceed the boundaries set by the output  
286 range of opamp A2, at a given value of feedback resistance  $R_T$

$$287 \quad V_{\text{OMIN}} - V_Y \leq I_W \cdot R_T \leq V_{\text{OMAX}} - V_Y \quad (4)$$

288 where  $V_{\text{OMAX}}$  and  $V_{\text{OMIN}}$  are the maximum and minimum  
289 output voltage of the TIA opamp, corresponding, for the  
290 LTC6082, to 3.3 V supply minus 30 and 30 mV, respectively.

291 A further constraint is introduced by the conversion range  
292 of the ADC, assumed from 0 to  $V_{\text{ADC}}$

$$293 \quad 0 < V_Y + I_W \cdot R_T \leq V_{\text{ADC}}. \quad (5)$$

294 Restricting the analysis to the case of a zero-mean cur-  
295 rent signal, the reading range is limited by the lower  
296 bound of the TIA-opamp (output) voltage in the case  
297 where  $V_Y$  is lower than the midpoint of the opamp output  
298 range, i.e.,  $V_{M-TIA} = 0.5 \cdot (V_{\text{OMAX}} - V_{\text{OMIN}})$ , as shown in  
299 Fig. 3(a). With  $V_Y$  higher than  $V_{M-TIA}$ , the reading range is  
300 bounded by the ADC conversion range, provided that  $V_{\text{ADC}}$  is  
301 lower than  $V_{\text{OMAX}}$ .

302 A further relevant design aspect is the signal-to-  
303 quantization-noise ratio (SQNR) of the readout channel. The  
304 maximization of the SQNR requires maximizing the signal  
305 swing at the ADC input as well. Still assuming a zero-mean  
306  $I_W(t)$  signal, the SQNR is estimated by the following formula:

$$307 \quad \text{SQNR} = 10 \cdot \log \left[ \frac{3 \cdot 2^{2 \cdot N_B} \cdot (\alpha_w \cdot \Delta I_W \cdot R_T)^2}{V_{\text{ADC}}^2} \right] \quad (6)$$

308 where  $N_B$  is the nominal resolution of the ADC,  $\Delta I_W$  is the  
309 peak-to-peak swing of the cell current, and  $\alpha_w$  is equal to one  
310 for a square-wave input current or to  $1/\sqrt{3}$  for a triangular  
311 waveform.

312 As shown in Fig. 3(b), the mismatch between  $V_Y$ ,  $V_{M-TIA}$ ,  
313 and the midpoint of the ADC range prohibits achieving the

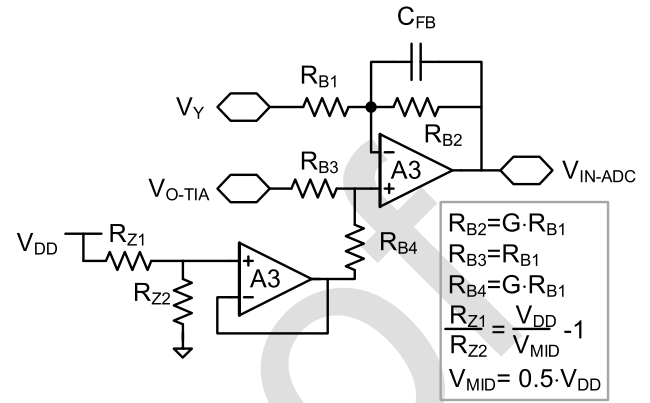


Fig. 4. BLS interfacing the TIA to the ADC.

314 maximum SQNR in (6) with  $\Delta I_W \cdot R_T = V_{\text{ADC}}$ . The problem  
315 has been addressed in the proposed design with a dedicated  
316 buffer-level shifter (BLS) placed between the TIA and the  
317 ADC. As shown in the schematic of Fig. 4, this buffer requires  
318 one additional opamp per channel (A3), whereas the opamp  
319 A4 is used to buffer a reference voltage ( $V_{\text{MID}}$ ) derived from  
320 the supply and shared over all the channels. Furthermore, since  
321 the bandwidth of A3 should be compatible with the settling  
322 time of the  $I_W(t)$  signal, and thus, smaller than the TIA-opamp  
323 GBW, the LTC6079 featuring a current consumption of 54  $\mu\text{A}$   
324 was used.

325 With the resistance settings as in Fig. 4, the buffer output  
326 voltage, corresponding to the ADC input signal, is centered at  
327 the  $V_{\text{MID}}$  voltage value and amplified by  $G$

$$328 \quad V_{\text{IN-ADC}} = G \cdot (V_{\text{O-TIA}} - V_Y) + V_{\text{MID}} = G \cdot I_W \cdot R_T. \quad (7)$$

329 The reference voltage  $V_{\text{MID}}$  must be equal to the midpoint  
330 of the ADC range, i.e., 1.25 V in our design.

331 The reading range  $\Delta I_W$  at the selected TIA gain  $R_T$  can  
332 be either limited by the TIA or the ADC, depending on the  
333 value of  $G$  and  $V_Y$ . If the ADC saturation occurs before the  
334 TIA-opamp saturation, the reading range is ADC-limited, as in  
335 cases 2 and 4 in Fig. 3(b). Depending on the setting of  $V_Y$  with  
336 respect to  $V_{M-TIA}$ , the condition for ADC-limited range is

$$337 \quad V_Y > V_{M-TIA} \rightarrow G \cdot (V_{\text{OMAX}} - V_Y) \geq \frac{V_{\text{ADC}}}{2} \quad (8)$$

$$338 \quad V_Y > V_{M-TIA} \rightarrow G \cdot (V_Y - V_{\text{OMIN}}) \geq \frac{V_{\text{ADC}}}{2}. \quad (9)$$

339 At the measurement startup, the  $\mu\text{C}$  will set the TIA gain at  
340 the maximum value. During the measurement, the gain will be  
341 decreased as soon as the ADC output approaches the upper or  
342 the lower saturation condition, corresponding to conditions (8)  
343 and (9), respectively.

344 If neither condition (8) nor (9) is fulfilled, the reading range  
345 is TIA-limited. In this case, the  $\mu\text{C}$  will decrease the gain  
346 during the measurement at the occurrence of the following  
347 condition for the ADC output code  $D_{\text{O-ADC}}$ :

$$348 \quad D_{\text{O-ADC}} = 2^{N_B-1} + k_O \cdot (2^{N_B} - 1) \cdot \frac{G \cdot \Delta V_O}{V_{\text{ADC}}} \quad (10)$$

## CURRENT CONS. BREAKDOWN

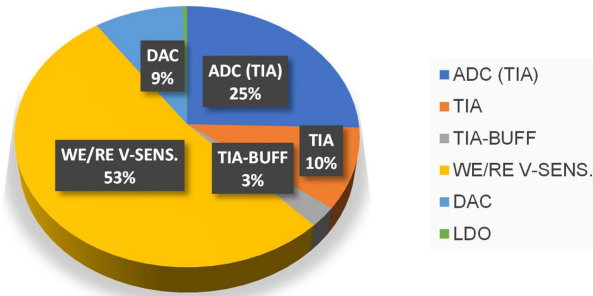


Fig. 5. Current consumption breakdown of the proposed four channels potentiostat. The consumption of the readout circuits for the WE and RE voltage (WE/RE V-SENS) includes the voltage buffer and the related ADC.

349 where  $\Delta V_O$  and  $k_O$  depends on the position of the  $V_Y$  level  
350 in the TIA output range, as in cases 1 and 3 in Fig. 3(b)

$$351 \quad V_Y > V_{M-TIA} \rightarrow \Delta V_O = (V_{OMAX} - V_Y), \quad k_O = +1 \quad (11)$$

$$352 \quad V_Y \leq V_{M-TIA} \rightarrow \Delta V_O = (V_Y - V_{OMIN}), \quad k_O = -1. \quad (12)$$

353 In the present design, we used the LTC1408 device inte-  
354 grating six A/D conversion channels. The device was selected  
355 for the simultaneous sampling capability, the low power  
356 consumption, and the 14-bit nominal resolution, which is  
357 suitable for the present potentiostat [36]. The  $\mu C$  provides  
358 the sampling and the clock signals to the ADC through  
359 a three-wire serial interface. The same interface allows the  
360 acquisition of the output data stream. It is worth noting that the  
361 outputs of the four channels have been connected to the same  
362 ADC. Hence, the sampling of the outputs of all the channels  
363 happens simultaneously. The results are then transmitted to  
364 the microcontroller with a serial interface, but this does not  
365 affect the timing of the measurements since the transmission  
366 is completed between two subsequent measurements.

367 Additionally, the potentiostat allows the acquisition of the  
368 voltages at the WE and RE pins for each cell. These readout  
369 channels (not shown in Fig. 1) add a diagnostic capability to  
370 the system since the virtual short circuit at the input of A1  
371 and A2 opamp can be continuously monitored during the cell  
372 current acquisition cycle. Furthermore, they allow measuring  
373 the voltage across WE and RE terminals, which is required in  
374 the DPV measurement procedure [29], [30].

375 The current consumption of the AFE (excluding the  $\mu C$ )  
376 is 3.3 mA per channel. A regulated 3.3 V is provided by an  
377 onboard low-dropout (LDO) regulator from either the battery  
378 or the 5 V USB supply. The consumption breakdown is  
379 obtained with circuit simulation (using the SPICE models pro-  
380 vided by the manufacturers) and from the current consumption  
381 values reported in the component data sheets [36]. The results  
382 are shown in the graph in Fig. 5.

383 With regard to the digital part of the system, a detailed  
384 analysis of the power consumption was discussed in [29].  
385 Here, considering the worst case of continuous operation with  
386 the  $\mu C$  always active, without low-power mode management  
387 between two readings, a current of 3.8 mA per channel should

TABLE I  
CURRENT RANGE, RESOLUTION, AND INPUT NOISE

$R_T$ (k $\Omega$ )	$\pm \Delta I_w/2$ ( $\mu A$ )	Resolution (nA)	$I_{IN-N}$ rms (nA)
10	60	7.6	1.72
100	6.0	0.76	0.173
1000	0.60	0.076	0.017

TABLE II  
CHANNEL MISMATCH RESULTS

$R_{RW}$ ( $\Omega$ )	$\epsilon_{A,max}$ (mV)	$\epsilon_s$ (%)	$\epsilon_i$ (%)	$r^2_{min}$
10k	9.2	1	0.5	0.9999
100k	5	0.9	0.4	0.9999
1M	3.1	0.07	0.3	0.9999

388 be considered. Hence, the total current required for both the  
389 analog and digital parts is 7.1 mA.

390 The reading range, the resolution, and the simulated  
391 input-current noise ( $I_{IN-N}$ ), root mean square (rms), for the  
392 minimum, i.e., 10 mV/mA, medium, i.e., 100 mV/mA, and  
393 maximum, i.e., 1 V/mA, TIA gain are reported in Table I.

## IV. RESULTS AND DISCUSSION

### A. System Characterization

394 Several tests were carried out to evaluate the performance  
395 of the proposed multichannel portable potentiostat. The single-  
396 channel performance was evaluated to assess both the linearity  
397 of the response and the output difference among channels with  
398 the same voltage signals and load (i.e., channel mismatch). For  
399 that purpose, a dummy cell was connected to the input of each  
400 channel. A schematic of the dummy cell is reported in the red  
401 box in Fig. 2.

402 Three different values of  $R_{RW}$  were taken into account (i.e.,  
403 10 k $\Omega$ , 100 k $\Omega$ , and 1 M $\Omega$ ). This allows considering differ-  
404 ent input current ranges (i.e., maximum currents of 62  $\mu A$ ,  
405 6.2  $\mu A$ , and 620 nA, respectively). The considered ranges are  
406 those usually needed in some common applications. For each  
407  $R_{RW}$  value, each DAC channel was configured to provide a  
408 voltage bias ranging from  $-0.6$  to  $0.6$  V, and the corresponding  
409  $V_{out}$ , i.e., the digital ADC output expressed in volts, was  
410 acquired. For each channel combination, an absolute error  
411  $\epsilon_A$  was evaluated, corresponding to the difference between  
412 the measured  $V_{out}$  of two considered channels. In Fig. 6, the  
413 measured absolute error between channels is shown, with an  
414  $R_{RW}$  value of 1 M $\Omega$ .

415 This analysis was repeated for each possible value of  $R_{RW}$ ,  
416 evaluating the maximum error in the measurement of the  
417 output voltage (Table II).

418 Moreover, a relative error in the slope and the intercept  
419 of the fit curve was evaluated as a further parameter of the  
420 mismatch between different channels. The relative error in the  
421 slope  $\epsilon_s$  was defined as

$$422 \quad \epsilon_s = \frac{m_{max} - m_{min}}{m_{avg}} \quad (13) \quad 424$$

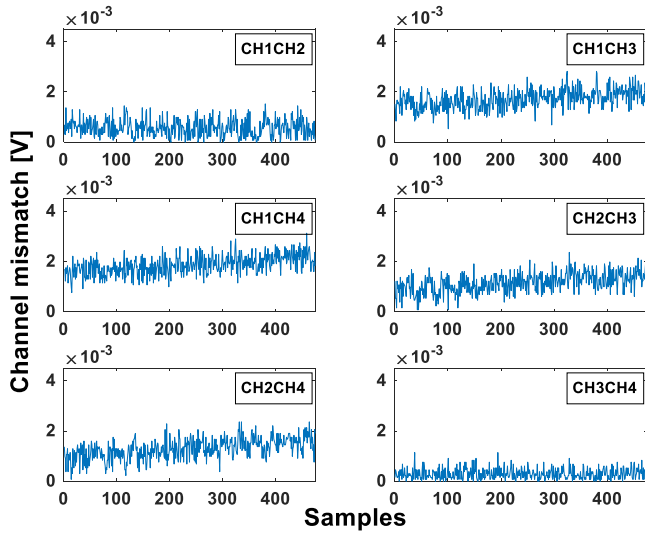


Fig. 6. Absolute error, corresponding to the difference between the measured  $V_{out}$  of the two considered channels, when the  $1\text{ M}\Omega$  resistor was selected into the dummy cell.

where  $m_{max}$  is the maximum value of the fit curve slope, among the four-channel data, given a dummy cell resistance value,  $m_{min}$  is the minimum value, and  $m_{avg}$  is the averaged value of the four-channel slopes.

Similarly, the relative error in the intercept  $\varepsilon_i$  was defined as

$$\varepsilon_i = \frac{q_{max} - q_{min}}{q_{avg}} \quad (14)$$

where  $q_{max}$  is the maximum value of the fit curve intercept, among the four-channel data, given a dummy cell resistance value,  $q_{min}$  is the minimum value, and  $q_{avg}$  is the averaged value of the four-channel intercepts. The overall results are reported in Table II.

Finally, the linearity in the  $V_{out}$  measurements was evaluated. An example of such an evaluation is reported in Fig. 7, when an  $R_{RW}$  value of  $1\text{ M}\Omega$  is selected.

These reported values show that each channel of the proposed multichannel potentiostat exhibits excellent linearity regardless of the resistance chosen on the dummy cell and then the current sensed by the device. The maximum error due to the channel design difference (e.g., components value and PCB design) is 1%. This is compatible with the tolerance chosen for the gain and feedback resistor.

Once the maximum mismatch among the channels was assessed, tests were carried out to quantify the channel-to-channel crosstalk. As the channels showed negligible mismatch, only channels 1 and 2 were considered for these tests. Two identical dummy cells (Fig. 2) were connected to the channels and configured with a resistance value of  $10\text{ k}\Omega$ . The onboard DAC was used to generate a sinewave signal with  $0.3\text{ V}$  amplitude and  $0.5\text{ V}$  offset at different frequencies (i.e., 1, 5, 10, 25, 50, and 100 Hz). This signal was applied to the  $V_X$  pin of channel 1 in Fig. 1 and thus replicated to  $V_{RE}$  (due to the virtual short circuit at the A1 input). The channel 1 TIA bias voltage  $V_Y$  was set to  $0.5\text{ V}$ . The  $V_X$  and  $V_Y$  inputs of

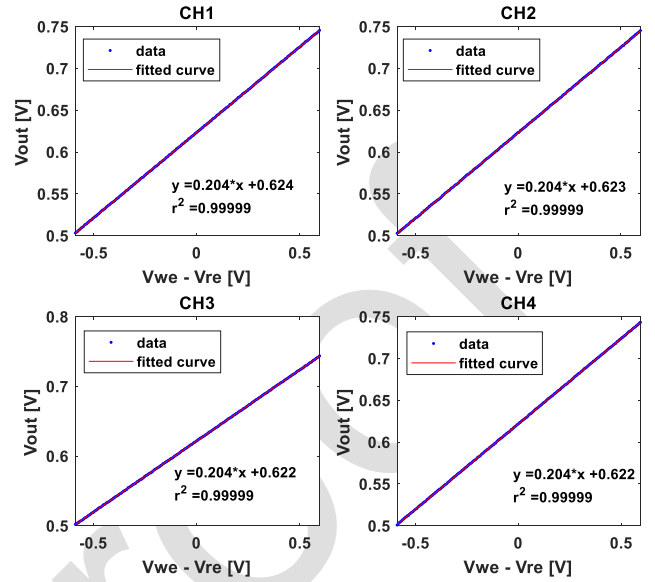


Fig. 7. Linearity in the four-channel measurements when a  $1\text{ M}\Omega$  resistor is selected in the dummy cell.

TABLE III  
CHANNEL CROSSTALK RESULTS

Input signal frequency (Hz)	FFT amplitude ratio (x1000)
1	2.6
5	4.2
10	5.1
25	4.6
50	4.5
100	2.4

channel 2 were conditioned with the same constant voltage of  $0.5\text{ V}$ . The output of each channel was sampled at a frequency of  $1\text{ kHz}$ . In Fig. 8, a period of the conditioning signals in the case of a signal frequency of  $1\text{ Hz}$  and the corresponding outputs for channels 1 and 2 are shown.

For each frequency considered, the fast Fourier transform (FFT) of both outputs was calculated, and the ratio of the amplitudes at the frequency considered was assessed. In Fig. 9, an example of this evaluation was shown considering a signal frequency of  $50\text{ Hz}$ .

The ratio between the FFT amplitudes of the two channels  $V_{out}$  at different input signal frequencies is summarized in Table III.

As can be seen from the results, the conditioning signal in a given channel does not influence the behavior of the circuit in another channel, as the ratios in Table III exceed two orders of magnitude for each considered frequency.

### B. Case of Study: Simultaneous Detection of IgG and IgA Anti-Tissue Transglutaminase Antibodies

To further demonstrate the behavior of the proposed multichannel potentiostat, the device was tested on the semi-quantitative detection of the anti-tTG antibodies directed

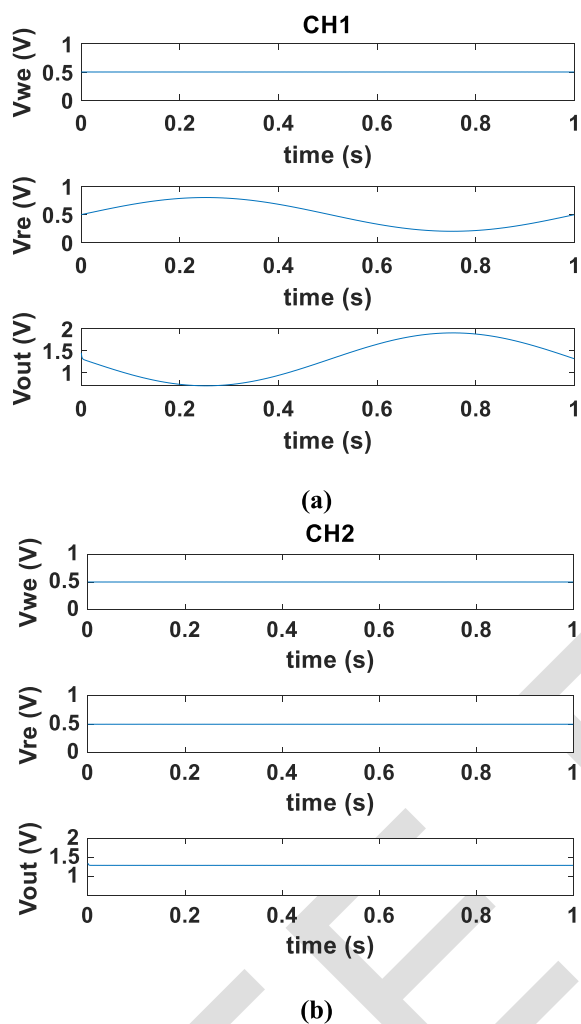


Fig. 8. Input configuration and output measurement in (a) channel 1 and (b) channel 2, for crosstalk evaluation.

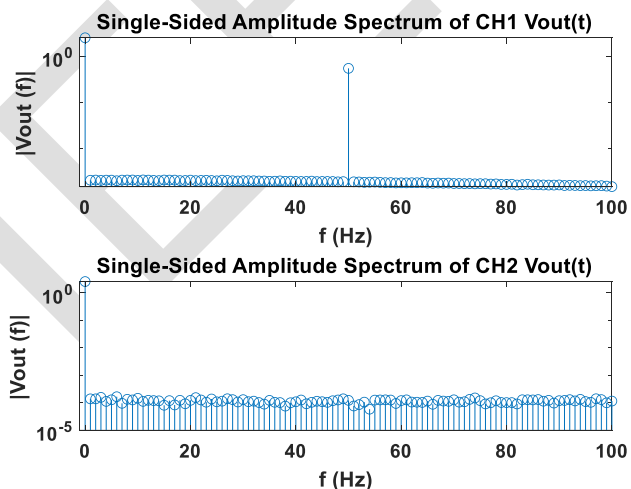


Fig. 9. FFT of the  $V_{out}(t)$  at channels 1 and 2 when a sinusoidal signal with a frequency of 50 Hz is applied to the channel 1.

481 against the transglutaminase enzyme in its open confor-  
 482 mation (Open-tTG). To this end, a previously optimized  
 483 protocol [21] was used for the functionalization of gold

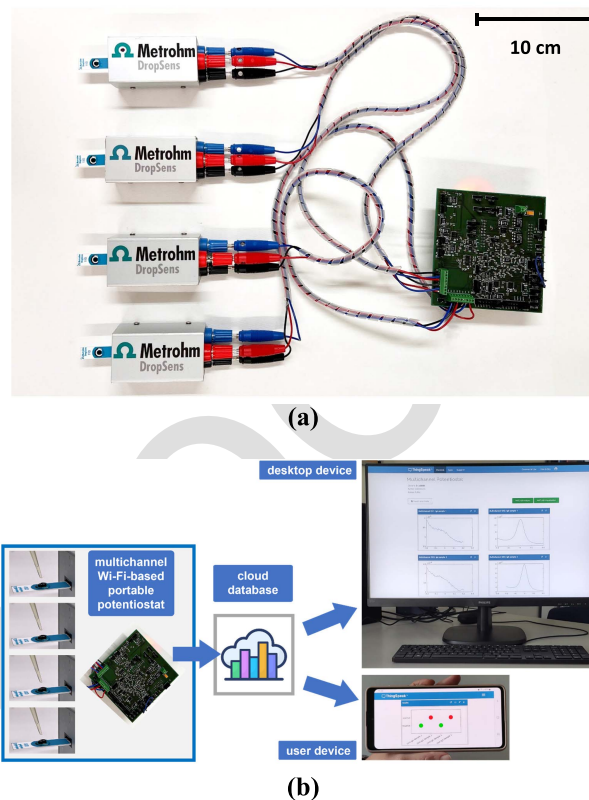


Fig. 10. (a) Prototype of the device designed and used for the tests performed for the semiquantitative analysis of IgG and IgA of the anti-tissue transglutaminase antibodies for celiac disease. (b) Schematic of the operations flow with images of the cloud visualizations.

nanoparticle-modified screen-printed electrodes (DropSens 484  
 DRP-110GNP, Metrohm). In particular, the Open-tTG<sup>1</sup> 485  
 (Zedira) enzyme was chemisorbed on gold nanoparticles, 486  
 thus allowing the recognition of anti-tTG antibodies by the 487  
 immobilized enzyme receptor. Detection was achieved using 488  
 secondary antibodies labeled with alkaline phosphatase, capa- 489  
 ble of selectively binding to IgG or IgA anti-tTG antibod- 490  
 ies (Thermo Fisher Scientific). After the addition of the 491  
 nonelectroactive hydroquinone diphosphate (Metrohm) sub- 492  
 strate, the enzymatic reaction yields the electroactive hydro- 493  
 quinone, the oxidation of which generates the signal output. 494  
 To test the performance of the multichannel device, positive 495  
 and negative controls of the ZediXclusive Open tTG<sup>1</sup>-Ab 496  
 (IgA/IgG) ELISA kits were used. These standard solutions 497  
 have a concentration, respectively, over and below the thresh- 498  
 old limit of 3 AU/mL for anti-tTG antibodies concentration. 499  
 These are recognized as specific biomarkers for celiac disease 500  
 [37]. Both IgA and IgG have to be monitored to avoid possible 501  
 false negative responses in the case of IgA deficiencies. 502  
 Considering that at least two replicates are required for each 503  
 analysis, the use of the proposed four-channel potentiostat 504  
 allows the entire analysis to be carried out with a single 505  
 parallel measurement, reducing the overall operation time. 506

Three replicates of positive and negative controls were 507  
 tested with the proposed device. In Fig. 10(a), the prototype of 508

<sup>1</sup>Registered trademark.

509 the device designed and used for the tests is shown. It is worth  
 510 noting that, in this prototype version, commercial-off-the-shelf  
 511 connectors have been used. In a future engineered version,  
 512 these components can be packed together, reducing the overall  
 513 dimensions of the device, without affecting its functionalities,  
 514 as they are only adapters without any circuitry in them.

515 In Fig. 10(b), a schematic of the operation flow is shown.  
 516 Once the data have been acquired by the device, they are sent  
 517 to the cloud service for storage and sharing. The cloud service  
 518 used is ThingSpeak [38], but other platforms can be exploited  
 519 as well. Data can be accessed for standard web browsers from  
 520 PC or mobile devices without the need for dedicated software  
 521 or APP.

522 The four electrochemical cells were conditioned with the  
 523 same parameters: after 3 min when the screen-printed elec-  
 524 trodes were left floating to allow for the enzymatic reaction  
 525 to occur and generate the electroactive hydroquinone, a  $V_{\text{bias}}$   
 526 voltage ranging from  $-0.2$  to  $0.2$  V was applied. To this  
 527 aim,  $V_Y$  was set at the constant voltage of  $1$  V, while a  
 528 variable voltage between  $1.2$  and  $0.8$  V was forced at the  
 529 RE pin through  $V_X$ . According to the DPV theory [30], the  
 530 resulting conditioning voltage should be a staircase waveform  
 531 with an increasing mean value [30]. It is worth to be noted  
 532 that the DPV technique has been used in these experiments,  
 533 but other techniques like chronoamperometry (CA) or cyclic  
 534 voltammetry (CV) are supported as well. Indeed, CA is based  
 535 on the application of fixed voltage and measurement of current  
 536 versus time, and the amplitude of the generated currents is  
 537 similar to the DPV case [29]. Regarding the CV, the triangular  
 538 voltage waveform required to bias the cell can be generated by  
 539 the 16-bit DAC independently for each channel. Furthermore,  
 540 the reading channel based on the combination of the TIA  
 541 and the proposed BLS, drives the ADC with a signal always  
 542 centered at the midpoint of the ADC range. Thus, both positive  
 543 and negative currents from the WE pin can be properly  
 544 converted and amplified.

545 Thanks to the 16-bit DAC, it was possible to set the  
 546 parameter of the conditioning voltage as those are normally  
 547 used on benchtop instruments (e.g., AUTOLAB PGSTAT 204  
 548 [13]), obtaining 319 measurement points. In particular, the  
 549 pulse amplitude was set to  $50$  mV, the step of the pulse low  
 550 level to  $5$  mV, the pulse duration to  $10$  ms, and the time  
 551 between pulses to  $200$  ms [30]. A preconditioning time of  $30$  s,  
 552 when the cell was kept at  $-0.2$  V, was introduced to precon-  
 553 centrate the reduced form of hydroquinone, thus increasing the  
 554 sensitivity of the analysis. To control the time intervals, the  $\mu\text{C}$   
 555 internal timers were exploited [31]. This ensures a total sample  
 556 acquisition time, for both IgG and IgA of  $4'17''$  with two  
 557 replicates against the  $17'10''$  that would occur using only one  
 558 channel and performing the measurements in sequence. The  
 559 last case was computed without considering the time needed  
 560 to change the electrodes at the input of the device.

561 Differential current waveforms are obtained by subtract-  
 562 ing from each other the measured currents at the beginning  
 563 and at the end of each pulse of the conditioning voltage,  
 564 as required by the DPV technique. The signals recorded  
 565 for the three-replicate for both IgG and IgA antibodies are  
 566 reported in Fig. 11(a) and (b), respectively. Since the peak of

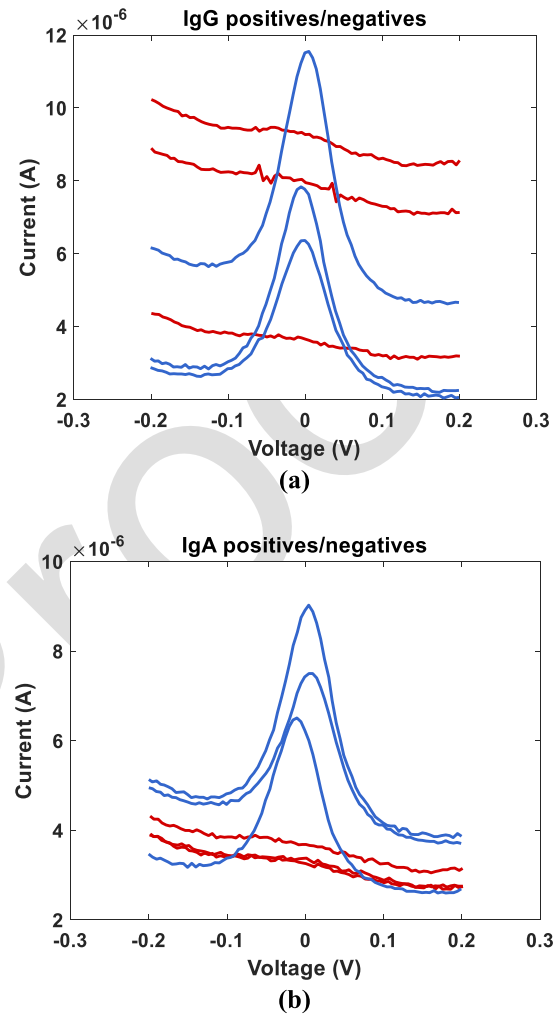


Fig. 11. Positive and negative data acquired for (a) IgG and (b) IgA. Positive samples are drawn with a blue line, and negatives with a red one.

567 the differential current is related to the concentration of the  
 568 analyte, it is possible to label a sample as positive or negative  
 569 through comparison with a predefined threshold.

570 From the recorded signals, the well-known baseline wan-  
 571 dering phenomenon is evident. To overcome this and correctly  
 572 estimate the current peak, a baseline correction algorithm was  
 573 performed on the  $\mu\text{C}$  platform after the signal acquisition  
 574 (Fig. 12). The actual baseline was estimated by computing  
 575 a linear interpolation of the ten first points of the differential  
 576 current (i.e., blue line in Fig. 12) plus ten relative minimum  
 577 points. The red line in Fig. 12 represents the baseline, and  
 578 the blue dots are the point exploited for computing it. The  
 579 current peak was then computed as the maximum value of the  
 580 distance between the differential current and the baseline.

581 As can be observed in Fig. 11, for both IgA and IgG the  
 582 output signals acquired for the negative controls (i.e., red  
 583 lines) are negligible, corresponding to current peaks within  
 584 hundreds of nanoamperes, while intense peaks are recorded  
 585 for the positive controls (i.e., blue lines).

586 The boxplot diagrams of the computed current peaks for  
 587 negatives and positives for IgG and IgA antibodies are reported  
 588 in Fig. 13(a) and (b), respectively. Again, the clear distinction

TABLE IV  PERFORMANCE COMPARISON

Device	Channels	Parallel channels	Max current range (resolution)	Min current range (resolution)	Voltage range (resolution)	Power Consumption per channel	Interface
[23]	32	4 (2 electrodes cell)	$\pm 1.5\mu\text{A}$	$\pm 1.650\text{nA}$ (125pA)	$\pm 4\text{V}$ (2mV)	-	Wi-Fi Ethernet
[24], [25]	128	Time multiplexed, 8 independent REs	$\pm 3.3\mu\text{A}$ (100pA)	-	$\pm 10\text{V}$ (305 $\mu\text{V}$ )	26.7 mW	UART
[27]	6	Shared CE, RE	-	$\pm 180\text{nA}$ (5.5pA)	$\pm 5\text{V}$ (153 $\mu\text{V}$ )	-	Bluetooth
[12]	2	Shared CE, RE	$\pm 5\text{mA}$ (300nA)	100nA (6pA)	$-2\text{V} \div 2.3\text{V}$ (537 $\mu\text{V}$ )	-	Bluetooth
[13]	8	Shared CE, RE	$\pm 80\text{mA}$ (40 $\mu\text{A}$ )	$\pm 1\text{nA}$ (1pA)	$\pm 4\text{V}$ (960 $\mu\text{V}$ )	-	Bluetooth
[28]	3	1 (sequential sampling)	$\pm 500\mu\text{A}$	-	$\pm 1.5\text{V}$ (700 $\mu\text{V}$ )	-	USB Bluetooth
Proposed	4	4	$\pm 62\mu\text{A}$ (7nA)	620nA (75pA)	$\pm 1.65\text{V}$ (50 $\mu\text{V}$ )	23.5 mW	Wi-Fi

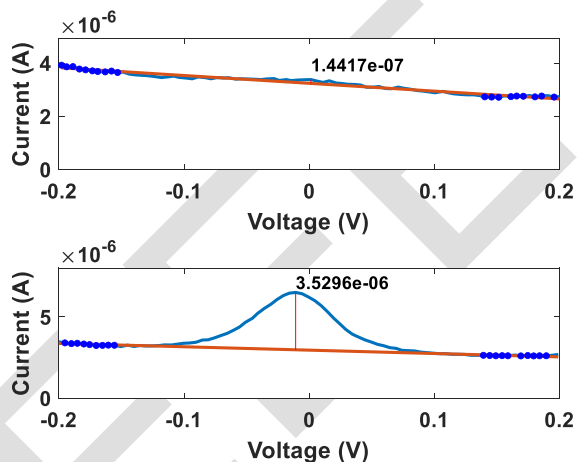


Fig. 12. Baseline correction algorithm with current peak computation for a negative sample (top) and a positive sample (bottom). The red line represents the baseline computed with the linear interpolation of 20 points (marked as blue dots). The actual peak is computed as the maximum value of the distance between the baseline and the differential current (blue line).

589 between positive and negative controls for both IgA and IgG  
 590 anti-tTG antibodies is evidenced in the boxplot charts, where  
 591 a significant difference ( $p < 0.001$ ) was observed.

592 *C. Comparison With Other Works*

593 The comparison of the proposed multichannel potentiostat  
 594 with other previously described devices is shown in Table IV.  
 595 As can be seen, the developed device is equipped with four  
 596 truly independent channels, with simultaneous acquisition and  
 597 independent conditioning voltages. The Wi-Fi connection,

thanks to a longer range than the Bluetooth radio, allows  
 improving portability without the need for an external device  
 nearby, acting as a gateway to the internet. Only a standard  
 Wi-Fi router, which is usually already present in a point of care  
 (PoC) or home environment, is needed to upload data to the  
 cloud. The Wi-Fi link is also exploited in [24]. In that solution,  
 the device was, however, connected to a local PC instead of a  
 cloud service, thus severely limiting the simultaneous sharing  
 of the results with multiple users. It is also worth noting that  
 the system described in [24] is based on a two-electrode cell,  
 waiving the protection of the RE against possible changes in  
 the WE potential [7].

Furthermore, as it can be seen in Table IV, the maximum  
 current range of the proposed device is higher than other  
 devices presented in the literature. The commercial Sensi-  
 BT [12] from PalmSens has a higher maximum current range,  
 but it has only two channels. Also, the commercial solution  
 from Metrohm [13] has some advantages in terms of the  
 number of channels and current ranges, but it requires a PC  
 and proprietary software for data processing and visualization.  
 Finally, the work presented in [28] has a higher current  
 range, but it features a shared ADC for the three channels,  
 so the sampling is sequential and not simultaneous. Moreover,  
 it requires a USB or Bluetooth communication to operate  
 and connect to a desktop or smartphone app for processing  
 and local visualization of the results. Regarding the power  
 consumption, most of the works considered do not report this  
 data. Our solution performs better with respect to [26]. The  
 power consumption of [12] is not reported; battery life is  
 reported instead; however, the battery capacity and a detailed  
 description of the measurement conditions are unknown. They  
 reported a battery life of 12 h at maximum power consumption.

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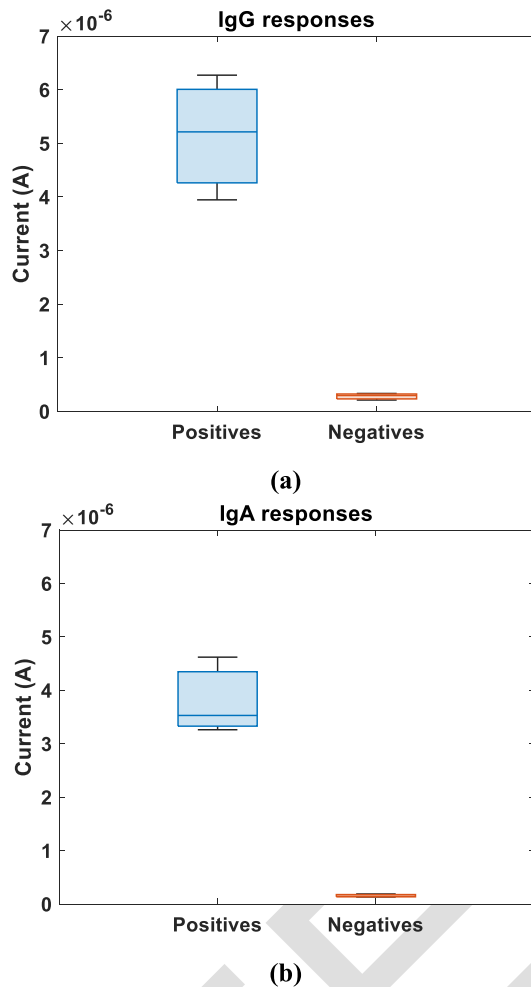


Fig. 13. Boxplot of the current peak of positive samples (blue) and negative samples (red) for the detection of the (a) IgG and (b) IgA antibodies.

For the sake of comparison, we can consider the worst case in which continuous measurements are performed and the microcontroller is always active with no power reduction techniques (e.g., sleep modes management [39]) implemented, and the network processor is idle and connected. Given two 1.5 V, 2700 mAh standard AA batteries, a battery life of 95 h is reached. Moreover, also considering the contribution of the transmission, in the unrealistic case of continuous data transmission, the battery life is reduced to 15 h, which is, in any case, better than the performance reported for the Sensi-BT [12].

## V. CONCLUSION

In this article, a multichannel potentiostat for electrochemical analysis with four truly independent channels that can be individually conditioned has been presented. The device is compact and portable, with limited power consumption (23.5 mW), and capable of both onboard processing and communication over a Wi-Fi protocol to eliminate the need for an external device nearby for data processing, viewing, and sharing. The maximum measurable current range is  $\pm 62 \mu\text{A}$ , with a resolution of 7 nA. The sensitivity is automatically tuned, defining the current range during the data acquisition

through a multiplexer and selecting the best gain of the transimpedance amplifier. The channel-to-channel mismatch has been evaluated, resulting in a maximum relative error in the gain of 1% when the maximum current range is selected. The channel crosstalk has been demonstrated to be negligible. The device shows characteristics that make it usable for different types of electrochemical analysis and then suitable for a large variety of contexts. As a case study, the device was applied for the parallel acquisition of two replicates of IgG and IgA anti-tissue transglutaminase antibodies showing analytical performance fulfilling the diagnostic purposes aimed at evaluating the onset of celiac disease. In this analysis, a reduction of the acquisition time of 76% with respect to the same measurements performed using only a channel is experienced.

In comparison with other commercial devices or published works, the proposed device, while maintaining compatible electrical characteristics, has good portability and low power consumption that makes it suitable for use outside laboratories in home and PoC contexts.

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