

# Wearable bioimpedance for continuous and context-aware clinical monitoring

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**Abstract**—Bioimpedance monitoring provides a non-invasive, safe and affordable opportunity to monitor total body water for a wide range of clinical applications. However, the measurement is susceptible to variations in posture and movement. Existing devices do not account for the variations and are therefore unsuitable to perform continuous measurements to depict trend changes. We developed a wearable bioimpedance monitoring system with embedded real-time posture detection using a distributed accelerometer network. We tested the device on 14 healthy volunteers following a standardized protocol of posture change and compared the obtained measurements with an existing commercial device. The impedance measured with both systems had a high correlation ( $r > 0.98$ ) and a Bland-Altman analysis revealed a bias of  $-4.5 \Omega$  and limits of agreement of  $-30$  and  $21 \Omega$ . Context-awareness was achieved with processing accelerometer data placed at the upper and lower leg with an accuracy  $> 95\%$ . The calculated current consumption is as low as 10 mA during continuous measurement operation, suggesting that without recharge the system can be used for multiple days. The proposed motion-aware design will enable the measurement of relevant bioimpedance parameters continuously over long periods and aid in informed clinical decision making.

**Keywords:** bioimpedance, wearable devices, fluid monitoring, actimetry, context-awareness, internet of medical things

## I. INTRODUCTION

A bioimpedance analyzer is a bioelectrical instrument to estimate body composition by measuring the electrical response of the tissues to the injection of a weak alternating current. Bioimpedance is sensitive to changes in total body water and fluid shifts, but is also used to estimate fat-free body mass. Bioimpedance has been suggested for a wide range of clinical applications ranging from impedance plethysmography [1], fluid status estimation [2], dengue management [3], up to wound healing [4]. Bioimpedance assessments are interesting techniques as they are non-invasive and lower cost compared to many other approaches, such as esophageal doppler monitoring or dual-energy X-ray absorptiometry (DEXA).

A typical bioimpedance setup consists of tetra polar configuration with 4 wires attached to the human body. The current is injected into the load (human body) with one pair of electrodes and the other pair is used to measure the voltage across the load. The signal is dependent on anthropometric factors such as age, gender and ethnicity. Postural changes may cause small shifts of fluids across body compartments and can alter the bioimpedance reading [5]. Furthermore,

bioimpedance measurements are distorted by motion artifacts [6]. Existing technologies do not account for the change in the signal due to posture variations or movement. In addition, the devices are rather bulky, unreasonably expensive, and inconvenient to use, in particular for continuous measurements. Continuous monitoring is an essential requirement in clinical environments to detect progression and transitions of physiological states [7]. All these factors explain the limited uptake of bioimpedance in clinical routines.

The goal of this work was to develop and test a novel integrated bioimpedance measurement system that is wearable, low power, affordable and enables context-based, continuous, and real-time bioimpedance assessments.

## II. DESIGN

We designed the device to include several functionalities to improve its possible clinical usage and overcome the limitations of existing systems. These contributions include an integrated low-power hardware design that includes internet of medical things functionality and a sensor network of distributed accelerometers. Furthermore, the software includes sensor fusion and context awareness that enables efficient continuous measurements independent of motion or position changes.

### A. Hardware

1) *CPU and User Interface:* The core of the system was the BLE654 (Laird Connectivity) module which was based on the nRF52840 chip set (Figure 1). It housed a 32bit ARM Cortex-M4 CPU which was used to acquire data from the bioimpedance sensor and accelerometers, storing data, controlling the display and establishing wireless communication with access points using Bluetooth 5 protocol. Four LEDs were available to notify the user for system power, charging status, battery level and data writing status without activating the display with high energy consumption. The button activated the display and could be used for other user interaction.

2) *Bioimpedance Sensing:* The AD5940 (Analog Devices Inc) analog front end was used to perform bioimpedance sensing. It was digitally configured via SPI to perform four-wire bioimpedance measurements from 1 kHz to 100 kHz. The impedance of the body segment was measured by applying an alternating signal to the body segment and measuring the potential difference between the body segment and the current that flows through the body segment. The signals were subsequently converted to digital data using

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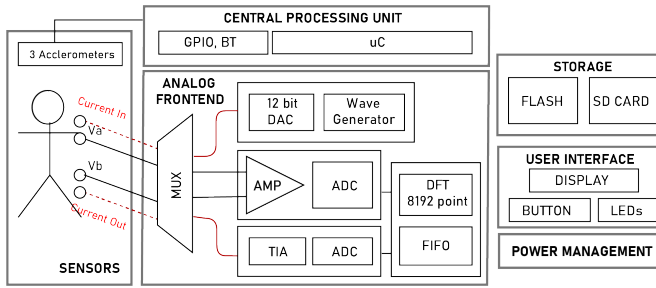


Fig. 1: System level description of the wearable bioimpedance system with main unit featuring the electronics and peripheral sensing units (accelerometers and bioimpedance electrodes).

a 16bit ADC. The internal DFT hardware accelerator extracted the real (resistance) and imaginary (reactance) parts of bioimpedance.

3) *Accelerometers*: Three MMA8451 (NXP Semiconductors) digital accelerometers were used to measure the linear acceleration with respect to positional changes. The accelerometers communicated with the CPU through the I<sup>2</sup>C interface. The sensor was configured with a 2 g dynamic range and a resolution of 14 bits.

4) *Data Storage*: The collected data was stored on either NAND Flash MT29F4G (Micron Technology) or a SD Card.

5) *Power Management*: A 3.7 V 660 mAh battery (Varta GmbH) was used to support the power requirements of the system. The device was either charged through USB or wirelessly through an inductive coil managed by a BQ51013 receiver IC (Texas Instruments). The BQ24073 (Texas Instruments) IC was used to manage the battery charging and power path. Dedicated TPS70933 (Texas Instruments) voltage regulators were used to ensure a 3.3 V power supply.

6) *Wearable*: The electronics were integrated into a wearable main unit that can be worn on the forearm (Figure 2). The size of the 3D printed unit was 122 x 43 x 14 mm. The unit provided access to touch proof electrodes and accelerometer connectors, SD card, and USB charging. A custom hand-cuff made of medical foam fixated the hand electrodes and the wearable main unit.

### B. Software

The bioimpedance signal was sampled at 20 Hz. Impedance and phase angles were calculated from the raw bioimpedance values. The accelerometer signals were sampled at 50 Hz and the raw 14 bit signal was converted to  $m/s^2$ . The length of the window used to determine the posture was 0.67 s with a 50% overlap between consecutive segments.

## III. METHODS

### A. Experiments

To perform functional testing of the device, we conducted a series of measurements on healthy young adults. The aim was to assess the accuracy of the measured signals, the capacity of performing real-time, continuous measurements and to demonstrate the context-awareness of the system.

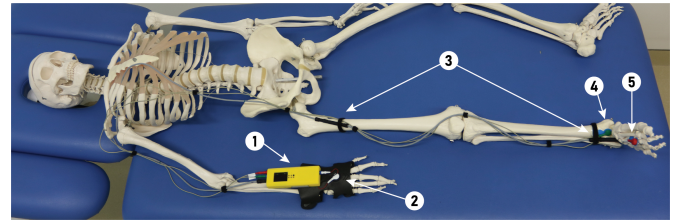


Fig. 2: Wearable, context-aware tetra-polar bioimpedance measurement system. 1) Main unit connecting to 2) wearable cuff the two hand electrodes. 3) upper and lower leg accelerometers, 4) and 5) are the foot electrodes. Upon measurement, this subject showed very high impedance suggesting a high state of dehydration.

This was achieved with two experiments: 1) Comparison of the novel device against a commercially available reference device in a stable supine posture and 2) continuous measurement during different postures, including the transitions between them. The experiments were conducted at ETH Zurich, Balgrist Campus, Zurich, Switzerland. The study was performed according to the Declaration of Helsinki and the protocol was approved by the institutional ethics committee of ETH Zurich (EK2018-N-69).

1) *Subjects*: After informed and written consent, 14 healthy volunteers (5 males) were recruited. The inclusion criteria were older than 18 years, no existing skin disorders or allergies, no metallic or electrical implants, no pregnancy, and provision of informed consent. Prior to clinical measurements, anthropometric data such as BMI, age, ethnicity, height and weight were collected. Eleven participants were from a European ethnic background, two were East Asians and one participant was South American. The mean age for the female participants was  $26.6 \pm 3.2$  y and the mean BMI was  $22.7 \pm 2.8$ . Mean age for the male participants was  $27.3 \pm 2.4$  y and the mean BMI was  $23.8 \pm 1.6$ .

2) *Placement of Electrodes*: Following the collection of anthropometric data, the electrodes and accelerometers were placed on the body. The subject was requested to lie down in a supine position for at least 5 min before starting the measurements in order to stabilize the hydrostatic pressure acting on the column of fluid in the body. The electrodes (EL500, BIOPAC Systems Inc) were placed on the subject after thoroughly cleaning the skin with medical ethanol. The sensing electrodes were placed at the wrist and ankle while the source electrodes were placed centrally on the metacarpals on the palm and metatarsals on the feet (Figure 2). The electrodes were connected to the measurement devices with long shielded medical instrumentation cables matching the respective device connectors. For all subjects, the electrodes were placed on the right limbs. To enable electrode switching in experimental protocol, the main unit was not placed directly on the arm, but kept adjacent to the subject.

3) *Placement of Accelerometers*: Two accelerometers were needed to faithfully reproduce positional orientation [8] and were placed using velcro on the ankle (adjacent to the sensing electrode) and on the anterior quadriceps region of the thigh (Figure 2). The third accelerometer was embedded

in the main unit and could not be used for the experiments. Therefore, a replacement accelerometer was placed on the upper limb.

4) *Experimental Protocol*: We performed two experiments: a) Accuracy and precision testing: During this part of the protocol, the subjects remained in supine position. The measurement cycle started with a measurement with the reference device (BIACorpus, Medi-cal Healthcare GmbH, Germany). After switching the electrodes to the wearable device, 6 frequency sweeps of 5 s were recorded (30 s). This cycle was repeated 3 times. b) Posture variation and continuous measurement testing: The wearable device was changed to continuously measure bioimpedance and accelerometer data. The subject followed specific positional changes that altered every 40 s. The fixed order supine-sitting-standing-sitting-supine-standing-supine ensured that every possible transition could be acquired.

### B. Post-processing

The data recorded from the reference and wearable device was transferred to a PC for further analysis using MATLAB (R2019b Mathworks Inc). The bioimpedance values retrieved from the system were used to calculate the resistance and reactance values through trigonometric relations. The values were filtered using a median filter to remove spiking artifacts from the signals. The accelerometer signal was filtered using a 2<sup>nd</sup> order Butterworth lowpass filter with a cutoff frequency of 3 Hz. The posture orientation angles were calculated such as

$$\theta = \arccos \frac{a_z^2}{\sqrt{a_x^2 + a_y^2 + a_z^2}} \quad (1)$$

The data from the upper and lower leg accelerometer pair was used to identify the angles associated with the body postures. Posture classification stages was performed using a decision tree. The data set was divided into training, validation, and test set. Initially a 90-10 hold out split was considered for training and testing data and the training set was further split into 10 training and validation subsets.

### C. Performance evaluation

Bland-Altman plots were generated to compare the performance between the wearable device and BIACorpus. The difference was plotted against the mean of the measured values. The bias, standard deviation of difference, limits of agreement (LoA) and confidence intervals were estimated. The limits of agreement were computed after testing for normality of the difference in measurements using the Anderson-Darling test for normality at a 5% significance level. The classification of postures was evaluated on the test set for accuracy, false negative rate and false positive rate.

## IV. RESULTS

### A. Performance

Four participants were excluded from the analysis. One was due to a glitch in the connectors and 3 due to the

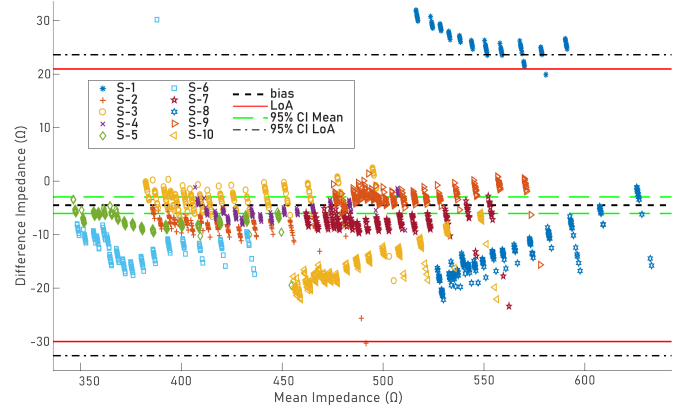


Fig. 3: Bland-Altman plot illustrating the measurement agreement clustered by subject.

reference device. The impedance measurements between the wearable and reference devices were highly correlated with ( $r > 0.98$ ,  $p < 0.0001$ ). When compared to the reference, the impedance measurements from the wearable device had a bias of  $-4.5 \pm 13.0 \Omega$ . The limits of agreement estimated were  $[-30.0, 21.0] \Omega$ , 95% confidence intervals for mean were  $[-6.1, -3.0] \Omega$  and 95% confidence intervals for LoA were  $[-32.6, 23.6] \Omega$  (figure 3).

### B. Continuous Bioimpedance

The simplified system power consumption during various system states was low with 0.8 mA during idle and 10.31 mA during measurement (Table I). Most power was consumed by the LCD display which can be turned off by the CPU for continuous measurements. As shown in figure 4, a continuous spectroscopic bioimpedance signal could be obtained, with a full spectrum sweep performed every 5 s.

TABLE I: Power consumption during different device states

Component	Current (mA) @ 3.3V			
	Sleep	Idle	Meas	Meas & Disp
MMA8451 (x3)	0.000 <sup>a</sup>	0.005	0.072	0.072
AD5940	0.000 <sup>a</sup>	0.009	9.100	9.100
BLE654	0.000 <sup>a</sup>	0.003	0.370	0.370
Display	0.020 <sup>b</sup>	0.780	0.780	25.000
<b>Total</b>	0.020	0.797	10.31	34.542

<sup>a</sup>negligible, <sup>b</sup>can be disconnected completely from power supply

### C. Context Awareness

For postural orientation analysis, two participants were excluded due to a poor data recording. Figure 4 illustrates angles estimated from the accelerometers attached to the upper and lower legs with the corresponding bioimpedance values. It was visually apparent that during the sitting posture there was a downward shift in the bioimpedance signal when compared to the standing posture. The accuracy on the posture detection was 95.68%. The false negative rate was almost negligible for supine ( $< 1\%$ ), below 18% for standing, and 9% for sitting. The false positive rates indicated that 10%, 11% and 1% of samples were falsely predicted into the class supine, sitting and standing respectively.

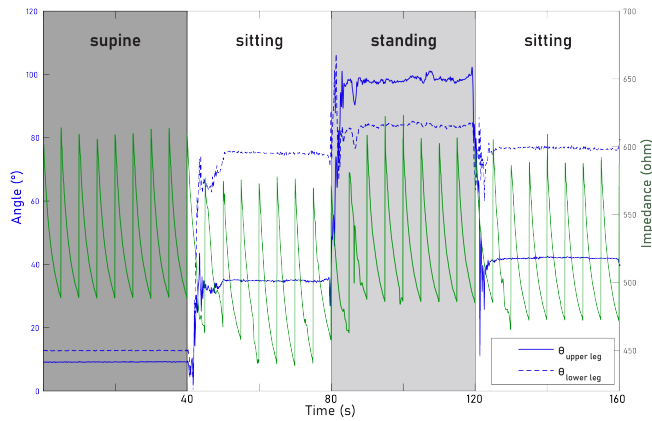


Fig. 4: Bioimpedance spectroscopy signal (green, right axis) and angle data from accelerometers placed at the upper and lower leg (blue, left axis) measured during the changing postures protocol. The baseline changes after altered posture is clearly visible.

## V. DISCUSSION

The wearable device developed in this work was a position-aware bioimpedance system. The low power consumption enabled up to 60 h of continuous measurement operation which could be extended with an optimized power plan. The operation of the wireless data modules will contribute to increased power consumption, but this feature was not yet implemented at time of testing. The system had control over the powering and writing sequence of the storage peripherals, thereby limiting power drain by these components. Considering these features, the device was able to perform recording continuously and in real time.

### A. Performance

The bioimpedance values measured by the wearable device showed high agreement with the values from BIACorpus. As expected, the measured impedance values decreased with an increase in frequency for both devices. Figure 3 indicated an appreciable degree of agreement between the wearable and reference. The 95% confidence intervals for the mean were below 0 indicating an overall negative bias except for S-1 where a strong positive bias was observed. A difference between subjects was observed. While many showed high agreement, subjects S-1, 6, 8, 10 displayed less agreement. However, the observed intra-subject variation was low. Furthermore, a trend between the difference measured and the mean values was observable. This indicated a relationship between the differences measured and the frequency used for measurements. As the frequency increased, the difference increased for most subjects. This trend could be attributed to a system drift during high frequency measurements. Further investigations and experiments will provide a better understanding of these observations.

### B. Context Awareness

The angles measured using the upper leg accelerometer data was crucial for differentiating supine and sitting position from standing, while the angles measured from the lower accelerometer data was crucial for differentiating between standing and sitting from supine. Therefore, the use of

an upper-lower leg pair for posture detection was well justified. Consistent information regarding motion during bioimpedance measurements is required to compensate posture related signal variability. A study by [9] indicated the influence of posture on the bioimpedance signal, which is an observation that we were able to reproduce. In this work, we explored context-aware sensing to provide more information regarding posture, time-locked with the acquisition of the bioimpedance signal. Therefore, we believe that the proposed context-aware system has the capacity to improve the interpretation of bioimpedance data that was obtained continuously over longer periods of time and making it suitable for use in clinical applications.

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