A Non-invasive Wearable Bioimpedance System to Wirelessly Monitor Bladder Filling

Markus Reichmuth*, Simone Schürle°, Michele Magno*
°Dep. of Health Sciences and Technology
*Department of Information Technology and Electrical Engineering
ETH Zurich, Zurich, Switzerland

Abstract— Monitoring of renal function can be crucial for patients in acute care settings. Commonly during postsurgical surveillance, urinary catheters are employed to assess the urine output accurately. However, as with any external device inserted into the body, the use of these catheters carries a significant risk of infection. In this paper, we present a non-invasive method to measure the fill rate of the bladder, and thus rate of renal clearance, via an external bioimpedance sensor system to avoid the use of urinary catheters, thereby eliminating the risk of infections and improving patient comfort. We design and propose a 4-electrode front-end and the whole wearable and wireless system with low power and accuracy in mind. The results demonstrate the accuracy of the sensors and low power consumption of only 80µW with a duty cycling of 1 acquisition every 5 minutes, which makes this battery-operated wearable device a long-term monitor system.

Keywords— Bioimpedance, Biomedical Application, Embedded Processing, Low power design, Wearable devices.

I. Introduction

The main objective of the insertion of urinary catheters is to drain urine when a person is impaired to do so naturally. However, approximately one-third of urinary catheters are solely used to monitor kidney performance without the actual need to release the bladder volume artificially. Given the increased risk of urinary tract infections (UTI) commonly associated with catheter insertion, alternative solutions to accurately monitor renal function in a non-invasive fashion are needed [1].

Sonography of the bladder volume is employed as an alternative technique in some cases. However, this approach is rather time-consuming, requires a professional and does not allow continuous measurements. In contrast, wearable devices for vital-sign monitoring are transforming the healthcare industry, allowing us to wirelessly monitor our vital signs and activity anytime, anywhere [3]. The most relevant information about some of these critical parameters can be obtained by measuring body impedance [4]-[7]. Sensing the electrical impedance of a specific region of the human body allows extracting biomedical information related to human's physiological and pathological conditions. The basic principle is to apply small alternating currents or voltages to the target body by using a dry or wet electrode to measure and monitor the corresponding electrical impedance changes [1]. The main advantage of bioimpedance systems is non-invasive long-term monitoring with high detection sensitivity, and the capability to provide rich functional information [7]-[14].

Evidently, such bioimpedance measurements pose an attractive solution to non-invasively monitor bladder volume. In particular, the effect of the different conductivities of bladder urine during natural bladder filling can be exploited. Previous studies have demonstrated a direct correlation between urinary bladder volume and electrical impedance [10] [12]. Additionally, it has been shown that the urinary bladder volume can be measured by detecting the corresponding electrical impedance changes of the bladder [13]. Finally, acquisition and transmission of bioimpedance data using electrodes have been used to monitor the bladder level [14]. Although the experimental evaluation and the system design is limited, [14] the work demonstrates that wireless bio-impedance can be used to monitor the bladder level accurately.

While previous work showed the potential of this method to be applied for renal function monitoring, for its effective usage in clinics, such wearable devices must be small, accurate, low cost, and low power [8][9]. Technology advancements in low-power integrated circuits(IC), wireless communication and sensors, and have enabled the design and the implementation of intelligent, lightweight and unobtrusive devices [15]. Smart wearables, where electronics are coupled tightly with the human body with sensors for biomedical applications are presented in recent interesting previous works [16]-[18]. Additionally, measuring bioimpedance faces challenges related to the use of dry electrodes and safety requirements.

This paper presents a novel and high accuracy embedded wireless bioimpedance system for bladder volume monitoring. The system includes the design of low power and low noise front-end with four channels to improve the accuracy of the acquired signals. Additionally, the system leverages a Bluetooth 5.0 System on Chip (SoC), which includes an ARM Cortex-M4F, both to processes the data on-board and sends the extracted information, i.e. level of urine in the bladder or alert events, wirelessly. The system has been designed and implemented with a wearable form-factor to evaluate the functionality and the accuracy of the bioimpedance sensors with experimental measurements. Lab measurements show the low power consumption, in mW range, of the designed system enabling in future the supply by only energy harvesting [19].

II. Wireless Wearable Biomedical Device

In Fig. 1 the architecture of the proposed bioimpedance system is shown, and the low power high accuracy wireless sensors system. The system consists of a SoC, which includes an ARM Cortex-M4F micro-controller and Bluetooth 5.0, power

978-3-9819263-4-7/DATE20/

management to supply the system, and several sensors for biomedical application. In particular, the system includes electrocardiography (ECG), a temperature sensor, an inertial measurement unit (IMU) and bioelectrical impedance analysis (BIA). All components are supplied by a Li-Ion battery (3.7 Volt). Finally, a buck converter regulates the voltage to a constant 3.3 Volt level, which is applied to the internal power plane. Subsequently, we introduce the main components of the system, giving more details on the bio-impedance front-end, as this is one of the main contributions of this work and our electronic design. Although the wearable device presented in this work is designed for bladder monitoring, it can be employed in many other biomedical applications.

The core of the system is the nRF52832 by Nordic Semiconductor [20], a 64 MHz ARM Cortex-M4F with a 32-bit processor, to process the data directly onboard with a BTLE 5.0 wireless interface. The system includes the IMU is the MPU-6050 by Invensense[4], which is designed for low power applications to detect movements and actions, and contains a gyroscope and an accelerometer to mitigate artifacts due to the movements of the human body. For ECG-EMG acquisition, the MAX30001 chip by Maxim Integrated was chosen as it is capable of measuring ECG and respiration, and is designed for low power applications.

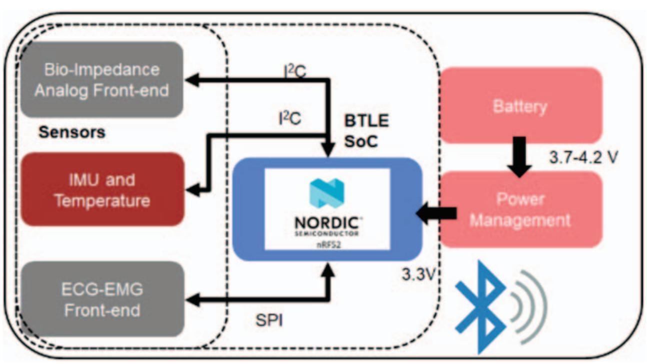

Fig. 1. Block diagram of the designed wireless sensor node.

A. Bio-impedance Sensors

Bioelectrical impedance is a widely used method for estimating body composition. The impedance is measured through electrodes attached to the skin. The electrodes are electrical transducers providing contact between a low power electronic circuit and a non-metallic object such as the human skin. The electronic interaction generates a voltage, called half-cell potential, which is measured with analog, digital converters in digital systems. One of the main challenges is that the half-cell potential varies with the electrode material and skin conditions [4]. The impedance of the tissue and the electrode circuit is shown in Fig. 2, where R_d and C_d represent the impedance associated with the electrode-skin interface and polarization at this interface [4]. R_s is the series resistance associated with the type of electrode materials. E_{hc} is the half-cell potential over-voltage discussed above. Due to the half-cell potential, a polarization can appear which reduces the electrode performance, especially if the measurements are performed during the human motion. This is the reason that most biomedical measurements use wet electrodes that are non-polarizable instead of dry electrodes. On the other hand, in wearable devices and consumer applications, dry electrodes are preferred due to their reusability, comfort and low cost, which make the acquisition more challenging.

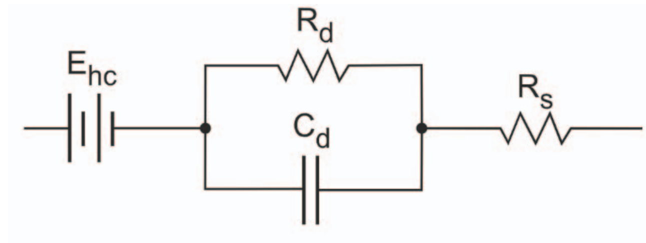

Fig. 2. Equivalent circuit model for bio-potential electrode E_{hc}

Fig. 3 shows the architecture of the designed analog front-end for bioimpedance monitoring. The main goal of the design is to minimize the power consumption keeping the accuracy of the acquisition high. To achieve this goal, our analog front-end is designed around the AD5933; state-of-the-art integrated circuits that guarantee both accuracy and low power for bio-impedance acquisition. The AD5933 is a high precision impedance converter system by Analog Devices [1].

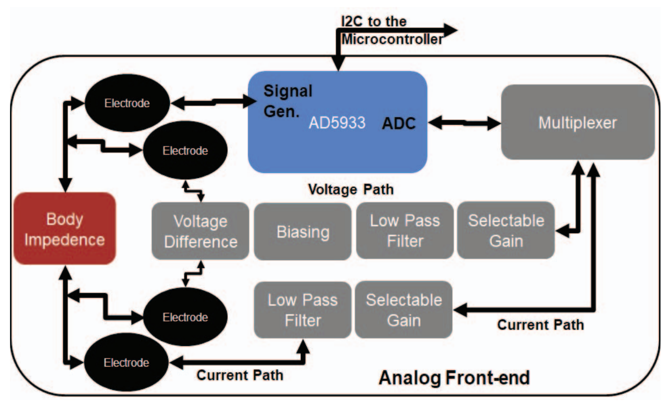

Fig. 3. Block diagram of the designed low power analog front-end.

We designed the analog front-end to be able to acquire both current and voltage to improve the accuracy further. The AD5933 on-board frequency generator produces a sinusoidal signal that is applied to the impedance under test (IUT). The frequency can be swept from 1 kHz to 100kHz. The output voltages range between 198 mV p-p to 1.98 V p-p. The signal is then sampled by an ADC and a Discrete Fourier Transform (DFT) is applied. The DFT computes the spectral power of the current at the output frequency by returning a real and imaginary value for each excitation frequency. The output resistant of the electrodes depends on the selected output voltage and ranges between 200Ω and 2.4kΩ. When the body has a small resistance below 500 Ω, the series resistant must be taken into account as explained in the previous section. Also, the voltage at the input voltage is hard biased to the voltage supply of the sensor node (3.3V/2 in our design). This biasing might lead to polarization of the electrodes. To mitigate this, a low pass filter and an adaptive gain have been employed. The components are selected with the lowest power possible. When measuring the current described above (two electrodes technique), the signal might vary significantly due to changes in the contact impedance of the electrode-skin interface when the patient is moving. This is because the output voltage amplitude of the AD5933 is constant, and the current feed into the IUT changes proportionally to the change of the ITU according to Ohm's Law. The impedance might change by more than 20% for small movements. To mitigate this, the additional measurement of the voltage is proposed in this work (four electrode technique). Fig. 3 shows the voltage path that includes a differential amplifier, a biasing circuit, a low pass filter, and an adaptive gain. Also, in this case, the design has kept power and accuracy in mind. The AD5933 is then connected via serial I^2C interface to the Nordic microcontroller

III. EXPERIMENTAL RESULTS

To evaluate the functionality and performance in terms of data acquisition and power consumption of the wearable devices we developed a working prototype of the bioimpedance system (Fig. 4).

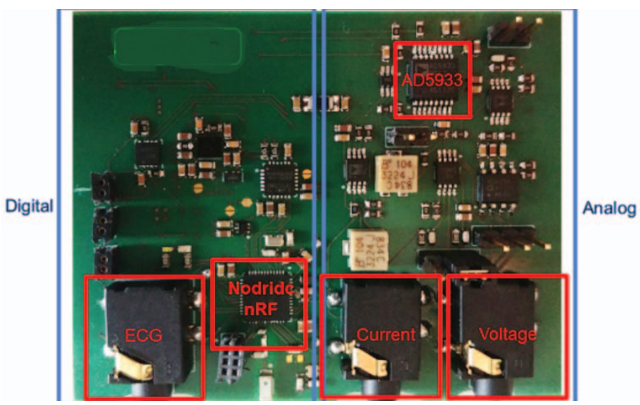

Fig. 4. Image of the developed prototype circuit board.

We did an experimental evaluation on an infusion bag to have repeatable and controlled measurements, as well as on the human body (Fig. 6). The infusion bag contained 400mL of sodium chloride (NaCl, 0.9mg/mL). The infusion bag was covered with adhesive copper foil on each side to have conditions closer to human skin. On each side, one current electrode and one voltage electrode was attached. The current flowed from the side of the bag through the sodium chloride into the other side. The voltage drop and the current was then measured. The liquid of the infusion bag then flowed into a second, empty infusion bag. The speed of the flow was not modified during the measurement and the time was measured until all the liquid had flowed from one bag to the other. We measured both the inflow and outflow scenarios. Fig. 5 shows the quality of the acquisition in both scenarios.

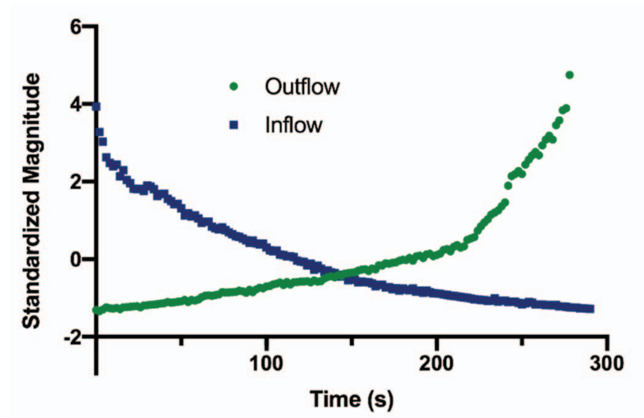

Fig. 5. In-Flow and Out-Flow measurements on the infusion bag

We tested our system also on a healthy 24-year-old male. The electrodes were attached on the skin above the bladder, 8 centimeters below the navel (Fig. 6, right). Fig. 6 illustrates the placement of the electrodes on the human body. The subject was drained of urine and then drank 500ml of water. The person then laid down and did not move intentionally until he had to go to the toilet. The subject then drank another 500ml of water and waited again. Measurements were taken every 5 seconds. The whole measurements took place over about 3 hours. Fig. 7 shows that is possible to identify the level of the bladder and the realizing of urine that confirms that our system is accurate and can be able to cover the target application scenario.

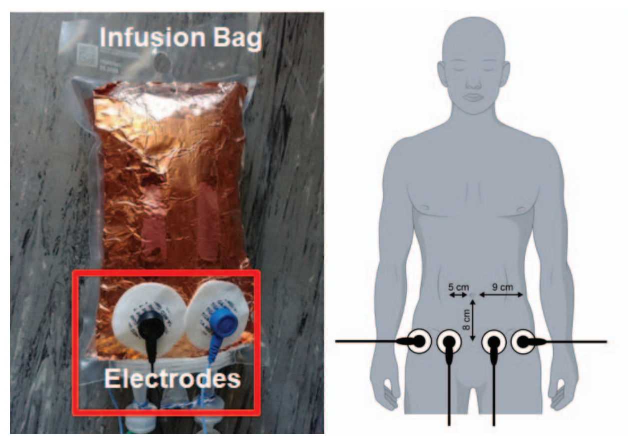

Fig. 6. Experimental setup on the infusion bag (left) and representation of the deployment on the human body (right).

The power consumption of the prototype board, collected using a lab power analyzer, is illustrated in TABLE I. The micro-controller needs significantly less power than the analog front-end. When doing a measurement, the AD5933 applys a voltage to the tissue and the current and voltage are measured. About 100mW is used by the analog part including the filtering and amplification stage of the front-end.

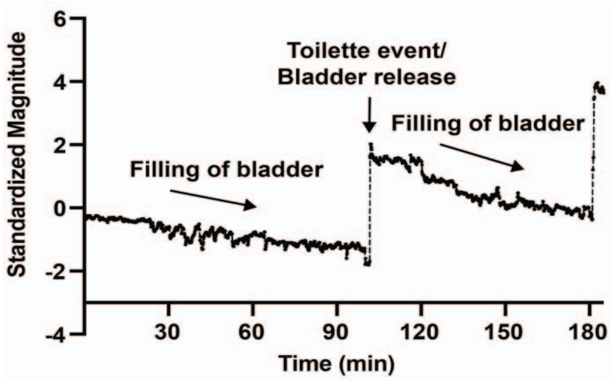

Fig. 7. Experimental evaluation of the system on a human body. The change of bioimpedanc aross the blaader is measured with four electrodes.

When not in use, the sleep mode can be activated, consuming only 16µW. After sitting idle for 5 minutes, measuring for 100ms, and repeating this duty cycling the device has an average power consumption of only around 80µW. One acquisition per minutes is a realistic duty cycling in our target scenario. This so low power consumption leads to a lifetime of several years with a small-size Li-Ion battery such 1000mAh.

TABLE I. MEASURED POWER CONSUMPTION OF THE SYSTEM

Component	*Measuring*	*Idle*	*Off*
Microcontroller	*7mW*	*5.8 µW*	*0.9 µW*
A. Front-End	*100mW*	*16µW*	*0.6µW*
Add. Sensors	*- (disconnected)*	*(disconnected)*	*(disconnected)*

IV. CONCLUSION

This paper presents the design and the implementation of a novel wearable, low power, and wireless system for bioimpedance monitoring. The core of the system is a low power Bluetooth 5.0 system on chip enabling on-board processing. The system includes a low power low-noise analog front-end to acquire data from two bio-impedance electrodes. Experimental results on an infusion bag and the human body have demonstrated the functionality and accuracy of the proposed systems. Low power consumption of 80µW for one acquisition every 5 minutes was also shown. Future works will focus on miniaturization of the devices and data acquisition with patients in hospital. In particular, the device will be compared with medical instrumentations to evaluate more precisely the accuracy.

REFERENCES

[1] T. M. Hooton, S. F. Bradley, D. D. Cardenas, R. Colgan, S. E. Geerlings,J. C. Rice, S. Saint, A. J. Schaeer, P. A. Tambayh, P. Tenke, and L. E. Nicolle, “Diagnosis, Prevention, and Treatment of Catheter-Associated Urinary Tract Infection in Adults: 2009 International Clinical Practice Guidelines from the Infectious Diseases Society of America,Clinical Infectious Diseases, vol. 50, no. 5, pp. 625663, 03 2010.

[2] Brouwer, Tammo A., Charina van den Boogaard, Eric N. van Roon, Cor J. Kalkman, and Nic Veeger. "Non-invasive bladder volume measurement for the prevention of postoperative urinary retention: validation of two ultrasound devices in a clinical setting." Journal of clinical monitoring and computing 32, no. 6 (2018): 1117-1126.

[3] Fan, Y., Xu, P., Jin, H., Ma, J., & Qin, L. (2019). Vital Sign Measurement in Telemedicine Rehabilitation Based on Intelligent Wearable Medical Devices. IEEE Access, 7, 54819-54823.

[4] Conchell, J. C.: Bio-Impedance Circuit Design for Body Worn Systems. Techn. Ber., Analog Devices, Inc., 2014

[5] Genton, L., Herrmann, F. R., Spörri, A., & Graf, C. E. (2018). Association of mortality and phase angle measured by different bioelectrical impedance analysis (BIA) devices. Clinical nutrition, 37(3), 1066-1069.

[6] Pevnick, Joshua M., et al. "Wearable technology for cardiology: an update and framework for the future." Trends in cardiovascular medicine 28.2 (2018): 144-150.

[7] O. G. Martinsen and G. Sverre, “Bioimpedance and bioelectricity basics,” Access Online via Elsevier, 2011.

[8] Tomasini, Marco et al 2016 Power Line Interference Removal for High-Quality Continuous Biosignal Monitoring With Low-Power Wearable Devices IEEE Sensors Journal 16 3887-3895

[9] Presti, D. L., Massaroni, C., D’Abbraccio, J., Massari, L., Caponero, M., Longo, U. G., ... & Schena, E. (2019). Wearable system based on flexible FBG for respiratory and cardiac monitoring. IEEE Sensors Journal.

[10] M. A. Talibi, et al., “A Model for Studying the Electrical Stimulation of the Urinary Bladder of Dogs,” British Journal of Urology, Vol. 42, No. 1, 1970, pp. 56-65. http://dx.doi.org/10.1111/j.1464-410X.1970.tb11908.x

[11] Elfaramawy, T., Fall, C. L., Arab, S., Morissette, M., Lellouche, F., & Gosselin, B. (2018). A Wireless Respiratory Monitoring System Using a Wearable Patch Sensor Network. IEEE Sensors Journal, 19(2), 650-657.

[12] F. M. Waltz, G. W. Timm and W. E. Bradley, “Bladder Volume Sensing by Resistance Measurement,” IEEE Transactions on Biomedical Engineering, Vol. 1, No. 1, 1971, pp. 42-46. http://dx.doi.org/10.1109/TBME.1971.4502788

[13] W. C. Liao and F. S. Jaw, “Noninvasive Electrical Impedance Analysis to Measure Human Urinary Bladder Volume,” Journal of Obstetrics and Gynaecology Research, Vol. 37, No. 8, 2011, pp. 1071-1075. http://dx.doi.org/10.1111/j.1447-0756.2010.01487.x

[14] Li, Rihui, et al. "Design of a noninvasive bladder urinary volume monitoring system based on bio-impedance." Engineering 5.10 (2013): 321.

[15] Mezzina, G., Aprigliano, F., Micera, S., Monaco, V. and De Venuto, D., 2019, June. EEG/EMG based Architecture for the Early Detection of Slip-induced Lack of Balance. In 2019 IEEE 8th International Workshop on Advances in Sensors and Interfaces (IWASI) (pp. 9-14). IEEE.

[16] Magno, M., Benini, L., Spagnol, C. and Popovici, E., 2013, October. Wearable low power dry surface wireless sensor node for healthcare monitoring application. In 2013 IEEE 9th international conference on wireless and mobile computing, networking and communications (WiMob) (pp. 189-195). IEEE.

[17] Magno, Michele, Michael Pritz, Philipp Mayer, and Luca Benini. "DeepEmote: Towards multi-layer neural networks in a low power wearable multi-sensors bracelet." In 2017 7th IEEE International Workshop on Advances in Sensors and Interfaces (IWASI), pp. 32-37. IEEE, 2017.

[18] Forooghifar, F., Aminifar, A. and Alonso, D.A., 2018, August. Self-aware wearable systems in epileptic seizure detection. In 2018 21st Euromicro Conference on Digital System Design (DSD) (pp. 426-432). IEEE.

[19] Magno, M., Spadaro, L., Singh, J. and Benini, L., 2016, June. Kinetic energy harvesting: Toward autonomous wearable sensing for Internet of Things. In 2016 International Symposium on Power Electronics, Electrical Drives, Automation and Motion (SPEEDAM) (pp. 248-254). IEEE.

[20] Mc Caffrey, C., Vaajoki, A., Halme, J., Sillanpää, T., Revuelta, A., & Puukko, P. (2018, October). Design and Verification of a Wireless Readout System for Integrated Motor Axle Condition Monitoring. In 2018 IEEE SENSORS (pp. 1-4). IEEE.

[21] C. Margo, J. Katrib, M. Nadi, and A. Rouane, “Four electrode embedded bioimpedance measurement system,” 2013 IEEE Faible Tension Faible Consommation, pp. 14, 2013.