

A Wrist-worn Respiration Monitoring Device using Bio-Impedance*

Kaan Sel¹, Alexander Brown¹, Hongwoo Jang⁴, Harlan M. Krumholz⁶, Nanshu Lu^{4,5}, and Roozbeh Jafari^{1,2,3}

¹Department of Electrical and Computer Engineering, ²Department of Biomedical Engineering,

³Department of Computer Science and Engineering, Texas A&M University, College Station, TX, USA

⁴Department of Texas Materials Institute, ⁵Department of Aerospace Engineering and Engineering Mechanics, University of Texas at Austin, Austin, TX, USA, ⁶Center for Outcomes Research and Evaluation, Yale University, New Haven, CT, USA

ksel@tamu.edu, alexandergbrown99@tamu.edu, hongwoo.jang@gmail.com, harlan.krumholz@yale.edu, nanshulu@utexas.edu, rjafari@tamu.edu

Abstract—In the US alone, 22 million individuals suffer from obstructive sleep apnea (OSA), with 80% of the cases symptoms undiagnosed. Hence, there is an unmet need to continuously and unobtrusively monitor respiration and detect possible occurrences of apnea. Recent advancements in wearable biomedical technology can enable the capture of the periodicity of the heart pressure pulse from a wrist-worn device. In this paper, we propose a bio-impedance (Bio-Z)-based respiration monitoring system. We establish close contact with the skin using gold e-tattoos with a 35 mm by 5 mm active sensing area. We extracted the respiration from the wrist Bio-Z signal leveraging three different techniques and showed that we can detect the start of each respiration beat with an average root mean square error (RMSE) less than 13% and mean error of 0.3% over five subjects.

Keywords – Respiration; e-tattoo; wrist-worn; obstructive sleep apnea (OSA); bio-impedance (Bio-Z).

I. INTRODUCTION

Sleep apnea is a potentially life-threatening disorder most often associated with weight gain and the buildup of soft tissues in the throat. The most common form of sleep apnea, obstructive sleep apnea (OSA), is a breathing disorder causing breathing to repeatedly stop and start during sleep. The occurrence of OSA degrades sleep quality leading to both cognitive and physical performance drop [1] and if left untreated could increase the risk of serious life-threatening conditions such as myocardial infarction and stroke [2], [3]. OSA can be difficult to detect, requiring overnight evaluation in a specialized clinic through continuous monitoring of the breathing. In the US alone it is estimated that 22 million people suffer from OSA, and it is estimated that 80% remain undiagnosed [4]. To facilitate a convenient diagnosis, a single continuous monitoring system with easy-to-wear sensors needs to be developed to capture the respiratory parameters such as the respiration rate (RR) and the pattern of breathing along with any others which may aid in the process such as heart rate (HR) and blood pressure (BP).

The most common methods preferred in clinics are capnography and nocturnal polysomnography, that monitor the respiration cycles by tracking the gas-exchange during ventilation or with chest-abdomen bands/belts that record the

upper abdominal wall movement. The former requires a face mask that is continuously worn over the course of the experiment and the latter requires the bands to be attached securely. Both cause significant discomfort and require a visit with trained clinicians in a sleep study room [5], [6]. Hence, these methods fail to provide an effective diagnostic tool for the higher percentage of the possible OSA sufferers. Current wearable technology allows us to provide certain physiological biomarkers like heart rate and its variability commonly through optical and capacitive sensing modalities. Additionally, these systems allow integration with smartwatches or other wrist-worn devices. On the other hand, capturing respiration from the wrist is not a direct measurement and requires extracting the influence of ventilation in the arteries, which possesses most of the blood circulation parameters at the wrist.

Photoplethysmography (PPG) is the most common example for the optical modalities, that depends on a light illumination on a specific part of the body and recording the resultant reflection/transmission to detect pulsatile activity [7]. The major drawbacks of PPG based systems in detecting respiration from the wrist is the weak penetration of the light into the skin that captures pulsatile activities from the capillaries rather than the arterial sites, which is not sufficient to detect slight variations in the blood flow parameters during each inhalation and exhalation accurately, essential for apnea detection [8]. In addition, the sensitivity of the PPG sensor is highly dependent on environmental and human-based factors such as humidity, skin color, skin temperature, hydration level and ambient light [9] that degrade the confidence in capturing respiration. Therefore, current wearable respiratory systems depending on optical sensing modalities are limited with the average RR and demonstrate low accuracy in detecting the sleep [10]. The capacitive [11] and radar/Wi-Fi-based [12], [13] systems suffer from similar problems, in addition to being extremely position-dependent and exhibiting low signal-to-noise ratios (SNR). Moreover, these methods report the dominant frequency component in the processed signals corresponding to the average respiration rate and this is not sufficient to detect a possible apnea, since apnea is not registered in the reported average per-minute respiration rate. There is a need to provide a new and robust sensing modality that still supports wearable and non-invasive wrist-worn applications, without trading off signal quality, fidelity and

*Research supported by the National Institutes of Health, under grant 1R01EB028106-01 and the National Science Foundation, under grant CNS-1738293.

sensitivity across the sensing area. In addition, the system should be able to provide breath-by-breath respiration rate analysis to detect the patterns of possible apnea.

In this paper, we introduce a technique based on bio-impedance (Bio-Z) to capture variations in the hemodynamic and respiratory biomarkers non-invasively. In our previous works, we showed that we can capture HR, pulse transit time (PTT) [14] as well as BP with high confidence [15] from the wrist, as well as the heart and lung activities from the chest [16] using Bio-Z. This paper focuses on capturing the respiration rate from the wrist with Bio-Z through ultra-thin gold e-tattoos as shown in Fig. 1. To achieve this, we capture signal modulations on the Bio-Z signal as possible influences from the respiratory cycle.

Our contribution in this paper can be summarized as follows:

- A novel method that leverages Bio-Z to extract breath-by-breath respiration rate from the variations in the heart pressure pulse-wave and the physical motion at the wrist.
- The system injects a non-invasive microamp AC current at high frequency and picks-up differential voltage induced by the Bio-Z on a 35mm by 5mm sensing area established by ultra-thin gold e-tattoos, which can be integrated into a wrist-worn device with ease.
- Since the system processes individual breaths to extract the breath-by-breath respiration rates, it can be extended to carry a respiration pattern analysis to detect possible hyperpnea and hypopnea along with apnea.

The rest of the paper is organized as follows. In Section II, we describe the developed measurement setup and our signal post-processing to extract respiration influence on Bio-Z. Then, we share the experimental results in Section III. Finally, our conclusion is given in Section IV.

II. METHODS

Biological tissues, cells, surrounding extracellular fluid and encapsulated intracellular fluid create a finite impeding path for any AC signal penetrating through the skin. Although cell membranes, as well as the skin, acts as a capacitive barrier, increasing the frequency of the applied AC signal leads to a lower attenuation while traveling its course. Therefore, a non-invasive bio-impedance (Bio-Z) signal can be acquired through injecting a very-small AC signal and picking up the accumulated voltage signal due to this injection. If the placement of the electrodes responsible for injection and sensing happens on the projection of the artery, the magnitude of the Bio-Z becomes a factor of physiological and physical changes occurring at the moment of signal injection. For instance, while blood volume changes at the vessels, due to periodic heart activity, additional changes in the momentary impedance of the vessels, muscle movements or gestures will also contribute to the Bio-Z variations. Therefore, continuous Bio-Z measurement provides significant and strongly correlated information about these phenomena. However, a

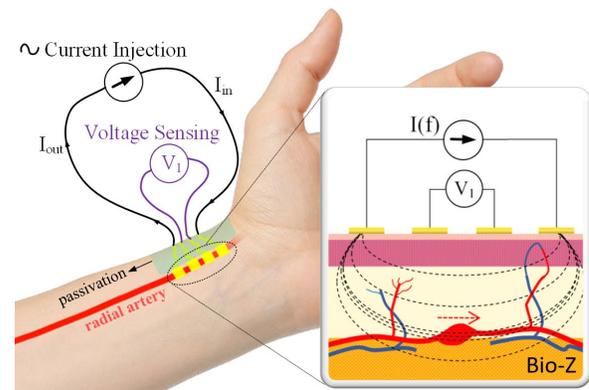


Fig. 1. A wrist-worn continuous respiration and blood flow monitoring device based on non-invasive bio-impedance (Bio-Z) sensing and ultrathin gold e-tattoos.

significant challenge in measuring the Bio-Z changes remains in building a system establishing good contact with the skin while preventing the skin impedance from dominating the dynamic range of the system. To obtain the former, we use ultrathin and flexible gold e-tattoos that form an intimate contact with the skin, and for the latter, we introduce Bio-Z XL that provides low-noise multichannel Bio-Z sensing based on four-terminal sensing (Kelvin sensing). In this study, the frequency of the injected signal is selected as 10 kHz due to the low electrode-skin impedance, the 1 mA allowance of current injection amplitude due to safety standards [17] and the low contribution of flicker noise ($1/f$ noise) at this frequency. This operation frequency is validated in our prior studies [18], [19].

A. Multi-Channel Bio-Impedance Sensing on the Wrist

In this work, we use ultra-thin gold e-tattoos with 25 mm² square shaped active surfaces that establishes an intimate contact with the skin. We previously shared the cut-and-paste manufacturing method of the tattoos as well as the acquisition of certain signal modalities such as ECG, SCG, and capacitive sensing [20], [21]. In order to improve the stability of the ultrathin tattoos, the traces are designed in a serpentine shape. Prior to each signal acquisition with the tattoos, we obtained LCR measurements (LCR Meter IM3536, Hioki, Japan) at the tattoo-skin interface. The interface brings 5-10 k Ω impedance at our operation frequency of 10 kHz, which is comparable to the state-of-the-art wet electrodes [18]. The tattoos are supported with a Tegaderm layer to provide additional flexibility, mechanical stability, and adhesion to the skin. The characterization of the tattoo integration, as well as the contact quality, is planned to be published as a separate work.

The Bio-Z XL board operates based on an ARM Cortex M4 MCU that triggers a 16-bit DAC converter (DAC8811, Texas Instruments, USA) to generate an AC signal with programmable gain and frequency. The output of DAC is fed to the negative terminal of a precision amplifier (OPA211, Texas Instruments, USA) through a resistor and a capacitor in series in order to prevent DC injection into the body and arrange the gain of V-to-I conversion. The injection electrodes are connected to the feedback loop of the op-amp, allowing a stable current injection into the epidermis. In order to sense the small variations in the Bio-Z, the board contains low-noise and

low-power instrumental amplifiers (AD8421, Analog Devices, USA) that amplify the small voltage signal induced in between the sensing electrodes due to the current injection with 20 dB. The amplified signal is then filtered with a low-pass anti-aliasing filter with cut-off frequency at 30 kHz and fed into a high-resolution and high-speed 24-bit delta-sigma ADC (ADS1278, Texas Instruments, USA) with a sampling frequency of 93,750 Hz with 4 simultaneous channels (0.3 μV sensing resolution, with 9 points per signal period) [18], [19]. The overall system is calibrated with a calibration circuit demonstrating less than 1 m Ω root-mean-square (RMS) error, which is sufficiently lower than the target Bio-Z variations (50 m Ω).

B. Signal Post-Processing for Respiration Extraction

The sampled data of all channels is sent to the MCU, which then sends it to the PC through Hi-Speed USB Bridge for signal post-processing in MATLAB and to extract the respiratory activity. First, the raw ADC readings are converted to the units of Ω using the calibration measurement and multiplied with the carrier signal, $V_c(t)$, which is also fed to the DAC for demodulating the changes in the Bio-Z. We apply a 2nd order Butterworth low-pass filter at 6 Hz cut-off frequency to measure the physiological dependent changes such as heart and respiration activities and to reject the image noise at twice the carrier frequency as well as the high-frequency oscillations. We then downsampled the signals from 93,750 Hz to 375 Hz without a significant loss of information to get the raw Bio-Z readings.

Respiration is the transport of oxygen and carbon dioxide in and out of the body, with lungs being the primary organs of the respiratory system. This vital process of the air movement is driven by lung muscles that cause a significant amount of volume change in the lungs during each breathing cycle. The volume changes of the lungs during respiration influences the intrathoracic pressure upon the circulation as well as the pulmonary arterial pressure [22]–[24], where during each positive-pressure insufflation of the lungs, the diastolic pressure increases [25]. This impacts the blood volume that is traveling through the arteries. At every heartbeat, the arrival of the blood volume to the sensing location due to the pressure pulse wave causes a sharp decrease in the Bio-Z signal from its first main peak (PK) to the first foot (FT) [15]. Therefore, for each hemodynamic cycle, the magnitude of the decrease in the impedance is a direct indicator of the blood volume travelling through the sensing location, and this can be calculated by recording the difference between PK and FT points, denoted as $\mathbf{Y}_{\text{Delta-Z}}$ in this manuscript.

In addition to blood volume changes, respiration causes a shift in the mean of the Bio-Z signal in both directions depending on whether the subject is inhaling or exhaling. This behavior is related to the heavy physical movements with the insufflation and inhalation as well as the change in the intrathoracic pressure. To monitor these shifts in the Bio-Z signal, we used maximum slope point (MSP), another important point defined at the descending slope from PK to FT. MSP provides more reliable information than PK and FT points about the changes in the baseband of the signal, \mathbf{Y}_{DC} , since it is insensitive to the noise that may occur at the peak or the foot of the signal [15]. Moreover, the difference between time locations for consecutive MSPs provides an accurate

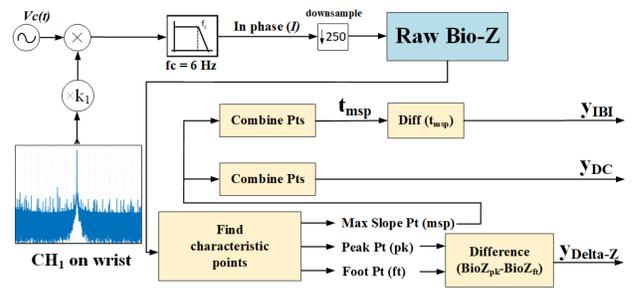


Fig. 2. Signal post-processing. k is channel calibration with units of Ω/Volts . $V_c(t)$ is digital sinusoidal (carrier) that feeds DAC and used for demodulation purposes. y is the resampled signal that passes through the corresponding characteristic point. t is the time stamp of the signal at the characteristic point.

estimate for the inter-beat intervals (IBIs). Monitoring the signal constructed by the serial combination of the consecutive IBIs, \mathbf{Y}_{IBI} , allow to detect the increase/decrease of the sinus rate with inspiration/expiration, revealing another interrelation between ventilation and circulation. This specific heart-rate variability is referred to as respiratory sinus arrhythmia and is a normal rhythm for healthy and young individuals, and becomes less apparent for the individuals that are likely to be older or have an underlying cardiac problem [26], [27]. Hence, the Bio-Z signal is modulated by the respiration in three main characteristic ways and the overall signal post-processing is shared in Fig. 2.

III. EXPERIMENTAL RESULTS AND DISCUSSION

All our experiments were conducted on healthy participants under the IRB approval IRB2017-0086D by Texas A&M University at room temperature with the participants being at rest. Gold e-tattoos were fabricated at UT Austin and were placed on the wrist, aligned to the arterial site. Prior to the tattoo placement, we located the artery on the wrist using an ultrasound doppler (Super Dopplex II, Huntleigh, USA). The tattoos have an array of six 25 mm² electrodes with a 3 mm separation that builds an intimate contact with the skin. The outermost electrodes were used for injecting a 0.4 mA sinusoidal electrical current at 10 kHz. The second most outer electrodes were used for sensing the Bio-Z changes, where we did not use the most inner electrode pair in this study. We employed a commercial Capnograph (RespSense II, Nonin, USA) that monitors the CO₂ exchange via an attached nose cannula as a gold standard to get the baseline respiration. We enrolled five participants, where we ran five sets of experiments each including two minutes of signal acquisition from each subject. Fig. 3 shows an example of the experimental setup and tattoo placement on the wrist. The first 10 seconds of each trial is used to synchronize the Capnography signal to the Bio-Z signal, where subjects were asked to hold their breath in this 10 second period.

Fig. 4 shows an example of the raw Bio-Z signal after the demodulation of the ADC readings recorded from the first subject. We also included the characteristic peak, foot and maximum slope points on the same plot, that are used to carry out the rest of the post-processing. Next, we estimated the respiratory influence in three different waveforms from the raw Bio-Z signal. Placing the electrodes at the arterial site leads to strong variations in the Bio-Z with circulation. This is mostly governed by the low impedance of the blood compared to the surrounding tissues [28] that allows Bio-Z to penetrate

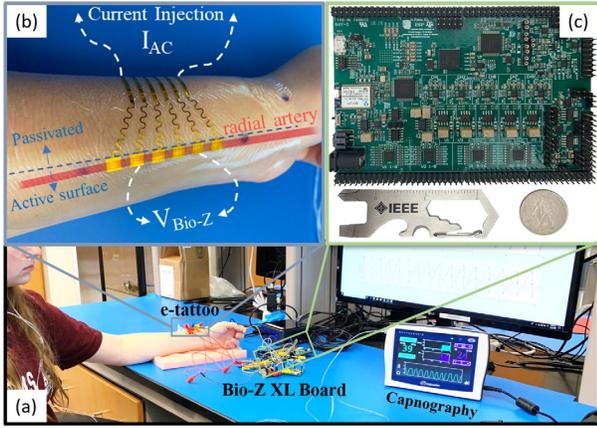


Fig. 3. (a) Experimental setup. (b) Gold e-tattoo placement at wrist. (c) Bio-Z XL board for low-noise multi-channel Bio-Z sensing designed in ESP Lab, TAMU, USA.

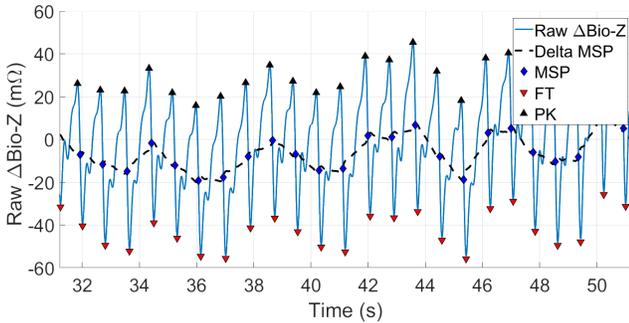


Fig. 4. Raw Bio-Z signal at wrist after demodulation. The plot includes the characteristic points as colored markers.

deeply. With the start of each circulatory cycle, oxygenated blood volume travels through the aorta to be distributed throughout the whole body via arteries. The amount of blood carried with the pulse pressure as well as its wave velocity are a factor of cardiac mechanics. Therefore, the Bio-Z signal acquired within a region containing the underlying arteries reflects all the variations in the cardiac cycle. As the ventilation process, governed by the lungs, is one of the factors that affect blood flow (and, consequently, Bio-Z), we built our analysis to extract the influence of the respiration on circulation using three different techniques. The first estimation, y_{DC} , focuses on the low-frequency trend of the Bio-Z signal, whereas $y_{\Delta Z}$ and y_{IBI} extract the amplitude and frequency modulation of the blood flow, respectively within a period of inhaling and exhaling. Fig. 5 shows the normalized fast Fourier Transform (FFT) of the estimated signals, the reference Capnography signal and the raw Bio-Z, with 8 mHz resolution. All of the signals carry a dominant frequency at 0.265 Hz in agreement with the reference signal.

To evaluate the performance of the proposed technique, we carried out a breath-to-breath analysis. Initially, we found the MSPs for each of the estimated signals, and zero-crossing of the Capnography signal, followed by calculating the corresponding inter-breath-intervals (IBrI). Fig. 6 shows the distribution of the IBrIs for each estimated signal. We shared the root-mean-squared error (RMSE) in locating the start of the breath (*i.e.* error between measured IBrI and true IBrI) as well as the error in the average breath-per-minute (BPM) in Table 1. The evaluation represents a total of five two-minutes

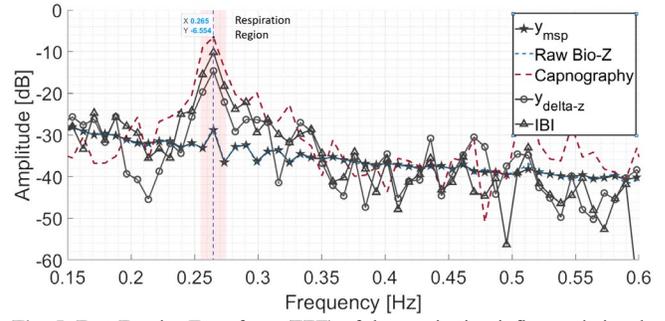


Fig. 5. Fast Fourier Transform (FFT) of the respiration influenced signals extracted from the raw Bio-Z and the corresponding capnography signal as reference.

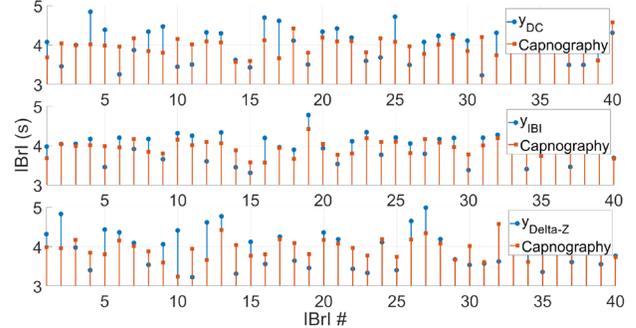


Fig. 6. Inter-breath-intervals (IBrIs) calculated from the estimated signals and the corresponding reference IBrI from the Capnography.

TABLE I
IBRI ESTIMATION PERFORMANCE PER SUBJECT

Subject	y_{DC}		y_{IBI}		$y_{\Delta Z}$	
	RMSE (%)	Mean Error (%)	RMSE (%)	Mean Error (%)	RMSE (%)	Mean Error (%)
1	12.5	0.62	9.7	0.03	13.3	0.48
2	6.6	0.23	10.9	0.74	-	-
3	13.25	0.04	17.09	0.05	22.81	1.35
4	11.64	0.21	15.48	0.40	21.57	0.36
5	18.28	0.27	12.11	0.14	25.77	0.87
Average	12.45	0.27	13.06	0.27	20.86	0.765

long signal acquisition and processing for each subject with a total of five subjects. All of the values in Table 1 are shared in percentages with respect to the average IBrI determined over the whole 10-minutes of data for each subject.

The average RMSE is less than 14% over five subjects for y_{DC} and y_{IBI} with the mean of average error appearing at 0.27%. OSA occurs with episodes of breath cessations. Since our system gives both the rate and pattern of the ventilation cycle, the device performance is well enough to detect a possible “apnea” [29]. In addition, the magnitude of the modulation of Bio-Z with the ventilation is possibly a factor of the lung capacity as well as the per-breath volume changes. Hence, a further investigation with our device offers a potential wearable and convenient approach to detect other abnormal patterns, such as Chyne-Stokes respiration, that are possible biomarkers of heart failure, strokes, brain tumors, traumatic brain injuries and hyponatremia [30], [31]. We will study the respiration pattern analysis with the proposed technique in the future.

IV. CONCLUSION

With growing concerns of obesity and increased stress levels, obstructive sleep apnea has become a significant life-limiting disorder. Most of the time the disorder stays hidden until the occurrence of a serious life-threatening implication. Therefore, continuous monitoring of the signs of OSA in the respiration pattern during sleep becomes a must for both diagnostic and prognostic care. We shared a new technique that depends on Bio-Z that can be integrated into a wearable wrist-worn device to non-invasively monitor ventilation. We showed the performance of our system in estimating respiration parameters. The proposed approach in this study carries a strong potential in improving the current practice of wearable respiratory monitoring due to its superior performance.

ACKNOWLEDGMENT

This work was supported in part by the National Institutes of Health, under grant 1R01EB028106-01 and National Science Foundation, under grant CNS-1738293. Any opinions, findings, conclusions, or recommendations expressed in this material are those of the authors and do not necessarily reflect the views of the funding organizations.

REFERENCES

- [1] A. Alberti, G. Mazzotta, E. Gallinella, and P. Sarchielli, "Headache characteristics in obstructive sleep apnea syndrome and insomnia," *Acta Neurol. Scand.*, vol. 111, no. 5, pp. 309–316, May 2005.
- [2] C. H. Lee *et al.*, "Severe obstructive sleep apnea and outcomes following myocardial infarction," *J. Clin. Sleep Med.*, vol. 7, no. 6, pp. 616–621, 2011.
- [3] K. Narkiewicz and V. K. Somers, "The sympathetic nervous system and obstructive sleep apnea: Implications for hypertension," in *Journal of Hypertension*, 1997, vol. 15, no. 12 II, pp. 1613–1619.
- [4] "Sleep Apnea Information for Clinicians – Sleep Apnea." [Online]. Available: <https://www.sleepapnea.org/learn/sleep-apnea-information-clinicians/>. [Accessed: 20-Jan-2020].
- [5] V. Ibáñez, J. Silva, and O. Cauli, "A survey on sleep assessment methods," *PeerJ*, vol. 6, p. e4849, 2018.
- [6] F.-T. Wang, H.-L. Chan, C.-L. Wang, H.-M. Jian, and S.-H. Lin, "Instantaneous respiratory estimation from thoracic impedance by empirical mode decomposition," *Sensors*, vol. 15, no. 7, pp. 16372–16387, 2015.
- [7] R. C. Hutchinson, "Mobile Monitoring with Wearable Photoplethysmographic Biosensors," *IEEE Eng. Med. Biol. Mag.*, no. June, 2003.
- [8] C. C. Y. Poon, Y. Zhang, T. Chinese, and H. Kong, "A Novel Biometrics Method to Secure Wireless Body Area Sensor Networks for Telemedicine and M-Health," *IEEE Commun. Mag.*, vol. 44.4, no. November 2013, pp. 73–81, 2006.
- [9] X. F. Teng and Y.-T. Zhang, "The effect of contacting force on photoplethysmographic signals," *Physiol. Meas.*, vol. 25, no. 5, p. 1323, 2004.
- [10] M. de Zambotti, L. Rosas, I. M. Colrain, and F. C. Baker, "The Sleep of the Ring: Comparison of the ÖURA Sleep Tracker Against Polysomnography," *Behav. Sleep Med.*, vol. 17, no. 2, pp. 124–136, Mar. 2019.
- [11] M. Chattopadhyay and D. Chakraborty, "A New Scheme for Determination of Respiration Rate in Human Being using A New Scheme for Determination of Respiration Rate in Human Being using MEMS Based Capacitive Pressure Sensor : Simulation Study," in *Proceedings of the 8th International Conference on Sensing Technology*, 2014, no. September, pp. 2–7.
- [12] P. Sharma, S. S. Member, X. Hui, E. C. Kan, and S. S. Member, "A Wearable RF Sensor for Monitoring Respiratory Patterns * Rx," *2019 41st Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, pp. 1217–1223, 2019.
- [13] F. Adib, H. Mao, Z. Kabelac, D. Katabi, and R. C. Miller, "Smart homes that monitor breathing and heart rate," in *Conference on Human Factors in Computing Systems - Proceedings*, 2015, vol. 2015-April, pp. 837–846.
- [14] B. Ibrahim, A. Akbari, and R. Jafari, "A novel method for pulse transit time estimation using wrist bio-impedance sensing based on a regression model," in *Biomedical Circuits and Systems Conference (BioCAS), 2017 IEEE*, 2017, pp. 1–4.
- [15] B. Ibrahim and R. Jafari, "Cuffless Blood Pressure Monitoring from an Array of Wrist Bio-impedance Sensors using Subject-Specific Regression Models: Proof of Concept," *IEEE Trans. Biomed. Circuits Syst.*, 2019.
- [16] K. Sel, J. Zhao, B. Ibrahim, and R. Jafari, "Measurement of Chest Physiological Signals using Wirelessly Coupled Bio-Impedance Patches," in *2019 41th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, 2019, p. (accepted).
- [17] *Medical electrical equipment Part 1: General requirements for basic safety and essential performance ANSI/AAMI ES60601-1:2005/A1:2012*.
- [18] K. Sel, D. Kireev, A. Brown, B. Ibrahim, D. Akinwande, and R. Jafari, "Electrical Characterization of Graphene-based e-Tattoos for Bio-Impedance-based Physiological Sensing," in *2019 IEEE Biomedical Circuits and Systems Conference (BioCAS)*, 2019, pp. 1–4.
- [19] B. Ibrahim, D. A. Hall, and R. Jafari, "Bio-impedance spectroscopy (BIS) measurement system for wearable devices," in *Biomedical Circuits and Systems Conference (BioCAS), 2017 IEEE*, 2017, pp. 1–4.
- [20] S. Yang *et al.*, "'Cut-and-Paste' Manufacture of Multiparametric Epidermal Sensor Systems," *Adv. Mater.*, vol. 27, no. 41, pp. 6423–6430, Nov. 2015.
- [21] T. Ha *et al.*, "A Chest-Laminated Ultrathin and Stretchable E-Tattoo for the Measurement of Electrocardiogram, Seismocardiogram, and Cardiac Time Intervals," *Adv. Sci.*, vol. 6, no. 14, Jul. 2019.
- [22] G. Wallgren, P. Karlberg, and J. Lind, "Studies of the Circulatory Adaptation Immediately after Birth," *Acta Paediatr.*, vol. 49, no. 6, pp. 843–849, Nov. 1960.
- [23] E. O. Attinger, "Some interrelations between ventilation and circulation," *Am. Heart J.*, vol. 54, no. 6, pp. 837–850, 1957.
- [24] D. F. Heiman, S. Rodbard, A. B. Shaffer, and G. L. Snider, "Respiratory factors affecting pulmonary arterial blood pressure and flow through the lungs," *J. Appl. Physiol.*, vol. 10, no. 1, pp. 31–36, 1957.
- [25] M. Harasawa and S. Rodbard, "Ventilatory air pressure and pulmonary vascular resistance," *Am. Heart J.*, vol. 60, no. 1, pp. 73–79, 1960.
- [26] Z. F. Issa, J. M. Miller, and D. P. Zipes, *Clinical Arrhythmology and Electrophysiology: A Companion to Braunwald's Heart Disease: Second Edition*. Elsevier Inc., 2012.
- [27] Z. F. Issa, J. M. Miller, and D. P. Zipes, *Clinical Arrhythmology and Electrophysiology: A Companion to Braunwald's Heart Disease*. Elsevier Inc., 2009.
- [28] B. Ibrahim, D. A. Hall, and R. Jafari, "Bio-impedance Simulation Platform using 3D Time-Varying Impedance Grid for Arterial Pulse Wave Modeling," *2019 IEEE Biomed. Circuits Syst. Conf.*, pp. 1–4, Oct. 2019.
- [29] R. B. Berry *et al.*, "Rules for scoring respiratory events in sleep: Update of the 2007 AASM manual for the scoring of sleep and associated events," *J. Clin. Sleep Med.*, vol. 8, no. 5, pp. 597–619, 2012.
- [30] T. Brack *et al.*, "Daytime Cheyne-Stokes respiration in ambulatory patients with severe congestive heart failure is associated with increased mortality," *Chest*, vol. 132, no. 5, pp. 1463–1471, 2007.
- [31] P. A. Lanfranchi *et al.*, "Prognostic value of Nocturnal Cheyne-Stokes respiration in chronic heart failure," *Circulation*, vol. 99, no. 11, pp. 1435–1440, Mar. 1999.