A compensation scheme for non-ideal circuit effects in biomedical impedance sensor

Hong, Yan; Goh, Wang Ling; Wang, Yong

2018

Hong, Y., Goh, W. L., & Wang, Y. (2018). A compensation scheme for non-ideal circuit effects in biomedical impedance sensor. *Analog Integrated Circuits and Signal Processing*, 95(3), 473-480. doi:10.1007/s10470-018-1182-9

<https://hdl.handle.net/10356/107572>

<https://doi.org/10.1007/s10470-018-1182-9>

This is a post-peer-review, pre-copyedit version of an article published in *Analog Integrated Circuits and Signal Processing*. The final authenticated version is available online at:
<http://doi.org/10.1007/s10470-018-1182-9>

Downloaded on 26 Apr 2023 20:31:38 SGT

A Compensation Scheme for Non-ideal Circuit Effects in Biomedical Impedance Sensor

Yan Hong^{1,2,3}, Wang Ling Goh², Yong Wang^{1,4}

Abstract The accuracy of an I/Q based biomedical impedance sensing sensor (IQBIS) suffers significantly from the PVT effects of the analog front-end, such as the amplitude errors of the stimulation signals, gain mismatches, amplitude and phase imbalances of in-phase (I) and quadrature (Q) signals, etc. These practical effects will severely impede the system performance if handled improperly. In this paper, the degradations of sensing performance by such imperfections are mathematically analyzed and quantified. Following theoretical studies, a training-based digitally controlled correction approach is proposed to finely alleviate these impairments. The performance of the proposed scheme had been verified using Simulink and MATLAB. With the proposed error correction scheme, the accuracy is improved by at least 17 times compared to that of the typical IQBIS, for both real and imaginary values of impedance. Thus, the proposed method is very useful for IQBIS, in resisting degradation in sensing accuracies due to the process-voltage-temperature (PVT) effects.

Keywords biomedical application • impedance sensor • non-ideal effects • compensation

I. INTRODUCTION

Micro-electrical stimulation and recording circuits are indispensable for neurological/physiological studies and clinical treatments. Typical examples can be found in deep brain stimulation [1], retinal/cochlear prosthesis [2,3], brain-to-brain interface, functional electrical stimulation [4], peripheral/vagus nerves [5,6], bladder control prosthesis [7], etc. The major problem accountable for performance degradation and function failure in biomedical stimulation applications is the variation of the electrode impedance, induced by tissue growth, inflammation, electrode erosion, etc. In a stimulation system, the change in electrode-tissue

interface impedance has weighty influence on the efficiency of the current delivery, the device battery life, the stimulation thresholds, and more [8,9]. Therefore, accurate monitoring of the electrode impedance is highly important in a stimulation system (see Fig. 1), to provide key and timely information for self-optimization of the measurement condition.

Impedance sensing [10-15] has been widely used in electrical bio-impedance measurement at the electrode-electrolyte interface in biomedical applications. The most widely used method is the I/Q demodulation based impedance sensing (IQBIS) [10-12]. A typical I/Q based impedance measurement circuit includes a sinusoidal wave synthesizer, a programmable amplifier (PA) and a local oscillator (LO).

Many remarkable works have been reported recently to solve various issues of IQBIS based systems. They are: the harmonic distortion reduction method in wide-range programmable sinusoidal wave synthesizer [16], the bio-impedance system that solves specific problems when IQBIS is used for wrist-worn heartbeat monitoring [17], the electrical impedance tomography technique that can acquire images of breast cancer [18], a battery-free bio-impedance sensor given for implantable application [19], a field-programmable gate array (FPGA) based electrical impedance tomography which can simultaneously measure different frequencies [20], etc.

There are however many circuitry imperfections that affect the accuracy when using IQBIS technique for bio-impedance measurement. Such imperfections are mainly due to the usual supply voltage variations, environmental temperature change and the variation in the fabrication process, etc. From circuits perspective, these imperfections are amplitude errors of sinusoid signals and the nonlinear characteristics of the PAs. In addition, the I/Q mismatch (i.g. I/Q amplitude and phase imbalances in LO signals) are important issues since they destroy the I/Q orthogonality and produce unwanted interference between signal paths. All these imperfections should be corrected to ensure system performance.

In this paper, the quadrature errors of LOs and other circuitry imperfections are studied. First, an I/Q demodulation model is created. The effects of sinusoidal current amplitude error, gain nonlinearity of PA and I/Q amplitude and phase imbalance to both its real and imaginary values of impedance are mathematically quantified. Thereafter, a training-based correction approach is proposed to compensate the bio-impedance measurement. The proposed structure is verified through simulation results obtained from Simulink analysis. With the proposed error correction

Author to whom correspondence should be addressed. Electronic mail: wangyongieee@gmail.com

1 University of Electronic Science and Technology of China, Chengdu, China

2 Nanyang Technological University, Singapore, Singapore

3 Singapore University of Technology and Design, Singapore, Singapore

4 IHP - Innovations for High Performance Microelectronics, Frankfurt (Oder), Germany

Fig. 1 Block diagram of a typical neural recording and stimulation system.

scheme, the accuracy for both the real and imaginary parts of the impedance values is improved by at least 17 times compared to that of the typical IQBIS. This work is hence very useful for IQBIS applications.

II. I/Q SIGNAL MODEL IN IMPEDANCE SENSING AND IMPERFECTIONS CORRECTION

A. Mathematic Model of Ideal IQBIS

Fig. 2 shows the typical I/Q demodulation impedance measurement scheme. Firstly, a small amplitude and variable frequency sinusoid current signal I_{stim} is generated by a voltage-controlled current source (VCCS). The sinusoid current is then injected to the electrode-electrolyte impedance interface. The amplitude of the ac current is maintained constant during the measurement. For biomedical applications, the maximum current amplitudes and operation frequencies need be optimized to comply with the medical safety guidelines. Passing through the impedance interface, the ac current is distorted, and a complex voltage signal (V_{in}) is obtained at the output of the impedance interface according to Ohm's law. The phase and amplitude of the voltage signal contain information on both the real and imaginary parts of the impedance. V_{in} can be very small; hence a voltage amplifier is needed to boost the signal strength so that the

voltage is strong enough for subsequent processing. To extract the real and imaginary parts of the impedance, V_{in} is split into two paths and fed into two mixers. The in-phase (I) and quadrature-phase (Q) signals are separately applied to two mixers as local oscillation (LO) in-phase and quadrature (I/Q) signals, V_{iLO} and V_{qLO} , where both are at the same frequency of the injected current. After mixing, the response voltage V_{in} is modulated into dc signal accompanied with higher frequency components. Only the dc signal is useful for the extraction of impedance values. Therefore, low pass filters (LPFs) are used to remove all other frequency components, leaving only the dc components. The obtained dc components are related to both the real or imaginary parts of the impedance. Finally, the dc components are digitized by the analog-to-digital converters (ADCs).

To make the derivation intuitively, the amplitude of the I/Q signal is assumed to be unity. The injected current, I_{stim} and I/Q components, V_{iLO} and V_{qLO} , in ideal case can be expressed as:

$$I_{stim} = I_0 e^{j\omega t} \text{ and } Z = |Z| e^{j\varphi} \quad (1a)$$

$$\begin{bmatrix} V_{iLO} \\ V_{qLO} \end{bmatrix} = \begin{bmatrix} \cos(\omega t) \\ \sin(\omega t) \end{bmatrix} \quad (1b)$$

where $\omega = 2\pi f_0$, I_0 and f_0 are the amplitude and frequency of the injected current, respectively. Z is the electrode-tissue interface impedance, φ is the phase angle of impedance and j is the imaginary number $\sqrt{-1}$.

After demodulation by the mixers (shown raw I/Q signal in Fig. 2), the real and imaginary parts of V_{in} can be derived as:

$$\begin{bmatrix} V_i(t) \\ V_q(t) \end{bmatrix} = I_{stim} \times Z \times k \times \begin{bmatrix} V_{iLO} \\ V_{qLO} \end{bmatrix} = I_0 |Z| k e^{j(\omega t + \varphi)} \times \begin{bmatrix} \cos(\omega t) \\ \sin(\omega t) \end{bmatrix} \quad (2)$$

where k is the gain of amplifier, $V_i(t)$ and $V_q(t)$ are the real

Fig. 2 Typical I/Q demodulation based biomedical impedance sensing system.

and imaginary parts of V_{in} after demodulation by the mixers, respectively.

A low pass filter (LPF) is used to filter $V_i(t)$ and $V_q(t)$ to remove the alias products and then an ADC is used to convert them to digital signals. In ideal case, the complex signals after filtering by the LPFs are

$$\begin{bmatrix} V_i \\ V_q \end{bmatrix} = \begin{bmatrix} \frac{1}{2} k I_0 |Z| \cos \varphi \\ \frac{1}{2} k I_0 |Z| \sin \varphi \end{bmatrix} \quad (3)$$

Hence, the real part, Z_r , and imaginary part, Z_i , of the impedance can be obtained as

$$\begin{bmatrix} Z_r \\ Z_i \end{bmatrix} = \begin{bmatrix} |Z| \cos \varphi \\ |Z| \sin \varphi \end{bmatrix} = \frac{2}{k I_0} \begin{bmatrix} V_i \\ V_q \end{bmatrix} \quad (4)$$

which are linear equations to the values of V_i and V_q . This implies that the errors of Z_r and Z_i can be represented by the errors of V_i and V_q , respectively.

B. Mathematic Study of Non-ideal IQBIS

In fact, many non-ideal effects exist in CMOS circuits, which lead to impairments of the measurement accuracy. The non-ideal effects can be broadly categorized into three aspects: the amplitude error in sinusoid current, the nonlinearity of PA, phase and gain imbalances introduced by the mixer between the LO I/Q signals. In this subsection, the effects of these imperfections have been modeled mathematically in IQBIS and are described as follow.

The I_{stim} and gain of PA in practical application can be expressed as:

$$I_{stim}' = (1 + \alpha) I_0 \cos(\omega t) + i_0 \quad \text{and} \quad k' = k_f \quad (5)$$

Here, α is the total ac amplitude deviation of injected current, i_0 is the dc biasing component of ac current and k_f is the gain of PA at frequency f_0 . After using β and θ to model the amplitude and phase mismatch of the LO I/Q signal, respectively, the I/Q components can be rewritten as V_{iLO}' and V_{qLO}' , where

$$\begin{aligned} \begin{bmatrix} V_{iLO}' \\ V_{qLO}' \end{bmatrix} &= \begin{bmatrix} (1 + \frac{\beta}{2}) \cos(\omega t + \frac{\theta}{2}) \\ (1 - \frac{\beta}{2}) \sin(\omega t - \frac{\theta}{2}) \end{bmatrix} \\ &= \begin{bmatrix} (1 + \frac{\beta}{2}) \cos(\frac{\theta}{2}) & -(1 + \frac{\beta}{2}) \sin(\frac{\theta}{2}) \\ -(1 - \frac{\beta}{2}) \sin(\frac{\theta}{2}) & (1 - \frac{\beta}{2}) \cos(\frac{\theta}{2}) \end{bmatrix} \times \begin{bmatrix} \cos(\omega t) \\ \sin(\omega t) \end{bmatrix} \end{aligned} \quad (6)$$

Similarly, after demodulation by the mixers and filtering by the LPFs, the output voltages in practical situation can be deduced as

$$\begin{aligned} \begin{bmatrix} V_i' \\ V_q' \end{bmatrix} &= (1 + \alpha) \frac{k_f}{k} \times \begin{bmatrix} (1 + \frac{\beta}{2}) \cos(\frac{\theta}{2}) & -(1 + \frac{\beta}{2}) \sin(\frac{\theta}{2}) \\ -(1 - \frac{\beta}{2}) \sin(\frac{\theta}{2}) & (1 - \frac{\beta}{2}) \cos(\frac{\theta}{2}) \end{bmatrix} \\ &\quad \times \begin{bmatrix} \frac{1}{2} k I_0 |Z| \cos \varphi \\ \frac{1}{2} k I_0 |Z| \sin \varphi \end{bmatrix} \end{aligned} \quad (7)$$

This expression can then be simplified to

$$\begin{bmatrix} V_i' \\ V_q' \end{bmatrix} = A \begin{bmatrix} V_i \\ V_q \end{bmatrix} \quad (8a)$$

where

$$A = (1 + \alpha) \frac{k_f}{k} \times \begin{bmatrix} (1 + \frac{\beta}{2}) \cos(\frac{\theta}{2}) & -(1 + \frac{\beta}{2}) \sin(\frac{\theta}{2}) \\ -(1 - \frac{\beta}{2}) \sin(\frac{\theta}{2}) & (1 - \frac{\beta}{2}) \cos(\frac{\theta}{2}) \end{bmatrix} \quad (8b)$$

Equations 8(a) and 8(b) represent the mathematical models for non-ideal situation, to define the imperfections effects that cannot be ignored. The dc bias current component, i_0 , is modulated to f_0 frequency by I/Q signals, and it can be easily cancelled after going through the LPF, hence it is not expressed in matrix of A. The total ac amplitude deviation α of injected current and amplifier gain fluctuation k_f play a linear relationship to the imperfections effects. The phase mismatch θ plays a significant role since it introduces crosstalk between the signals in I and Q signals. Hence, a correction method is needed so that these practical effects can be compensated. A training-based structure that considers all four effects of imperfections is proposed in the next section.

Fig. 3 Schematic diagram of proposed correction approach for non-ideal effects in IQBIS.

C. Proposed Correction Method

It can be found that

$$A^{-1} \begin{bmatrix} V_i' \\ V_q' \end{bmatrix} = A^{-1} A \begin{bmatrix} V_i \\ V_q \end{bmatrix} = \begin{bmatrix} V_i \\ V_q \end{bmatrix} \quad (9)$$

To calibrate the errors described in section B, the raw I/Q signal matrix under small imperfections can be multiplied by the inverse matrix of A:

$$A^{-1} = \frac{A^*}{|A|} \approx \frac{k}{k_f(1+\alpha)(1-\frac{\beta}{2})(1+\frac{\beta}{2})} \begin{bmatrix} (1-\frac{\beta}{2}) & \frac{\theta}{2} \\ \frac{\theta}{2} & (1+\frac{\beta}{2}) \end{bmatrix} \quad (10)$$

Since $\frac{\theta}{2}$ is very small, here we assume that

$$\sin(\frac{\theta}{2}) = \frac{\theta}{2} \quad (11a)$$

$$\cos(\frac{\theta}{2}) = 1 \quad (11b)$$

The calibrations of the four major imperfections (i.e. α , k_f / k_f , β , θ) are implemented by the matrix A^{-1} . The above situation is observed again in the following simulation results in Section III.

Thus, the calibration coefficient $C = A^{-1}$, can be represented as

$$C = A^{-1} = \begin{bmatrix} C_{ii} & C_{qi} \\ C_{iq} & C_{qq} \end{bmatrix} \quad (12)$$

where,

Fig. 4 MATLAB/Simulink diagram of proposed correction approach for non-ideal effects in IQBIS.

$$\begin{bmatrix} V_{ic} \\ V_{qc} \end{bmatrix} = \begin{bmatrix} C_{ii} & C_{qi} \\ C_{iq} & C_{qq} \end{bmatrix} \begin{bmatrix} V_i' \\ V_q' \end{bmatrix} \quad (13)$$

For each existing chip, the four imperfections (i.e. α , k_f , β , θ) are fixed at a certain value. This implies that a matrix of four PAs can be used for the correction, as shown in Fig. 3. The gain of these PAs can be controlled by four digital signals. In practical application, we can use a training-based approach to obtain the suitable gain of four PAs with a known device under testing since the non-idea effects are fixed for a certain chip after tapeout. Derived from Eqs. (3), (7) and (12), the percentage error of the real and imaginary parts of the impedance can be described as:

$$\begin{bmatrix} E_i \\ E_q \end{bmatrix} = \begin{bmatrix} \left| \frac{V_i'}{V_i} - 1 \right| \\ \left| \frac{V_q'}{V_q} - 1 \right| \end{bmatrix} \quad \text{and} \quad \begin{bmatrix} E_{ic} \\ E_{qc} \end{bmatrix} = \begin{bmatrix} \left| \frac{V_{ic}}{V_i} - 1 \right| \\ \left| \frac{V_{qc}}{V_q} - 1 \right| \end{bmatrix} \quad (14)$$

Here, E_i and E_q are the errors before correction, while E_{ic} and E_{qc} are the errors after correction.

III. RESULTS AND DISCUSSION

To verify the performance of the correction method of the proposed impedance measurement system as discussed in Section II, we analyzed the errors of the real and imaginary impedance values within the bio-impedance range of the neural stimulation/recording. The equivalent simplified

Fig. 5 Errors in both the impedance real and imaginary parts related to imperfections: (a) impedance real part before correction, (b) impedance real part after correction, (c) impedance imaginary part before correction, (d) impedance imaginary part after correction.

Randle model [21, 22] of the neural-electrode bio-impedance is shown in Fig. 2, whereby it contains R_s , R_p and C_s . It can be found that the impedance angle for the neural-electrode bio-impedance is at the range of -90° to 0° .

Different impedance values within the impedance range have been selected for simulation. Due to the process variation and device mismatch in IC manufacturing, the maximum gain error is less than 10% and the maximum phase mismatch is less than 5° [23]. All the parameters used in the simulation are given in Table I. The parameters selected are for the worst case in practical biomedical impedance sensors and are provided merely as an example.

Table 1 Parameters of Simulink Model

Test	α	k_f/k	β	θ	f
	0.1	1.1	0.1	5°	1 kHz

Two IQBIS structures are set up in MATLAB/Simulink. They are the ideal IQBIS without any non-ideal effects, the non-idea IQBIS without/with correction (shown in Fig. 4), respectively. The errors are obtained at different angles of the Randle model. The LPF used in the simulation is with bandwidth of 10 Hz.

As shown in Figs. 5 (a) and (c), the accumulated effects of these imperfections play a huge role in the accuracy of impedance sensing, and thus, it has immense potential compensating the non-ideal effects. Meanwhile, the error ratios of both the real and imaginary parts are not the same at different impedance angles. For example, the error ratio of the real part is even more pronounced for higher impedance angle.

The corresponding errors, both with and without corrections are also shown in Fig. 5. Real and imaginary part errors are reduced greatly after deploying the correction method. For the real part, the error percentage reduced from 27.3%~65.2% to 0.26%~3.8%, (see Figs. 5(a) and 5(b)), the error has significantly decreased by 105 times to 17.2 times at different impedance phases. When comes to the imaginary part (refer to Figs. 5(c) and 5(d)), the percentage error falls from 14.5%~80.3% to 0.28%~3.12%, and this cut down the percentage error by at least 26 times. This indicates that the proposed compensation technique has alleviated the imperfection impairments to the real and imaginary parts of impedance sensing, to improve the accuracy significantly.

IV. CONCLUSION

In this paper, a mathematical model of the non-ideal effects in IQBIS is analyzed. The non-ideal effects include the magnitude error of the injected current, gain mismatch in PA and the I/Q signal imbalance from LO. Following that, a simple training-based digitally controlled correction approach for compensation is explored, which uses four PAs to correct the non-ideal effects in IQBIS. The suitable gain of

these four PAs can be obtained with a known device under testing. MATLAB/Simulink results showed that the explored model can improve the accuracy of both the real and imaginary parts by at least 17 times in all impedance phases. This demonstrates that the proposed digitally controlled correction approach structure has good compensation performance for impedance sensing in biomedical applications.

References

1. J. Lee, H. G. Rhew, D. R. Kipke and M. P. Flynn. (2010). A 64 Channel Programmable Closed-Loop Neurostimulator With 8 Channel Neural Amplifier and Logarithmic ADC. *IEEE J. Solid-State Circuits*, 45(9), 1935 - 1945.
2. K. Chen, Z. Yang, L. Hoang, J. Weiland, M. Humayun and W. Liu. (2010). An Integrated 256-Channel Epiretinal Prosthesis. *IEEE J. Solid-State Circuits*, 45(9) 1946 - 1956.
3. Kwok Roberta. (2013). Neuroprosthetics: Once more, with feeling. *Nature*, 497 (7448), 176–178.
4. Matei, E. Lee, J. Gord, P. Nercessian, P. Hess, H. Stover, T. Li and J. Wolfe. (2009). A Biomedical Implantable FES Battery-Powered Micro-Stimulator. *IEEE Transactions on Circuits and Systems I: Regular Papers*, 56(12) 2583 – 2596.
5. S. Lee, S. Sheshadri, Z. Xiang, I. Delgado-Martinez, N. T. Sun, N. Thakor, S. Yen, C. Lee. (2015). Selective stimulation and neural recording on peripheral nerves using flexible split ring electrodes. *Sensors and actuators B-chemical*, 242, 1165-1170.
6. D. R. Hulsey, J.R. Riley, K. W. Loerwald, R. L. Rennaker, M. P. Kilgard, S. A. Hays. (2017). Parametric characterization of neural activity in the locus coeruleus in response to vagus nerve stimulation. *Experimental Neurology*, 289, 21-30.
7. Tao W.-J., Li F., Wang C.-H., Tu G.-M., Wang Z.-G., Lu X.-Y. (2017). Design of integrated neural stimulating and recording frontend for bladder control prosthesis. *Analog Integrated Circuits and Signal Processing*, 91(3), 403-416.
8. D. Loi, C. Carboni, G. Angius, G. N. Angotzi, M. Barbaro, L. Raffo, S. Raspopovic and X. Navarro. (2011). Peripheral Neural Activity Recording and Stimulation System. *IEEE J. transactions on biomedical and systems*, 5(4), 368 - 379.
9. Y. Hong, Y. Wang, W. L. Goh, Y. Gao and L. Yao. A current-excited triple-time-voltage oversampling method for bio-impedance model for cost-efficient circuit system. In *IEEE 2015 37th International Conference of the Engineering in Medicine and Biology Society*.
10. Jia Xu, Xiaomeng Gao, Alexander Lee, Shuhei Yamada, Ehsan Yavarim, Victor Lubecke, and Olga Boric-Lubecke. Wrist-worn heartbeat monitoring system based on bio-impedance. In *IEEE 2016 38th Annual International Conference of the Engineering in Medicine and Biology Society*.
11. P. Kassanos, I. F. Triantis, and A. Demosthenous. (2013). A CMOS magnitude/phase measurement chip for impedance spectroscopy. *IEEE Sensors Journal*, 13(6), 2229-2236.
12. Uranga, J. Sacristan, T. Oses, N. Barniol. (2007). Electrode–Tissue Impedance Measurement CMOS ASIC for Functional Electrical Stimulation Neuroprostheses. *IEEE Transaction of Instruction and Measurement*, 56(5), 2043-2050.
13. Panagiotis Kassanos, Loucas Constantinou, Iasonas F. Triantis, and Andreas Demosthenous. (2014). An integrated analog readout for multi-frequency bioimpedance measurements. *IEEE Journal Sensors*, 14(8), 2792–2800.
14. T. Motomura, Y. Kasashima, O. Fukuda, F. Uesugi, H. Kurita, and N. Kimur. (2014). Practical monitoring system using characteristic impedance measurement during plasma processing. *Review of Scientific Instruments*, 85 (2).
15. T. Dargent, K. Haddadi, T. Lasri, N. Clément, D. Ducatteau, B. Legrand, H. Tanbakuchi, and D. Theron. (2013). An interferometric scanning

microwave microscope and calibration method for sub-fF microwave measurements. *Review of Scientific Instruments*, 84(123705).

16. Chia-Ling Wei, Yi-Wen Wang, and Bin-Da Liu. (2014). Wide-Range Filter-Based Sinusoidal Wave Synthesizer for Electrochemical Impedance Spectroscopy Measurements. *IEEE J. transactions on biomedical and systems*, 8(3), 442–450.
17. Jia Xu, Xiaomeng Gao, Alexander Lee, Shuhei Yamada, Ehsan Yavari, Victor Lubecke and Olga Boric-Lubecke (2016). Wrist-worn heartbeat monitoring system based on bio-impedance analysis. *IEEE 38th Annual International Conference of the Engineering in Medicine and Biology Society (EMBC)*, 6294–6297.
18. Yongsu Lee, Unsoo Ha, Kiseok Song, and Hoi-jun Yoo. 3.8 mW electrocardiogram (ECG) filtered electrical impedance tomography IC using I/Q homodyne architecture for breast cancer diagnosis. *2014 IEEE International Symposium on Circuits and Systems (ISCAS)*, 2389–2392.
19. Rodriguez, S. Ollmar, M.Waqar, and A.Rusu. (2016). A Batteryless Sensor ASIC for Implantable Bio-Impedance Applications. *IEEE Trans. on Biomedical Circuits and Systems*, 10(3), 533–544.S.
20. A. Santos, A. Robens, A. Boehm, S. Leonhardt and D. Teichmann. (2016). System Description and First Application of an FPGA-Based Simultaneous Multi-Frequency Electrical Impedance Tomography. *Sensors*, 16(8).
21. Y. Hong, Y. Wang, W. L. Goh (2017). Note: Hybrid- π model and parameter extraction method for electrode-electrolyte interface characterization with superbly accurate reactance. *Review of Scientific Instrument*, 88, 086106.
22. D. Boinagrov, X. Lei, G. Goetz, T. Kamins, K. Mathieson, L. Galambos, *et al.*, "Photovoltaic Pixels for Neural Stimulation: Circuit Models and Performance," *IEEE Transactions on Biomedical Circuits and Systems*, vol. 10, pp. 85-97, Jan. 2016.
23. X. Zhang, L.E. Larson, P.M. Asbeck, P. Nanawa (2001). Gain/phase imbalance-minimization techniques for LINC transmitters. *IEEE Transactions on Microwave Theory and Techniques*, pp. 2507 – 2516.

student and 43 M.Sc. students. She is currently supervising 10 PhD students, 1 M.Eng. student, and 2 M.Sc. student.

Yong Wang Yong Wang obtained B.Eng from Harbin Institute of Technology, China, in 2010; M.Eng and Ph.D degrees from the Nanyang Technological University (NTU), Singapore, in 2012 and 2015, respectively. While working on his thesis, he was also an attachment research engineer at Institute of Microelectronics, Agency for Science, Technology and Research (A*STAR), Singapore. From 2012 to 2017, he was with NTU as research associate, research fellow and a leader for a FMCW radar system on chip project. From 2017 to 2018, he was a scientist and leader for terahertz sensing system-on-chip project at IHP, Innovations for High Performance Microelectronics, an institute of the Leibniz Association, Germany. He is now serving as a full professor at University of Electronic Science and Technology of China. His research interests include millimeter-wave FCMW radar transceivers, sensor ICs, terahertz wireless transceiver ICs, FMWC radar systems and algorithms, biomedical sensor ICs, device modelling, surface acoustic resonators, etc. He has authored and co-authored over 30 conference and journal papers and served as reviewers for many prestigious journals and conferences. He received the 2016 ISSCC Student Travel Grant Award (STGA), and the Best Speaker Award at IEEE Circuits and Systems Singapore Chapter Graduated Students Workshop 2015. In 2018, he got enrolled into the Recruitment Program for Young Professionals of “The Thousand Talents Plan” in China.

Yan Hong received the B.Eng. and M.Eng. degrees from the Southeast University, Nanjing, China, in 2008 and 2011, respectively. She worked at ZTE Corporation, China, from 2011 to 2012 and then at National University of Singapore from 2012 to 2013. Since 2013, she has been working toward the Ph.D. degree at School of Electrical and Electronic Engineering, Nanyang Technological University (NTU), Singapore. She has been also attached to the Institute of Microelectronics (IME) during this period. Her

research focuses on CMOS biomedical Sensor design for personalized diagnosis and treatment. She received the 2016 Symposium VLSI Circuit Student Travel Grant Award (STGA) and 2017 IEEE CAS Travel Grant Funds.

Wang Ling Goh received the B.Eng. degree in electrical and electronic engineering and Ph.D. degree in microelectronics from Queen's University of Belfast, Belfast, U.K. in 1990 and 1995, respectively. She was a Research Engineer at the Northern Ireland Semiconductor Research Centre while working toward the Ph.D. degree. She joined the school of electrical and electronic engineering, Nanyang Technological University, Singapore, as a Lecturer in 1996, and became an Associate Professor in 2004. Her research interests

include digital/mixed-signal IC design, telemetry circuits, neural recording ICs, and 3-D IC.

Dr Goh has co-authored 1 international professional technical reference text, filed 16 patents, and published ~160 research papers in international journals and conferences. Since 1996, Dr Goh has trained a total of 84 graduate students: 17 Ph.D. students, 3 Ph.D. Interns, 20 M.Eng. students, 1 M.Phil.