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14. ABSTRACT The primary objective of this research is to combine advanced integrated electronics with transrectal electrical impedance tomography (TREIT) and electrical impedance sensing biopsy (EIS-Bx) devices to improve TRUS-guided prostate biopsy procedures and help to address the challenge of distinguishing aggressive from indolent disease in men newly diagnosed with prostate cancer. The novelty in this program falls into two categories: 1) we aim to couple our single point EIS-Bx measurement device with an array of electrodes adhered to a standard end-fired TRUS probe (Halter's team) and 2) we aim to develop a custom application specific integrated circuit (ASIC) that will enable us to measure these electrical properties over a wide frequency range (optimal for discriminating high from low-grade prostate cancer) and at high frame rates (required for real-time prostate biopsy procedures) (Odame's team). During the first two years of this study, we focused on designing, implementing and validating (1) a small form-factor EIT based on a custom ASIC chip (2) an EM-tracking approach to capture the surface of the prostate during TRUS, (3) finite element method meshing approaches and reconstruction algorithms for imaging the electrical properties of prostate during an TREIT + EIS-Bx procedure. Year 3 focused on further phantom studies and the development of a second version (20-channel) of a minaturized EIT system. A no-cost-extension period of Year 4 will focus on clinical data acquisition from a series of ex vivo human prostates.					
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INTRODUCTION

No clinical test or imaging modality is currently able to accurately discriminate between aggressive and indolent prostate cancer. As a result, there are a significant number of men unnecessarily treated with radical, life-altering, therapies including external beam radiation, brachytherapy, and radical prostatectomy. The primary reason for this is that the definitive method for diagnosing prostate cancer and determining treatment strategies depends primarily on histological assessment of tissue cores obtained during a transrectal ultrasound (TRUS) guided biopsy; a *typical procedure extracting 12 cores samples less than 0.5% of a typical prostate gland (~50cc)*. Because of the small volume of tissue sampled, it is difficult to determine with certainty how aggressive or indolent the disease is.

Electrical properties of tissue are largely morphology dependent. We have shown significant ($p < 0.001$) electrical property contrast between benign and malignant prostate in a number of published studies. This contrast arises because of large structural difference in tissue architecture between benign and malignant tissues. We have also shown significant electrical property differences between cancers of different Gleason grades. The sensitivity (SN) and specificity (SP) for distinguishing high (Gleason score > 7 , SN=0.93, SP=0.81) and low (Gleason score < 7 , SN=0.7, SP=0.8) prostate cancers are similar to that of magnetic resonance imaging based alternatives, currently the most promising technologies for grading disease.

This program is focused on enhancing a standard clinically used TRUS-guided biopsy procedure by imaging the electrical properties of prostate. The hypothesis is that these electrical property images will provide surrogate measures of disease aggressiveness. Our specific objective is to couple a dual-modality TRUS and electrical impedance tomography (EIT) system with an electrical property sensing biopsy instrument to better guide biopsy sampling and to provide improved staging criteria with more specific imaging studies. This program is specifically designed to augment clinically accepted devices (TRUS-guided biopsy) with the aim of rapidly translating this highly cancer specific technology to the clinic.

Much of the first year of this program has focused on designing and fabricating the appropriate hardware (electrode array, custom biopsy needles), implementing strategies to capture TRUS images and produce anatomically accurate meshes of the prostate, developing and optimizing image reconstruction algorithms for prostate imaging, and designing custom ASIC device specifically designed for transrectal EIT (TREIT) imaging of the prostate. The second year of the program primarily focused on finalizing our custom hardware, refining and optimizing our image reconstruction algorithms, and evaluating both the software and hardware in a series of electrical benchtop tests and phantom studies. Year 3 focused on further phantom studies and the development of a second version (20-channel) of a miniaturized EIT system. Personnel changes and the IRB approval process necessitated produced project delays and the necessity for a no-cost-extension. The no-cost-extension period of Year 4 will focus on clinical data acquisition from a series of ex vivo human prostates. We will continue to prepare publications and proposals focused on our technical developments and clinical data acquisition and evaluation.

BODY

The following research summary is presented in terms of the approved Statement of Work, with each task being discussed separately. When appropriate, detailed discussion is referenced to manuscripts published, submitted, or in preparation which are provided in the Appendix. All tasks have been accomplished at Dartmouth College's Thayer School of Engineering. Major Tasks are marked as KO (Kofi Odame) or RH (Ryan Halter) to denote the responsible PI. Note that future task and objectives to be completed are marked as **TBC**.

SPECIFIC AIM 1: TO DEVELOP NOVEL TREIT HARDWARE FOR PROSTATE IMAGING

Major Task 1: Design and fabricate an ASIC for EIT applications (KO)

a) Design and fabricate ASIC v1 w/ ADC, current drive, and multiplexing

Completed in 2015-16 reporting period.

b) Design and fabricate an optimized ASIC v2 w/ additional digital control circuitry

Completed in 2016-17 reporting period.

Task 1 Milestones:

1. ASIC v1 designed and fabricated – **Completed**
2. ASIC v2 designed and fabricated – **Completed**
3. First complete ASIC (v2) optimized for EIT – **Completed**

Major Task 2: Integrate EIT-ASIC into a complete small form-factor EIT system (KO)

a) Design small form-factor PCB to house the ASIC and support circuitry.

Completed in 2016-17 reporting period.

b) Write microcontroller code to control small form-factor EIT system

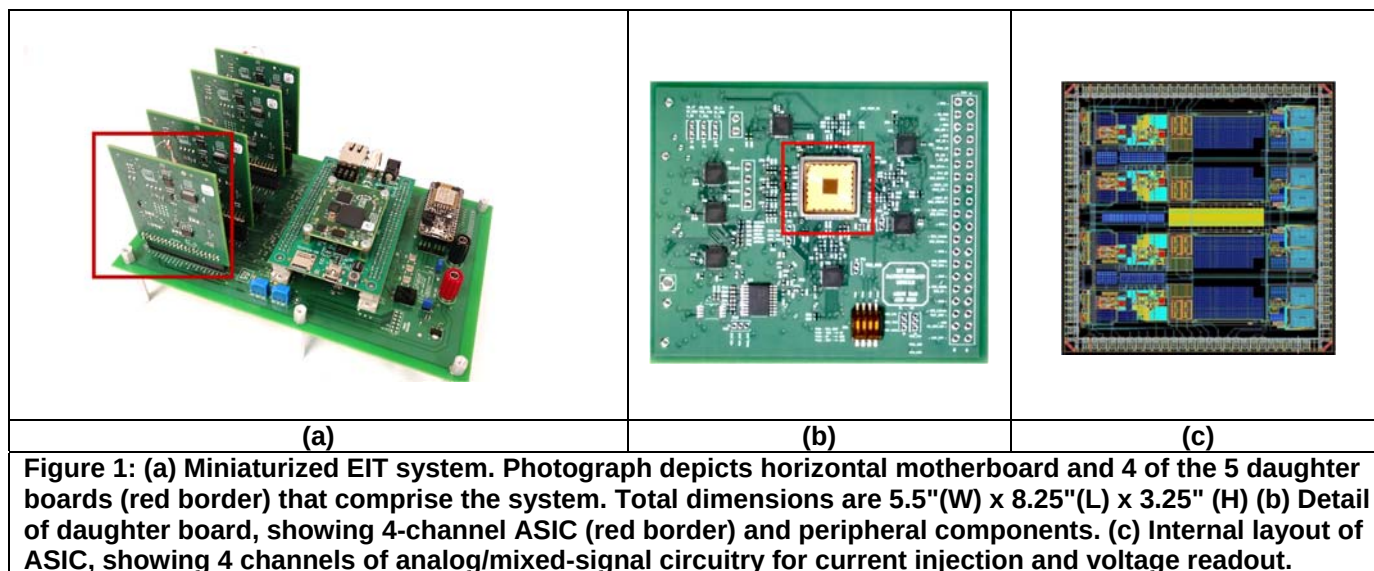
Microcontroller replaced with FPGA. Completed in 2016-17 reporting period.

c) Write code to communicate between EIT-system and laptop

Completed in 2016-17 reporting period.

d) Construct an enclosure to house the EIT system integrated to the TRUS probe

Enclosure will be constructed after the electronics of the 20 channel system have been validated. The dimensions of the envisioned enclosure are: 5.5"(W) x 8.25"(L) x 3.25" (H) (see Fig. 1).



Task 2 Milestones:

1. Fully functional small form-factor EIT system (month 18) – **Completed**

Major Task 3: Validate ASIC-based EIT system (KO)

a) Bench-top electrical evaluation of ASIC v1

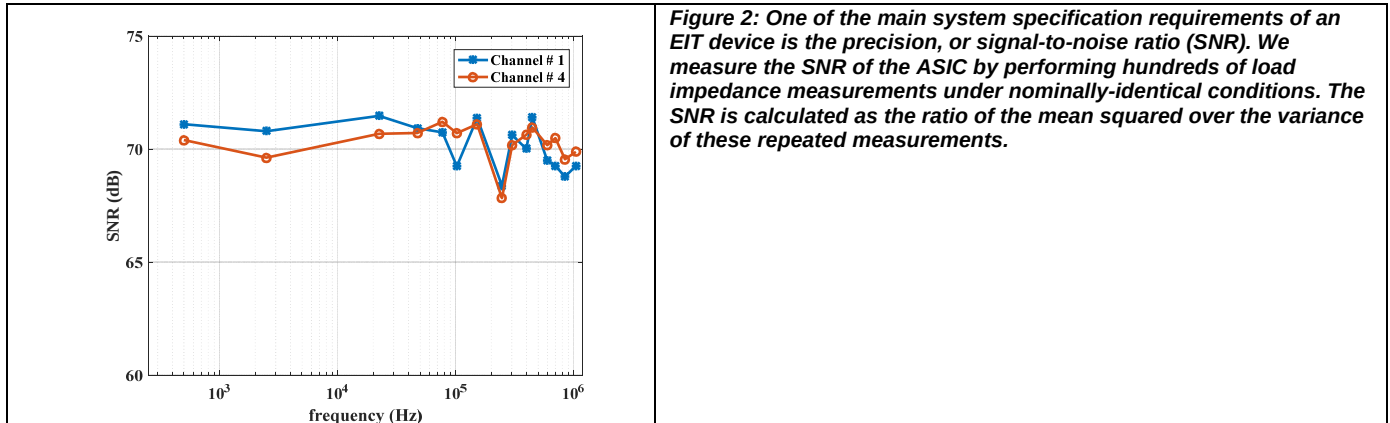
Completed in 2015-16 reporting period.

b) Bench-top evaluation of ASIC v2

ASIC v2 (4-channel readout/current injection) chip samples arrived in October 2017 and were evaluated using a similar testing protocol as for previous ASIC v1 (single channel readout/current injection), described in 2016-2017 report.

c) Bench-top evaluation of small form-factor ASIC-based EIT system

End-to-end evaluation of small form-factor ASIC based system was performed to measure the signal-to-noise ratio (SNR) of the system. Figure 2 depicts SNR for two representative channels, from 200 Hz to 1 MHz.



Task 3 Milestones:

1. Validation that our small form-factor ASIC-based EIT system provides sufficient electrical characteristics to enable EIT imaging – **Completed**
2. 1 peer-reviewed publication submitted – **Completed**

SPECIFIC AIM 2: TO DEVELOP TREIT IMAGING ALGORITHMS FOR PROSTATE IMAGING

Major Task 1: Optimize and manufacture TREIT electrode arrays and EIT-Bx needles (RH)

a) Design, fabricate, and evaluate TREIT electrodes arrays

Completed in 2015-16 reporting period.

b) Construct EIS-Bx needles

Primary tasks completed in 2015-16 reporting period.

Manufacturing optimizations completed in 2016-2017 period.

Task 1 Milestones:

1. TREIT electrode arrays fabricated for use in phantom and ex vivo experiments – **Completed (additional electrode array will be fabricated as needed)**
2. EIS-Bx needles fabricated for use in phantom and ex vivo experiments. – **Completed (additional EISBx devices will be fabricated as needed)**

Major Task 2: Develop 3D prostate meshing through EM-tracked TRUS (RH)

a) Integrate EM tracking with TRUS image acquisition

Completed in 2015-16 reporting period.

b) Write software to extract prostate point cloud from 3D tracked US images

Completed in 2015-16 reporting period.

c) Developing meshing algorithms to create 3D prostate mesh for TREIT reconstruction
Completed in 2015-16 reporting period.

Task 2 Milestones:

1. Hardware and software capable of producing a 3D FEM mesh of the prostate based on EM-tracked TRUS images – **Completed**

Major Task 3: Customize image reconstruction software for TREIT-based prostate biopsy (RH) a)

Write code to specify EIS-Bx needle electrodes within the 3D prostate mesh
Completed in 2016-2017 reporting periods.

b) Write code to couple probe surface and EIS-Bx electrodes in TREIT reconstructions
Completed in 2015-16 reporting period.

c) Validate the new meshing and reconstruction algorithms through software simulations
Completed in 2015-16 and 2016-2017 reporting periods.

Task 3 Milestones:

1. Validation of mesh generation code and TREIT + EIS-Bx image reconstruction code – **Completed**
2. 1 peer-reviewed publication submitted – **Completed – manuscript published in Physiological Measurement**

Major Task 4: Submit documents for IRB and MPMC HRPO approval (RH)

a) Draft and submit IRB protocol revisions for ex vivo study
Human studies protocol was submitted to our local IRB in September 2017.

b) Draft and submit documentation for MPMC HRPO approval
Documents were submitted to MPMC HRPO in February 2018.

Task 4 Milestones:

1. Obtain IRB and MPMC HRPO approval for ex vivo and in vivo cohorts – **Completed**

SPECIFIC AIM 3: TO VALIDATE TREIT THROUGH PHANTOM STUDIES

Major Task 1: Evaluate TREIT through use of liquid phantoms (RH)

a) Construct a tank for phantom studies
Completed in 2015-16 reporting period.

b) Perform saline based phantoms with bench-top EIT system (software validation)
Completed in 2016-17 reporting period.

c) Perform saline-based phantoms with new small form-factor EIT system (hardware validation)
We have leveraged our previously-developed EIT system to carry out these saline phantom experiments. The small form-factor EIT system phantom test will be completed during the no-cost-extension performance period.

Task 1 Milestones:

1. Validation that meshing and image reconstruction software functions as expected – **Completed**
2. Validation that ASIC-based small form-factor EIT system performs better than bench-top EIT systems – **TBC**

Major Task 2: Evaluate TREIT through use of anatomically accurate phantoms (KO/RH)

a) Perform anatomically accurate phantom imaging with bench-top EIT system (software validation)

System has been evaluated with anatomically-accurate phantoms studies.

b) Perform anatomically accurate phantom imaging with new small form-factor EIT system (hardware validation)

The small form-factor EIT system testing with anatomically-accurate phantom will be completed during the no-cost-extension performance period.

Task 2 Milestones:

1. *Validation of EIS-Bx + TREIT + TRUS procedure that will ultimately be used in ex vivo studies – **Completed***
2. *Validation that meshing and image reconstruction software functions as expected – **Completed***
3. *Validation that ASIC-based small form-factor EIT system performs better than bench-top EIT systems – **TBC***

SPECIFIC AIM 4: TO EVALUATE TREIT USING A SAMPLE OF EX VIVO PROSTATES

Major Task 1: Evaluate TREIT/EIT-Bx in a cohort of ex vivo prostates (RH)

The following tasks are to be completed during Year 4 no-cost-extension period of this program:

a) Record TRUS/TREIT (+ EIS-Bx) images of ex vivo prostates

b) Perform post-imaging pathological assessment of ex vivo prostates and biopsy cores

c) Statistically analyze TREIT-based images and pathological metrics

Task 1 Milestones:

1. *Preliminary clinical statistics defining utility of TREIT+EIS-Bx combined with TRUS – **TBC***
2. *Initial parameter threshold values for use in detecting and staging prostate cancer – **TBC***
3. *1 peer-reviewed publication submitted – **TBC***

KEY RESEARCH ACCOMPLISHMENTS

- Implemented EM-tracking and 3D surface extraction of prostate TRUS images
- Developed and implemented approaches to incorporate prior anatomic information into EIT reconstruction algorithms – 1st time this has ever been done.
- Developed and implemented approaches that enables us to incorporate electrical data recorded from multiple electrode positions and demonstrated that incorporating this additional data improves image resolution.
- Developed and validated a dual mesh approach for estimating the electrical properties of prostate.
- Designed and fabricated a custom ASIC that includes a current drive and measurement chain capable of driving and sensing signals up to 10 MHz – this represents the best-in-class device of this type.
- Designed and fabricated a 20-channel version of the small form-factor EIT system. End-to-end functionality has been validated, and phantom testing is underway.

PRODUCTS

Manuscripts

Rao, Arun, Yueh-Ching Teng, Chris Schaef, Ethan K. Murphy, Saaïd H. Arshad, Ryan J. Halter, and Kofi M. Odame. "An Analog Front End ASIC for Cardiac Electrical Impedance Tomography, A. Rao, Y. Teng, C. Schaef, E.," in IEEE Transactions on Biomedical Circuits and Systems, vol. 12, no. 4, p. 729-738, Aug. 2018.

Presentations/Abstracts

Murphy, Ethan K., Xiaotian Wu, and Ryan J. Halter. "Absolute Phantom Reconstructions using Fused-Data EIT," in Proceedings, 19th International Conference on Biomedical Applications of Electrical Impedance Tomography (EIT2018), Edinburgh, 2018.

Murphy, Ethan, Amoh, Justice, Arshad, Saaïd, Halter Ryan and Kofi Odame. "Deep Neural Networks for Noise-Robust Cardiac Output Monitoring," in Proceedings, 19th International Conference on Biomedical Applications of Electrical Impedance Tomography (EIT2018), Edinburgh, 2018.

PARTICIPANTS & OTHER COLLABORATING ORGANIZATIONS

Name:	Kofi Odame
Project Role:	Principal Investigator
Nearest person month worked:	1 (off-term); 0.36 (academic)
Contribution to Project:	Dr. Odame has overseen the design and testing of the EIT system (ASIC, printed circuit board, FPGA control), and has maintained communication with Dr. Halter and Dr. Halter's lab to coordinate on design requirements and performance specifications.

Name:	Arun Rao
Project Role:	Graduate Student
Nearest person month worked:	5 (calendar)
Contribution to Project:	Mr. Rao has designed, taped-out and tested the ASIC v1 and v2 chips. He has also constructed the small form-factor EIT system and integrated the various parts of the FPGA controller.

CONCLUSION

It is a daily challenge for clinicians to determine whether a man recently diagnosed with prostate cancer has aggressive disease requiring immediately radical therapy or indolent disease requiring a more passive watchful waiting or active surveillance approach. This program is focused on developing Transrectal Electrical Impedance Tomography (TREIT) specifically for prostate imaging. Over the past two years, we have developed several new image reconstruction algorithm that will be used to optimize the resolution and accuracy of TREIT. We have implemented a high-speed solution to capture prostate shape information during routine TRUS imaging. A number of phantom studies and experiments have been conducted to demonstrate feasibility of image acquisition and to validate and refine our image reconstruction algorithms. We have made significant progress in designing our custom ASIC for EIT imaging and have thus far produced and validated a 4-channel device that exceeds the specifications of any other impedance sensing device on the market or reported in literature. A prototype of our small form-factor, 20-channel EIT has been constructed, based on the success of our ASIC design. We are will validated and use this small form-factor EIT for phantom and clinical studies. In addition, we have refined our reconstruction algorithms to incorporate biopsy needle sensors to improve resolution and sensitivity, conducted several anatomically accurate phantom models to validate both hardware and software, and manufactured electrical impedance sensing biopsy needles. Preparation has begun for the ex vivo clinical studies planned for year 4.

Appendices

Appendix 1: IEEE TBioCAS Publication

Appendix 2: EIT2018 Publication 1

Appendix 3: EIT2018 Publication 2

An Analog Front End ASIC for Cardiac Electrical Impedance Tomography

Arun Rao , *Member, IEEE*, Yueh-Ching Teng , *Student Member, IEEE*, Chris Schaefer, *Member, IEEE*, Ethan K. Murphy , Saaid Arshad, Ryan J. Halter, *Member, IEEE*, and Kofi Odame, *Senior Member, IEEE*

Abstract—In this paper, an end-to-end CMOS application specific integrated circuit (ASIC) for readout channel in a cardiac electrical impedance tomography system is presented. The ASIC consists of an integrated current driver for current injection, an instrumentation amplifier, variable gain amplifier at the analog front end for voltage readout from electrodes, and an on-chip 10-bit successive approximation register analog to digital converter with serial peripheral interface. The ASIC is fabricated in the CMOS 0.18 μm process with a supply voltage of 3.3 V. Amplitude and phase extraction of the voltages is performed in the digital domain with a matched filter. A fully integrated solution for use in multiple electrode system is demonstrated. The readout chain in the ASIC achieves a minimum signal-to-noise ratio of 71 dB over the frequency range of 500 Hz–700 kHz, while maintaining an average accuracy of 99.7%. Frame rates of 21 frames per second for a 32 electrode system is feasible, and the ASIC has an overall power consumption of 11.8 mW.

Index Terms—ASIC, cardiac monitoring, current driver, electrical impedance tomography, instrumentation amplifier, matched filter, variable gain amplifier.

I. INTRODUCTION

SELF-MANAGEMENT of chronic heart disease is hindered by three critical barriers: (i) the symptoms that patients monitor today are insensitive, non-specific and late indicators of heart failure [1]; (ii) this lack of predictive strength forces management of the condition to be reactive rather than proactive [2], [3]; (iii) single, once-a-day measurements cannot capture the patient's hemodynamic response to the stresses of daily life activities [4], [5], or the diurnal variation in sympathetic activation [6], [7], thus hindering a full understanding of the patient's condition [8], [9].

One possible solution is to use an implanted device for on-demand telemonitoring of pulmonary artery pressure [10]. Unfortunately, this approach is invasive, expensive, and risks in-

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Color versions of one or more of the figures in this paper are available online at <http://ieeexplore.ieee.org>.

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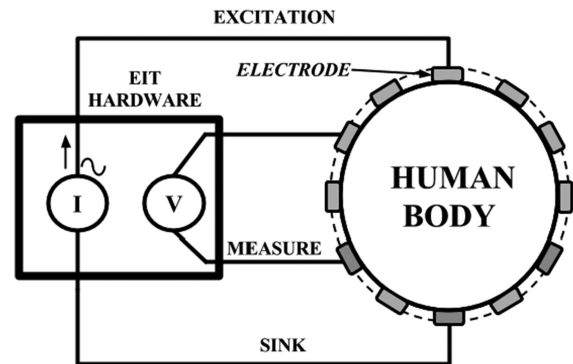


Fig. 1. Block diagram of a typical EIT system. AC current excitation is performed using a pair of electrodes and the voltages induced on the remaining electrodes are measured.

fection, bleeding, arterial embolism, prolonged hospitalization, thrombus and cardiogenic shock, among others [11].

A promising, non-invasive alternative is dynamic electrical impedance tomography (EIT) of the thorax, from which left ventricular volume changes and cardiac output can be extracted. In an EIT system the tissue excitation is performed by injecting alternating current through a pair of electrodes and measuring the boundary voltages induced at the remaining electrodes as shown in Fig. 1. The excitation frequencies can be swept over a pre-determined range depending on the tissue being probed. This idea has been demonstrated previously, including in our own recent work [12] using the EIT Pioneer Set (Swisstom AG, Landquart, Switzerland). The challenge with these systems is that they are large and bulky. They may be appropriate for hospital bedside use, but they are unsuitable for telemonitoring, where physical obtrusiveness and ease of use are critical to ensuring patient adherence [13], [14].

Application specific integrated circuits (ASICs), like the one depicted in Fig. 2, could dramatically miniaturize the cardiac EIT instrumentation system, making its use more feasible in a telemonitoring environment. Unfortunately, the state-of-the-art ASICs for EIT do not meet the performance requirements necessary for cardiac output monitoring. For example, the ASICs described in [15]–[17] do not provide adequate signal-to-noise ratio.

In this paper, we present a novel analog front end ASIC that provides the signal-to-noise ratio, interrogation frequency and power consumption performance needed for a cardiac output monitoring application. Section II presents the system level

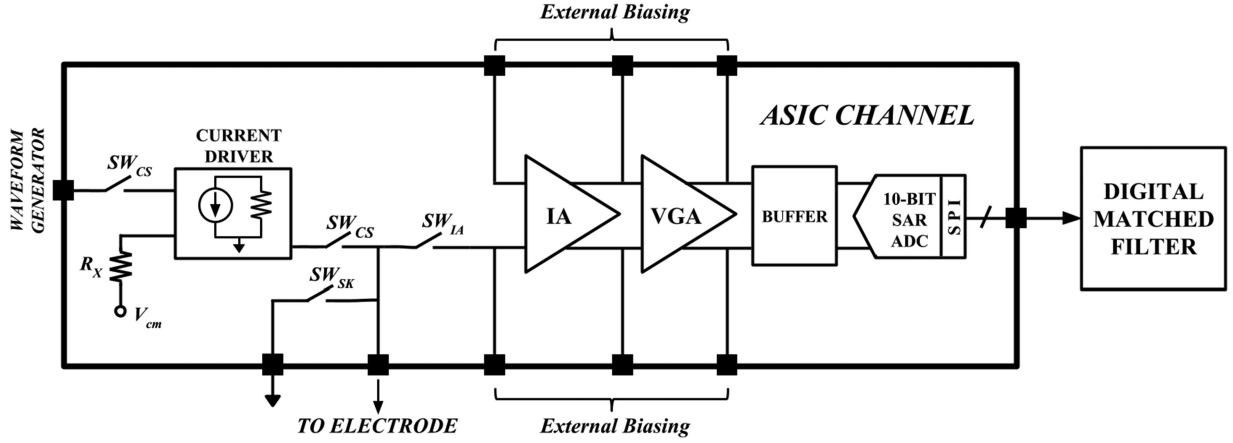


Fig. 2. Block diagram of the cardiac EIT ASIC. The on-chip switches are used to configure the channel in a current-source, current-sink or voltage-read mode. R_X is an on-chip resistor and V_{cm} is the common mode voltage of the current driver. The IA, VGA and the ADC form the voltage readout chain. The digital data from the ADC is sent off-chip using a SPI interface to a digital matched filter for amplitude extraction. Circuit blocks of the ASIC were measured individually with custom built PCBs.

specifications necessary for cardiac monitoring. Section III discusses ASIC building blocks in detail and provides a summary for measured performance. Section IV discusses the system design using the ASIC and provides measured results. Section V summarizes the performance achieved and provides concluding remarks.

II. EIT SYSTEM LEVEL SPECIFICATIONS

Image reconstruction in dynamic EIT aims to estimate the spatiotemporally-varying electrical properties (conductivity, σ and permittivity, ϵ) within the thorax, provided a set of boundary measurements. These measurements (small-amplitude, imperceptible AC currents that are injected into the thorax, and the resulting voltages that develop on the skin surface), carry information about the internal electrical properties. By solving the inverse problem, it is possible to reconstruct a tomography-like spatial distribution of σ and ϵ .

Over the frequency range 10 Hz to 1 MHz, the conductivity of the blood in the heart is about $\sigma_h = 0.7$ S/m, while that of the background (heart muscle, lungs, fat, bone, etc.) is less than 0.4 S/m [18]. This contrast in electrical properties makes the blood-filled heart visible as a high conductivity region in an EIT image of the thorax. In practice, the frequency of interrogation for cardiac output measurement is at least 20 kHz, in order to avoid the high skin resistivity that would be encountered at lower frequencies [19]. Looking beyond cardiac output measurement, there are some other cardiac EIT applications that might require interrogation frequencies up to 1 MHz. These include bioimpedance spectroscopy for detecting cardiac ischemia [20], measuring fluid index ratio due to edema in congestive heart failure [21], and frequency differencing EIT for anatomic characterization, such as ventricular wall thickness [22]. Also, the entire process of making boundary measurements and constructing an EIT image (a “frame”) must be performed at a rate of 14 frames per second (fps) or higher. This way, enough points are captured over the ~ 1 s cardiac cycle such that heart volume change and cardiac output can be calculated.

To understand the remaining performance requirements of a cardiac EIT system, we made use of a highly detailed digital phantom of the human thorax. The phantom comprises the MRI and CT-based 4D XCAT model [23], mesh generation using distmesh [24] and gmsh [25], a perfusion model [26], ex-vivo tissue values of conductivity and permittivity from multiple frequencies [18], and a 3D finite element (FEM) implementation of the complete electrode [27]. We analyzed the digital phantom assuming a 32 electrode EIT system. As shown in Fig. 3, the phantom produces an image that is qualitatively similar to one constructed from measurements of a human volunteer.

Figure 4(a) shows the boundary voltage amplitudes that were generated by the digital phantom for a typical electrode position, with measurements taken at end diastole and also at end systole. Depending on the anatomy of the subject, the boundary voltages can range in amplitude from tens of millivolts to several hundred millivolts. Interestingly, the voltage measured at end diastole (when the heart is full of blood) is different from that at end systole (just after ejection of blood) by less than 1%. This small change is due to the large amounts of insulating air that are inhaled into the lungs, which confound the relatively smaller signals of blood flow in and out of the heart and cardiac muscle motion.

Since the EIT system must detect the 1% change in voltage amplitude across the systolic cycle in order to extract cardiac output, this would suggest a minimum required signal-to-noise ratio (SNR) of 40 dB. In fact, the EIT image reconstruction problem is poorly conditioned and ill-posed, and demands SNR > 70 dB for this application. Fig. 4(c) illustrates how the EIT-derived cardiac output would deviate from ground truth if the SNR fell too low.

In summary, an EIT system for cardiac output monitoring must cover a dynamic range from 20 mV_{pp} to 1 V_{pp}, operate at a frame rate of 14 fps, and maintain an SNR of > 70 dB. In addition, it must support an interrogation frequency of a few tens of kHz, although frequencies up to 1 MHz would be useful for more generalized cardiac EIT applications. Accuracy and linearity requirements are less stringently defined, especially

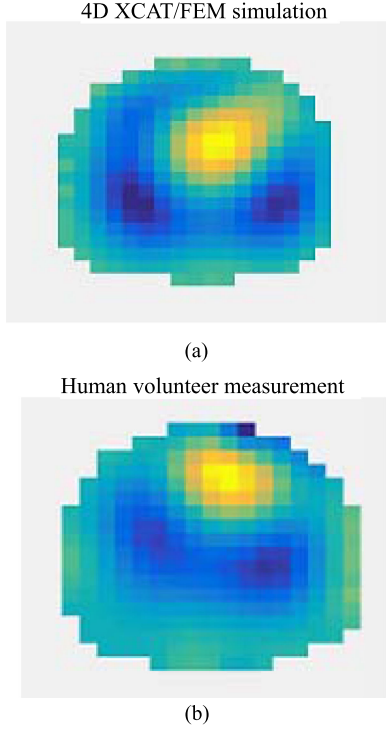


Fig. 3. Cross-sectional conductivity profile of the thorax produced from (a) Digital phantom model; and (b) Bioimpedance boundary voltage measurements of a healthy human subject. The digital phantom model and the in vivo measurements result in qualitatively very similar images, where the heart and lung (from perfusion) are very clearly seen when dynamic EIT imaging is used between end-diastole and end-systole.

with the use of the calibration techniques that we introduced in [28]. Still, it is typical to design for $\geq 99\%$ accuracy and ≤ -40 dB total harmonic distortion [28]–[30].

III. ASIC DESIGN AND DESCRIPTION

In this section, we describe the different components that comprise the EIT cardiac output analog front end shown in Fig. 2.

A. Current Driver

System level specifications derived earlier indicate that cardiac monitoring requires an accuracy $\geq 99\%$ and ≤ -40 dB total harmonic distortion. The current driver in the ASIC is capable of generating 1 mA_{pp} sinusoidal current while maintaining total harmonic distortion of $< 1\%$ for frequencies up to 1 MHz. The accuracy of current delivered over this frequency range is $> 99.6\%$. Accuracy is achieved by designing the current driver with high output impedance, whereas linearity is obtained by using a resistor for voltage to current conversion. The details of the implementation are below.

The current driver for this design is based on the CCII current conveyor [32] shown in Fig. 5. To obtain high linearity, the voltage to current conversion is performed using a 2-stage op-amp in unity gain configuration with a resistor at the terminal X generating current, $I_X = \frac{V_X}{R_X}$ [33]. A class AB buffer at the output of the 2-stage op-amp creates a low impedance node, and

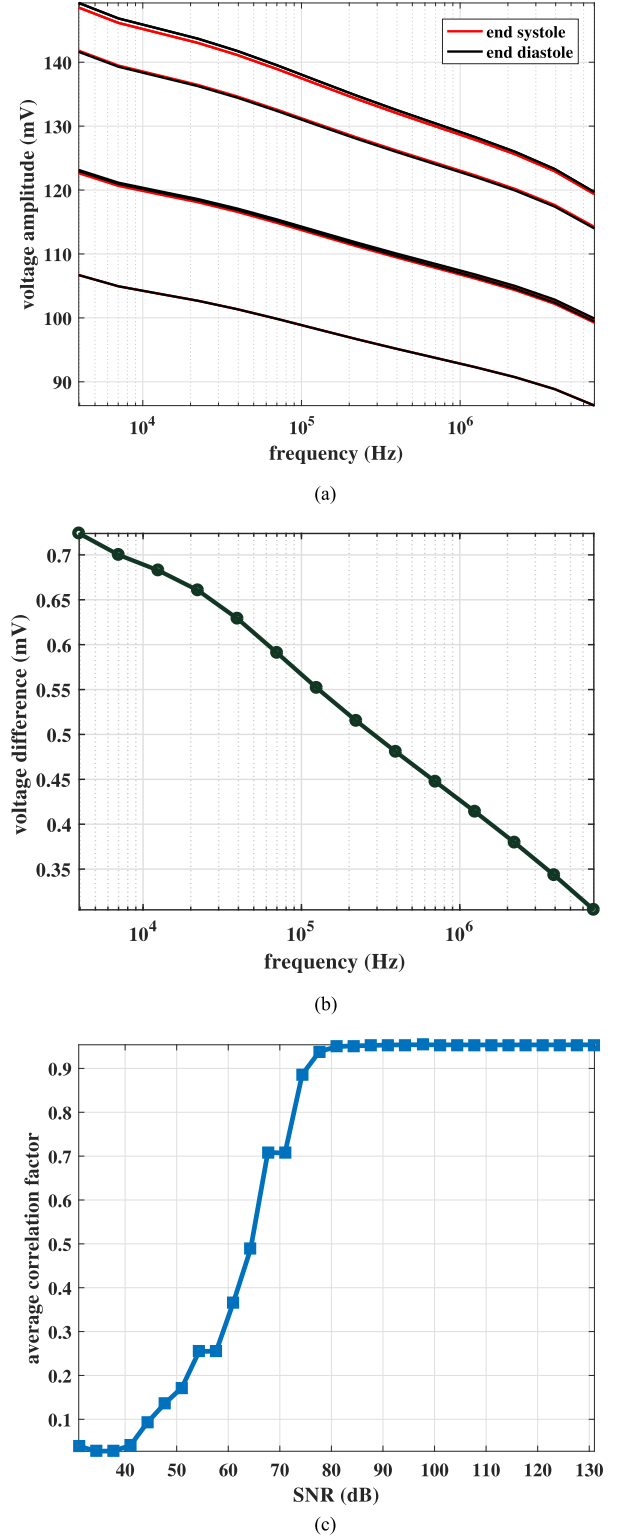


Fig. 4. (a) Boundary voltage amplitude versus frequency plot for some selected electrodes and current source-sink patterns. (b) The voltage difference between end systole and end diastole measured at the same electrode for a representative source-sink pattern from Fig. 4(a). The difference in voltage at these two time points is less than 1 mV, and decreases with frequency. (c) Effect of measurement SNR on accuracy of the extracted heart area [31]. The y-axis is the correlation factor between actual versus EIT-extracted heart area, averaged across the cardiac cycle. For accurate calculation of cardiac output, the correlation factor should ideally be 1. The plot shows that a measurement SNR of at least 75 dB is necessary for sufficiently-high correlation.

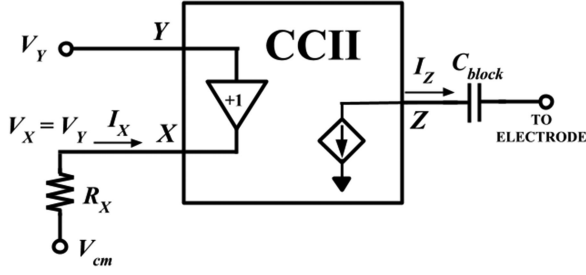


Fig. 5. Block diagram of the current conveyor architecture [32]. Current is generated at terminal X using an on-chip resistor such that $I_X = \frac{V_X}{R_X}$, where $R_X = 500 \Omega$ and V_X follows the voltage at terminal Y . Current I_X is delivered to terminal Z with a current gain of unity.

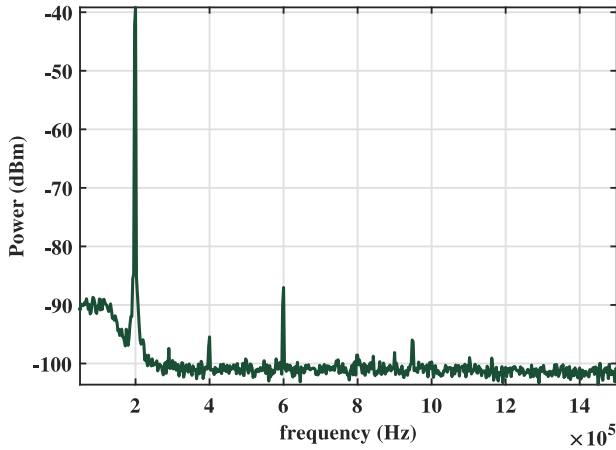


Fig. 6. FFT plot of the current driver delivering $500 \mu A_{pp}$ current to a fixed load of 500Ω . The total harmonic distortion measured is $\sim 0.5\%$ at an input frequency of 200 kHz .

TABLE I
MEASURED PERFORMANCE OF CURRENT DRIVER

Parameter	Value
Area	0.26 mm^2
Bandwidth	$100 \text{ Hz} - 10 \text{ MHz}$
Output Impedance	$101 \text{ k}\Omega @ 1 \text{ MHz}$
Max. Output Current	1 mA_{pp}
THD	$< 0.1 \% @ 50 \text{ kHz}$
Accuracy	$99.6 \% \text{ up to } 1 \text{ MHz}$
Power	1.65 mW

the use of indirect compensation technique [34] helps achieve wide-band operation up to 10 MHz . The generated current is mirrored to the output terminal Z through a gain enhanced cascode current structure to provide high output impedance [35]–[37].

Figure 6 shows an example of the current driver's measurement results. When injecting a 200 kHz , $500 \mu A_{pp}$ current onto a 500Ω load, the measured total harmonic distortion is 0.5% . Table I summarizes the other key performance metrics of the current driver.

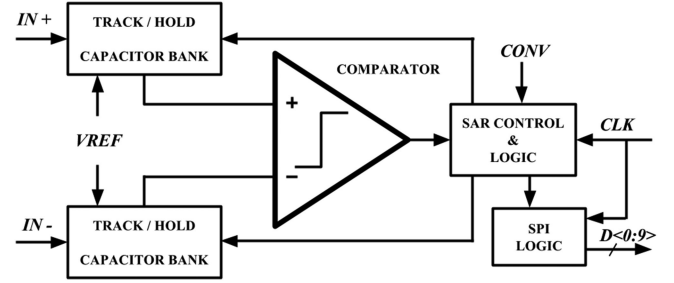


Fig. 7. Block diagram of the differential 10-bit SAR ADC. The maximum sampling rate of the ADC is 250 kps . Serial data output is achieved by incorporating a SPI interface.

B. 10-b SAR ADC

The ADC samples the voltage readout chain and provides the digital data to an off-chip matched filter to extract the amplitude and phase. A signal-to-noise ratio of 80 dB is desirable to meet the system SNR specifications. Assuming an output-referred noise of IA, VGA chain of 1 mV_{rms} [38], 80 dB SNR can be achieved with a 10 bit ADC and 100 tap matched filter [39].

Next we discuss the sampling rate of the ADC based on the the desired frame rate of 14 fps . The frame rate is calculated by the following formula,

$$FR = \frac{1}{P_N \cdot \left(\frac{N_{tap}}{f_s} + t_s \right)} \text{ (fps)} \quad (1)$$

where, FR represents the frame rate in frames per second, N_{tap} is the number of taps in the matched filter for amplitude extraction, P_N is the number of current source-sink patterns in a multi-electrode system, f_s is the sampling rate of the ADC and t_s is the settling time after each pattern is switched. With $N_{tap} = 100$, $P_N = 30$ for a 32 electrode system, and ideal settling time of zero, the minimum required sampling speed of the ADC is 42 kps . In reality, the sampling rate should be higher to account for the settling time during pattern switching to obtain desired precision (see detailed discussion on frame rate in Section IV). The ADC is designed with a maximum sampling rate of 250 kps .

An SAR ADC architecture was chosen as it exhibits lower power consumption in comparison to other ADC architectures [40]. The block diagram of the ADC designed is shown in Fig. 7. The track and hold comprises of a capacitor bank and CMOS switches controlled by the SAR logic block. An SPI interface was designed to achieve serial data output which simplifies system level design in terms of input/output infrastructure necessary when the ASIC is incorporated in a multiple electrode EIT system. The 10-bit data is available at the SPI interface after 14 clock cycles when the $CONV$ signal is asserted high. The ADC consumes an average current of $77.23 \mu A$ with a sampling rate of 250 kps and input clock frequency of 3.5 MHz . The dynamic characteristics and measured performance parameters of the 10-bit ADC are shown in Table II.

The figure of merit for the ADC is given by (2) [41].

$$FOM_{ADC} = \frac{P_{avg}}{f_s \cdot 2^{ENOB}} \text{ (J/conv-step)} \quad (2)$$

TABLE II
MEASURED PERFORMANCE OF 10-BIT SAR ADC

Symbol	Parameter	Value
V_{DD}	Process $0.18\ \mu\text{m}$	1.8 V
V_{IN}	Differential Input Voltage	0 - 1.2 V
$F_s(\text{max})$	Sampling Rate	250 ksp/s
SINAD	Signal to Noise and Distortion	54.98 dB (1 kHz)
ENOB	Effective number of bits	8.84
P_{avg}	Average Power	139 μW
FOM_{ADC}	Figure of Merit	1.25 pJ/conv-step

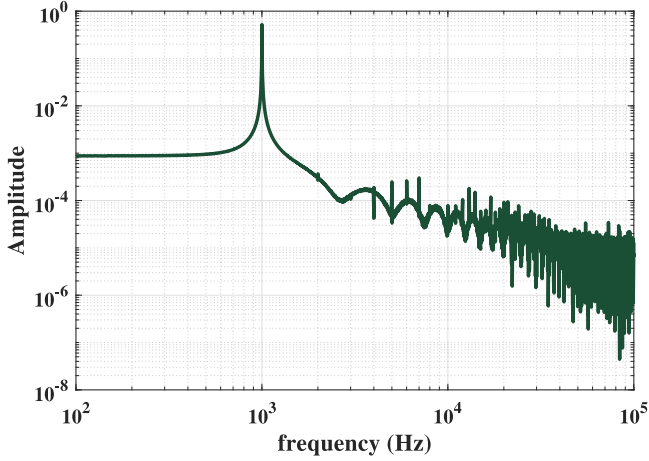


Fig. 8. FFT plot of the ADC for a 1 kHz input signal and sampling rate of 250 ksp/s.

where, P_{avg} is the average power, f_s is the sampling rate and ENOB represents the effective number of bits for the 10-bit SAR ADC. The ADC achieves an FOM_{ADC} of 1.25 pJ/conv-step.

The FFT plot for a 1 kHz input signal at 250 ksp/s sampling rate for the ADC is shown in Fig. 8.

C. Voltage Sensing Amplifiers (IA and VGA)

Boundary voltages on electrodes for a few selected current source-sink patterns are shown in Fig. 4(a). The readout chain should amplify these voltages without saturating the channel and provide high linearity (≤ -40 dB) for frequencies up to 1 MHz. The IA is designed with a fixed gain of 3 V/V, as we want to minimize noise, and prevent saturation of the readout chain. The VGA provides three additional gain settings of 1, 3 and 12 V/V. Thus, to amplify the smallest possible voltage (20 mV_{pp}), the readout chain can provide a gain of 36 V/V. The target output-referred noise specification of the IA and VGA chain is 1 mV_{rms}. With a 10-bit ADC and 100 taps in the matched filter, the circuit noise from the voltage sensing amplifiers will provide the required SNR of 80 dB (discussed in the ADC section).

The IA shown in Fig. 9 is a fully differential amplifier based on current conveyors [42]. It has a balanced structure, with high input impedance and linearity. The IA is designed for low noise operation and provides a fixed voltage gain determined by the ratio of two on-chip polysilicon resistors, R_2/R_1 . The IA has

a common mode rejection ratio (CMRR) over 86 dB, and a measured input-referred noise of $14\ \text{nV}_{\text{rms}}/\sqrt{\text{Hz}}$. The VGA is a fully differential amplifier which has a similar architecture and performance as the IA reported in [38]. Additionally, it has a reconfigurable gain of 0 dB, 9.5 dB and 21.5 dB to fully utilize the input dynamic range of the ADC.

For speed and power constraints, the amplifiers G in Fig. 9 only consists of a few transistors which provide proper level shifting incorporated, for the flipped source follower to increase its dynamic range. It also lowers the output impedance of the source follower, such that the transconductance stage can drive a low value gain resistor R_1 to reduce the overall amplifier noise. The V_{ref} bias voltages are set by on-chip current sources and the implementation of level-shifted source follower concept is similar to our design presented in [38], [43]. The common mode feedback (CMFB) circuit consists of a differential source follower and resistors [44] to control the gate voltage of transistors M_6 and M_6' .

The magnitude response of cascaded voltage sensing amplifiers which include IA and VGA is shown in Fig. 10. The digitally selectable gains of 9.3, 19 and 30 dB are set by the external control bits. Table III summarizes the measured performance of the IA and VGA. The measured voltage gains deviate less than ± 0.5 dB across the 3dB bandwidth. A total of 18 chips were used for this measurement. In addition the variation in gain across the frequencies of the readout chain can be calibrated out for each channel incorporated into the EIT system.

The ASIC is fabricated in the XFAB $0.18\ \mu\text{m}$ CMOS process with a supply voltage of 3.3 V. The micro-photograph of the ASIC is shown in Fig. 11. The current driver, and the readout channel which contains the IA, VGA and the ADC are clearly indicated.

IV. SYSTEM DESIGN AND RESULTS

The system level block diagram of the ASIC channel is shown in Fig. 12. Necessary bias currents and reference voltages needed for the building blocks in the ASIC are provided by using current source IC's and voltage references. The input to the integrated current driver for voltage-to-current conversion is provided with a waveform generator IC, followed by an anti-aliasing filter with a cut-off frequency of 2 MHz. The output current is varied by changing the amplitude of the sinusoidal input voltage to the current driver. To sense the accurate current injected into the electrode, external sense resistors with switching mechanism is designed on the board. External ADC with anti-aliasing filter is also included as a backup. The output from the on-chip ADC and external ADC's is interfaced to a field programmable gate array (FPGA) using SPI communication protocol. The board is powered with linear dropout regulator (LDO) IC's which provide two voltage levels of 1.8 V and 3.3 V.

A. System Signal-to-Noise Ratio (SNR)

The SNR of the system indicates the noise in the readout signal chain of the EIT system. According to our analysis, the acceptable SNR for accurate cardiac output monitoring should

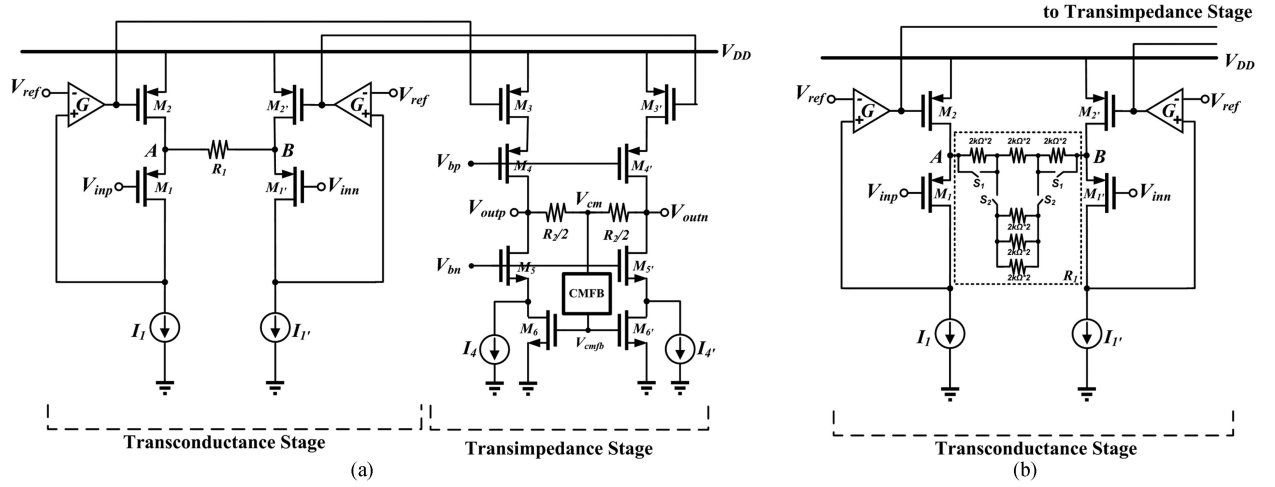


Fig. 9. (a). Simplified schematic of the current conveyor architecture used to implement the IA. The fixed gain is determined by the ratio of on-chip polysilicon resistors $R_1 = 10 \text{ k}\Omega$ and $R_2 = 30 \text{ k}\Omega$. (b). The transconductance stage of the VGA is shown, while the transimpedance stage has a similar structure as the IA. Variable gains for the VGA is achieved by the on-chip resistor ladder shown in the figure. Resistor $R_2 = 15 \text{ k}\Omega$ in the transimpedance stage is fixed while resistor R_1 in the transconductance stage is variable with values $15 \text{ k}\Omega$, $5 \text{ k}\Omega$ and $2.5 \text{ k}\Omega$. Digital control of variable gains is achieved using switches S_1 and S_2 .

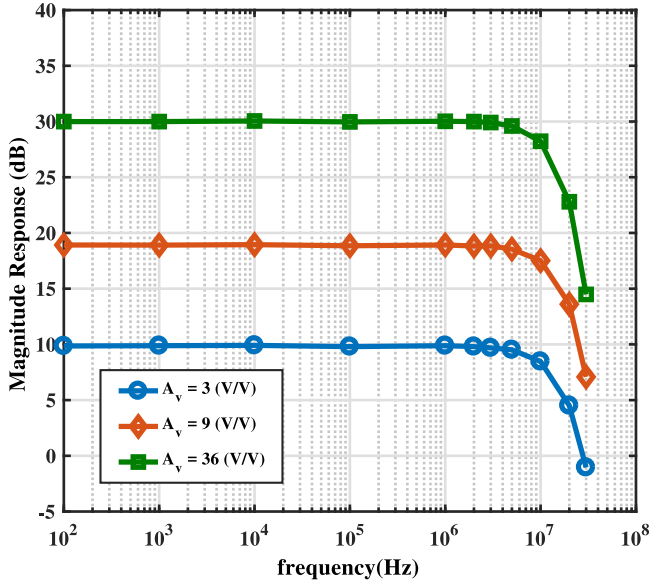


Fig. 10. Measured magnitude response of the readout channel (IA and VGA cascaded). The IA provides a fixed gain of 9.3 dB, whereas the VGA can be programmed to achieve a gain of 0, 9.3 and 21.5 dB with external control bits.

TABLE III
MEASURED IA AND VGA PERFORMANCE SUMMARY

	IA	VGA
V_{DD}	3.3 V	3.3 V
Power	2.07 mW	4.66 mW
Bandwidth	>10 MHz	>10 MHz
Input Impedance (DC)	>1 G Ω	>1 G Ω
Input Noise (nV _{rms} /√Hz)	14	80 / 27 / 6.7
Gain (dB)	9.3 (1 kHz)	0 / 9.3 / 21 (1 kHz)
THD (dB) ($V_{out,diff} = 2V$)	-63 (1 kHz)	-70 (1 kHz, $A_v = 21 \text{ dB}$)

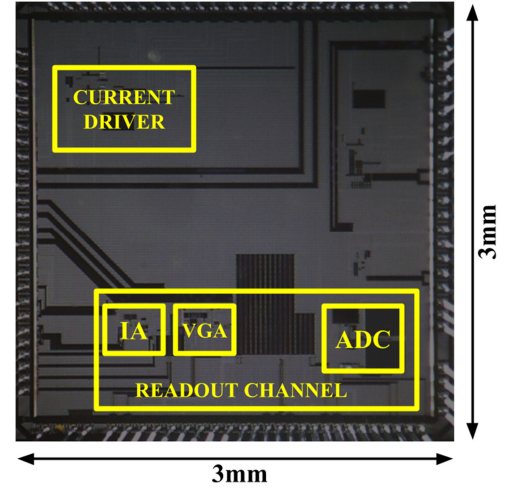


Fig. 11. Chip Micro-photograph. The current driver and the readout channel are shown.

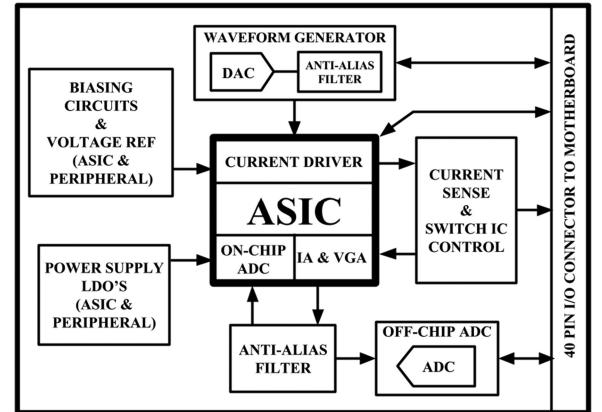


Fig. 12. Block Diagram of the PCB module. The core ASIC is shown with additional circuitry for power supply, biasing, waveform generation, current sensing, switch control, off-chip ADC, and anti-aliasing filters. The 40 pin I/O headers mounts onto a motherboard PCB and contain a large number of ground pins for signal integrity.

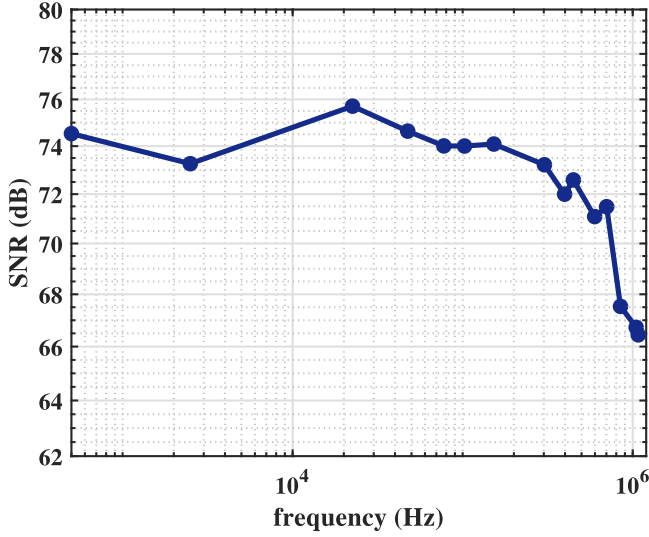


Fig. 13. Signal-to-noise ratio (SNR) at the output of the matched filter that follows the ASIC. SNR was measured using Eq. 3.

be ≥ 70 dB. The SNR formula is given by [28].

$$\text{SNR} = 10 \cdot \log_{10} \cdot \left\{ \frac{\sum_{n=1}^N V_n^2}{\sum_{n=1}^N (V_n - \bar{V})^2} \right\} \quad (3)$$

where N is the repeated number of SNR measurements, V_n represents the measured SNR value of the signal chain, and $\bar{V}$ represents the mean of all measurements. The ASIC was configured in the voltage readout mode, and the readout channel gain settings chosen such that the ADC input dynamic range was utilized. The ADC sampling rate was set at 250 ksp/s and the input frequencies swept from 500 Hz to 1 MHz. The digital data from the ADC was sent off-chip and a matched filter used to extract the amplitude. To obtain the SNR of the readout signal chain, $N = 128$, repeated number of measurements were performed. The minimum SNR measured is 71 dB over a frequency range of 500 Hz–700 kHz. At frequencies > 700 kHz the input amplitude had to be reduced to prevent distortion and accounts for the drop in SNR value. The measured SNR plot over frequencies of 500 Hz to 1 MHz is shown in Fig. 13.

B. Cole-Cole Plot

An R-C circuit combination is used to model the complex tissue impedance of the thorax. The representative load is chosen from simulations using the 4D XCAT model described earlier, where the voltages induced on the electrodes from select current injection patterns are analyzed across frequencies, (see Fig. 4(a)). The block diagram for the measurement of the representative complex impedance is shown in Fig. 14 which incorporates two ASIC channels. One channel is configured in the current source mode and the accurate current injected into the R-C load is determined by measuring the voltage across the 100Ω sense resistor. The second ASIC channel is configured in voltage read mode and measures the voltage induced on the R-C load circuit due to the injected current. In order to obtain high accuracy, a calibration method is adopted where a known value

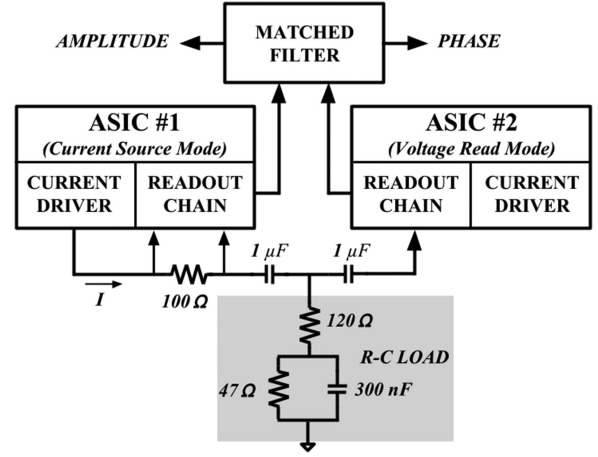


Fig. 14. Setup for obtaining Cole-Cole plot. Representative R-C load is selected via 4D XCAT simulations. Measurement is performed by two channels of the ASIC. Amplitude and phase extraction is performed in the digital domain with a matched filter. Calibration is incorporated to obtain accurate measurements.

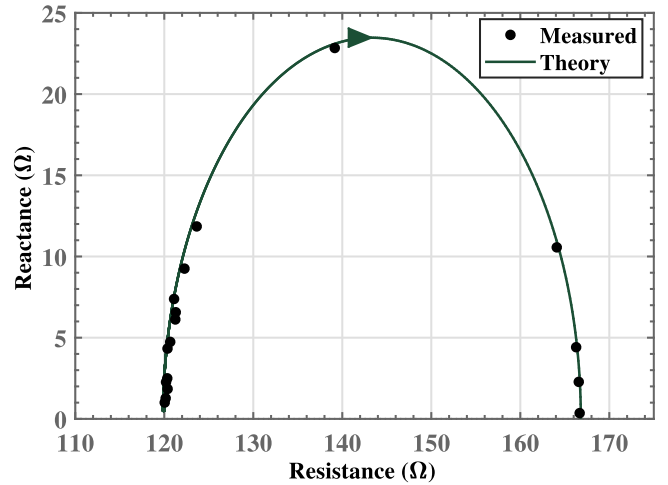


Fig. 15. Cole-Cole plot for a parallel RC load shown in Fig. 14. The measured results are compared with theory for frequencies from 250 Hz–1 MHz.

of current ($500 \mu\text{A}$) is injected into a fixed resistance (200Ω). The R-C circuit components, sense resistor and associated parasitics are measured using a LCR meter. To ascertain the exact gain of the readout channel, gain values are incorporated from the measured results of the IA, VGA specifications over frequencies of interest. This data is saved as reference for actual measurement of load and used for calibration. The measured Cole-Cole plot for frequencies of 250 Hz–1 MHz is shown in Fig. 15. An average of 50 measurements are performed for each frequency point and the amplitude and phase are extracted using a matched filter.

Cardiac EIT systems typically have accuracies $> 99\%$ as it is necessary to measure small impedance changes. The accuracy of the ASIC readout chain is calculated using Eq. 4.

$$\text{Accuracy (\%)} = \left(1 - \frac{V_{\text{meas}} - V_{\text{sim}}}{V_{\text{sim}}} \right) \cdot 100 \quad (4)$$

TABLE IV
PERFORMANCE COMPARISON OF ASIC'S FOR ELECTRICAL IMPEDANCE TOMOGRAPHY

Reference	[15]	[16]	[17]	This Work
Process	0.18 μm CMOS	0.18 μm CMOS	65 nm Mixed CMOS	0.18 μm CMOS
Frequency	up to 90 kHz	10 - 200 kHz	10 - 256 kHz	500 Hz - 700 kHz
Current Injection (peak-peak)	100 - 350 μA	0.1 - 1 mA	0.1 - 1 mA	up to 1 mA
Voltage Gain of Readout Chain (dB)	18 - 40	18 - 60 (6 dB step)	20 - 70 (6 dB step)	9.3, 19, 30
ADC (Sampling Rate)	10-bit (N/A)	10-bit (1 - 200 kpsps)	10-bit (1 - 256 kpsps)	10-bit (12.5 kpsps - 250 kpsps)
SNR (dB)	40	56.3 ¹	N/A ²	71 ³
Power (mW)	6.3	1.733	6.69	11.8 ⁴
FOM (pW/Hz)	700	13.27	N/A	4.75

¹ SNR averaged over measurement index [16].

² SNR was not reported. However, the analog matched filter approach used in [17] typically provides an SNR of less than 60 dB.

³ The minimum achieved SNR is 71 dB over frequency range of 500 Hz - 700 kHz, whereas for the frequency range of 500 Hz - 1 MHz, the minimum achieved SNR is 66.7 dB.

⁴ Breakdown of power is as follows. The current driver consumes 1.65 mW, IA and VGA consume 6.73 mW, and the ADC and its driver consume 3.44 mW.

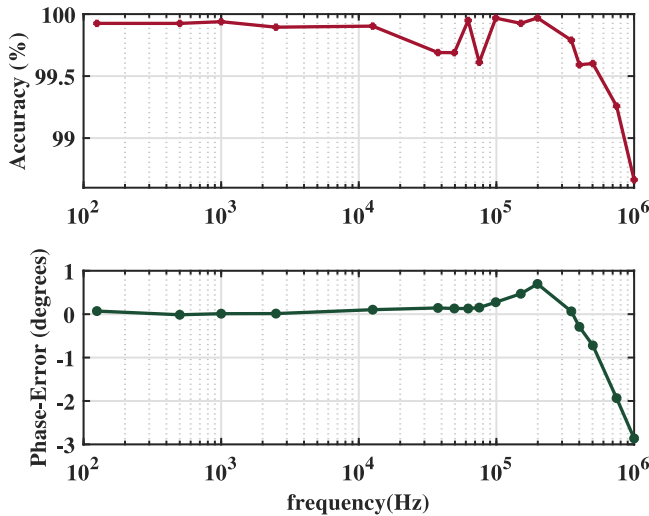


Fig. 16. Accuracy of measured amplitude and phase error over frequencies.

where V_{meas} is the measured voltage and V_{sim} is the voltage obtained from theory. The measured amplitude accuracy and phase error over frequencies is plotted in Fig. 16. An average accuracy of 99.7% is obtained for the measurement of complex impedance amplitude.

C. Frame Rate

To detect the change of heart volume and for accurate cardiac output monitoring the frame rate should exceed 14 fps as shown in our discussion earlier. The frame rate for the system is calculated by (1). It is typical in Swisstom systems and EIDORS modeling [45] to utilize skip patterns for single-ended voltage measurements. In a 32 electrode system using skip patterns, $P_N = 30$. The settling behavior after pattern switching can be studied with the help of the circuit used for measuring the

complex impedance shown in Fig. 14. The RC time constant is determined by the DC blocking capacitor (1 μF) and the resistive load (167 Ω). To achieve a 10 bit resolution the voltage at the load requires 7 time constants [46] to settle. N_{tap} is set to 100 and the ADC is sampled at 250 kpsps. Frame rates of ~ 21 fps is feasible with the designed ASIC.

D. Comparison

Table IV summarizes the performance of this work with previously reported literature on EIT ASIC's. A figure of merit for the ASIC channel is defined by (5) [47],

$$\text{FOM} = \frac{\text{Power}}{\text{BW} \cdot 10^{\left(\frac{\text{SNR}}{20}\right)}} \text{ (W/Hz)} \quad (5)$$

where, Power is the average power consumption per ASIC channel, BW is the bandwidth of the system, and SNR is the minimum signal to noise ratio in decibels over frequency range of 500 Hz–700 kHz. As seen in Table IV the ASIC used for cardiac output measurements in this work is more power efficient than other designs with the lowest FOM of 4.75 pW/Hz. This is achieved by designing an IA and a VGA with wider swing, lower noise and high linearity [38], [43].

V. CONCLUSION

The cardiac EIT system consists of several components such as the analog front-end for signal processing, the digital domain for single point impedance computation, FPGA block for control and communication strategies and reconstruction software to obtain the final image of the thorax. Some of the challenges associated are the signal processing on the ASIC up to 1 MHz, control strategies for electrode switching using FPGA, and robust board design for maintaining signal integrity. This paper measures the performance of the analog chain alone, without it being confounded by other components of the EIT system.

An ASIC with all analog front end components of a cardiac EIT system is fabricated in the CMOS 0.18 μm technology and individual blocks are characterized. The integrated current driver delivers currents up to 1 mA_{pp} while maintaining a THD of <1%. The voltage sensing amplifiers provide gains of 9.3, 19 and 30 dB to process the dynamic range of voltages incident on the electrodes. The readout chain in the ASIC achieves a minimum SNR of 71 dB over frequency range of 500 Hz–700 kHz at the output of the matched filter. An average accuracy of 99.7% is measured by incorporating a representative R-C circuit and calibration scheme. Frame rates of 21 fps are feasible for a 32 electrode system for single ended measurements. The use of power efficient SAR ADC with serial interface for data acquisition enables the ASIC to be used in a multiple electrode EIT system for cardiac output monitoring. The ASIC achieves the desired SNR, frame rate and accuracy specifications with a power consumption of 11.8 mW.

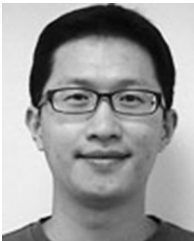
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Absolute Phantom Reconstructions using Fused-Data EIT

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Abstract: Absolute reconstructions from gelatine prostate phantoms using fused-data transrectal electrical impedance tomography (fd-TREIT) incorporating biopsy electrode measurements are produced. The successful reconstructions are important steps toward ex-vivo and in-vivo evaluation of the technology.

1 Introduction

Detecting prostate cancer non-invasively is clinically challenging. Low threshold PSA-based screening has a high sensitivity, but low specificity due to numerous benign conditions elevating PSA levels [1]. Men with elevated levels of PSA are typically subject to an image-guided biopsy protocol for more accurate diagnosis. Unfortunately, transrectal ultrasound (TRUS)-guided biopsies miss 10-30% of all cancers [1]. A number of ex-vivo studies have shown that electrical properties exhibit significant differences between benign and cancerous prostate [2]. We have previously developed an fd-TREIT system for prostate imaging that fuses tetra-polar impedance data recorded from 18 sonolucent electrodes adhered to a TRUS probe and 4 electrodes integrated near the tip of a biopsy needle. The work presented here advances our prior efforts in fd-TREIT [3]; specifically, improvements include using 1) a realistic number of measurements that could be recovered during a standard 12-core TRUS-guided biopsy, 2) prostate phantoms with metal, plastic, and gelatine inclusions, 3) electromagnetically (EM)-tracked TRUS and biopsy probes, and 4) a parallelized-software implementation providing 8x speed-up for absolute reconstructions.

2 Methods

Gelatine phantoms were produced with either 3 inclusions (2 metal and 1 plastic bead, Fig. 1B) or one large gelatine/graphite inclusion (Fig. 1A). TRUS images were used to segment the prostate and gelatine inclusion. EM-tracking using an NDI Aurora V2 system provided 6 degree-of-freedom states of the TRUS probe and biopsy needle for each EIT measurement and of the true bead locations. For each experiment, 10 x/y-locations (needle insertion sites) were measured at 3 depths yielding a total of 30 measurement locations.

The fd-EIT approach was based on a standard Gauss-Newton algorithm using a regularization scheme optimized for open domains [4]. The fusion process utilizes a single finite element method (FEM) mesh (776k nodes, 4.4M elements) that has the TRUS probe and biopsy needle encoded. The mesh was transformed for each state and related to a fixed coarse inverse mesh using the dual-mesh method. The measurements and Jacobians are concatenated so the problem can be solved as if it were a standard EIT problem. Absolute reconstructions required 90 (30x3) forward solves and 30 Jacobian calculations for

each iteration. Three forward solves per state are due to the standard update and a parabolic linear search algorithm. Matlab's *parfor* with 16 workers solved the forward problems using Pardiso yielding a 8x speedup compared to serial runs while using up to ~150 GB of RAM. Each iteration took approximately 40 minutes, yielding total times of ~3.3 hours for 5 iterations.

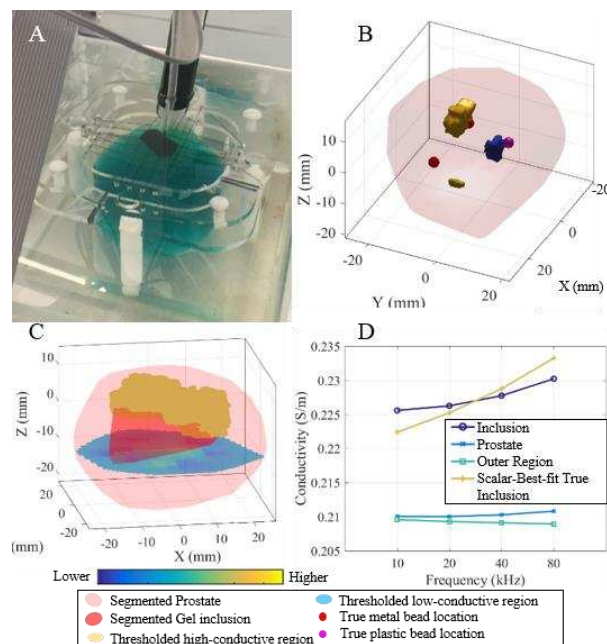


Figure 1: A. Setup for a prostate and gel inclusion experiment, absolute reconstructions of B. a 3 inclusion test and C. a gel inclusion test, and D. average values of each region from C.

The 3-inclusion experiment resulted in average position errors of 6.4 mm. The gel inclusion reconstruction resulted in 44% of the thresholded volume within the true inclusion volume and yielded frequency-dependent conductivities resembling the true inclusions.

3 Conclusions

This study represents significantly steps towards bringing this technology towards pre-clinical evaluation. Further study of EM data and TRUS images can likely improve the results and further speed-up can likely be accomplished.

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Deep Neural Networks for Noise-Robust Cardiac Output Monitoring

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Abstract: A neural network (NN) approach is presented for estimating cardiac output from noisy bioimpedance data. Our innovation is to train the NN with a subject-specific database of synthetic training examples. The NNs are shown to be robust to poorly-contacting electrodes on a measured tank and simulated thorax experiment.

1 Introduction

Adverse cardiac events account for more than 1 million hospitalizations per year in the United States [1]. Thus the ability to pro-actively monitor cardiovascular health is crucial. EIT is a promising technology to non-invasively and continuously monitor cardiac output (CO), an important cardiac diagnostic metric. Although EIT appears more promising than commercial electrical bioimpedance systems, there remain significant concerns of errors due to several sources, including, electrode contact issues. This study presents results from an algorithm based on deep neural networks (DNN) that is robust to faulty electrodes.

2 Methods

The steps of the proposed CO monitoring system (Fig. 1) are 1) Collect good data from a series of patients in a controlled clinical setting and perform EIT reconstructions and metric extraction, 2) Correlate metrics with stroke volumes, 3) Synthesize a database of training samples that realistically models various measurement errors, such as electrodes with partial contact, and 4) Train a DNN on the synthetic data so that it can accurately estimate stroke volume or CO when presented with actual, noisy measurements.

A measured tank experiment (Fig. 2A) was performed where two different DNNs were trained on synthetic data, and then used to estimate the area of an inclusion from surface impedance measurements. Three of the electrodes used for measurements were partially occluded, as might be encountered in a realistic, non-ideal setting. In a second experiment, the DNN were used to estimate stroke volume from bioimpedance measurements of a simulated chest phantom, which again included partially-occluded electrodes. Simulations and reconstructions used Dartmouth's custom software [2].

One can clearly see that DNN1 works well in both experiments regardless of knowledge of which electrodes are bad (Figs. 2B-C), and that by simply removing bad

electrodes EIT reconstructions themselves can work rather well (Fig. 2B).

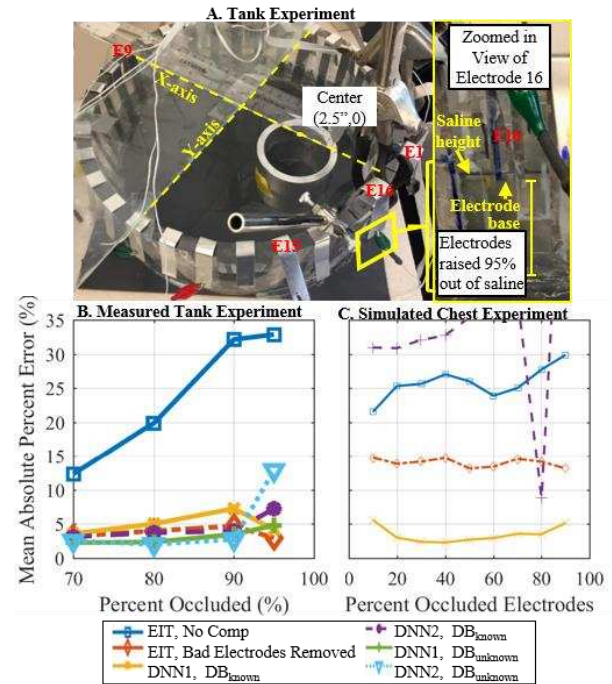


Figure 2: A. The cylindrical tank setup, and the mean absolute percent error for tank and simulated chest experiments. The DNNs were trained on databases that either assumed or did not assume knowledge of which electrodes were ‘bad’. DNN1 is a denoising autoencoder followed by a Gaussian process regression model, while DNN2 is a multilayer feedforward neural network.

3 Conclusions

Although, there are challenges to implement the full proposed CO monitoring approach, the results appear promising and highlight the robustness of DNNs.

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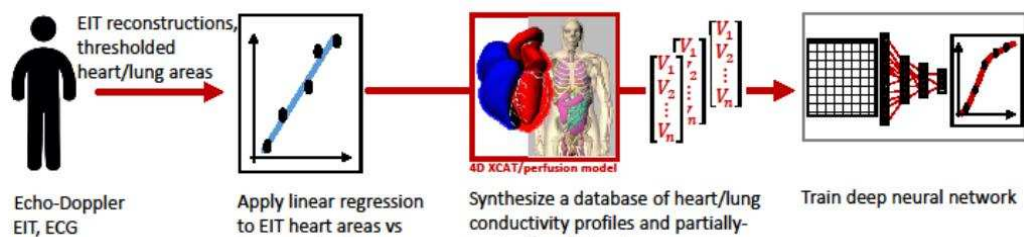


Figure 1. Flow chart of training scheme for the proposed cardiac output algorithm.