Multi-Frequency Electrical Impedance Tomography System With Automatic Self-Calibration for Long-Term Monitoring

Hun Wi, Student Member, IEEE, Harsh Sohal, Student Member, IEEE, Alistair Lee McEwan, Senior Member, IEEE, Eung Je Woo, Senior Member, IEEE, and Tong In Oh, Member, IEEE

Abstract—Electrical Impedance Tomography (EIT) is a safe medical imaging technology, requiring no ionizing or heating radiation, as opposed to most other imaging modalities. This has led to a clinical interest in its use for long-term monitoring, possibly at the bedside, for ventilation monitoring, bleeding detection, gastric emptying and epilepsy foci diagnosis. These long-term applications demand auto-calibration and high stability over long time periods. To address this need we have developed a new multi-frequency EIT system called the KHU Mark2.5 with automatic self-calibration and cooperation with other devices via a timing signal for synchronization with other medical instruments. The impedance measurement module (IMM) for flexible configuration as a key component includes an independent constant current source, an independent differential voltmeter, and a current source calibrator, which allows automatic self-calibration of the current source within each IMM. We installed a resistor phantom inside the KHU Mark2.5 EIT system for intra-channel and inter-channel calibrations of all voltmeters in multiple IMMs. We show the deterioration of performance of an EIT system over time and the improvement due to automatic self-calibration. The system is able to maintain SNR of 80 dB for frequencies up to 250 kHz and below 0.5% reciprocity error over continuous operation for 24 hours. Automatic calibration at least every 3 days is shown to maintain SNR above 75 dB and reciprocity error below 0.7% over 7 days at 1 kHz. A clear degradation in performance results with increasing time between automatic calibrations allowing the tailoring of calibration to suit the performance requirements of each application.

Index Terms—Automatic self-calibration, electrical impedance tomography, long-term monitoring, system stability.

I. INTRODUCTION

Electrical Impedance Tomography (EIT) is a non-invasive and radiation-free imaging technique to provide valuable diagnostic information on physiological and pathological conditions of biological tissues and organs [1]–[4]. In EIT, we measure induced voltages using multiple surface electrodes when injecting a small and safe amount of current into the human body. It produces functional tomographic images of the admittivity distribution by computing the inverse problem using a finite-element model (FEM) of the human body [5]–[8]. Most clinical applications in EIT monitor the time variation of admittivity distributions in the reconstructed images, caused by physiological changes [9]–[11].

The requirements for long term stability differ between applications. A widely used, EIT system from Sheffield, was able to measure impedance changes in the lungs, esophagus, stomach and heart with 30–40 dB SNR [12], [13]. An earlier version of this system was used in two long term studies. In the first, 20 patients were imaged on four occasions over 5 days for lung impedance changes found to be related to left ventricular failure [14]. Secondly, regional lung volume was monitored in 8 infants over several hours on consecutive days with the outcome that the EIT images are potentially useful for monitoring ventilation and planning treatments such as the use of surfactants [15]. Bleeding detection is more difficult with Xu et al. able to image the location of impedance changes of at least 5 ml in a piglet model of retroperitoneal bleeding using a setup with 0.2% or 54 dB SNR [16]. However, Tang et al. found an SNR of at least 75 dB was required in a computational model of EIT for screening and monitoring cerebral intraventricular hemorrhage (IVH) in neonates where the skull is not completely closed [17]. While Tehrani et al. used modeling to find that the requirement increased to at least 86 dB in the case of monitoring a 1 ml reduction in bleeding in the adult brain [18]. Pre-surgical monitoring of epilepsy may last up to 1 week and Fabrizi et al. suggest a required sensitivity of 0.1% to detect epilepsy related impedance changes on the human scalp [19].

In order to apply EIT in clinics, a high performance and stable EIT system is essential with improved image reconstruction algorithms and cooperation with existing complementary clinical devices. Saulnier explained how to design such an EIT system including its architecture, current source, and voltmeter [20]. In...
the wide range of literature available regarding previous EIT systems from RPI [21], [22], Sheffield [13], UCL [23], [24], Goettingen [25], Dartmouth [26], [27] and IIRC [28]–[30], there is a lack of details on the system performance variation and calibration effect over long periods of time. To support the clinical applicability of the EIT procedure we have designed and tested the stability of a new EIT system over 24 hours and 7 days continuous use through automatic self-calibration [31].

Oh et al. developed a fully-parallel multi-frequency EIT system called the KHU Mark2 with a scan speed of 100 scans per second using a pipelined architecture [32]. The KHU Mark2 was a flexible system and could be configured to tailor any electrode configuration needed for a specific application since it is based on an impedance measurement module (IMM) comprising of an independent current source and an independent voltmeter. It was descended from the KHU Mark1 [28]–[30] in terms of technical details such as digital waveform generation, Howland current source with multiple generalized impedance converters (GIC), and digital phase-sensitive demodulators.

The KHU Mark2.5 EIT system builds on the previous KHU Mark2 system with the addition of automatic self-calibration, additional control and timing signals to interface with external medical devices such as patient monitors, ventilators and stimulators, and improved system software using script files. Each IMM includes a current source calibrator for automatic self-calibration to maximize output impedance, to minimize the DC current output, and to balance the current output between source and sink channels. We installed a resistor phantom inside the system for intra-channel and inter-channel calibration of all voltmeters in multiple IMMs. A number of modifications and improvements were required to support calibration and long term stability. These included changes to the system architecture, external device interface unit, IMMs, analog circuit design for calibration, analog backplane, power supply and pipeline structure which are all fully described in this paper.

II. METHODS

A. System Architecture

Fig. 1 shows the architecture of the KHU Mark2.5 EIT system. It is based on a DSP (TMS320F2812, Texas Instruments, USA) as the main controller with an isolated USB interface to a PC. The main controller talks to each IMM through an intra-network controller implemented on the digital backplane. Multiple IMMs are sandwiched between the digital backplane and the analog backplane which includes connectors to lead wires and optional switches for a serial or semi-parallel systems. In order to support long term operation the power supply and intra-network controller were updated with an improved pipelined architecture described in the relevant section below. Channel independent automatic calibration was implemented by updating the analog backplane and IMM as described in the sections below. Fig. 2 shows a 16-channel KHU Mark2.5 EIT system with its internal views. Dimensions of the 16-channel system are 130 × 140 × 130 mm³. A customized medical power supply for safety standards is located underneath the KHU Mark2.5 EIT system in Fig. 2(b). One may use any kind of lead wires and we adopted triaxial cables (SML 50 Triaxial cable MPR, Habia cable, Sweden) in our standard system. The device is also able to use conventional EEG or ECG electrodes, cables and connectors through a new external electrode interface unit [Fig. 2(c)].

B. External Device Interface Unit

The external device interface unit provides an input port for receiving an event trigger signal and two output ports to send out an EIT data acquisition timing signal and a trigger out signal to other medical devices. These triggers are interfaced to the digital backplane and intra-network controller which are implemented in a similar way to the previous system [32]. The intra-network...
controller generates a start-scan signal to commence a certain number of scans defined in a script file. It also controls measurement timing and the number of measurements depending on operating modes described in Section II-I.

Biological signal-gated conductivity imaging can be implemented with the event trigger input signal enabling the capture of fast cardiac events. This may improve images and the signal-to-noise ratio (SNR) by using signal averaging methods at the same point in cardiac or respiration cycles. One may need to synchronize EIT scans with other biosignals or images acquired by other biosignal equipment that do not have proper trigger signals. The KHU Mark2.5 EIT system can provide an EIT data acquisition biosignal-gated imaging technique that does not necessitate EIT scans with other biosignals or images acquired by other biosignal equipment that do not have proper trigger signals. The KHU Mark2.5 EIT system can provide an EIT data acquisition timing signal to external devices for such cases.

C. Impedance Measurement Module (IMM)

Fig. 3 shows a picture of an IMM with its functional blocks. It includes a single-ended constant current source, a differential voltmeter, and a current source calibrator.

Since Oh et al. provided details of the current source and voltmeter designs, we explain only the new embedded current source calibrator [32]. A schematic of the calibrator is shown in Fig. 4(a). The current output can be directly connected to the input of the calibration circuit through a known resistor \(R_{\text{Cal}}^H = 1 \, \Omega\) or \(R_{\text{Cal}}^L = 0 \, \Omega\). The current source calibration circuit is an implementation of the droop method proposed by Cook et al. [21].

While Cook used a negative impedance converter (NIC) to generate a negative capacitance to reduce the effect of stray capacitance, we used the generalized impedance converter (GIC) that generates a tunable inductance at the output of the current source to maximize its output impedance over a wide frequency range [29]. In each IMM we included four GIC circuits to meet the LC resonance condition. We designed the GIC circuit to allow a variation of \(\pm 50\%\) of these estimated values and overlapped these to cover a wide calibration range.

D. Analog Circuit Design for Calibration

In order to reduce the loading effect of the current source, we need to maximize the output resistance, and minimize the output capacitance, for high output impedance, at all operating frequencies. The Howland current pump (HCP) can provide the calibration method to maximize the output impedance, as it is based on balancing the resistance of two feedback loops. Even though we matched the resistance balance well in the HCP, this balance is different at each operating frequency, due to the output capacitance of the cable, PCB layout and components, such as \(C_4\), which used for preventing oscillation of the HCP [33]. In order to reduce the output reactance of the current source, we synthesize a correcting inductance using the four GIC circuits to meet the LC resonance condition.

To design these circuits we used the droop method to find that output capacitance of the current source in each IMM varied from 550 pF to 1 nF, due to the combined variations in the capacitances from the cable, PCB layouts and components. This range could not be covered by a single GIC circuit. By SPICE simulation we found that a minimum of four GIC circuits are required to cover the frequency range of 10–500 kHz, with 1% tolerance resistors and 5% tolerance capacitors, in all GICs.

We then tuned the circuits during the calibration using two digitally programmable potentiometers, \(R_{3B}\) in the HCP, and \(R_9\) in the GIC (Fig. 4). This enabled us to calibrate the output impedance of current source from 10 Hz to 500 kHz. These resistors were chosen as they do not appear in any feedback loop and are grounded. This was a precaution to avoid introducing additional frequency responses from the parasitic reactive components of real potentiometers and switches. The specific digital potentiometers were DS1267E-010, -050, from Maxim Integrated Products, Inc., USA, and switches MAX4545CAP1009, from Maxim Integrated Products, Inc., USA, and switches MAX4545CAP1009, from Maxim Integrated Products, Inc., USA, and switches MAX4545CAP1009, from Maxim Integrated Products, Inc., USA. The tuned GIC circuits generated inductances of 196 mH (10 kHz), 11.1 mH (50 kHz), 4.49 mH (100 kHz), 734 \(\mu\)H (250 kHz) and 191 \(\mu\)H (500 kHz), using multiple GIC circuits, in order to meet the LC resonance condition. We designed the GIC circuit to allow a variation of \(\pm 50\%\) of these estimated values and overlapped these to cover a wide calibration range.

An important feature found in this calibration process was that the tuning parameters of the HCP and GIC affect each other. This implied that the output impedance can not be determined independently from the output capacitance. Put another way, the parameters able to produce the highest output resistance were not always the optimum parameters to generate the highest output impedance. We did find that the optimal setting...
obtained from the combinational sweeping of two potentiometers sustained the maximum output impedance for longer, and was more stable. Fig. 5(a) shows how the inductance range can be generated by changing the $R_{3b}$ at each operating frequency. We also simulated the operating margins caused by the error of components and the temperature variation, shown as error bars in the figure. As an example when we changed $R_{3b}$ in the HCP and $R_{3a}$ in the GIC, the output impedance could be confirmed by the droop method proposed by Cook et al. [21]. Fig. 5(b) is an example of the output impedance variation by tuning the parameters at 100 kHz.

E. Analog Backplane and Electrode Interface Unit

The KHU Mark2.5 EIT system can accommodate any data collection protocol by use of a custom analog backplane with appropriate voltage buffers and switches [32]. Since the calibration circuit is implemented in each IMM, and each IMM independently operates the data acquisition, no additional changes are required in the IMM for use in different configurations. This means that the system is able to maintain operating speed and calibration speed, when the number of channels is increased.

The electrode interface unit includes D-sub connectors attached near the patient in order to connect the EIT system with conventional touch-proof type ECG (MEB2.0 and MLK2.0, Multi-contact, Switzerland) or EEG electrodes (EL254, Biopac systems Inc., USA). We can measure induced voltages relative to the patient ground potential or differential voltages from neighboring electrode pairs selectively. There exist capacitances between electrode-electrolyte interfaces in the imaging subject. When we repeatedly inject current with a non-zero DC current, even though we remove DC through calibration, charge may be accumulated in the capacitors. If we do not discharge them before injecting a subsequent current, the stored charge may produce erroneous voltage measurements. In order to avoid this memory effect, we momentarily ground the chosen pair of electrodes after each current injection. We found that we needed to implement this grounding step to achieve the system performances with calibration presented in this paper.

F. Isolated DC Power Supply

To improve system performance for long-term monitoring, we also found that we needed to improve the power supply, as a high performance and stable power supply is important to establish the measurement system. The KHU Mark2.5 EIT system is powered by a custom-designed isolated DC power supply consisted of 3 switching mode power supplies (SMPS), approved for medical use (ECM100US07, ECM100US09, XP Power Limited, Singapore), and precision regulators (TL431, Texas Instruments, USA), arranged in a double stage to remove the switching noise and obtain high thermal stability [34]. It provides $\pm 5$ V analog power and $+5$ and $+3.3$ V digital power. The output power of SMPS had 70 kHz switching noise of $-60$ dB. The switching noise was reduced below $-90$ dB by a voltage stabilization circuit. Other frequency components are smaller than $-110$ dB up to 100 kHz. Noise spectral density was 9.42 $\mu$V/$\sqrt{Hz}$. These results were obtained with a load for full current output using an audio analyzer (U8903A, Agilent, USA). The overall size of the power unit was decreased by 34.3%. Noise was less than 295 $\mu$V. In order to ensure that we had a suitable power supply for long-term operation we measured less than $0.006\%$ variation in output voltage over 24 hours.

G. Pipeline Structure

To increase the speed of the system for higher frame rates, we adopted the pipeline structure described by Oh et al. [32]. We define the frame rate as the number of scans per second, which is determined by the lowest operating frequency and the amount of averaging. When the lowest operating frequency is 10 kHz, the maximal frame rate is 100 scans per second (no averaging). If the lowest frequency is below 10 kHz, the frame rate is reduced. For example, the lowest frequency of 100 Hz has a period of 10 ms. When we use 16 projections (current injections) per scan, the scan time is 54.18 ms. It means that the maximum frame rate is less than 2 scans per second if 100 Hz is included as the lowest operating frequency.

In the KHU Mark2 EIT system, each IMM automatically commences data acquisition, repeatedly, based on a specific time segment in the pipeline structure, after receiving a start scan command from the main controller. This is problematic for continuous operation for several hours, as we directly receive data without checking the synchronization among IMMs. This configuration is also difficult due to the requirement to commence acquisition based on an input trigger from any number of diverse sources.

To address this problem we modified the intra-network controller to generate and transmit a start-scan signal to all IMMs for synchronous data acquisition. This can be initiated by the command from the main controller called the ‘Command Mode’ or an event trigger input signal from other medical equipment as the ‘Event Mode’. Each trigger signal can commence a certain number of scans defined in a script file. Also, we provide a real EIT data acquisition timing signal to confirm the synchronization among IMMs. The intra-network controller computes the operating parameters of number of scans, scan time and interval, and data size based on the instructions of desired frequency and other settings from the user. This enables flexible control of acquisition timing.

H. Automatic Self-Calibration

To maintain the KHU Mark2.5 EIT system at its maximal performance, we need to calibrate it regularly. Oh et al. suggested...
an external calibrator for current sources and voltmeters and described how to perform the intra-channel and inter-channel calibrations of multiple voltmeters using a resistor phantom [29], [32]. However its use requires manual operation by trained personnel and as an EIT system has usually many channels, each including a current source or voltmeter, calibration can take a long time and needs constant attention to change a measurement configuration by the operators. Also, the operators need to analyze the calibration data to find the optimal factors. In the new Mark2.5 system we automated the calibration process, by embedding a current source calibrator in each IMM and a resistor phantom inside the system enclosure. The resistor phantom is a simplified version of the Cole-Cole phantom [7] with 32 91 Ω resistors, one between each channel and another between each channel and a common terminal. By reducing the calibration tasks, the operator can execute the calibration procedure automatically and use their time for preparing other setups for measurement such as preparing the subject. This is faster and more convenient for the operator which we feel is vital for the application of EIT clinically or in research studies.

Fig. 6 shows the automatic self-calibration procedure in the KHU Mark2.5 EIT system. The procedure consists of the following steps:

1. Calibration start
2. DC offset calibration
3. Phase compensation for CCS calibration
4. CCS output impedance calibration
5. Intra- and inter-channel calibration
6. Finish

Fig. 6. Automatic self-calibration procedure in the KHU Mark2.5 EIT system.

We developed the system software to support calibration as a Windows application program providing a graphical user interface shown in Fig. 7(a). It has two independent windows. One to control the EIT system, and another to display reconstructed images in real time. The user can follow four steps in the control window. The first step is to initialize USB communication to the KHU Mark2.5 EIT system and to check the status of all IMMs. A script (text) file describing the data collection protocol is selected in step 2. In step 3 the electrode contact is checked and any bad electrodes are replaced or adjusted before starting data acquisition. In the final step, we set the storage folder and required number of scans or frames. The scan is defined in the script file. There are controls for setting variable and application specific impedances. This cable connection is made in calibration steps A to C to calibrate the current source; in steps D to F we simulate normal voltage measurement by connecting the current sources to an internal resistor phantom, and measure the resulting voltage via the switching network on the analog backplane. In step A, we remove the DC offset in the current output by adjusting a 10-bit DAC which subtracts or adds the required offset current. Then in step B we compensate for any phase difference of the measured current at each operating frequency. These may be caused by various components and analog circuits. In order to perform steps A and B, we set the calibrator in each IMM by setting the switches at ‘S2’ and ‘S3’ in Fig. 4(a) to measure the current. We then initiate the current source output impedance calibration. For given values of the digital potentiometers in the Howland circuit and GICs, we measure the current at the two different switch settings of ‘S1’ and ‘S2’. This provides the difference due to the series connected calibration resistor $R_{DC}$. By repeating the measurements for all possible combinations of the digital potentiometers, we choose the best digital potentiometer settings that produced the largest output resistance and the smallest output capacitance, which is thereby the largest output impedance. This step C is an automation of the semi-manual process described in Oh et al. [29]. We assume that the EIT system will be used in the conventional way with balanced current source pairs connected to adjacent or opposite electrodes. So before step D, we connect the two current source outputs to the internal resistor phantom and the switch set at ‘S4’ instead of ‘S3’.

I. PC Software Based on Script

We developed the system software to support calibration as a Windows application program providing a graphical user interface shown in Fig. 7(a). It has two independent windows. One to control the EIT system, and another to display reconstructed images in real time. The user can follow four steps in the control window. The first step is to initialize USB communication to the KHU Mark2.5 EIT system and to check the status of all IMMs. A script (text) file describing the data collection protocol is selected in step 2. In step 3 the electrode contact is checked and any bad electrodes are replaced or adjusted before starting data acquisition. In the final step, we set the storage folder and required number of scans or frames. The scan is defined in the script file. There are controls for setting...
Script files are plain text files prepared by a user to dictate how to collect the EIT data using the KHU Mark2.5 EIT system. All abbreviations used in the script file are explained in the Appendix of this paper. Before the detailed explanation of the script file, we define the basic terminology of ‘Projection’ for data acquisition as the measurement of induced voltages from all IMMIs simultaneously when injecting current between a pair of electrodes. The ‘Scan’ consists of a series of projections for reconstructing an image (i.e., an image frame).

As an example, the my_script.txt file defines a scan with self-explanatory text. It can be divided into 4 sections. The first section starts with some comments such as the file name and includes a second script file with predefined information about the projections. Here, this file is called my_projection.txt and it describes three projections as shown in Fig. 7(c). The second part of the script file explains the data acquisition method with variables such as the operating mode, the total scan number, the total number of channels in the EIT system, the number of sinusoidal cycles used in each voltage measurement for data averaging, the time segment for the pipeline operation, and the acquisition delay time after starting the current injection. We have two operating modes of ‘Command’ and ‘Event’ which are explained in Section II-G. The maximum number of successive scans is limited to 255. The third part is commented out because we use calibration factors obtained from a previous calibration. If we need to collect new calibration factors using an automatic calibration process explained in Section II-H, the third part is activated when removing the comment marks (‘%’). It consists of DC offset calibration, output impedance maximization for the current source, amplitude calibration for the current output, and intra-channel and inter-channel calibrations for voltmeters. The final part calls multiple projection lists defined in the included projection file. Using multiple independent current sources, we may implement various current injection methods in each projection. The my_projection.txt explains how to implement a projection using the neighboring, diagonal, and trigonometric protocols at the chosen frequency of 10 kHz using 8 channels.

J. Experimental Evaluation

1) Performance of Automatic Calibration: We compared the amount of DC current, before and after automatic calibration, from all current sources, and at all operating frequencies. The output impedance of the current sources, and the percentile unbalanced factors between current output pairs, were examined at all frequencies. Reciprocity is a key indicator to assess EIT measurement quality as the EIT reconstruction algorithm relies on the Geselowitz relationship [37] and the reciprocal assumption. Reciprocity implies that the same impedance signal will be found if we apply voltages and measure currents, or apply currents and measure voltages, from the same electrodes [7]. Therefore we measured the reciprocity error (RE), which is the ratio of the impedance measured with the voltage and current electrode pairs swapped, in a four terminal measurement. Since the performance of the voltmeter calibration affects reciprocity data collection process are prescribed in script files of which examples are shown in Fig. 7(b) and (c).
error, we computed the reciprocity error from data recorded on a homogenous saline tank. The conductivity of the saline tank (0.07 S/m), used in this paper, was lower than the conductivity of muscle and fat, found in the literature, because we wanted to examine the long term time temporal performance of the system clearly [2], [3]. The current output variation can be seen as the induced voltage multiplied by the resistivity of testing material.

2) Long-Term Evaluation: We assessed the stability of the power supply over 24 hours. We then obtained the calibration factors using the automatic self-calibration function, in a 16-channel KHU Mark2.5 EIT system, every day for a week. After applying calibration factors, we measured the DC offset current and output impedance of each constant current source, the difference between source and sink current, and RE to evaluate the performance of the system. We were interested in reporting the effect of calibrating at intervals of more than 1 day. Therefore we repeated the measurements with calibration factors obtained 1, 2, and 3 days previously.

III. RESULTS

The KHU Mark2.5 EIT system has the same basic performance as the previous Mark2 system in terms of the waveform fidelity of the current output, CMRR and gain error of the voltmeter. Since these are already described in Oh et al. [32], in this paper, we will show results related to the new features. Before testing the calibration features we ensured that the system SNR was optimal. This required a warm up time of 5 minutes as shown in Fig. 8.

A. Automatic Calibration

We measured the DC offset current of each channel before and after calibration. The average value of the DC offset current was reduced by at least 170 times from less than 5.68 μA to 0.033 μA. The variation of DC offset current between channels was also decreased by 9.26% as shown by the error bars in Fig. 9(a). Fig. 9(b) shows the output impedance of the current source after applying calibration values to the digital potentiometers in the Howland current circuit and GIC circuits.

The output impedance values of all current sources were greater than 5.75 MΩ at the operating frequencies. The average value was 41.6 MΩ. The error bars reveal that the variation in output impedance between the 16 current sources is very dependent on the operating frequency. The amplitude difference between source and sink currents produces the residual current at each frequency shown in Fig. 9(c). The averaged percentage error between source and sink currents was 0.21% after CCS amplitude calibration. Fig. 9(d) shows measured REs for all frequencies. It was always less than 0.52% with a mean value of 0.4%.

B. 24 Hours Continuous Operation

Fig. 10 shows how the SNR and RE at each frequency varied over 24 hours continuous operation. There was a slight slope of deterioration in both SNR and RE at all frequencies and the variation in performance over time differed substantially among frequencies with 100 Hz varying by over 20 dB in a single hour. However the SNR at 100 Hz dropped below 100 dB only once and all frequencies remained above 80 dB SNR except the highest frequencies of 500 kHz. The behaviour of the RE was similar with variation greatest in the lowest frequencies. Overall the RE was less than 0.52% over the full 24 hours.
C. Long-Term Evaluation of 1 Week Continuous Operation

During the week long evaluation of continuous operation, calibration data was applied daily and from the previous 1, 2, and 3 days. This was performed to figure out a characteristic of how the system performs with greater intervals between calibrations, with the intent to inform users of the system about how often they need to perform calibration. From Fig. 11 it is evident that the key figure of interest is the minimum SNR which decreased with the number of days since calibration, as 83.3, 79.6, 74.8 and 76.5 dB, for 0, 1, 2 and 3 days due to the calibration, respectively. For example, if we applied the calibration factors measured two days ago, the minimum SNR decreased by 8.5 dB. The key figure of interest here is the deviation and increase in the maximum RE, this increased from 0.6% to 0.7% when the calibration period was increased to 3 days.

IV. DISCUSSION AND CONCLUSION

We have developed a new EIT system aimed at long-term monitoring applications. We included a current source calibrator inside each IMM channel and modified the analog backplane to allow calibration of its performance to a standard level according to the calibration procedure described in Section II-H. Within the testing frequency range (50 Hz to 500 kHz), the DC offset current was lower than 0.033 μA, the output impedance of the current source was higher than 5.75 MΩ, and the difference between source and sink currents was less than 0.2%. The REs were lower than 0.5% and the SNRs were higher than 90 dB when averaging 64 periods. The calibration process was performed every day for a week. We examined the results after using calibration data obtained 0, 1, 2, and 3 days previously. Minimum SNR and maximum RE were decreased and increased over time, respectively. We can estimate the pre-calibration SNR and RE as 65.7 dB and 0.82% after 1 week (7 days) by assuming that the performance of the 16 channel EIT system would degrade at the same rate as found in the experimental results. To communicate the benefits of that our calibration scheme will confer in reconstructed images, we show simulated images in Fig. 12, when SNRs are 83.28, 76.5, 65.7, 40 dB and REs are 0.57, 0.65, 0.82, 4%. This is a common EIT testing example of a cylindrical saline tank with an anomaly of 0.1% conductivity difference. We compared the GREIT parameters to assess the quality of the reconstructed images of simulated degradation in SNR and RE [36]. After 1 week all parameters degraded, the image amplitude response decreased and other parameters (position error (PE), resolution, shape deformation, ringing artifact) increased each day. When the SNR decreased from 83.28 dB to 40 dB, the image amplitude response was reduced by 0.3%, and PE, resolution, shape deformation and ringing artifact was increased by 0.56%, 0.18%, 0.4%, and 1.14%, respectively. When the RE increased from 0.57% to 4%, the image amplitude response was reduced by 0.15%, and PE, shape deformation and ringing artifact was increased by 0.04%, 0.008%, and 0.04%, respectively. This example is informative but has limitations in addition to being a single example. EIT images are also strongly dependent on other aspects of the setup such as the reconstruction algorithm and parameters, the boundary size and shape, the contrast and the electrode locations among other variables. Based on our knowledge, this is the first time performance changes over time have been related to repeated calibration in EIT systems. This clearly motivates the need for automatic self-calibration in EIT systems. Also, the time it took to stabilize the amplifiers and the temperature of the system were a concern from the perspective of performance. We found a recommended warm-up time of 5 minutes. We achieved and maintained high-quality performance for clinical applications through simple and automatic calibration. When we compared the execution times for calibration in the Mark 1, 2, and 2.5 16 channel EIT systems, they were 52.22, 220.83, and 28.58 minutes, respectively. These times depended on how many current sources each system has (i.e., the Mark 1 has only one current source pair), and how many operating frequencies were used. For the system presented here, we calibrated with the nine operating frequencies from 50 Hz to 500 kHz at multiple gain steps. In many applications the lower frequencies may not be useful and should not be calibrated as these contribute significantly to the calibration time with 50 Hz and 100 Hz contributing to over half of the calibration time, simply due to their long periods. Overall there is a significant improvement in calibration time for the fully parallel Mark2.5 system over the fully parallel Mark 2 system. When we increase the number of channels we will need to include a larger resistor phantom within the unit and the calibration step D will increase as it involves the comparison between current source pairs. However the time of step D is negligible compared to the other calibration steps, and these will not increase with more channels, as each IMM channel includes its own calibration circuit. Therefore the overall calibration time will be maintained.
The KHU Mark2.5 multi-frequency EIT system is suitable for clinical applications because it can be operated with other medical devices, has high performance, and long term stability. We plan to use it in imaging experiments to detect stroke, bleeding detection, neural and cardiac activity, and ventilation. The information presented in this paper can be used to interpret which applications are currently feasible with EIT and the requirements for calibration. For example, the system is able to meet the expected requirements of 40–60 dB SNR in most abdomen and thorax measurements with calibration at least once every 3 days. However, daily calibration would be recommended for screening or monitoring IVH in neonates, based on the simulations by Tang et al. which suggested a 75 dB or more SNR. The performance is not yet good enough for bleeding detection, epilepsy and neural activity monitoring in the adult human as the SNR, even with daily calibration, is as low as 80 dB and the reciprocity error at least 0.1%. These are at the same level as the impedance change signal that is desired to be measured in these challenging applications.

APPENDIX A
ABBREVIATION FOR SCRIPT FILE
OpMode, Operating mode; NumScan, Number of scan; NumChannel, Total number of IMM channels; NumAverage, Number of cycles for data averaging; TimeSegment, Time for one section in pipeline operation (μs unit); AcqDelay, Delay time for data acquisition after injecting current (μs unit); DcOffset, DC offset calibration in constant current source; OutputImpedance, Output impedance calibration in constant current source; Amplitude, Amplitude calibration in constant current sources; Voltmeter, Phase compensation and intra-, inter-channel calibration in voltmeter; Projection, Measuring induced voltages from IMMs when injecting current between a pair of electrodes; Scan, Combination of projections for reconstructing an image.

APPENDIX B
ABBREVIATION FOR PROJECT FILE
Channel, IMM channel number; Amplitude, Amplitude of injecting current (mA unit); Frequency, Operating frequency (kHz unit); Gain, Gain for voltmeter (x unit).

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Mr. Wi is a member of the Korea Society of Medical and Biological Engineering.

Harsh Sohal (S’09) received the B.Tech. degree in electronics and instrumentation engineering from Punjab Technical University, India, in 2005, and the M.Tech. degree in very large scale integration design and automation from the National Institute of Technology, Hamirpur, India.

Since 2009, he has been working toward the Ph.D. degree in the Biomedical Engineering Department at Kyung Hee University, Yongin-si, Gyeonggi-do, Korea, under the supervision of Dr. Eung Je Woo. He is working on the development of impedance imaging systems. His research interests include FPGA design, biomedical instrumentation, biomedical signal processing, and computing.

Alistair Lee McEwan (M’89–SM’09) received the B.E., B.Com. and M.Phil. degrees in economics and electrical engineering from The University of Sydney, Sydney, Australia, in 1999 and 2001, respectively, and the D.Phil. degree in microelectronics from Oxford University, Oxford, U.K., in 2005.

He is a Senior Lecturer of Computer Engineering at The University of Sydney and International Scholar at Kyung Hee University, Yongin-si, Gyeonggi-do, Korea. He performed postdoctoral research at University College London, London, U.K., and then at Philips Research Labs in Germany as a Marie Curie Research Fellow. His research interests include medical instrumentation, integrated circuit design, and bio-inspired systems. He has authored or coauthored more than 80 papers in these research areas and is the inventor on nine patents.

Dr. McEwan is a member of the BioCAS Technical Committee of the IEEE CAS Society.

Eung Je Woo (M’83–SM’09) received the B.S. and M.S. degrees in electronics engineering from Seoul National University, Seoul, Korea, and the Ph.D. degree in electrical and computer engineering from the University of Wisconsin-Madison, Madison, WI, USA, in 1983, 1985, and 1990, respectively.

From 1990 to 1999, he was an Assistant and Associate Professor in the Department of Biomedical Engineering, Konkuk University, Seoul, Korea. Currently, he is a Professor in the Department of Biomedical Engineering, Kyung Hee University, Yongin-si, Gyeonggi-do, Korea. Since 2002, he has been the Director of the Impedance Imaging Research Center. He teaches undergraduate and graduate courses on bioinstrumentation, bioimaging, and inverse problems. His research interests include electromagnetic tissue property imaging of EIT, MREIT, MREPT and QSM, biomedical instrumentation, biomedical signal processing, and computing.

Dr. Woo is a senior member of the IEEE Engineering in Medicine and Biology Society and a member of the Korea Society of Medical and Biological Engineering.

Tong In Oh (M’10) received the B.S. and M.S. degrees in electronics engineering, and the Ph.D. degree in biomedical engineering from Kyung Hee University, Yongin-si, Gyeonggi-do, Korea, in 1999, 2002, and 2006, respectively.

From 2007 to 2009, he was a Postdoctoral Research Fellow in medical physics and engineering, University College London, London, U.K. Currently, he is an Assistant Professor in the Department of Biomedical Engineering, Kyung Hee University. His research interests include analog and digital circuit design, biomedical instrumentation, electrode and smart sensors, cell culture monitoring, biomedical signal processing, and computing.

Dr. Oh is a member of the IEEE Engineering in Medicine and Biology Society and the Korea Society of Medical and Biological Engineering.