Portable multi-parameter electrical impedance tomography for sleep apnea and hypoventilation monitoring: feasibility study

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Abstract

Objective: Quantitative ventilation monitoring and respiratory event detection are needed for the diagnosis of sleep apnea and hypoventilation. We developed a portable device with a chest belt, nasal cannula and finger sensor to continuously acquire multi-channel signals including tidal volume, nasal pressure, respiratory effort, body position, snoring sound, ECG and SpO2. The unique feature of the device is the continuous tidal volume signal obtained from real-time lung ventilation images produced by the electrical impedance tomography (EIT) technique. Approach: The chest belt includes 16 electrodes for real-time time-difference EIT imaging and ECG data acquisitions. It also includes a microphone, accelerometer, gyroscope, magnetometer and pressure sensor to acquire, respectively, snoring sound, respiratory effort, body position and nasal pressure signals. A separate finger sensor is used to measure SpO2. The minute ventilation signal is derived from the tidal volume signal and respiration rate. Main results: The experimental results from a conductivity phantom, four swine subjects and one human volunteer show that the developed multi-parameter EIT device could supplement existing polysomnography (PSG) and home sleep test (HST) devices to improve the accuracy of sleep apnea diagnosis. The portable device could be also used as a new tool for continuous hypoventilation monitoring of non-intubated patients with respiratory depression. Significance: Following the feasibility study in this paper, future validation studies in comparison with in-lab PSG, HST and end-tidal CO2 devices are suggested to find its clinical efficacy as a sleep apnea diagnosis and hypoventilation monitoring tool.

1. Introduction

The prevalence of obstructive sleep apnea (OSA) is high, especially in elderly and obese groups (Hiestand et al 2006). Polysomnography (PSG) is recommended as a gold standard for its diagnosis (Berry et al 2017). During a PSG test, a comprehensive set of physiological signals including electroencephalography (EEG), electrooculography (EOG), chin and leg electromyography (EMG), oronasal thermistor, nasal pressure, respiratory effort, oxygen saturation (SpO2), body position and electrocardiography (ECG) are measured from a subject while sleeping overnight in a sleep lab (Kushida et al 2005). The subjects often have difficulty falling asleep because of the first night effect and stress due to many attached sensors (Newell et al 2012). As a cost-effective alternative to the in-lab PSG test, home sleep test (HST) has been used for patients without comorbidity. For HST, a reduced set of signals are acquired from a subject sleeping at home. Nasal pressure, SpO2 and respiratory effort signals are most commonly measured in both PSG and HST to estimate the apnea hypopnea index (AHI) defined as the number of apnea and hypopnea events per hour of sleep (Flemons et al 2003, Berry et al 2017).

The oronasal thermistor signal measures temperature changes of inhaled and exhaled airflows. Since the temperature signal is not proportional to the airflow, its use is mainly to detect apnea. The nasal pressure is a
better estimate of the airflow and has been widely used to detect both apnea and hypopnea in HST as well as PSG. The nasal pressure signal could be contaminated by motion artifacts and it cannot compensate for errors due to oral breathing.

In a different approach, the peripheral arterial tone (PAT) has been used together with SpO₂ to eliminate other sensors at the face and chest (Penzel et al 2004). Alternative methods with simpler human interface have been also suggested using only ECG (Khandoker et al 2009), snoring sound (Penzel et al 1990), actigraph (Hedner et al 2004) or wireless chest motion detection (Lai et al 2011). Though millions of PSG and HST tests are conducted every year, there still exist clinical needs for more quantitative analyses of respiratory signals especially to better determine the severity of sleep apnea (Khalyfa et al 2016). In this paper, we hypothesize that continuous nonintrusive tidal volume measurements can supplement existing PSG and HST devices for more quantitative analyses of respiratory events.

Hypoventilation or inadequate lung ventilation causes hypercapnia and may lead to brain damage or death (Lee et al 2015). Continuous monitoring of hypoventilation is needed for patients with obesity hypoventilation syndrome (OHS), neuromuscular diseases and chronic obstructive pulmonary disease (COPD) (Nowbar et al 2004, Nava and Hill 2009, Mokhlesi 2010). Patients with respiratory depression due to residual anesthetics, opioid or Propofol also needs continuous monitoring of hypoventilation (Eichhorn et al 2010). Monitoring of SpO₂ and end-tidal CO₂ has been used to detect hypoventilation but their efficacy is not validated due to delayed responses and insufficient accuracy (Lynn and Curry 2011, Boing and Randerath 2015). Nonintrusive continuous monitoring of tidal volume could be a practical solution for these clinical needs in hypoventilation monitoring (Williams et al 2017).

A spirometer can accurately measure tidal volume using a face mask or mouthpiece. Though widely used for pulmonary function tests, its adoption for continuous monitoring especially during sleep is not expected. Since 1960s, impedance pneumography has been applied to continuously measure tidal volume. However, its clinical acceptance is slow since it produces erroneous tidal volume signals during obstructive respiratory events (Brouillette et al 1987). Currently, it is mostly used to measure the respiration rate in a patient monitor.

On the other hand, electrical impedance tomography (EIT) has been extensively studied over the last three decades to produce cross-sectional images of an internal conductivity distribution. In EIT, electrical currents are injected and induced voltages are measured through multiple surface electrodes attached around a chosen imaging domain such as the chest (Holder 2004). EIT images are reconstructed from the measured boundary current–voltage data using an image reconstruction algorithm. Though linearized image reconstruction algorithms using a sensitivity matrix are widely used, nonlinear algorithms as well as data-fitting approaches have been also proposed (Adler et al 2009, Seo and Woo 2013, Adler and Boyle 2017, Lee et al 2017). Lately, time-difference EIT has been clinically used for regional lung ventilation imaging during mechanical ventilation for lung protective ventilation (Frerichs et al 2017). In EIT imaging of lung ventilation, pixel values of time-difference EIT images represent air volume changes at that pixel.

In this paper, we suggest adopting the EIT technique for sleep apnea diagnosis and hypoventilation monitoring. Unlike existing EIT devices designed for regional lung ventilation imaging of mechanically ventilated patients, the EIT device for sleep apnea diagnosis should provide typical PSG signals as well as the tidal volume signal. Our approach is to use a portable multi-parameter EIT device with a patient interface including not only multiple electrodes for EIT and ECG data acquisitions but other sensors for signals such as nasal pressure, respiratory effort, body position, snoring sound and SpO₂. The device should produce real-time EIT images of lung ventilation, from which the tidal volume signal is extracted. To be compatible with in-lab PSG tests or used as a HST device, the portable multi-parameter EIT device should be smaller and lighter than currently available EIT devices.

We will describe the design and construction of the device simultaneously acquiring EIT images of lung ventilation and typical PSG signals of nasal pressure, SpO₂, respiratory effort, body position, snoring sound and ECG. To the best of our knowledge, this is the first portable multi-parameter EIT device. After describing the results of its basic hardware tests, we will discuss the results of animal experiments using four pigs for its performance validation against a mechanical ventilator. From the results of a preliminary volunteer study, we will show the feasibility of the proposed technique to supplement existing PSG and HST devices in determining the severity of sleep apnea. Future clinical studies of sleep apnea diagnosis and hypoventilation monitoring using the developed portable multi-parameter EIT device will be suggested.

2. Methods

2.1. Device design

Figure 1 shows the structure of the portable multi-parameter EIT device for sleep apnea diagnosis and hypoventilation monitoring. The main body is connected to the chest belt with 16 active electrodes via the multi-parameter module including switching circuits, pressure sensor, microphone, SpO₂ module, motion
sensor and ECG amplifier. A nasal cannula is connected to the pressure sensor and a separate finger sensor is connected to the multi-parameter module for SpO\textsubscript{2} measurements. Respiratory effort and body position signals are obtained from the motion sensor comprising a three-axis accelerometer, three-axis gyroscope and three-axis magnetometer. The microphone captures the snoring sound signal. ECG is measured using three among 16 electrodes in the chest belt and the heart rate is derived from the ECG signal.

The main body includes the constant current source and four-channel voltmeters to be used for EIT imaging. Their designs are largely based on the previous EIT system designs by Oh et al (2011, 2007a, 2007b, 2008) and Wi et al (2014). In these previous EIT systems, however, 16 separate triaxial cables were used to connect the passive electrodes to the EIT hardware for best signal quality. Since such thick and heavy cables are troublesome for sleep studies, we designed the chest belt to include active electrodes and used thin and light cables to connect the active electrodes to the current source and voltmeters.

For EIT imaging, we need to attach multiple electrodes around the imaging domain. As the number of electrodes is increased, the spatial resolution can be improved (Holder 2004, Seo and Woo 2013). The voltage difference between an adjacent pair of electrodes, however, decreases as the gap between them is decreased for a larger number of electrodes. The number of electrodes should be, therefore, limited considering the signal-to-noise ratio (SNR) of an EIT system to avoid unnecessary hardware complexity. Furthermore, the number of electrodes should be chosen depending on a targeted clinical application. Considering that the lungs occupy relatively large volumes inside the thorax, 16 electrodes seem to be an optimal choice for the portable multi-parameter EIT device design.

During EIT imaging, sinusoidal currents of 1 mA\textsubscript{rms} at 11.25 kHz are injected between a chosen pair of electrodes and induced voltages are measured between other electrode pairs. At 10 kHz, the electrical safety standard limits the patient auxiliary current as 1 mA\textsubscript{rms}. The current amplitude should be decreased to 0.01 mA\textsubscript{rms} at dc and can be increased up to 10 mA\textsubscript{rms} at 100 kHz (ISO 2005). The choice of 11.25 kHz was a tradeoff between the allowed amplitude of injection current and the adverse effects of stray capacitances at higher frequencies.

The current-injection and voltage-sensing operations are repeated for all 16 electrode pairs to collect a set of current–voltage data for one EIT image or frame. The measured current–voltage data around the chest are then used to reconstruct cross-sectional EIT images of the chest with the temporal resolution of 25 frames s\textsuperscript{-1}. Following numerous previous studies confirming that the pixel values are proportional to changes in regional air volumes (Holder 2004, Frerichs 2000, Adler and Boyle 2017), the tidal volume signal is extracted from the reconstructed EIT images. The tidal volume signal meets the American Academy of Sleep Medicine (AASM) recommendation of a minimum sampling frequency of 25 Hz for respiratory signals (Berry et al 2017).

2.2. Chest belt

The chest belt is placed at the 5th or 6th intercostal space of the subject for EIT imaging of lung ventilation (Karsten et al 2016). Inside the chest belt, there is a flexible printed circuit board (fPCB) with 870 mm length, 25 mm width and 0.32 mm thickness covered by a thin sheet of silicone as shown in figure 2(a). For continuous operation of several hours during sleep, the electrodes attached to the chest belt must maintain stable contacts.
with the skin. To use commercially available Ag/AgCl electrodes with adhesive, 16 eyelet connectors are soldered on the fPCB.

Figure 2(b) shows a differential buffer amplifier module connected to each electrode. Sixteen of them are soldered on the fPCB near the eyelet connectors. Two operational amplifiers (OPA2140, Texas Instrument, USA) and one difference amplifier (AD8139, Analog Device, USA) are used to implement each differential buffer amplifier with a high common-mode rejection ratio (CMRR) and low output impedance. Its high-pass cutoff frequency is 100 Hz to reduce low-frequency noise and motion artifacts at the electrode–skin interface. The outputs of 16 differential buffer amplifiers are routed to the multi-parameter module described in the next section for the measurements of voltages induced by injected currents. All 16 electrodes are separately routed on the fPCB to the multi-parameter module so that currents can be injected between any chosen pair of electrodes.

2.3. Multi-parameter module
The multi-parameter module in figure 2(c) is located at the middle of the chest belt. It contains analog switches for EIT data collections. It also contains necessary sensors and circuits for nasal pressure, SpO$_2$, respiratory effort, body position, snoring sound and ECG measurements. All functions of the multi-parameter module are controlled by an FPGA (EP3C10F256C8N, Altera, USA). The size of the multi-parameter module is $45 \times 65 \times 20$ mm$^3$ and it has three connectors for the nasal cannula, finger SpO$_2$ sensor and main body. A twin-axial ribbon cable (HQDP series, SAMTEC, USA) is used for the connection of the multi-parameter module to the main body.

2.3.1. Switching of analog signals for EIT data collections
One set of switches are used to choose a pair of electrodes for current injection. Other sets of switches are used to connect four pairs of neighboring electrodes to four voltmeters. For a given current injection, each voltmeter is switched four times to measure 16 voltage data from all 16 adjacent electrode pairs. Among those 16 voltage data, three voltage differences measured from two current-injection electrodes are discarded since they are affected by electrode–skin contact impedances. The switching circuits are implemented using T-bar switches (MAX4545, Maxim Integrated, USA).

2.3.2. Nasal pressure measurements
Nasal pressure is measured using a pressure sensor (HSCDRRD001NDSA3, Honeywell, USA). The measured gage pressure is represented by two bytes of digital data including the sensor status. Adding two bytes of header data, four bytes of digital data are transmitted to the main body through the twin-axial ribbon cable. The maximum sampling frequency of the nasal pressure signal is 2 kHz.

2.3.3. SpO$_2$ measurements
A disposable SpO$_2$ sensor (6000CA, Nonin Medical Inc., USA) is attached to the finger of a subject and connected to the pulse oximeter module (OEM III, Nonin Medical Inc., USA). The pulse oximeter module transmits the photoplethysmography (PPG) signal and SpO$_2$ data with a maximum sampling frequency of 3 Hz.

2.3.4. Respiratory effort and body position measurements
A motion chip (MPU-9250, InvenSense, USA) including a three-axis gyroscope, three-axis accelerometer and three-axis magnetometer is used to measure respiratory effort and body position signals. The acquired data are transmitted to the main body with a maximum sampling frequency of 250 Hz. We adopted the method proposed by Pires et al (2016) to extract the respiratory effort and body position signals using all of these three sensors.
2.3.5. Snoring sound measurements
An audio sensor (MP23AB02B, STMicroelectronics, Switzerland) including a microphone and 12-bit ADC (ADS7042, Texas Instruments Inc., USA) is used to capture snoring sound. Sound data with a sampling frequency of 1 kHz are transmitted to the main body.

2.3.6. ECG measurements
One-channel digital ECG amplifier (ADS1291, Texas Instruments Inc., USA) is used to measure the ECG signal with the maximum sampling frequency of 200 Hz. The ECG signal is used to extract the heart rate signal.

2.4. Main body
The main body includes the analog backplane and communication module that are connected to the analog signal and digital data lines, respectively, from the multi-parameter module. It also includes modules of EIT hardware, microprocessor with I/O devices and isolated power supply.

2.4.1. EIT module
The EIT module consists of one balanced constant current source and four digital voltmeters. Technical details of the current source and voltmeters are available in Oh et al (2011) and Wi et al (2014). Each output of the balanced current source (source or sink) can be connected to any one of 16 electrodes through the analog backplane and analog switches in the multi-parameter module. Similarly, four chosen output signals from 16 differential buffer amplifiers are connected to four voltmeters in the EIT module.

2.4.2. Communication module
The communication module is implemented using an FPGA (EP3C10F256C8N, Altera, USA), which arbitrates all data communications between the main body and multi-parameter module. The speed of the serial communication is 11.25 Mbps. It also controls timing of the EIT module for sequential current injections and voltage measurements (Wi et al 2014).

2.4.3. Microprocessor module
The main body is designed using a Raspberry Pi 3 processor module (Raspberry Pi 3 Model B, Raspberry Pi Foundation, UK). The microprocessor module is equipped with various input and output ports including a color LCD, USB, buttons and LEDs.

2.4.4. Isolated power supply module
The portable device is line-powered using an isolated switching regulator power supply (ECM40UT31 & ECM40UT35, XP Power, Singapore). The output of the switching regulator power supply is linearly regulated for better performance.

2.5. Embedded software
The embedded software runs in the Raspbian operating system (Raspbian Stretch, Raspberry Pi Foundation, UK). The QT library (Qt, The Qt Company, Finland) is utilized to implement all graphics and user interface functions. EIT image reconstructions and subsequent real-time digital signal processing tasks are implemented in the embedded software.

   The measured EIT data are influenced by lung ventilation, cardiac blood flow, movements of the heart and large blood vessels, breathing-related chest movements and other motion artifacts. We used a sensitivity matrix derived from the lead field theory to extract only the influence of lung ventilation and reconstructed images of air distributions in the lungs (Malmivuo and Plonsey 1995, Seo and Woo 2013). For EIT image reconstructions, the fidelity-embedded regularization (FER) algorithm is used with the subspace-based motion artifact rejection method (Lee et al 2017). Repeating the image reconstructions 25 times per second, a time series of lung ventilation images is produced at 25 frames s$^{-1}$.

   The tidal volume signal is extracted from the time series of the EIT images as a pixel sum of the lung regions. From the extracted tidal volume signal, we computed tidal volume, inspiration time, expiration time and respiration rate for every breath. Minute ventilation is calculated by adding tidal volumes within the most recent one-minute moving window. The tidal volume signal is processed together with other signals of nasal pressure, respiratory effort, body position, $\text{SpO}_2$, snoring sound and ECG for real-time data storage and display.

3. Results
We conducted a series of basic performance tests of the developed portable EIT device using agar objects and a saline phantom. Animal experiments using four pigs were followed to validate the accuracy of the measured tidal
volume compared with the amount of supplied air volume from a mechanical ventilator. Finally, a volunteer study was conducted to show the feasibility of the device to detect respiratory events using the tidal volume signal extracted from reconstructed time-difference EIT images.

### 3.1. Evaluation of basic performance

Figure 3 shows the developed portable device and its screen capture. The performance of the differential buffer amplifiers inside the chest belt was tested by measuring the SNRs in acquired EIT voltage data with and without using the differential buffer amplifiers. The audio analyzer (U8903A, Agilent Technologies, USA) was connected to the analog outputs of the switching circuits using cables with three different lengths of 0, 1.5 and 3 m. The measurand was a cylindrical agar object of 0.25 S m$^{-1}$ conductivity with 16 Ag/AgCl electrodes (Red Dot, 3M, USA) attached around its surface. Table 1 shows that the SNR decreased significantly without using differential buffer amplifiers. This indicates that the differential buffer amplifiers at backsides of electrodes are essential for high-quality EIT image reconstructions. High-quality images in turn produce tidal volume signals with high accuracy.

To check if the differential buffer amplifiers can be replaced by simpler single-ended buffer amplifiers, CMRRs were measured. The CMRR using the differential buffer amplifier was at least 5 dB higher than that using the single-ended buffer amplifier. To test the long-term stability, the developed portable device was connected to a resistor phantom for 8 h of continuous data collections. The minimum SNR was maintained at about 80 dB throughout the 8 h test.

### 3.2. Phantom experiment

A cylindrical phantom with 15 cm diameter was filled with a saline of 0.25 S m$^{-1}$ conductivity. Sixteen stainless steel electrodes were attached to the boundary of the phantom and connected to the chest belt. Figure 4(a) shows the reconstructed EIT image of the phantom when a circular acrylic insulator with 1 cm diameter was placed inside the phantom. Figure 4(b) is a signal of the sum of all pixel values before and after inserting and removing the acrylic insulator. Note that the increased amount of fluctuations after inserting and removing the insulator stemmed from movements of water inside the phantom. The developed portable multi-parameter EIT device showed similar performances in terms of the SNR, CMRR and reciprocity error (RE) compared with our previously developed EIT systems using thick tri-axial cables (Oh et al 2011, Wi et al 2014, Oh et al 2007b). The RE was evaluated as the percentage difference between two measured voltage data acquired by swapping two electrode pairs for current injection and voltage measurement.

### 3.3. Animal experiments

The animal experiments using four pigs were conducted in accordance with all the regulations of the Institutional Animal Care and Use Committee (KBIO IACUC 2017035). Each animal was intravenously anesthetized using a
syringe pump (Ketamine & Xylazine 4:1 mixture, 0.5 ml kg$^{-1}$, intravenous injection). The animal was connected to a mechanical ventilator (Hamilton-G5, Hamilton Medical, Switzerland) by tracheal intubation. The status of the animal was continuously monitored using a patient monitor (IntelliVue MP50, Philips, Netherlands). The chest hair was removed and Ag/AgCl electrodes were attached at the fifth intercostal space. The portable device and the chest belt were connected with a 1 m twin-axial ribbon cable. EIT voltage data were acquired by injecting 1 mA rms current at 11.25 kHz. To obtain EIT images with proper anatomical structure information, the chest shape and electrode positions were measured using a 3D scanner (Sense, 3D systems, USA). During animal experiments using the mechanical ventilator, nasal pressure and snoring sound signals were not measured.

Figure 5 shows acquired signals from a mechanically ventilated pig with controlled airway volumes of 100, 200, 300, 400 and 500 ml. The tidal volume signal in figure 5(a) was extracted from the reconstructed EIT images in (f) and calibrated by using the linear relation in figure 6(b) that is explained later in this section. Figures 5(b)–(e) are the chest motion (respiratory effort), airway pressure from the ventilator, airway volume from the ventilator and ECG signals, respectively. The reconstructed EIT images and tidal volumes extracted from the images closely followed the controlled airway volumes by the mechanical ventilator.

Figure 5 shows acquired signals from a mechanically ventilated pig with controlled airway volumes of 100, 200, 300, 400 and 500 ml. The tidal volume signal in figure 5(a) was extracted from the reconstructed EIT images in (f) and calibrated by using the linear relation in figure 6(b) that is explained later in this section. Figures 5(b)–(e) are the chest motion (respiratory effort), airway pressure from the ventilator, airway volume from the ventilator and ECG signals, respectively. The reconstructed EIT images and tidal volumes extracted from the images closely followed the controlled airway volumes by the mechanical ventilator.

The signals in the dotted box in figure 5 are magnified in figure 6(a) to show their details at 300 ml airway volume from the mechanical ventilator. The tidal volume signal is synchronized with the airway volume signal provided by the mechanical ventilator. Figure 6(b) plots the linear relation between the controlled airway volume from the mechanical ventilator and the tidal volume extracted from the reconstructed EIT images. The $R^2$ value using all data from four pigs was 0.98.

Figure 7 shows the tidal volume signal during a simulated apnea condition for about 16 s by setting the mechanical ventilator to produce zero airway volume. The reconstructed EIT images of the animal at three time moments marked by the red circles are shown in figures 7(a)–(c). Note that the end-expiratory lung volume (EELV) increased after the apnea as marked by the red arrows and dotted line. The image in figure 7(d) shows the changes in the EELV after the apnea.

3.4. Volunteer experiment

We conducted a pilot study of one volunteer with severe OSA. Figure 8(a) shows the acquired signals for 100 s before falling asleep. Though some noticeable changes in the tidal volume and respiratory rate occurred, no respiratory event was observed in figure 8(a) since the subject was still awake. Figure 8(b) shows the acquired signals for 100 s during sleep. The uncalibrated tidal volume change (TVC) signal clearly shows two respiratory events, that can be classified as obstructive sleep apneas by observing the respiratory effort signal. Figure 9 shows three different cases of apneic events including obstructive apnea with oxygen desaturation in (a), obstructive hypopnea without oxygen desaturation in (b) and central apnea without oxygen desaturation in (c).
4. Discussion

We developed the portable multi-parameter EIT device for potential applications in sleep apnea diagnosis and hypoventilation monitoring. Its unique feature is real-time EIT imaging of lung ventilation at 25 frames s$^{-1}$ and extraction of the tidal volume signal from the time-series of reconstructed EIT images. This tidal volume signal provides information about changes of the tidal volume instead of absolute values. To calibrate the EIT-derived tidal volume signal, one may use a spirometer or a bag with a fixed amount of air. From animal experiments using four swine subjects, we found that the mean error between the calibrated tidal volume and the controlled airway volume from the mechanical ventilator was less than $\pm 2.6\%$ with $R^2$ value of 0.98.
For sleep apnea diagnosis, the tidal volume signal extracted from EIT images can be accompanied by other signals such as nasal pressure, respiratory effort, body position, snoring sound, $\text{SpO}_2$ and ECG. Though the tidal volume signal might be able replace the nasal pressure and oronasal thermistor signals, future studies are required to validate this possibility. The device in its current form can be adopted for HST without sleep staging. In this figure:

Figure 7. Tidal volume signal during a simulated apnea by the mechanical ventilator. Reconstructed EIT images (a) before, (b) during and (c) after the apnea. (d) is the image of the increased end-expiratory lung volume (EELV) between two end-expiratory times before and after the simulated apnea.

Figure 8. Results of a preliminary volunteer study: (a) acquired signals before falling asleep without any respiratory event and (b) during sleep with two obstructive apneic events accompanied by oxygen desaturation. The tidal volume change (TVC) signal is plotted with an arbitrary unit (AU) since it was not calibrated.
case, the AHI could be underestimated since the monitoring time is always longer than the sleep time (Berry 
et al 2017). We note that the simultaneously measured ECG signal from the chest belt allows the heart rate variability (HRV) analysis that can be used to estimate the sleep time (Penzel 
et al 2003, Willemen 
et al 2015, Yoon 
et al 2017, 2018). The body movement signal from an accelerometer can also be added to the sleep time estimation for better performance. For in-lab PSG tests, a simplified version of the device providing only the tidal volume signal can supplement existing PSG devices.

Personalized treatment of a sleep apnea patient considering causes and severity of the disease and long-term comorbidity risk has been emphasized for better outcomes. We speculate that the quantitative measures of tidal volume, minute ventilation, inspiration time and expiration time provided by the developed device could be useful to determine severity of sleep apnea. When the developed portable multi-parameter EIT device is used together with a positive airway pressure (PAP) device, those quantitative measures may be used to estimate the loop gain (Wellman 
et al 2011). In addition, PAP titrations at home could be pursued by using the developed portable device.

Since we showed in this paper only the feasibility of using the EIT technique for sleep apnea diagnosis and hypventilation monitoring, future studies of performance evaluation are needed to show its clinical efficacy. Especially for hypventilation monitoring, comparison of the tidal volume signal with the end-tidal CO$_2$ and SpO$_2$ signals should be investigated. The following questions could be addressed in future studies.

- Is the calibration of the tidal volume signal required to detect apnea and hypopnea?
- Is it possible to improve the accuracy of automatic apnea and hypopnea detections by using the tidal volume signal and thereby reduce the subsequent manual scoring time?
- Is it possible to detect apnea and hypopnea using the tidal volume signal alone without nasal pressure, oronasal thermistor and SpO$_2$ signals?
- Can the tidal volume signal quantify hypventilation without end-tidal CO$_2$ and SpO$_2$ signals?
- Can the tidal volume signal detect hypventilation earlier than end-tidal CO$_2$ and SpO$_2$ signals?

Such future studies should consider the fact that obesity is prevalent among sleep apnea patients. Though the current belt design includes two expandable parts, it will be necessary to prepare belts with a few different lengths. We should also consider the limitations of the proposed method in designing a future validation study. Since the injected current spreads three-dimensionally inside the chest, the slice thickness in EIT is not clearly defined. Using 16 electrodes around the chest with equal gaps among them, the slice thickness could be a few centimeters considering the sensitivity map explained by Adler and Boyle (2017). Since the tidal volume signal is extracted from single-slice EIT images, there could occur errors when the air distribution in the lungs is highly inhomogeneous along the longitudinal direction. The accuracy of the tidal volume signal extracted from single-slice EIT images may depend on the position of the chest belt in such cases.

Figure 9. Three different cases of apneic events from the preliminary volunteer study: (a) obstructive apnea accompanied with oxygen desaturation, (b) obstructive hypopnea without oxygen desaturation and (c) central apnea without oxygen desaturation. The tidal volume change (TVC) signals are plotted with an arbitrary unit (AU) since the signals were not calibrated.
The EIT technique utilizes a sensitivity matrix derived from the lead field theory to produce cross-sectional images of lung ventilation (Malmivuo and Plonsey 1995, Holder 2004, Seo and Woo 2013, Adler and Boyle 2017). Since motion artifacts can be associated with the columns of the sensitivity matrix corresponding to the boundary pixels, we removed motion artifacts using the subspace-based method proposed by Lee et al (2017). During overnight sleep, however, a patient may change body position producing severe motion artifacts. If properly managed by the image reconstruction algorithm, such motion artifacts result in baseline shifts in the tidal volume signal. Most of these signal transients can be handled by post-processing methods to recover stable baselines. If severe motion artifacts during position changes can not be handled either by the image reconstruction algorithm or post-processing methods, such transient data should be discarded.

If a patient sweats during overnight sleep, there could be shunting currents among surface electrodes even though those Ag/AgCl electrodes are well isolated by adhesive form pads. In the worst case of excessive sweating, reconstructed EIT images may fail to provide information about air volume changes in the lungs. Future studies are needed to produce a patient selection guideline for the proposed device to be used in sleep apnea diagnosis and hypoventilation monitoring. The guideline should exclude the patients with electrical stimulators such as cardiac pacemakers and implantable cardioverter defibrillators, impaired skin in the chest area, spinal lesions or fractures and others. The EIT device should not be used with other devices measuring bioimpedance. If ECG, EEG, EMG and EOG are simultaneously measured, their signal qualities should be checked since injected high-frequency currents from the EIT device may have influenced the biopotential signals.

5. Conclusions

The developed portable multi-parameter EIT device could be used as a diagnostic tool to supplement existing in-lab PSG devices and also as a stand-alone HST device. Most significant contribution is the continuous and robust quantification of the tidal volume signal using multiple electrodes around the chest. Clinical applications of the device may also include nonintrusive continuous monitoring of hypoventilation from non-intubated patients with obesity hypoventilation syndrome (OHS), neuromuscular diseases and chronic obstructive pulmonary disease (COPD). Monitoring of respiratory depression during opioid administration and procedural sedation can also be tried. Following the feasibility study described in this paper, future studies should compare the performance of the developed device with existing PSG and HST devices. Carefully designed clinical trials should be followed to evaluate the clinical efficacy of the quantitative measures of tidal volume, minute ventilation, inspiration time and expiration time in sleep apnea diagnosis and hypoventilation monitoring.

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