

Investigation of Bladder Volume Measurement Based on Fringe Effect of Electrical Impedance Tomography Sensors

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This work was supported in part by the National Natural Science Foundation of China under Grant 61701550, and in part by the Fundamental Research Funds for the Central Universities.

ABSTRACT Patients with bladder dysfunction are unable to perceive bladder volume and urinate spontaneously, and hence usually require catheterization. Personalized catheterization can effectively reduce the risk of urinary tract infections caused by catheterization. Electrical impedance tomography (EIT) can provide an effective tool for personalized catheterization by monitoring bladder volume. This study exploits the fringe effect of a 2D EIT sensor to derive the 3D volumetric information about bladder and improve the measurement consistency and accuracy of bladder volume under varied urine conductivities. The EIT sensor is optimized regarding the specified application of fringe effect. The parameters to be optimized include the electrode arrangement, the distance of electrode plane from bladder bottom, and the number of electrodes. Three evaluation criteria are proposed for the optimization. Simulation and experiment confirm the feasibility of the fringe effect-based method for bladder volume measurement, and the performances of the optimized sensor are illustrated. It is concluded that the fringe effect-based method has a better measurement consistency and accuracy against urine conductivity than the widely-used global impedance-based method, and hence provides a new promising alternative for bladder volume measurement.

INDEX TERMS Bladder volume measurement, fringe effect, electrical impedance tomography, sensor optimization.

I. INTRODUCTION

BLADDER dysfunction is very common for the spinal cord injury patients caused by cerebral hemorrhage and stroke. It would cause two critical issues, i.e. unawareness of bladder volume and urinary dysfunction. Currently, catheterization is a useful method to help the patients to urinate. However, it still has some drawbacks such as urinary tract infections and urinary reflux. The most serious are urinary tract infections due to inappropriate catheterization interval, which have intolerably adverse impacts on patients' health and personal lives. Subsequently, this would impose a heavy burden on public medical resources and patients' families. Bladder volume monitoring for

personalized catheterization renders possible to reduce those drawbacks [1], [2], [3], [4].

Traditional techniques for bladder volume measurement are computed tomography (CT) and ultrasound. CT can provide accurate results, but it is expensive and radioactive. For clinic applications, it is generally used for diagnosing diseases such as bladder cancer, rather than the daily monitoring of bladder volume. Ultrasound methods usually require the probe to be pressed against the abdomen to image. It generally requires professionals to operate on a regular basis, so it is inconvenient to be used for continuous monitoring. Electrical impedance tomography (EIT) is a technique which can image the electrical impedance distribution in the

target sensing field by stimulating and measuring on the electrodes mounted on the body surface. It is a promising medical modality since it has the advantages of being non-invasive, portable, radiation free, inexpensive and high temporal resolution [5], [6]. Besides, it can enable continuous measurements because the electrodes can be attached on the surface of human body. Exploiting the fact that urine conductivity is explicitly different from other tissues around the bladder, the volume change of urine can generate obvious conductivity change in the pelvic region. EIT can measure the conductivity changes and derive the corresponding changes of bladder volume.

Over the past decade, many methods have been attempted to measure bladder volume based on EIT [7]. Dunne *et al.* suggested that the urination of a patient can be determined by the bladder status. They proposed a method based on machine learning by training the measured voltages [3] and the reconstructed EIT images [8] to determine the bladder status. Schlebusch *et al.* found that urine conductivity significantly affects the measurement accuracy of bladder volume by EIT [9]. He *et al.* simulated a planar electrode array to image the bladder in three-dimension (3D), which obtained the information about the bladder location in the third dimension [10]. Compared to 3D-imaging, 2D imaging only needs to solve fewer unknowns, which reduces the ill posed of the inverse problem [11]. Then a single-layer electrode sensor can be used, which leads to less hardware and software costs. Leonhäuser *et al.* compared EIT with ultrasound method regarding bladder volume monitoring [12]. The results showed that EIT is a promising tool and the electrode arrangement of an EIT sensor has an impact on the measurement accuracy. Liu *et al.* proposed a spatiotemporal structure-aware sparse Bayesian learning (SA-SBL) framework for solving the time-continuous EIT inverse problems [13]. This method can reduce artifacts and improve image quality in EIT imaging. Schlebusch *et al.* compared seven electrode arrangements of a specified EIT sensor with respect to bladder volume measurement [14]. Among those methods, the most widely used is by measuring the global impedance [15], which is similar to the average conductivity index [16], i.e. the average of all the pixel values in the reconstructed EIT image. There are two issues with the global impedance-based approach: 1) The fringe effect due to the simplification of 3D-imaging into 2D is not considered, which would bring measurement errors; 2) The global impedance is affected by both the bladder volume and urine conductivity. When the bladder volume and urine conductivity change simultaneously, there is an inevitable measurement error with this method. In fact, urine conductivity is not the same between different individuals or for the same individual at different times, affected by one's diet and many other factors [12]. These two issues prevented the EIT measurement of bladder volume from being applied to clinical applications.

The purpose of this study was to apply fringe effect for bladder volume measurement and to explore a robust method

which can make use of the 3D information contained in a conventional 2D EIT image to enhance the measurement accuracy and is immune to the changes of urine conductivity. The proposed method can obtain the information about the bladder location in the third dimension to estimate the bladder volume. On this basis, three geometric parameters of the adopted EIT sensor that may affect the measurement accuracy are optimized, which are the electrode arrangements, the distance between the electrode plane and bladder bottom and the number of electrodes. Fringe effect will be explained with details in Section II. In Sections III and IV, EIDORS-based simulations and in-vitro experiments are conducted to illustrate the sensor optimization process and verify the feasibility of the proposed method. Finally, conclusions are drawn.

II. METHOD

A. WHAT IS FRINGE EFFECT

Unlike the hard-field sensing in CT, EIT is a soft-field sensing technique. The electric field between a pair of injection electrodes diffuses in the 3D space [17], and fringe effect is referred to as the spread of electric field into the third dimension near the top and bottom edges of electrodes in a 2D EIT sensor for cross-sectional imaging. Consequently, object distribution in the third dimension also has an influence on the 2D imaging result, which would cause inaccuracies in 2D imaging. In general, the fringe effect needs to be reduced to improve 2D imaging [17], [18], [19]. However, it can also be used to derive the object location in the third dimension [11], [20], [21], and its characterization is redefined and adapted to bladder volume measurement in this paper.

EIT generally adopts small “pin” electrodes. A pair of adjacent “pin” electrodes in an EIT sensor with the popular adjacent excitation strategy can be approximated as an electric dipole. When excited with low-frequency electrical current, the electric potential at an observation point sufficiently far away from the two adjacent “pin” electrodes (or one dipole) can be expressed as:

$$V = \frac{ql\cos\theta}{4\pi\epsilon_0r^2} \quad (1)$$

where q is the amount of charge, l is the distance between the two electrodes, ϵ_0 is the vacuum permittivity and θ is the angle as defined in Fig. 1. The electric potential distribution is illustrated in Fig. 1 (a), in which the fringe effect is shown as the distorted electric field lines above or below the two electrodes.

Based on above approximations, the influence of object location in the third dimension (i.e., axial location) on its 2D reconstructed image can be explained in details. For illustration, an object was assumed to move vertically from point ‘a’ to ‘b’, as in Fig. 1 (a). Because the reconstructed object location in the 2D image is determined by the intersection of the electric field lines crossing through the object and the

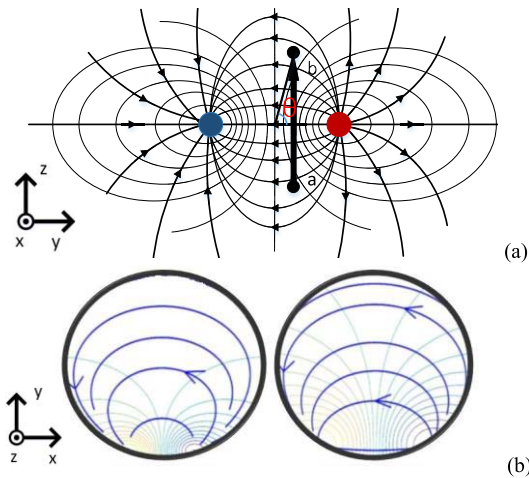


FIGURE 1. Electric field and equipotential distributions when a pair of electrodes is excited. (a) Vertical section. (b) Observation planes at different heights above electrode plane (left $z = 0.1$ cm and right $z = 3$ cm).

electrode plane [21], it would be close to the sensor center when it is located at point ‘a’ or ‘b’ due to the severe distortions of electric field lines as shown in Fig. 1 (a), even though it is actually not. This can be further explained by the potential distributions in observation planes at different heights above the electrode plane, as shown in Fig. 1 (b). It can be observed that the electric field lines in the observation plane far from the electrode plane (i.e., when $z = 3$ cm) clearly diffuse towards the sensor center. The intersection of the electric field lines through the object and the electrode plane keeps moving when the ball moves vertically (or axially) from point ‘a’ to ‘b’ as in Fig. 1 (a), which would result in a changing location of the ball in the reconstructed 2D image.

Simulations were performed to verify the above theoretical analysis. In the simulation, 16 circular electrodes with radius of 0.05 cm were evenly distributed. The blue-colored ball with radius of 0.1 cm was not in the cross-sectional center and its vertical location was changed from $z = 0.5$ to 3.5 cm, as shown in Fig. 2 (a). The vertical location of EIT electrodes was 2cm at the z axis. In each specified vertical location, a 2D image was reconstructed. The reconstructed object location is close to the sensor center ($(x, y) = (0, 0)$) at the beginning and moves away when the ball approaches the sensor plane (i.e., $z = 2.0$ cm in Fig. 2 (a)). It moves towards the sensor center again when the ball moves away from the sensor plane, as shown in Fig. 2 (b). When the ball is inside the sensor plane, the reconstructed location of the ball is the same as its actual location ($(x, y) = (0, 0.5)$). Those simulation results are consistent with the above theoretical analysis.

B. BLADDER VOLUME MEASUREMENT BASED ON FRINGE EFFECT

When the bladder volume increases, the bladder extends upward along the abdominal wall [22]. Additionally, the

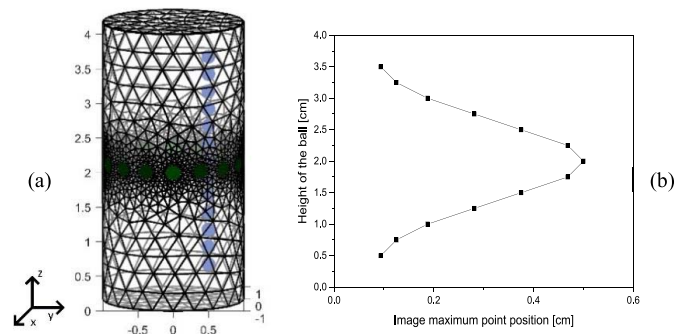


FIGURE 2. EIT simulation with a ball at different vertical locations. (a) Simulation setup. Coordinates of ball: $(x, y) = (0, 0.5)$, $z = [0.5:0.25:3.5]$. (b) Reconstructed locations of ball at different heights.

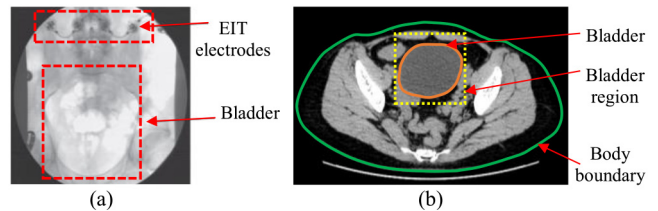


FIGURE 3. (a) CT image of bladder (radial section). (b) Bladder region based on CT image (transverse section).

location of bladder bottom has ignorable changes when its volume increases [3], [14]. Therefore, the location change of bladder center in the axial direction can reflect the change of bladder volume. Due to the fact that the bladder bottom is located in the lower ventral region, as shown in Fig. 3 (a), EIT sensor is usually installed above this region (i.e., the bladder is outside the sensor plane when its volume is small) for convenience and reliable contacts between electrodes and body skin. In this case, the bladder volume is measured by the fringe field excited by the EIT sensor. On the other hand, the cross-sectional location of the bladder is not in the center of abdomen and its location change in the axial direction would induce the change of fringe effect, i.e., the reconstructed bladder center in the 2D EIT image would move accordingly as described above. Therefore, the bladder volume can be estimated by measuring the location change of bladder center in the 2D EIT image. For this purpose, the change in the gravity center of the reconstructed bladder region was utilized as a measure of the bladder volume.

This approach has several advantages. Firstly, it exploits the 3D information about the bladder location contained in the reconstructed 2D EIT image for the measurement of bladder volume, which can enhance the measurement accuracy since the 3D location of bladder is explicitly related to its volume. Secondly, since the change in the urine conductivity does not affect the geometry and thus reconstructed location of bladder, it does not affect the location of the gravity center in the reconstructed 2D image. This makes the proposed method superior to the global impedance-based method since the latter relies on the grey levels in the reconstructed image which is subject to the urine conductivity.

C. OPTIMIZATION OF EIT SENSOR

When measuring the bladder volume based on fringe effect, the structural and geometric parameters of EIT sensor can affect the measurement accuracy. This study selects three critical ones, i.e., electrode arrangements, the distance of the electrode plane from the bladder bottom and the number of electrodes, to optimize. The reasons are as follows:

- 1) The influence of the electrode topology on the measurement of bladder volume has been explored by researchers [12], [14], but it has not been specifically optimized for the fringe effect-based measurement. In the study of partial boundary imaging by Cao and Xu [23], the imaging quality near the dense-electrode side can be significantly improved. These findings shed an inspiration on the current study. Since the bladder is closer to the front wall of abdomen, namely ventral side, it may increase the bladder sensitivity by increasing electrode density in the ventral side.
- 2) The distance between the electrode plane and the bladder bottom would affect the change trend of fringe effect with increasing bladder volume and the detection sensitivity in the bladder region. If an inappropriate distance is chosen, it would lead to a non-monotonous change of fringe effect. If the distance is too far, the measurement sensitivity would be too small.
- 3) The number of electrodes has an effect on the image reconstruction. In general, the more the number of electrodes, the better the imaging resolution would be. But it would cause additional issues, such as, more electrodes would bring inconveniences for use and more issues of contact instability in practical applications. The purpose of this parameter optimization is to simplify the sensor design while still maintaining sufficient measurement accuracy.

The abovementioned three parameters are optimized sequentially by evaluating their influences on the reconstructed image and derived gravity center of the bladder. The electrode arrangement was believed to be the very important and has been studied by many researchers [12], [14], then the appropriate distance of the EIT sensor plane from bladder bottom and finally the number of electrodes.

III. SIMULATION

A. SIMULATION SETUP

Simulation was conducted using an open-source software package, i.e., Electrical Impedance Tomography and Diffuse Optical Tomography Reconstruction Software (EIDORS) [24]. The EIT reconstruction adopts time-difference imaging due to the distinct impedance change in the bladder region with increasing bladder volume. The measurement for the empty bladder is taken as the reference. The reconstructed EIT image shows the change in electrical impedance distribution relative to the reference.

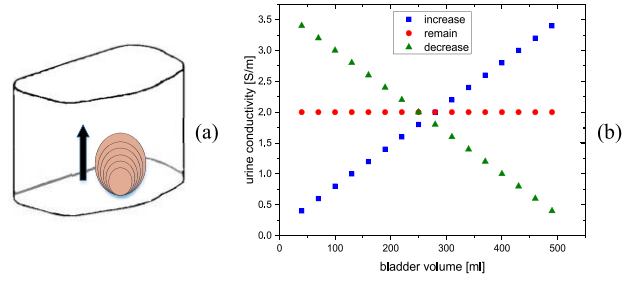


FIGURE 4. (a) EIT model in simulation. (b) Combinations of different bladder volumes and urine conductivities in simulation.

The EIT model simulates the shape of human abdomen by referring to the Visible Human Project [25], as shown in Fig. 4 (a). The contour model of abdomen was built using the `ng_mk_extruded_model` in EIDORS. The bladder is approximated as an upwardly extending ellipsoid with a constant bottom location [14]. The center of the ellipsoid is located at the half way in the anterior region of the model [3], [7]. The ranges of bladder volume and urine conductivity were chosen based on the data set used by Dunne *et al.* [3], [26]. The height of bladder varied from 0 to 11.6 cm in the simulation. The urine conductivity varied from 0.4 to 3.4 S/m, and the conductivity of surrounding tissues, i.e., the background medium, was set to be 0.2 S/m. In order to verify the immunity of the proposed method to the change of urine conductivity, three changing trends of urine conductivity were set in the simulation, which can be denoted as increase, remain and decrease, respectively. There are 16 simulation points for each changing trend, as shown in Fig. 4 (b).

B. EVALUATION CRITERIA

To optimize the EIT sensor design, three evaluation criteria were proposed.

Fringe Effect Sensitivity: This criterion was proposed to evaluate the sensitivity of measured fringe effect to volume changes. The fringe effect sensitivity (FES) is defined as:

$$FES = \frac{\left| \sum_{i=1}^3 g(V_{max,i}) - \sum_{i=1}^3 g(V_{min,i}) \right|}{3 \times \varnothing_{V_{max}}} \quad (2)$$

where g represents the ordinate of the gravity center of the reconstructed bladder region. v_{max} and v_{min} represent the maximum and minimum bladder volumes to be detected, respectively. $\varnothing_{V_{max}}$ represents the cross-sectional diameter in pixels at the maximum volume, and $i = 1, 2, 3$ represent the changing trends of bladder conductivity, i.e., increase, remain and decrease, respectively. This parameter indicates the maximum change of fringe effect during the changing process of bladder volume. FES should be the most important evaluation criteria. A larger FES means that the measurement is more sensitive to the changes in bladder volume.

Measurement Consistency at different Urine Conductivities: Urine conductivity is easily affected by water intake, exercise and other factors and thus it would fluctuate within a certain range [9]. As mentioned

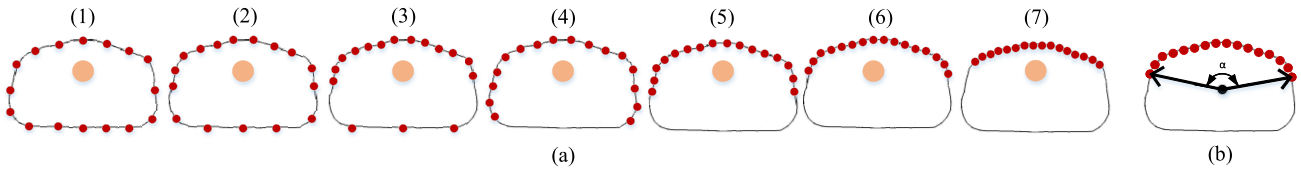


FIGURE 5. (a) Seven typical electrode layouts. (b) Open angle covered by electrode region.

previously, the variation in the urine conductivity would cause the failure of globe impedance-based method. Thus, it is critical to study the influences of urine conductivity on the performance of proposed method. To evaluate the measurement consistency, the deviation of each estimated gravity center of bladder from its mean value at different urine conductivities for the same volume can be normalized to be a volume error:

$$\begin{aligned} \text{Conductivity Consistency} &= \frac{g(V,i) - \bar{g}_V}{\Delta g} \cdot \Delta V \\ \Delta V &= (V_{max} - V_{min}) \\ \bar{g}_V &= \frac{\sum_{i=1}^3 g(V,i)}{3} \\ \Delta g &= \bar{g}_{Vmax} - \bar{g}_{Vmin} \end{aligned} \quad (3)$$

This criterion should be as small as possible, otherwise the estimated volumes would be very different at different urine conductivities.

Measurement Sensitivity in Bladder Region: The mean measurement sensitivity in the bladder region is defined as:

$$\text{SBR} = \frac{\sum_{i=1}^N \sum_{j=1}^M Z_{ij}}{MN} \quad (4)$$

The Z_{ij} is the sensitivity value at i_{th} pixel in the bladder region regarding the j_{th} measurement by the EIT sensor. $M = n \times (n - 3)/2$ is the number of independent measurements by the EIT sensor with n electrodes, and $M=104$ in EIT sensors with 16 electrodes, for example. The 2D cross-sectional plane for sensitivity analysis is selected to be at around half the axial height of bladder with the maximum volume, e.g., 5 cm from the bladder bottom.

C. SIMULATION RESULTS

Optimization of Electrode Arrangement: To optimize the arrangement of 16 electrodes in a typical EIT sensor, several typical layouts, as showed in Fig. 5 (a), were set up as follows:

- 1) uniformly distributed in the sensor periphery;
- 2) 11 electrodes uniformly distributed in the front half of the periphery with 5 electrodes in the rear half;
- 3) 13 electrodes uniformly distributed in the front half with 3 in the rear half;
- 4) uniformly distributed in the front three quarters;
- 5) uniformly distributed in the front half;
- 6) uniformly distributed in the front 42.6%;
- 7) uniformly distributed in the front 34.5%.

Noted that the abovementioned percentage refers to the ratio of the opening angle covered by the electrode region

to that by the whole periphery (i.e., $\alpha/360 \times 100\%$), as shown in Fig. 5 (b).

The sum of all the sensitivity maps with each electrode layout is shown in Fig. 6 (a), in which the yellow-colored region indicates the region of higher sensitivity. It can be observed that as the electrode density increases in the ventral side, the region of higher sensitivity also moves towards the front side of abdomen. The measurement sensitivity in the bladder region presents a trend that is increasing first and then decreasing with the specified change of electrode layout as shown in Fig. 6 (b). Fig. 6 (c) shows the reconstructed images with the seven electrode layouts when the urine conductivity is 3.4 S/m and bladder volume is 490 ml. In Fig. 6 (c), the seventh layout gives a deformed image of bladder, which is elongated in the longitudinal direction, while the other layouts reconstruct the bladder well. Fig. 6 (d) shows that the FES increases as the electrodes are concentrated to the front abdomen. The measurement error at different urine conductivities with different electrode layouts are compared in Fig. 6 (e), and the error difference between urine conductivity increase, remain and decrease represents the measurement inconsistency. In general, the inconsistency is larger when the bladder volume is the largest or the smallest. The electrode layouts 1, 2 and 4 have a greater inconsistency when the volume is small, while all the layouts have an acceptable consistency, with a maximum inconsistency of 20 ml when the volume is large.

According to the above discussion, the sixth electrode layout was chosen for further optimization. This is due to the following reasons: (1) it has prominent fringe effect, which is the most important; (2) its mean sensitivity within the bladder region is sufficiently high; (3) the reconstructed image with this array layout is not severely deformed; (4) it has reasonable resistances to the changes in urine conductivity.

Optimization of Distance between Electrode Plane and Bladder Bottom: Nineteen distances between the electrode plane and the bladder bottom were simulated for optimization, which ranges from 1 cm to 19 cm with 1cm as the interval. The sum of all the sensitivity maps with each layout is shown in Fig. 7 (a) and the mean sensitivity in the bladder region is shown in Fig. 7 (b). The red circle represents the bladder boundary in Fig. 7 (a). Since the plane used to observe the sensitivity is 5 cm from the bladder bottom, sensitivity is the highest at 5 cm in Fig. 7 (a) and (b) and approximately symmetric relative to 5 cm. When the electrode plane moves down or up relative to 5 cm, sensitivity drops rapidly. If the electrode plane is too

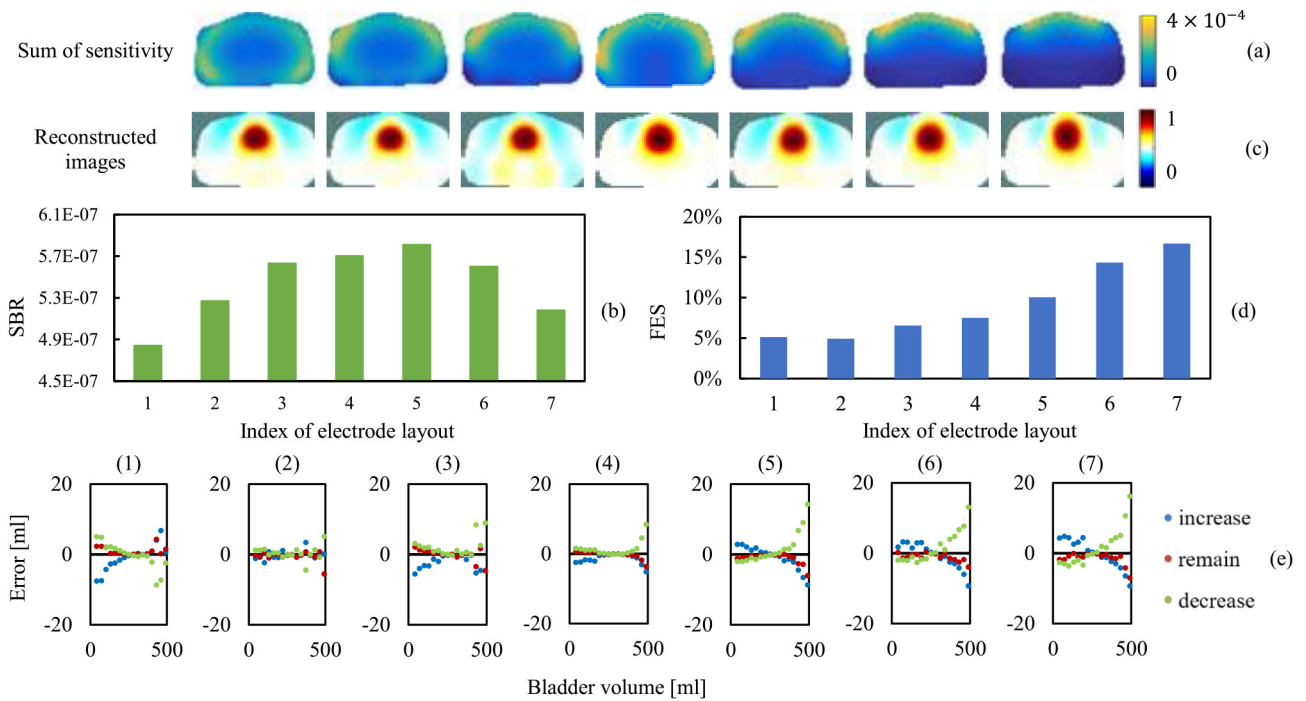


FIGURE 6. Optimization of electrode arrangement. (a) Sum of sensitivity maps. (b) Measurement sensitivity in bladder region. (c) Reconstructed images. (d) FES. (e) Measurement consistency at different urine conductivities.

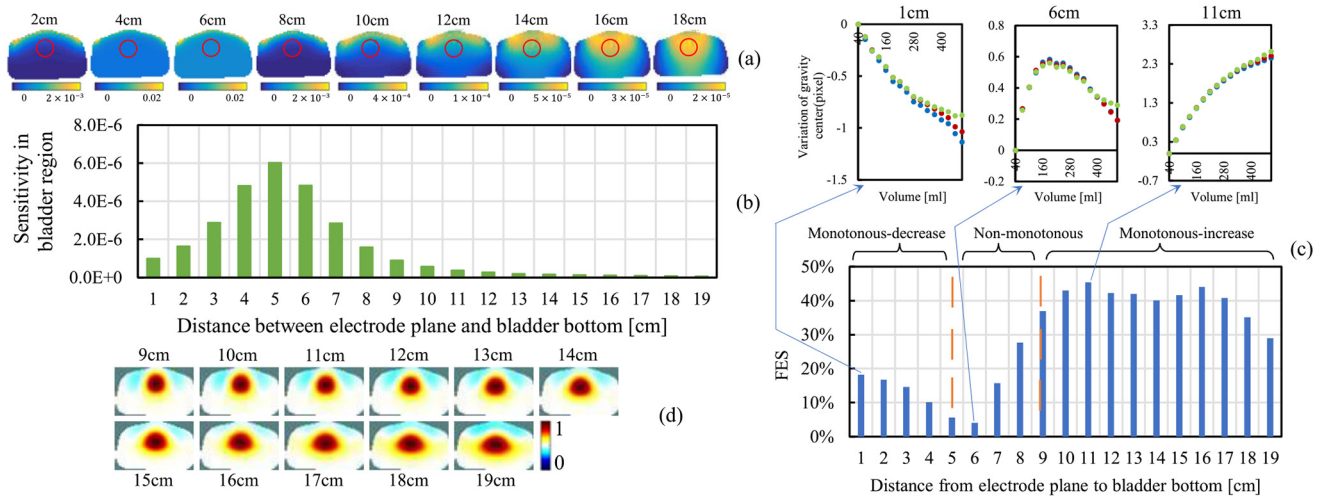


FIGURE 7. Optimization of distance between electrode plane and bladder bottom. (a) Sum of sensitivity maps. (b) Sensitivity in bladder region. (c) FES. (d) Reconstructed images.

far from the bladder (e.g., 19 cm), it would result in low sensitivity and reduce the quality of reconstructed images. The calculated FESs against distance are shown in Fig. 7 (c). In Fig. 7 (c), the first peak appears at the distance of 11 cm, beyond which the FES fluctuates within 3%. The three parts separated by the red segmented lines in Fig. 7 (c) indicate different monotonicity properties of FES change with increasing bladder volume. These monotonicity properties are denoted as monotonous-increase, non-monotonous and monotonous-decrease, examples of which are illustrated on the top of Fig. 7 (c) for the distances of 1, 6 and 11 cm, respectively.

The monotonous-decrease region in Fig. 7 (c) was pre-selected for bladder volume measurement since the FES change is monotonous and sufficiently high (more than 15%). Therefore, a suitable distance can be selected in this region. Images of bladder were reconstructed in Fig. 7 (d) regarding the distances in this region. Obvious deformations occur when the distance is larger than 12 cm. Among 19 distance settings, the Conductivity Consistency is similar. The optimized distance was chosen to be 11 cm due to the same four reasons as in the *Optimization of Electrode Arrangement* part.

Optimization of Number of Electrodes: In order to optimize the number electrodes, 5 settings, i.e., 8, 10, 12, 14 and 16 electrodes, were simulated. It is found that the region of higher sensitivity shrinks to the ventral edge as the number of electrodes increases. The sensitivity in the bladder region decreases as the number of electrodes increases. The FES roughly increases with the number of electrodes. However, the reconstructed images of bladder with the five settings are not significantly deformed and the measurement consistencies at different urine conductivities with the five settings are similar. According to the considerations in previous optimizations, 8 or 10 electrodes would be preferred.

Comparison with Global Impedance-based Method: In order to test the fringe effect-based method for bladder volume measurement, it was compared to the most commonly used method, i.e., the global impedance-based method. The global impedance-based method adopted a common 16-electrode sensor with uniform electrode layout [12], and the fringe effect-based method used an optimized sensor design, i.e., 10 electrodes were uniformly distributed in the front 42.6% of the abdomen periphery. Both sensors are located 11 cm away from the bladder bottom. It is found that the fourth power of gravity center change has a linear relationship with the increase in bladder volume. This enables a measurement calibration with only two known points. The estimated results of bladder volume using these two methods are shown against their real values in Fig. 8. It can be clearly seen that the fringe effect-based method has a good measurement linearity and consistency at different urine conductivities, which confirms the above conclusions. In contrast, the global impedance-based method has poor measurement linearity and consistency towards conductivity changes, especially when the bladder volume is large. Furthermore, the change of global impedance is not monotonic with the change of bladder volume when the urine conductivity decreases with increasing volume, as shown by the green curve in Fig. 8 (a).

IV. EXPERIMENTAL STUDY

A. EXPERIMENTS SETUP

To validate the findings in the simulation, an in-vitro experiment was conducted. Fig. 9 shows the EIT system employed in the experiment, which mainly consists of an AC current source (KEITHLEY 6221), a multiplexer (NI PXI 2530B) and a data acquisition unit (NI PXI 5105). The frequency of excitation current is set to be 10 kHz with an amplitude of 1 mA.

To ensure that the experimental phantoms are similar to the human organs in electrical properties, the biomaterial, i.e., agar, was used to make the phantoms of bladder with varied volumes in the experiment. Molten agar was poured into molds and cooled to form the desired shape, as shown in Fig. 9. Variations in the phantom conductivity were achieved by adding a certain amount of salt into the agar [27]. Conductivities of agars were obtained by measuring the impedance between two electrodes located on the

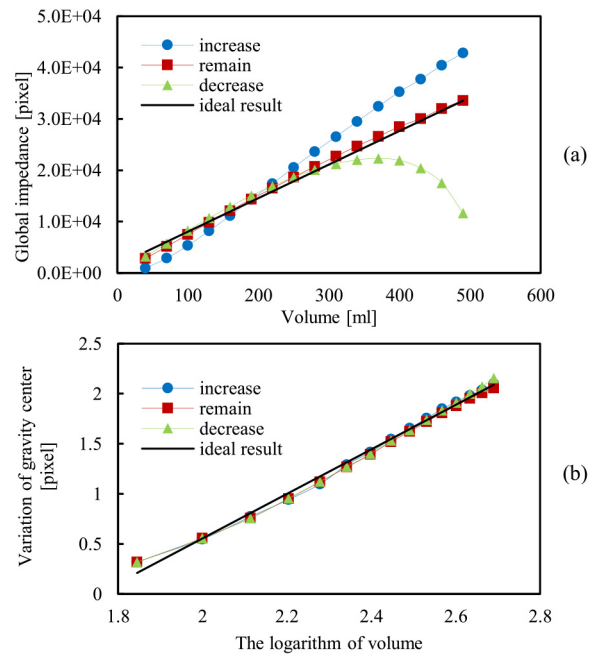


FIGURE 8. Comparison between two methods for measuring bladder volume. (a) Global impedance-based method. (b) Fringe effect-based method.

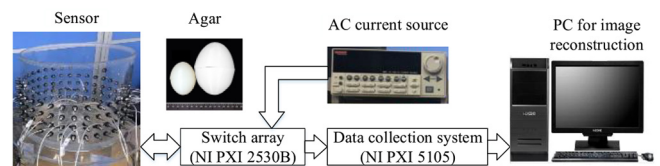


FIGURE 9. EIT system.

TABLE 1. Conductivities of agar ellipsoids.

Agar ball size	Conductivity (S/m)		
130 ml	(a) 0.6	(c) 2.0	(e) 3.4
490 ml	(b) 3.4	(d) 2.0	(f) 0.6

top and bottom of a cylindrical container of known geometry filled with sampled agars. The tissues surrounding the bladder were simulated with a salt solution of 0.2 S/m, and its conductivity was measured with a conductivity meter. Six agar ellipsoids were papered in the experiment. Table 1 shows their respective sizes and conductivities.

A multi-plane EIT sensor was used in the experiment. Note that the fourth electrode plane (from bottom) was selected for the measurement in which 32 electrodes are uniformly mounted on the tank periphery, as shown in Fig. 9. In the beginning, experiments were carried out to validate the simulation results regarding the optimization of electrode layout, the distance between the electrode plane and bladder bottom and the number of electrodes. For the electrode layout, two different layouts, i.e., 16 electrodes uniformly distributed in the entire (Uniformly-Entire) or front half (Uniformly-Half) of periphery, were adopted for comparison. In term of the distance between the electrode plane and bladder bottom, five

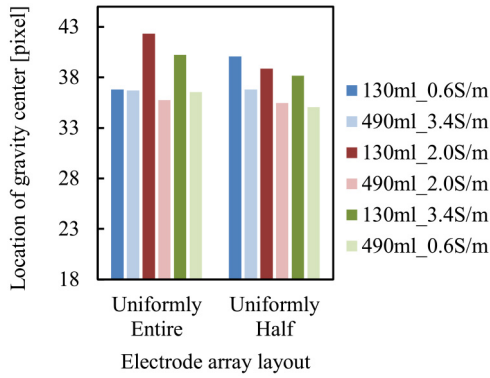


FIGURE 10. Comparison of two electrode layouts.

distances were setup: 1, 4, 7, 10 and 13 cm. As for the number of electrodes, five different settings were investigated: 8, 10, 12, 14 and 16 electrodes.

B. EXPERIMENTS RESULT

The detected gravity center of reconstructed bladder region with each electrode layout is shown in Fig. 10. It can be observed that the Uniformly-Entire layout has a smaller change in the location of gravity center and greater measurement inconsistency, which confirms the simulation results.

As the distance of the electrode plane from the bottom of the agar ellipsoid increases, the location of gravity center (LGC) (i.e., pixel index) increases. This phenomenon indicates that the gravity center moves toward the sensor center, which is consistent with the previous analysis in Section II. When the agar ellipsoid is too close to the electrode plane, there was no consistent change trend of LGC with increasing bladder volume. The reason may be that the FES values at these locations are too small. When the electrode plane is too far from the agar ellipsoid distance, although the change trend is correct, some values are abrupt. It would be due to that the quality of reconstructed image is degraded as analyzed in Section III. The distance of 10 cm shows a good result which is close to the optimal distance of 11 cm in the simulation.

Among 5 electrode settings, the 10-electrode sensor is slightly better than others in FES and measurement consistency against conductivity, which partially confirms the simulation findings regarding the number of electrodes.

To compare the fringe effect-based method with the global impedance-based method, the measured values by the two methods were normalized using the real volume and the measurement error is given in Fig. 11. The measurement error is defined as follows:

$$Error = g_{(V,\sigma)} \cdot \frac{V}{\bar{g}_V} - V \quad (5)$$

where $g_{(V,\sigma)}$ represents the measured LCG or global impedance under volume V and conductivity σ . \bar{g}_V represents the average of the measured LCG or global impedance values under different conductivities at a certain volume V .

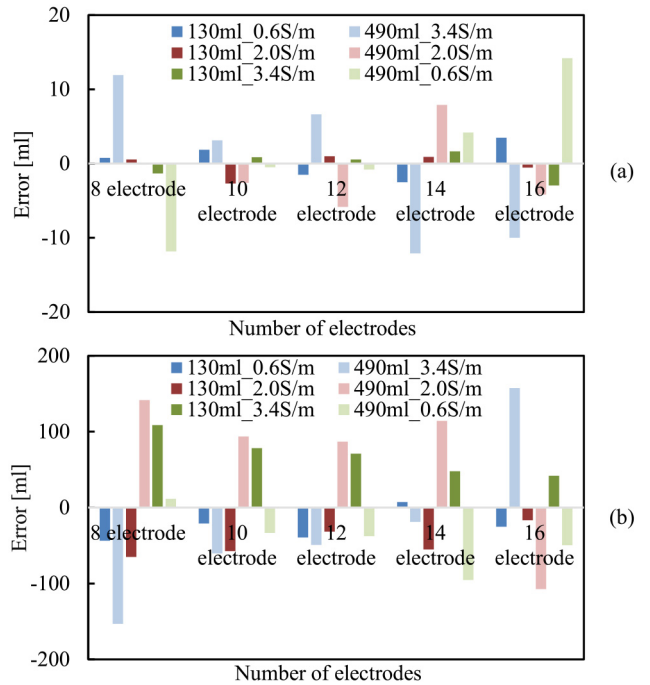


FIGURE 11. Measurement errors with two methods. (a) Fringe effect-based method. (b) Global impedance-based method.

Fig. 11 shows that the measurement error of fringe effect-based method is within 20 ml, which is significantly smaller than the global impedance-based method, having a maximum error more than 100 ml [7].

To further verify the simulation results, experiments were performed on the agar balls. In addition to the agar balls in Table 1, a 310ml agar ball with a conductivity of 2S/m was made to form three sets of experimental sequences with different conductivity trends, similar to those in simulation. Specifically, the change trends of urine conductivity with increasing bladder volume in the simulation, i.e., increase, remain and decrease, were simulated using the seven agar balls with three of them used in each trend. EIT measurements were carried out under each change trend and the data fitting results of volume measurements using both the global impedance-based and fringe effect-based methods are shown in Fig. 12. The experimental results show that the fringe effect-based method has better measurement consistency and linearity, which validates the simulation results. Compared with the simulation, the measurement error of the method based on edge effect is larger when the agar volume is small, which may be caused by the lower SNR of the equipment when the effective signal is small.

V. CONCLUSION AND FUTURE WORK

Fringe effect-based method for bladder volume measurement was first proposed and investigated in this paper. This method has distinct advantages over the global impedance-based method in measurement consistency and accuracy under different urine conductivities. In the study, both simulation and

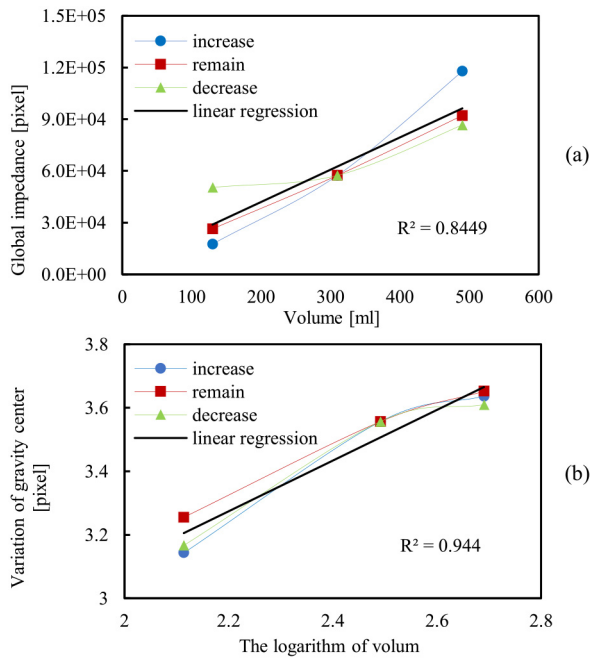


FIGURE 12. Comparisons under three change trends of agar conductivity. (a) Global impedance-based method. (b) Fringe effect-based method.

in-vitro experiment were carried out to validate the feasibility of the proposed method and optimize the design of the employed EIT sensor. Simulation and experimental results can lead to the following conclusions:

- 1) The dense electrode arrangement in the ventral side is beneficial for improving FES. However, too dense an arrangement would cause the sensing region of EIT sensor fail to sufficiently cover the bladder region, and thereby causing deformations of the reconstructed image. It is recommended that the cover ratio of electrode array in the ventral side to the abdomen periphery should be between 40% and 50%.
- 2) The further the electrode plane is from the bladder bottom, the larger the FES. But the sensitivity of the sensor to conductivity change in the bladder region decreases rapidly with increasing distance. Simulation and experimental results showed that a distance around 10 cm was satisfactory.
- 3) The influence of the number of electrodes on the measurement is relatively small compared to other two sensor parameters. Experimental results show that a 10-electrode sensor is acceptable regarding measurement sensitivity and consistency against conductivity.

To sum up, the fringe effect-based method provides a promising alternative for the monitoring of bladder volume which has a great potential for clinical use. In future, in-vitro experiments using a more realistic model of human abdomen will be conducted for further investigation of the proposed method. A miniaturized wearable measurement system will be developed and clinically tested in volunteers.

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