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Bladder volume estimation from electrical impedance tomography

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Abstract— **Ubiquitous knowledge of bladder volume is of great interest to patients whose perception of bladder volume is impaired. A promising approach to provide frequent bladder volume estimates to the patient are automatic and non-invasive measurements by electrical impedance tomography (EIT). Previous studies have shown a linear correlation of abdominal electrical impedance and bladder volume. In this article, we present two methods to extract a volume estimate from EIT measurements. One method is based on the global impedance from a reconstructed image, the second method is based on a singular value decomposition of the raw voltage measurement vector. A performance evaluation in presence of noise is performed.**

I. INTRODUCTION

A. Motivation

Patients suffering from paraplegia or patients with age-related diseases like diabetic neuropathy often show an impaired perception of bladder filling level. Therefore, they miss to go to the toilet in time, resulting in incontinence or even damages to the urogenital system. For patients suffering from paraplegia, even frequent intermittent self-catheterizations due to a fixed time scheme are necessary. To advance from a fixed emptying time scheme to a more demand-driven approach, regular bladder volume measurements are necessary. But also the light cases could benefit from a frequent bladder volume measurement, since this could help these patients to train the capacity of their bladder.

B. Approaches to bladder volume estimation

The most accurate bladder volume measurement available today is probably a CT scan, but this is only done in very rare cases due to the high radiation exposure and not suited for frequent, automatic measurements. Today, ultrasound examinations are most widely used for bladder volume calculations in hospitals, but a skilled and experienced examiner is necessary for high accuracy [1], [2]. Although there are ultrasound devices for patient home-use available – like BladderScan for example –, it is common to all available ultrasound devices that a manual placement of the probe is necessary. Frequent automatic measurements are not possible at the moment.

A promising alternative enabling unobtrusive measurements is electrical impedance. Talibi et al. [3] attached electrodes directly to the bladder wall, showing a linear correlation of electrical impedance and bladder volume. Denniston and Baker [4] showed that this linear correlation is retained also for non-invasive measurements in dogs, even in a very unspecific setup by band electrodes around the thorax and the thighs. They also placed spot electrodes ventrally to the abdomen in the bladder region, which showed an increased sensitivity. Following these experiments, Doyle [5] reproduced the measurements in humans, but mentions a strong influence of body posture and impedance changes not related to bladder volume changes. Hua et al. [6] used multiple electrodes in a matrix-like alignment to get increased sensitivity to bladder-related impedance changes. This approach has been continued by Leonhardt et al. [7] who applied electrical impedance tomography with sixteen electrodes around the pelvis. The new aspect is to first compute an impedance distribution image and to estimate the bladder volume based on the impedance image in a second step.

In this paper, we would like to focus on the second step, the estimation of bladder volume from EIT measurements. To be able to specify the volume error, simulated EIT measurements from a finite element model have been used. The rest of the article is structured as follows: First, we would like to introduce in brief the basic principle of EIT measurement and image reconstruction (sec. II-A) and present the finite element model which has been used for simulation (sec. II-B). It follows a description of the two methods for volume estimation (sec. II-C and sec. II-D) and a comparative evaluation of both methods in sec. III.

II. MATERIALS AND METHODS

A. Electrical impedance tomography for bladder volume monitoring

The principle of electrical impedance tomography has been first described by Webster and Henderson in 1978 [8], but the first medical application was by Barber and Brown a few years later [9]. The fundamental principle of EIT systems is the injection of a small current (hundreds of microampere to milliamperes) to a pair of electrodes and the recording of the resulting differential voltages between all other electrode pairs spanning a cross section of the body. In case of 16 electrodes, this results in 13 voltage measurements per injection and 208 measurements in total. Although the injection and measurement in adjacent electrodes may not be state-of-the-art [10], it has been used in this work to be compatible to the Goe MF II EIT-device as a reference.

For a known conductivity distribution $\sigma(\vec{x})$ at position $\vec{x}$ in the body Ω, the resulting electrode voltages $\vec{U}$ on the body surface $\partial\Omega$ are defined explicitly. This is called the forward problem of EIT, which has been solved using a COMSOL Multiphysics model in our case:

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$$\vec{U} = f(\sigma(\vec{x})) \quad (1)$$

For medical applications, the conductivity distribution $\sigma(\vec{x})$ is usually unknown and only the surface electrode voltages $\vec{U}$ are known. To compute $\sigma(\vec{x})$, $f(\sigma(\vec{x}))$ has to be inverted, which is an ill-posed problem widely addressed in literature. Usually, the impedance changes in biomedical problems are rather small. To limit the influence of measurement channel mismatch, the measured voltages are often scaled to a reference measurement $\vec{U}_0$:

$$\vec{U}^* = \frac{\vec{U} - \vec{U}_0}{\vec{U}_0} \quad (2)$$

In our case, the reference $\vec{U}_0$ is a recording with empty bladder. Since we are then only reconstructing small changes in the region of the reference impedance, it is possible to find a linearised reconstruction matrix R, such that:

$$\Delta\sigma(\vec{x}) = R \cdot \vec{U}^* \quad (3)$$

Several methods to find the reconstruction matrix R are available. In this work, the GREIT algorithm [11] has been used, which is available in the EIDORS (Electrical Impedance Tomography and Diffuse Optical Tomography Reconstruction Software) Matlab Toolbox [12].

B. Simulation model

To be able to specify the volume estimation error, a finite element simulation (FEM) model with known impedance distribution has been used instead of human measurement data. The FEM simulation has been done in COMSOL Multiphysics using the AC/DC module. The simulation model has been built from a pelvic bone segmented from a CT dataset, an elliptical cylinder is used to approximate the surrounding body and a parametrizable ellipsoid as the bladder. For the simulation, tissue parameters supported as a COMSOL parameter file from Pettersen and Høgetveit [13] have been used. The cylinder volume has been assigned to the conductivity of intestine and the bladder conductivity has been set to 30 mS/cm, which is based on exemplary urine conductivity measurements. As depicted in Fig. 1, sixteen copper electrodes for current injection and voltage measurements have been placed around the body on the level of the bladder.

Automatic rotation of the current injection electrodes in the COMSOL model has been realized from Matlab via the COMSOL LiveLink for Matlab interface, which also has been used to load the measurement voltages to a Matlab vector for further processing by EIDORS.

C. Volume estimation by global impedance

For general EIT images, every resulting pixel represents the change of impedance relative to a reference impedance. An exemplary image, recorded during an urodynamic examination at maximum bladder filling level referenced to an empty bladder is shown in Fig. 2. Since urine has higher conductivity than the surrounding tissue, a filled bladder gets visualized as a negative impedance change (white blob in the upper middle of Fig. 2). A summation of all pixels in the image P yields the global impedance value. In literature it is also sometimes referred to as resistivity index [14]:

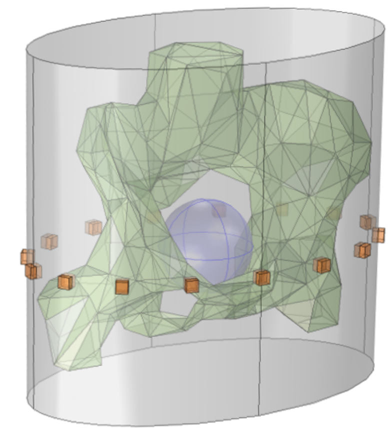

Fig. 1. COMSOL model for bladder impedance EIT

$$\Delta Z_{glob} = \sum_{i,j} P_{i,j} \quad (4)$$

The global impedance is in good correlation with the bladder volume, as shown in Fig. 3. For real measurement data as presented in Fig. 2, evaluation of the global impedance in a region of interest can be necessary to limit influences of the surrounding tissue.

A calibration of the global impedance method is possible by recording reference impedances at several bladder filling levels. In the example shown, six calibration points have been chosen. With this calibration data, a volume estimation error of 6.4% can be reached. During simulation, it is easy to acquire calibration datasets. In a set up involving a patient, simultaneous impedance and urine flow measurements or impedance measurements during regular urodynamic examinations can be used to gather calibration data.

D. Volume estimation by singular value decomposition

As mentioned in sec. II-A, the inverse EIT problem is ill-conditioned. In the reconstruction process, only 104 linear independent boundary voltage measurements are used to reconstruct 1024 pixels of a $32 \cdot 32$ pixel image. Applying the global impedance method, the values of 1024 pixels are reduced to one volume estimate again.

The motivation for the singular value method was to find a direct mapping of 104 independent boundary voltage measurements to one volume estimate. Avoiding the regularization process necessary for the ill-posed image reconstruction, the method is more stable to noise. Further, the regularization process has low pass character. By eliminating the need for regularization in the direct mapping, full information from the voltage measurement vector is kept.

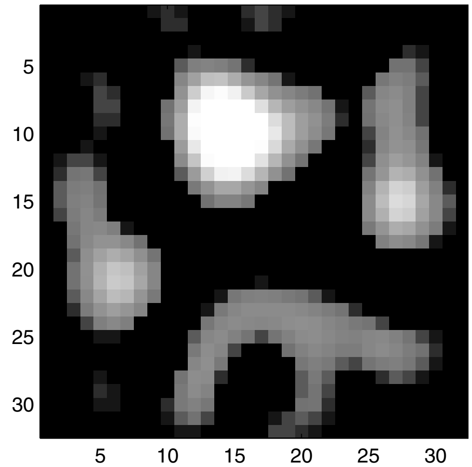

Fig. 2. Reconstructed EIT image of full bladder referenced to empty bladder

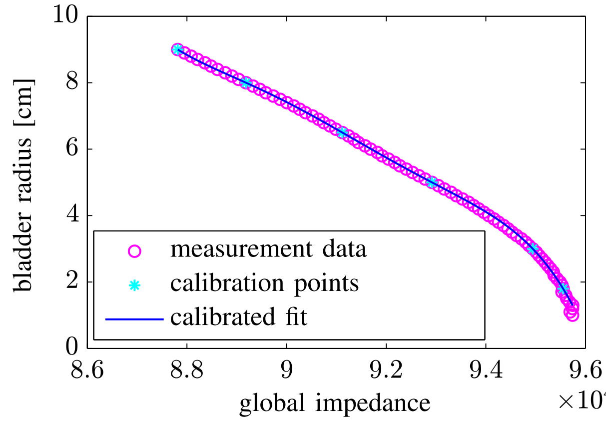

Fig. 3. Volume estimates from global impedance

Different methods of correlating changes in the raw voltage measurement vector to the volume change have been considered, for example maximum norm, minimum norm or a summation of all voltage measurements as presented by You et al. [15]. Due to space limitations only the singular value method is shown here, as it presents superior noise stability compared to all other algorithms taken into consideration.

For the singular value volume estimation method, the 208 boundary voltage measurements are ordered as a 13×16 matrix M. Each column holds the voltage measurements for one current injection position.

By use of the singular value decomposition, this matrix M can be factorized into three matrices

$$M = U \Sigma V^T \tag{5}$$

with the $m \times n$ matrix Σ of form

$$\Sigma = \left(\begin{array}{ccc|ccc} \sigma_1 & & & & \vdots & \\ & \ddots & & \cdots & 0 & \cdots \\ & & \sigma_r & & \vdots & \end{array} \right) \tag{6}$$

The positive diagonal elements $\sigma_1 \geq \ldots \geq \sigma_r$ are the singular values, describing the properties of a matrix like eigenvalues for square matrices. For bladder volume estimation, only the difference of the first two singular values is being used:

$$V \propto \sigma_1 - \sigma_2 \tag{7}$$

Like for the global impedance method, also for the singular value method few calibration points are necessary to reach a good performance.

III. RESULTS

In Fig. 3, the radius of the blue bladder sphere in Fig. 1 is shown for several values of the global impedance. This is in good agreement with a linear correlation of volume and impedance, as shown by several authors before (e. g. [7], [3]). During computations, it has been noticed that not all reconstruction algorithms are suited equally well for volume estimations by the global impedance method: while the Gauß Newton reconstruction produced the best bladder radius estimate, the GREIT algorithm produced much better global impedance results. Main reason are strong ringing artefacts in the GaußNewton reconstruction for great bladder sizes. Finally, the GREIT algorithm has been used for all computations, since it provided better correlations of global impedance and bladder volume. The results of the singular value method are very comparable to the global impedance method for the noise free case and provides good results.

To analyse the performance of both methods in the presence of noise, all computations have been repeated with a signal to noise ratio of 20, 10, 5 and 2.5. To add noise to the measurement voltages, the noise figure option by EIDORS has been used, which defines SNR as

$$SNR = \frac{||signal||_2}{||noise||_2} \tag{8}$$

where

$$|| \cdot ||_2 = \sqrt{\sum_{i=1}^{N} |x_i|^2} \tag{9}$$

In Table I, the relative error of the bladder size estimation is shown. As clearly visible from Fig. 5, the presented singular value method is very stable to applied noise. The measurement error is relatively constant in the range of 3%, while the error for the global impedance method increases rapidly due to amplifications in the process chain of reconstruction and integration of image pixels. Fig. 4 visualizes that for SNR of 5 or less, bladder size estimation is infeasible.

Since applied noise appears mainly in the smaller singular values, its influence on the volume estimation by using the two first singular values is reduced. On the other hand, applied noise has a great influence on image reconstruction and appears as artefacts in the reconstructed image. When

SNR	global impedance	singular value
∞	6,5 %	2,7 %
20	20,1 %	2,8 %
10	210,1 %	2,9 %
5	–	3,1 %
2,5	–	4,3 %

TABLE I
RELATIVE ERROR OF BLADDER SIZE ESTIMATION

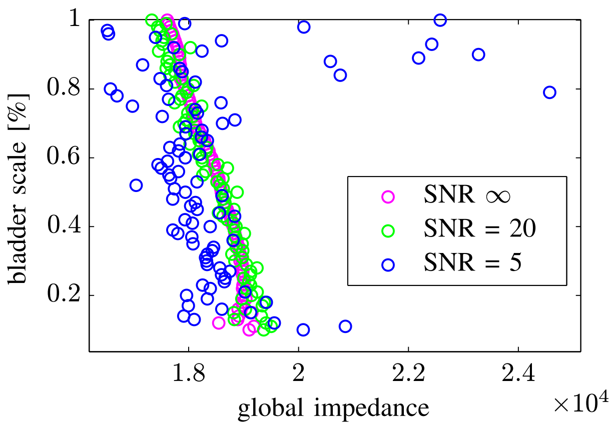

Fig. 4. Influence of noise on global impedance method

calculating the global impedance, noise and resultant artefacts are summed up.

IV. CONCLUSION

Based on a finite element simulation using COMSOL Multiphysics and Matlab, two different methods to estimate bladder volume from electrical impedance measurements have been shown. The first method is based on the integration of relative impedance changes for all pixels in a reconstructed EIT image (global impedance), while the second method uses the singular value decomposition to extract a feature correlating with bladder volume directly from the raw voltage measurement vector, avoiding the need to reconstruct an EIT image first. It could be shown that both methods perform well for the noise free case. In the presence of noise, the singular value method outperforms the global impedance method clearly. Even for a signal to noise ratio of 2.5, the singular value method still provides good performance.

V. ACKNOWLEDGEMENT

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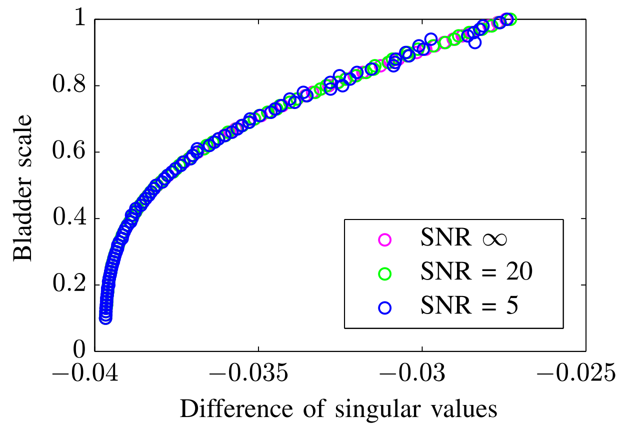

Fig. 5. Influence of noise on singular value method

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