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Electric impedance tomography for monitoring volume and size of the urinary bladder

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Abstract

A novel non-invasive technique for monitoring fluid content in the human bladder is described. Specifically, a precommercial electric impedance tomograph (EIT) was applied to measure and visualize impedance changes in the lower torso due to changes in bladder volume. Preliminary measurements were conducted during routine urodynamic tests of nine male paraplegic patients, in whom a contrast agent was slowly infused into the bladder for diagnostic purposes. In some patients, a good correlation between bladder volume and EIT measurements was found, whereas in others the correlation was still good but inverted, presumably due to a poor electrode positioning. These preliminary results indicate that a sufficiently accurate finite element modeling of the impedance distribution in the abdomen, and proper electrode positioning aids, are important prerequisites to enable this technology to be used for routine measurement of bladder volume.

Keywords: bladder volume estimation; electrical impedance tomography; urodynamic.

Introduction

Urinary incontinence is a frequent and often distressing condition, which may have an adverse impact on a patient's well-being and quality of life. The main underlying physiological cause is an imbalance between proximal urethral and bladder pressure (Figure 1), which may be caused by various pathophysiological conditions, including muscular weakness, nervous disorders, an enlarged prostate gland, and/or aging.

One of the diagnostic procedures for patients with disturbed urinary continence is the so-called urodynamic testing. This procedure is well established and allows physicians to evaluate how the bladder and urethra perform their job of storing and releasing urine [27]. During the test, the bladder is artificially filled with an external fluid (a specific X-ray contrast agent), and pressures inside the bladder and the rectum are measured simultaneously. X-ray images are made at certain intervals to evaluate bladder size and form. Disorders (e.g., urinary incontinence, frequent urination, urge incontinence, problems with emptying the bladder, and other symptoms) can be quantified and the underlying causes, such as dysfunction or overactive function of the detrusor muscle, can be revealed and diagnosed.

In paraplegics, the urinary condition is particularly complicated because they generally lack control over urinary function and cannot voluntarily discharge the bladder. Depending on the severity of the spinal cord injury, micturition reflexes are often blocked and patients with impaired or destroyed nervous pathways may exhibit bladder-sphincter dyssynergia, which causes functional outlet obstruction [5, 25]. Typically, these patients have no natural desire to urinate. To discharge their bladder, paraplegic patients have to manually activate spinal micturition reflexes by, e.g., suprapubic compression, or the use of intermittent catheterization, which (on the long-term) can cause infection or kidney damage. In addition, unnoticed elevated bladder pressure can lead to autonomic dysreflexia, a potentially life-threatening condition characterized by severe paroxysmal hypertension, and to several other disorders, such as severe headache and profuse sweating [18]. Particularly for this paraplegic population, proper monitoring of the bladder volume is very important. One option is to use one of several instruments that are commercially available to measure bladder volume based on ultrasound technology (see e.g., [1, 4, 8, 13, 14, 23]). However, their use requires partial disrobement and application of a gel (which generally requires privacy), and also obliges the patient to estimate the appropriate time for a control measurement. Thus, a method that continuously and unobtrusively monitors fluid volume in the bladder would be a valuable asset for these patients, especially when coupled with a threshold-based alarm function. In principle, the measurement of tissue bioimpedance seems to be a promising technique for this task, as it specifically relates to tissue fluid content [10, 15, 21] and has been proposed for the determination of bladder volume [6, 22, 24] or pressure [11]. However, until now, such measurements have been catheter based and invasive.

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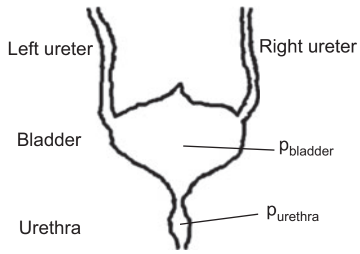

Figure 1 Sketch of the anatomy of the human bladder. Incontinence results from a disbalance of $\Delta p = p_{bladder} - p_{urethra}$. For more details on the human bladder, see, e.g., *Prometheus, Hals und Innere Organe*, Georg Thieme Verlag.

The present study extends the idea of applying classic extracorporeal bioimpedance measurements (e.g., bioimpedance spectroscopy) by proposing the use of external electric impedance tomography (EIT), as this provides spatial information that may increase measurement accuracy. EIT is a non-invasive clinical monitoring technique that, although has been investigated for many years [3], has not yet become an established clinical tool. It is envisaged that ventilation imaging may be the first clinical application of EIT. As explained below, EIT measures the spatial distribution of resistivity in a specific body volume and maps that information on a 2D image. EIT offers several advantages over other established methods of imaging and monitoring the body, i.e.,

- it is relatively inexpensive
- there are no known hazards attached to its use
- it has a high temporal resolution allowing dynamic changes in function to be measured.

However, despite the high temporal resolution, several drawbacks still exist. These include the relatively low number of independent transfer impedance measurements resulting in a low spatial resolution compared with computed tomography or magnetic resonance tomography, and the numerical difficulty of solving the non-linear and badly posed inverse problem. Governed by quasi-static Maxwell equations, the resulting potential differences on the surface of the measured object are a function not only of the applied current pattern but also of the internal and boundary conductivity distribution and geometrical shape of the body [16].

There are some indications that absolute EIT imaging may become a reality [2, 10]. However, we believe that (at this stage) it is more reasonable to follow the established approach of functional EIT (fEIT) imaging [7, 9, 19] because this is numerically much simpler and its clinical potential for, e.g., lung monitoring, has been demonstrated in several studies. Note that fEIT imaging is a relative technique based on deviations from a baseline value, thereby avoiding the problem of knowing the exact geometric shape of a measured object. However, this introduces the disadvantage that only changes of the conductivity distribution can be measured. There are, however, clinical applications where fast and distinct changes of the tissue conductivities in the examined area occur, such as in acute lung injury or the acute respiratory distress syndrome during mechanical ventilation [17, 20, 26]. The present study examines whether this assumption is also valid for the filling and emptying of the human bladder.

Methods

Electric impedance tomography

EIT uses the general principle of bioimpedance measurements, in which a small alternating electric current is injected into a biological tissue through two electrodes and the corresponding voltage drop is measured across another pair of electrodes. In principle, this four-electrode arrangement has the advantage of theoretically eliminating the effect of the unknown skin electrode resistances. In EIT, this principle is expanded by continuously rotating the location of current injection and voltage measurements around the body on a plane, thus spanning a cross section. Typically, 16 electrodes are equidistantly attached to the skin around the subject. It is easy to show that ($N\cdot(N$-3)/2) (here, 104) independent transfer impedance measurements can be obtained by sequentially supplying all adjacent electrodes with a small alternating current (e.g., 5 mA_{rms} at 50 kHz) using a digital-to-analog converter while measuring the transfer voltages at the (N-3) remaining electrode pairs with an analog-to-digital converter. The switching of current injection electrodes and corresponding voltage measurements is generally controlled by a digital signal processor. Figure 2 shows, for example, how to acquire an EIT image from the lower human torso. From the 104 voltage measurements, for each frame a row vector g_n with 104 rows is created, where n is the frame number. The corresponding EIT image vector z_n may be obtained by either using a classic back-projection algorithm or a modified Newton-Raphson algorithm, see, e.g., Reference [15]. In either case, a B matrix (dim 912×104) is constructed, mapping the transfer voltages g_n to image space (32×32=1024 pixels, out of which 912 are used).

$$Z_n = B\cdot g_n$$

The frame rate may vary depending on the device. In our case, an EIT evaluation kit (EEK2; Draeger Medical AG & Co. KG, Luebeck, Germany) was used and the frame rate was set to 40 per second. As mentioned above, it is very difficult to obtain accurate absolute images because this requires highly detailed knowledge on geometry and absolute electrode positioning. Therefore, to obtain better EIT imaging results, the so-called functional imaging only uses the changes in tissue impedance [9]. Here, for any frame, the appropriate relative row vector dg_n is gained from a comparison with a reference vector g_{ref} for each measurement:

$$dg_n(i) = \frac{g_n(i) - g_{ref}(i)}{g_{ref}(i)}$$

where the reference vector g_{ref} is obtained by averaging over a certain number of EIT image frames N (typically 20 s→N=800) and i indicates the number of the voltage measurement. Then, the associated relative EIT image vector dz_n is given by:

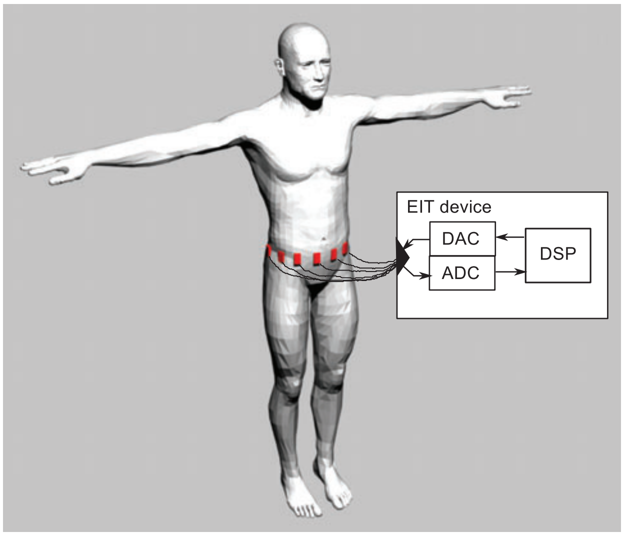

Figure 2 Image showing the principle of EIT measurements of the lower human torso.

$$dz_n = B \cdot dg_n$$

An fEIT image vector f_n can simply be calculated as the standard deviation for each pixel p of a series of M consecutive EIT images:

$$f_{n,M}(p) = \sqrt{\frac{\sum_{k=n-M+1}^{n} [dz_k(p) - \overline{dz_{n,M}}(p)]^2}{M-1}}$$

with

$$\overline{dz_{n,M}}(p) = \frac{\sum_{k=n-M+1}^{n} dz_k(p)}{M}$$

The resulting fEIT image provides a steady-state description of any physiological process causing impedance changes in the corresponding body cross section, be it periodical or aperiodical. For illustration purposes, Figure 3 shows an fEIT image in which large impedance changes are in red, resulting from lower torso measurements (right-hand image). As shown by our group, the change of impedance distribution occurring over a certain period can be well visualized by subtraction of the corresponding functional images. This method has been named difference fEIT (dfEIT) and was able to quantify the effects of different ventilator settings in acute lung injury [17, 20].

In the lower torso, the bladder is surrounded by various tissues, including rectum/prostatic gland tissue or uterus/vagina tissue, and pubic bone, muscles, etc. These tissues have various resistivities and it is difficult to predict their size and individual thickness. However, if only differences of functional image (dfEIT) are examined [17, 20], the hypothesis is that all static resistances cancel out and only the local resistance change due to bladder volume changes are amplified (contrast amplification). To correlate the EIT images with the contrast fluid volume (Peritrast infusion, 31% 13.35 mS/cm ≈ 74.9 Ω m; Dr. Franz Koehler Chemie GmbH, Bensheim, Germany) injected into the bladder, the functional image was mapped to a numerical value [the so-called functional global impedance (fGI)] by summing the functional activity of all image pixels.

$$fGI = \sum_{p=1}^{912} f_{n.M}(p)$$

Clinical measurement scenario

An experimental EIT device (manufactured by Draeger Medical) was used for EIT measurements (Figure 4, left). Since this is the only available system with a clinical license, we could not use alternative setups, e.g., a multifrequency EIT mode. A silicone belt with 16 conductive silicone electrodes was attached to the lower torso of the patient (Figure 4, right, and X-ray image in Figure 5).

Proper skin contact was tested by measuring the electrode resistances, and EIT measurements were only made if the electrode resistance was found to be within the appropriate range of 100–300 Ω. An urodynamic test with ordinary contrast fluid [Peritrast infusion 31% (Dr. Franz Koehler Chemie GmbH); conductivity 13.35 mS/cm at 20.0°C] was performed. The bladder was slowly filled with contrast fluid and the corresponding pressures were recorded as usual. For the present study, however, there were two deviations from the normal procedure: (i) filling of the bladder was interrupted several times to acquire an EIT image and note the corresponding bladder volume, and (ii) directly after the evacuation of the bladder before the urodynamic test, the EIT baseline image was recorded. Nine male paraplegic patients

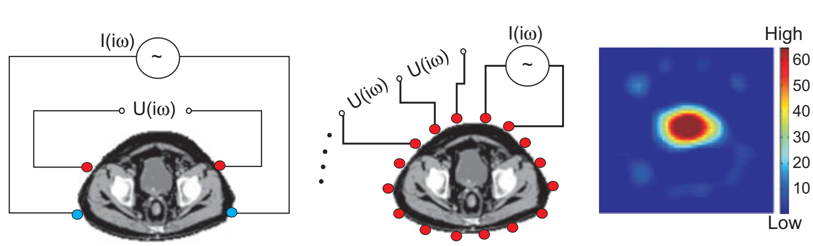

Figure 3 Classic bioimpedance electrode arrangement with four electrodes for the lower torso (left-hand image) and extension of this four-electrode principle to EIT of the lower torso (middle image). The right-hand image shows the resulting impedance distribution as a false-color functional EIT image, with red indicating high impedance changes and blue the lower impedance changes.

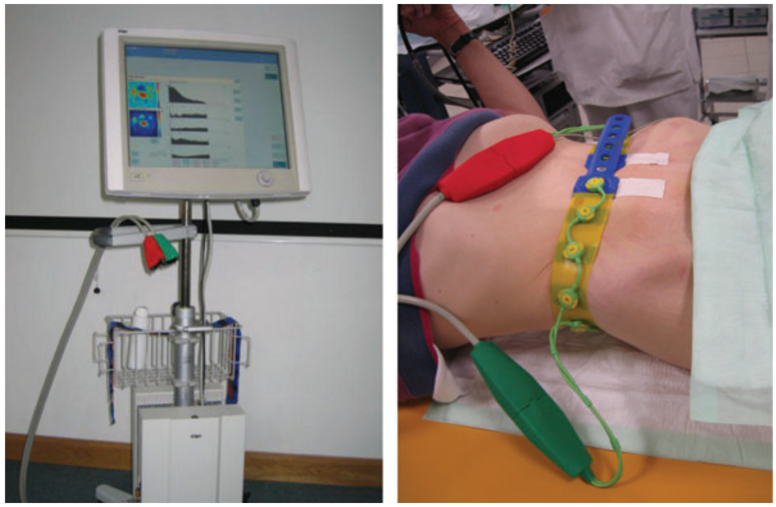

Figure 4 Left: Experimental EIT device (EEK2, Draeger Medical); Right: EIT electrode belt attached to the waist of a patient. Note that in some very thin patients (such as this patient), the belt has to be positioned relatively cranial from the bladder.

were enrolled in this preliminary study. All measurements of these patients took place at the Orthopaedic Department of Heidelberg University Clinic during routine urodynamic testing procedures. Approval from the Local Ethics Committee was obtained (approval no. S-458/2007). Informed written consent was obtained from all patients.

Results

When infusing a fluid with good conductivity into the bladder, one expects an image indicating a ventral signal in the dfEIT image and the global impedance over the infused volume, which should decline with an increasing volume of fluid. This was found to be the case in several of the patients. For example, Figure 6 (left graph) shows the corresponding results, with the red line indicating the result of a least-squares regression, while the right-hand image gives the corresponding dfEIT image, produced by subtracting an fEIT image produced at the beginning of the urodynamic test (i.e., bladder empty) from an image produced at the end of the test (bladder filled to the maximum). Note that in the chosen false-color coding, blue indicates no change in impedance, whereas yellow (and red to an even greater extent) indicates a change in impedance.

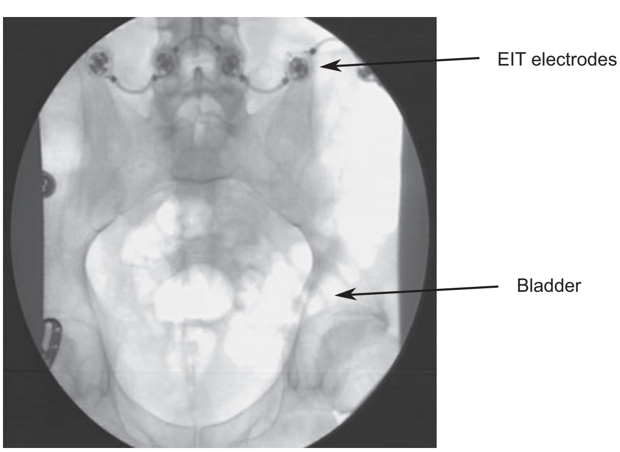

Figure 5 X-ray image of the human bladder with attached EIT electrode belt.

For another patient, the corresponding findings are presented in Figure 7. Table 1 presents the results obtained from the nine patients. The number of data points indicates the fluid volumes at the moment the infusion was briefly interrupted and EIT images were made. However, these plots demonstrate the correlation between the infused volume and the GI of the EIT images, and as explained above, an absolute estimation of the bladder volume without a calibration method is not possible.

Discussion

In this preliminary study, the expected behavior – a decline in overall resistance (negative correlation) with increasing bladder volume during urodynamic testing – was demonstrated in four of the nine patients. Interestingly, in five patients, the global resistance of the lower torso increased with increasing bladder volume during infusion of the contrast agent. However, in all cases, the impedance changed linearly with the infused volume, which is represented by the high values of correlation. This linear relationship indicates that the different arithmetic signs of the impedance-volume relationship are not attributed to a measurement problem. Even though it is not entirely clear what mechanisms are involved in these unexpected findings, some factors have been identified with a potential influence on the measurement results.

In our study, we did not take into account any urine production by the patients. However, since the overall measurement time was kept below 30 min, we think that the influence of the produced urine can be neglected. Additionally, a standardized

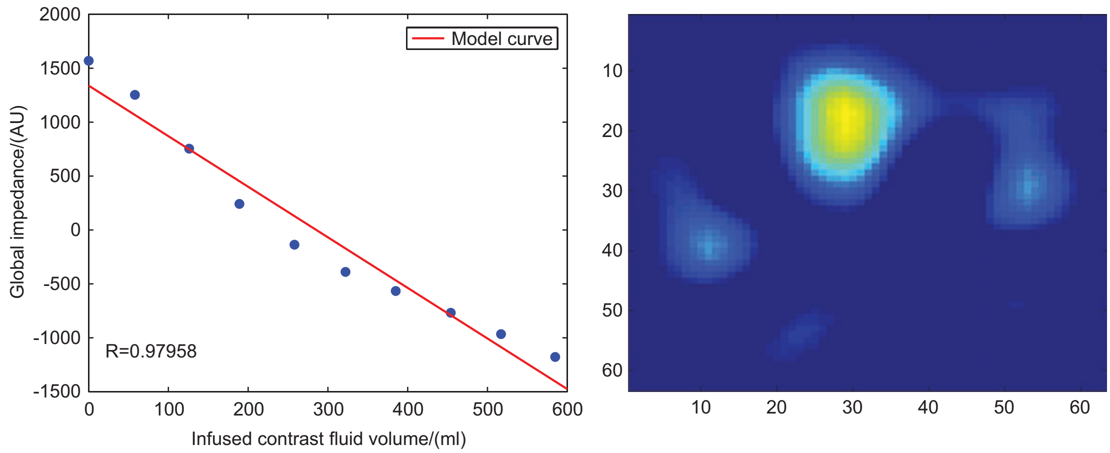

Figure 6 Left: GI of the lower torso as a function of infused volume for patient no. 5. The overall GI decreases as the infused fluid volume increases; the red line (model curve) indicates the result of the least-squares fit. Right: Corresponding dfEIT image (color-coded in absolute units), with the yellow spot indicating an area of high impedance change (presumably the location of the bladder).

medical-grade infusion medium has been used, for which a fixed conductivity value with a very low tolerance is guaranteed by the manufacturer.

The positive sign of the correlation in some patients may be explained by the fact that during infusion with contrast fluid, the bladder may have moved away from the measurement region, thereby imitating an increasing overall resistance in the region of interest. Another factor is that there was a wide variation in the body mass index of the study patients. For the thinner patients, it was relatively difficult to achieve good electrical contact; in these patients, the belt with electrodes was probably placed somewhat higher above the iliac bones and therefore not in the same position as used for more obese patients (compare Figure 5). Therefore, we believe that the EIT method itself worked, but that structural changes within the measurement region are responsible for different arithmetic signs of the impedance. We were not able to control the placement of the EIT electrodes in relation to bony landmarks, since X-ray images with the exposure radiation dose would not be justified in a feasibility experiment.

In summary, the present study provides some evidence that EIT can give an indication of bladder volume changes. The reason for the differences in the correlation coefficients needs to be resolved before it could be considered a potential application of the technique. However, by using an individual calibration, the differences in the course of the impedance between users may be compensated together with the non-linear effects caused by the changes of the impedance of the urothelium at high volumes [12].

Conclusion and future perspective

On the basis of the findings of this preliminary study, we conclude that it seems feasible to monitor the filling and emptying

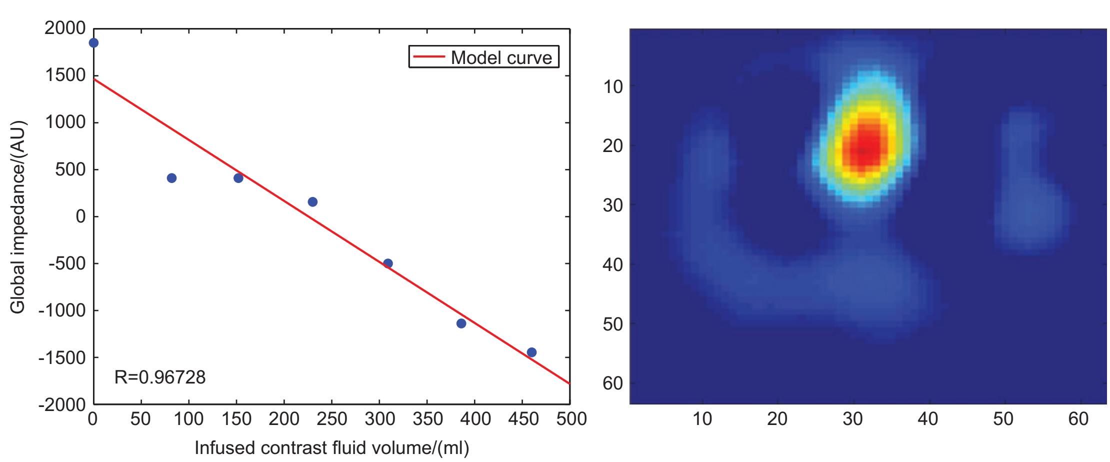

Figure 7 Left: GI of the lower torso as a function of infused volume for patient no. 6. The overall GI decreases as the infused fluid volume increases; the red line (model curve) indicates the result of the least-squares fit. Right: Corresponding dfEIT image (color-coded in absolute units), with the yellow spot indicating an area of high impedance change (presumably the location of the bladder).

Table 1 Data of nine paraplegic male patients.

Patient no.	Diagnosis	Weight (kg)	Age (years)	Infused contrast fluid volume (ml)	No. data points	Pearson's correlation coefficient R between global impedance and fluid volume
1	Compliant bladder	105	38	536	12	0.831341
2	Hyperactive bladder	95	53	282	5	0.79072
3	Hyperactive bladder	85	63	99	3	-0.95147
4	Hyperactive bladder	85	63	454	8	-0.87756
5	Compliant bladder	59	30	585	10	-0.97958
6	Compliant bladder	62	32	460	7	-0.96728
7	Hyperactive bladder	65	34	217	5	0.63795
8	Compliant bladder	75	58	550	11	0.82732
9	Compliant bladder	60	23	500	10	0.53187

of the human bladder using a two-dimensional arrangement of electrodes and an EIT-related reconstruction algorithm. We believe that use of a three-dimensional electrode arrangement, with additional electrodes placed on the buttocks, will provide even better results; this setup would allow different schemes for current injection, and reinforcement of the bladder, which is subject to stronger fluxes in current. Meanwhile, because these preliminary findings are promising but need further elucidation, more studies are needed, including a finite element modeling study to reveal the underlying reasons for the different signs of the correlation coefficients.

Our study did not take into account any urine production by the patients. In fact, since the urodynamic tests were conducted rather quickly, we felt that this assumption is justified. However, depending on the fluid status and fluid intake of a patient, this assumption may prove not to be valid in all patients. We think this influence should be further investigated. The future goal of this research project is to investigate whether this approach may lead to a new bladder volume monitoring technique. However, EIT only measures impedance distributions in cross-sectional areas. Since in principal the link between bladder volume and impedance is strongly influenced by ion concentration, the effect of the often strongly variable pH value of the urine should be further examined.

Acknowledgments

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