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A Small 8-Electrode Electrical Impedance Measurement Device for Urine Volume Estimation in the Bladder

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***Abstract*— Urinary incontinence is prevalent among elderly people. Recent studies have demonstrated the effectiveness of continence care based on urine volume measurement for elderly people who maintain their urinary storage function, but have difficulty feeling bladder fullness owing to dementia or neurological disorders. Electrical impedance measurement is a feasible technique that can be adopted in the diaper or underwear for continuous and unobtrusive urine volume measurements. We developed a small sensor device that can measure electrical impedance with a resolution of 0.017 Ω, which is sufficiently small to capture abdominal impedance alterations triggered by urine accumulation. The results obtained from a preliminary feasibility test in a young healthy volunteer suggested that the 8-electrode electrical impedance measurement with linear regression can estimate urine volume in the bladder in humans for the first time.**

***Clinical Relevance*— Continence care for elderly people is essential; however, it is a huge burden for nurses and caregivers, because it involves taking patients to the toilet or changing diapers. This study proposes a continuous and unobtrusive measurement device for urine volume in the bladder. Via continuous monitoring and bladder fullness alert, the device will enable nurses and caregivers to provide personalized continence care without hindering their routine care.**

I. INTRODUCTION

Urinary incontinence is prevalent among elderly people [1]. It increases the risk of adverse events such as falls [2] and is related to a decreased quality of life [3]. Therefore, the management of urinary incontinence is one of the most critical issues in the care for elderly people. Ultrasound-assisted prompted voiding (USAPV) is an effective method for the management of urinary continence in elderly people living in nursing homes [4]. In these populations, although some maintain urinary storage function, they have difficulty feeling bladder fullness owing to dementia or neurological disorders. In USAPV, the urine volume in the bladder, which is measured by an ultrasound device, is used as a substitute for the desire to void. The patient is prompted to void when the urine volume is more than 75% of the individual pre-fixed bladder capacity, or when the patient has a desire to void. Through this process, the patients regain the sensation of the desire to void the bladder at the appropriate time.

Studies on USAPV demonstrate the importance of continence care based on urine volume measurements in the bladder. However, portable ultrasound devices have disadvantages in terms of size (too large for a person to wear in daily life) and inaccuracy [5]; hence there is a need for a novel wearable device that continuously and unobtrusively measures urine volume.

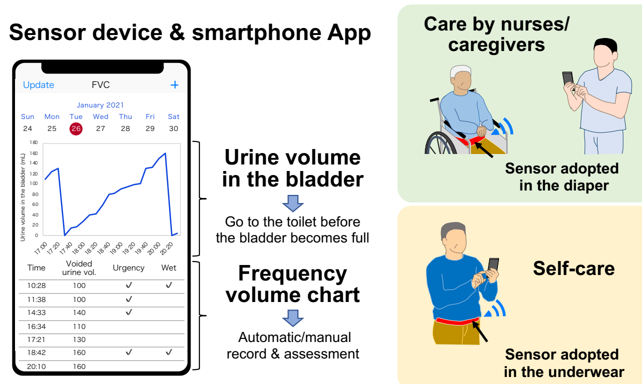

Figure 1. Concept of continence care utilizing the urine a new wearable urine volume sensor device with smartphone application

Fig. 1 illustrates the concept of continence care utilizing a novel wearable device with a smartphone app. The device will be adopted in the diaper or underwear to alert the nurses or caregivers for patients to urinate at the appropriate time via the application. The smartphone app displays the estimated urine volume in the bladder in real time, and a frequency volume chart, which is used to assess lower urinary tract symptoms, is implemented in the same application. The combination of the novel device and smartphone app will enable nurses and caregivers to provide comprehensive continence care.

Electrical impedance (EI) measurement is a feasible technique that can be adopted in the diaper or underwear for continuous and unobtrusive urine volume measurements. The electrodes used for EI measurement can be woven into underwears (textrode), thereby overcoming the limitations of the size faced by existing ultrasound devices. The miniaturization of EI measurement devices is crucial for implementing them as a clinically applicable urine volume estimation tool, and several studies have developed wearable sensors [6], [7]. However, a tradeoff exists between portability, wearability, and estimation accuracy. Because the number of electrodes is limited (i.e., three or four electrodes), such wearable sensors solely obtain one EI value and capture the decreasing trend and steep increase in abdominal EI with the

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accumulation of urine and during urination, respectively. In contrast, the equipment for electrical impedance tomography (EIT), which adopts many electrodes (i.e., 16 or 32 electrodes) can acquire a significant number of voltage measurements and estimate urine volumes by reconstructing cross-sectional images of the lower abdomen [8]. However, the size of such equipment is too large to be applied in the daily lives of elderly people [9], [10]. The large size is primarily attributed to the complexity of channel switching for current injection and voltage measurement with many electrodes using multiplexers. Active electrode (AE) [11] or dual mode driver, which enables a analog front-end to work as both current injection and voltage readout circuits [12], have been used to obviate the need for multiplexers. However, AE is not suitable for textrode and the number of measurement pairs is limited in the measurement using dual mode driver. Therefore, it is necessary to reduce the number of electrodes with using multiplexers, for implementation as a wearable sensor.

To examine whether EI measurement can accurately estimate urine volume with fewer electrodes, we performed a finite element model simulation and confirmed that an 8-electrode EI measurement combined with linear regression accurately estimated urine volume [13]. This study was aimed at developing a small-size 8-electrode EI measurement device. The device is equipped with a wireless communication function to enable its application as a wearable sensor. We also conducted a preliminary feasibility test in a healthy volunteer to examine the possibility of long-term measurements in daily life and urine volume estimation.

II. Sensor Device for 8-electrode EI measurement

A. Hardware

The developed sensor device comprises two circuit boards: main and measurement boards, as illustrated in Figs. 2(a) and (b). The HUZZAH32 board (Adafruit Industries, New York, NY), which included an ESP32 microcontroller (Espressif Systems, Co. Ltd., Shanghai, China), was adopted as the main board. On the measurement board, an AFE4300 analog front-end (Texas Instruments Inc., Dallas, TX) and ADXL362 3-axis accelerometer (Analog Devices Inc., Norwood, MA) were implemented for EI measurements and body movement detection, respectively. Although the 3-axis accelerometer was not used in this study, it was implemented for future applications. Generally, EI measurement in daily life is susceptible to variations in posture and movement; therefore, we will exclude noisy EI data based on changes in acceleration. The entire device was powered by a lithium-ion polymer battery (DTP502035 (PHR), DTP Battery, Dongguan, China) connected to the HUZZAH32 board.

AFE4300 is an integrated analog front-end that supports three channels of EI measurements using the tetrapolar I-V method. Mosquera *et al*. reported the application of AFE4300 to 8-electrode EI measurements [14]; and we adapted their method. The amplitude and frequency of the output AC current of AFE4300 were set to 833 $\mu A_{p\text{-}p}$ (295 μA_{rms}) and 50 kHz, respectively.

The EI value in the vicinity of the lower abdomen is generally less than 100 Ω, and the signals at the measurement electrodes are approximately 80 mV or less. In addition, the EI alterations caused by urine accumulation are of the order of a few ohms, which are vulnerable and result in a poor signal-to-noise ratio (SNR). In our previous simulation, we determined that a low noise level of < 0.025 mV_{rms} in the voltage measurement was required for accurate urine volume estimation [13]. However, the gain of AFE4300 is not large enough to realize this low noise level; therefore, analog signals were amplified before the analog-to-digital conversion. Output ports for signal rectification were connected to the input port of the analog-to-digital converter in the AFE4300 via an amplifier (Fig. 3). Consequently, the signal at the measurement electrodes was amplified by 40, including the internal amplifier in the AFE4300.

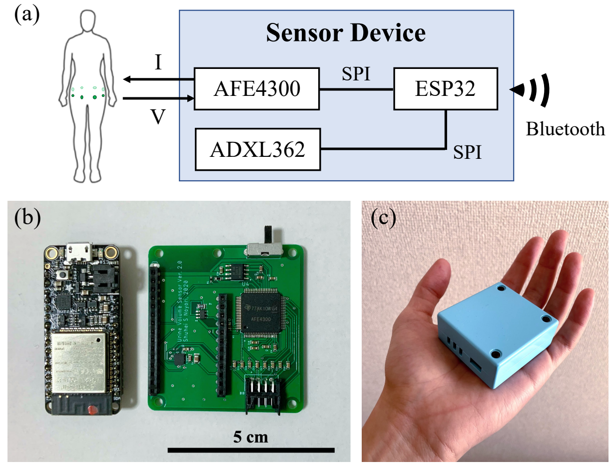

Figure 2. (a) System configuration, (b) circuit boards, and (c) assembled sensor device

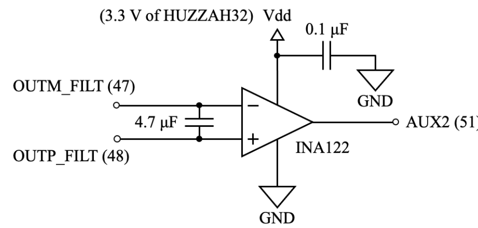

Figure 3. External amplifier connected to AFE4300
The numbers in parentheses are port numbers of AFE4300. The gain of INA122 was set at 20.

All circuit boards and batteries were placed inside a small 3D-printed plastic case with dimensions of 57 × 54 × 24 mm (Fig. 2(c)). The weight of the sensor device was 60 g without cables and electrodes.

B. Software

EI was measured using eight electrodes with a skip-of-four current injection and voltage measurement patterns; and 40 EI values were recorded. This process included channel switching for injection and measurement, and it was necessary to wait until the voltage between the electrodes stabilized. The waiting time was set at 50 ms, based on [14]. The 8-electrode EI measurement with a skip-of-four pattern comprised eight switches for the current injection channels and five switches for the measurement channels per current injection. Therefore, one measurement takes approximately 2.4 s (50 ms × (8 + 5 × 8) switches). EI was measured every 5 s and AFE4300 was turned off in the remaining 2.6 s to save battery life. During this period, the data were transferred to a smartphone via Bluetooth.

III. EI Measurement Experiment

The noise level, repeatability, and linearity of the measurements were examined for single-channel tetrapolar measurements. The EI of the resistors between 10 and 100 Ω (1% tolerance) was measured over 5 min, and procedure was repeated three times. The root mean square (RMS) noise and SNR were calculated after converting the raw output values into voltages. The SNR was defined by the following equation:

$$\mathrm{SNR} = 10\log_{10}\frac{\sum x^2}{\sum (x - \bar{x})^2}, \tag{1}$$

where x is a measured value.

Because the raw output values contained spike noise, a median filter with a window size of N was applied. Table 1 presents the RMS noise and SNR values for single-channel measurements with different window sizes of the median filter. The results filtered using a median filter with window sizes of 12, 18, and 24 (60, 90, and 120 s, respectively), achieved the required RMS noise level of 0.025 mV [13], and we decided to adopt the 90-s window.

After filtering, linear regression was performed on the output data from our sensor device and resistor values to derive a conversion equation from the raw output values to the EI values. Although we could directly estimate the urine volume from the output, to evaluate the measurement accuracy and precision bases on the required specification, we converted the values to voltage or resistance values. The conversion equation and R^2 derived from the linear regression were (EI) = 0.017 × (sensor output) − 82.136 and 0.998, respectively; therefore the resolution is theoretically 0.017 Ω. Fig. 4 presents the variances of the three measurements for the 10 resistor values. The variances were larger in the case of smaller resistor values, and approximately 1 Ω or more for resistor values of 10, 20, and 30 Ω.

Table I. RMS Noise and SNR in the Single-Channel Measurement

Data/window size of median filter	Root mean square noise (mV)	Signal-to-noise ratio (dB)
Raw data	0.318 (0.518)	52.5 (7.0)
15 s	0.181 (0.496)	61.9 (9.0)
30 s	0.070 (0.202)	70.4 (9.5)
60 s	0.013 (0.010)	76.4 (6.6)
90 s	0.010 (0.010)	79.1 (7.9)
120 s	0.008 (0.008)	80.6 (8.5)

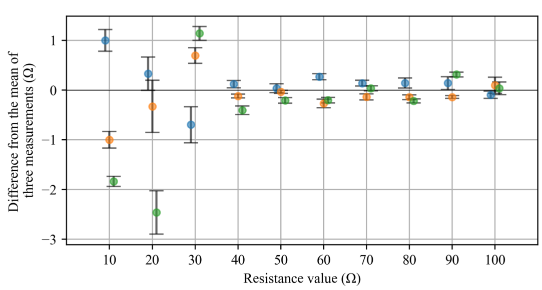

Figure 4. Means and standard deviations of differences from the overall mean values for three measurements
The variances of the three means for the measurements for 10 resistance values from left to right are 2.126, 2.130, 0.918, 0.069, 0.016, 0.087, 0.020, 0.037, 0.052, and 0.010 Ω (10 to 100 Ω in increments of 10 Ω).

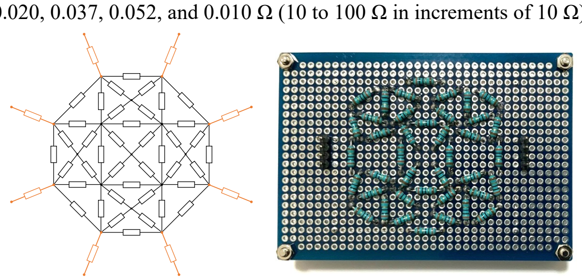

Figure 5. Schematic (left) and implemented circuit board (right) for a resistive mesh phantom for 8-electrode measurement
The black and orange resistors in the schematic are 30 Ω and 1 kΩ, respectively. The 1 kΩ resistors are mounted on the bottom of the circuit board.

The noise level of the multichannel measurements over a long period of time was also investigated using a resistive mesh phantom (Fig. 5). The phantom consisted of 48 resistors (Forty 30-Ω resistors and eight 1-kΩ resistors emulating contact impedance) and emulated measurements in the human body. Because the proposed sensor device was developed for continuous measurement during daily life with switching current injection and measurement channels, the absence of drift error and stability in multichannel long-term measurements was investigated. The sensor device was connected to the phantom, and multichannel measurements were conducted for 12 h. The difference between the maximum and minimum values within 12 h and the standard deviations (SD) of each measurement index over 12 h were adopted as indices for the existence of drift error and stability, respectively. The difference between the maximum and minimum values of one channel was as small as 0.100 ± 0.113 and 0.017–0.542 mV (mean ± SD, min–max), which is theoretically equal to the difference of 0.340 ± 0.384 and 0.057–1.838 Ω. Regarding stability, the SD over 12 h was as small as 0.010 ± 0.007 and 0.002–0.038 mV.

IV. Preliminary Feasibility Test in a Healthy Human

We conducted a preliminary feasibility test to examine the possibility of long-term measurements in daily life and urine volume estimation. The study was approved by the institutional review board of the University of Tokyo (11674-(4)). A 27-year-old healthy male with a height, weight, and body mass index of 173 cm, 55.3 kg, and 18.5 kg/m^2, respectively, wore eight electrodes connected to the device around his lower abdomen (Fig. 6) and recorded when and how much he urinated using a frequency volume chart for 12 h (7:54 a.m.–7:55 p.m.). The electrical conductivity of urine, which is known to affect the accuracy of urine volume estimation [13], was measured using LAQUAtwin-EC-33B (HORIBA, Ltd., Kyoto, Japan). He urinated seven times in the 12-h measurement (Table II), and the relationship between the EI values just before urination and the voided urine volume was evaluated. The urine conductivity in the seven urination instances varied between 7.02 and 16.02 S/m.

The urine volume was estimated via linear regression. The average EI values of the 1-min measurement immediately before urination (40 values) were adopted as independent variables. The regression model was evaluated via leave-one-out cross-validation (six training data and one test data). Fig. 7 presents a scatter plot of the voided and estimated urine volumes. The root mean square error (RMSE) and relative error in all data were 73 mL and 37 (SD, 122)% respectively. When we excluded a small amount of voided urine case (45 mL) as an outlier, the RMSE and relative error were 45 mL and -12 (SD, 14)%, respectively.

V. Discussion

We developed a portable device that can measure EI with a resolution of 0.017 Ω, which is sufficiently small to capture abdominal EI changes triggered by urine accumulation. The sensor device is small and has an internal battery for users to carry it in their pockets. These features validate the clinical applicability of our device.

Although the SNR of the raw outputs was not large, we applied a 90-s median filter and denoised the values with an SNR of 79.1 dB. Because this sensor device was developed to measure slowly changing urine volumes, a filter with a prolonged window size does not pose a challenge. When the measured EI values were small, a significant variation was observed between the measurements. However, by averaging multiple measurements, the measurement accuracy for small values can be increased. For long-term measurements in the resistive mesh phantom over 12 h, the values changed slightly with time. However, considering that this result was obtained over 12 h, a maximum of 0.542 mV and an average of a 0.100-mV difference was sufficient for stable measurement.

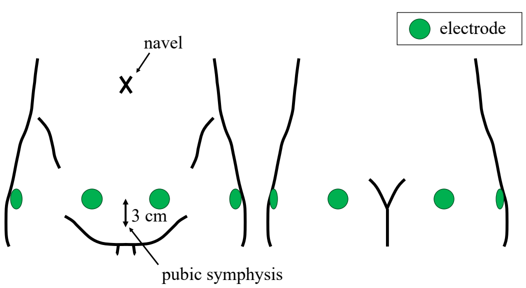

Figure 6. Electrode placement at the lower abdomen

Table II. Frequency Volume Chart in the healthy volunteer study

Time	Voided urine volume (mL)	Electrical conductivity of urine (S/m)
7:59	153	15.70
8:56	45	15.75
10:53	230	10.05
12:10	220	6.05
14:52	230	15.91
15:51	370	7.02
18:26	250	16.02

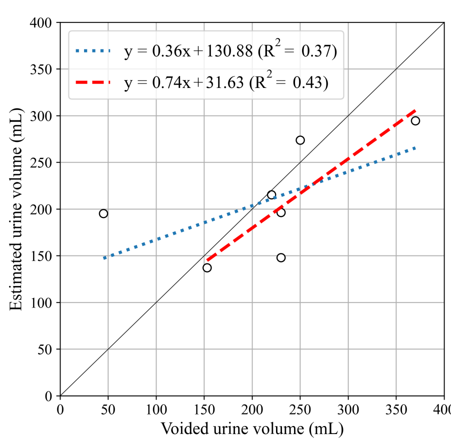

Figure 7. Scatter plot of voided and estimated urine volume. Blue dotted and red dashed lines are the regression lines with all seven points and six points excluding small amount of urine (45 mL), respectively.

The preliminary feasibility test in a young healthy volunteer indicates that the sensor device can safely measure abdominal EI in daily life. Moreover, the results suggest that the 8-electrode EI measurement can estimate urine volume in the bladder in human for the first time, although the estimation error was large when the voided volume was small. The variation in urine conductivity among the seven urination conditions might have affected the estimation accuracy (Table II). Further studies exploring the effect of body characteristics, including urine conductivity, on estimation accuracy are required.

VI. Conclusion

Our proposed small-size EI measurement device realized an SNR of 79.1 dB with a resolution of 0.017 Ω. The preliminary feasibility test in a healthy volunteer indicates that the 8-electrode EI measurement is feasible in daily life and can estimate urine volume in the bladder.

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