Advanced Biomedical Engineering
2: 17-24, 2013.

Original Paper

Involuntary Measurement System for Respiratory Waveform for Prevention of Accidental Drowning during Bathing

Kazuki NAKAJIMA,*, # Katsuhisa SEKINE,** Katsuya YAMAZAKI,*** Atsushi TANPO,* Yasushi TSUBOSAKA,* Juhyon KIM,* Kazuyuki TOBE,*** Masashi KOBAYASHI,*** Kazuo SASAKI*

Abstract Death rate of accidental drowning in the bathtub was the highest among casualties occurring at home, according to the annual report of the Japanese Ministry of Health, Labour and Welfare in 2007. To prevent accidental drowning during bathing at home, we obtained respiratory waveforms from bioelectric impedance (BEI) measurement using non-contact electrodes. The BEI measurement is an involuntary measurement method, from which respiratory waveform during bathing can be extracted. In the present study, to find the most appropriate electrode configuration as well as the optimal measuring frequency, we calculated the frequency dependence of impedance amplitude by numerical technique based on a three-dimensional finite difference method for a composite system consisting of a human body submerged in bath water. The results of model calculation agreed with the experimental results. Next, to obtain respiratory waveforms with large amplitudes, we investigated the optimal frequency experimentally. The frequency of 1 MHz was suitable for involuntary measurement of respiratory waveform during bathing.

Keywords : bathing, respiratory waveform, bioelectric impedance, involuntary measurement.

Adv Biomed Eng. **2**: pp. 17-24, 2013.

1. Introduction

Drowning in the bathtub, the major site of drowning at home, has been reported in UK[1], US[2, 3], and Japan[4, 5]. According to the annual report of the Japanese Ministry of Health, Labour and Welfare in 2007, the death rate of accidental drowning in bathtub was the highest among the casualties occurring at home. Especially, more than 60% of the deaths by accidental drowning in the bathtub were persons aged 65 years and older[4]. In most cases, death during bathing is attributed to respiratory arrest with unknown cause, cerebral stroke, ischemic heart disease, or drowning[6]. Continuous vital monitoring is expected to prevent accidental drowning during bathing at home. However, clinical monitors to measure vital signs are inconvenient for bathing at home because these monitors employ electrodes or probes that are attached to the subject. As a result, the probes and wires disturb the subjects' movements during bathing. Unconstrained and involuntary vital monitoring is required to prevent accidental drowning during bathing at home. Bathtub-electrocardiograph (bathtub-ECG) detects ECG through the bath water (tap water) during bathing, using electrodes that are placed on the bathtub wall, and heart rate can be evaluated without disturbing the subjects' movements during bathing[7, 8]. Moreover, respiratory waveform can be extracted from the ECG[9, 10]. The bathtub-ECG may serve as a monitor to detect not only heart disease but also apnea during bathing. Several respiratory monitors have been developed and used, such as bioelectric impedance (BEI) measurement[11], nasal thermistor[12], and capnography[13]. These conventional monitors employ probes and wires that might disturb the subjects' movements during bathing. Compared with other methods, BEI measurement is influenced by many different factors including geometry, tissue conductivity, and blood flow[11]. Therefore many applications have been developed. As one of these applications, we developed a BEI technique to monitor respiratory waveform during bathing, which does not require the subject's cooperation. We already reported the respiratory waveform of BEI measurement recorded using non-contact electrodes during bathing[14]. We placed electrodes near the chest in order to increase breathing movement sensitivity. However, such electrode placement restricts the subject's movements. Therefore, we sought to obtain respiratory waveform by placing the electrodes on the inner walls of the bathtub to allow free movements during bathing. Electrodes were placed on the bathtub wall so as not to disturb the subjects' movements during bathing, and BEI was monitored through the bath water[15]. In the present study, to find

This study was presented at the Symposium on Biomedical Engineering 2012, Suita, September, 2012.
Received on July 27, 2012; revised on October 13, 2012, November 25, 2012 and December 15, 2012; accepted on December 29, 2012.

* Division of Bio-Information Engineering, Faculty of Engineering, University of Toyama, Toyama, Japan.
** School of Health Sciences, Faculty of Medicine, Kanazawa University, Kanazawa, Japan.
*** The First Department of Internal Medicine, University of Toyama, Toyama, Japan.
# 3190 Gofuku, Toyama 930-8555, Japan.
E-mail: kazukin@eng.u-toyama.ac.jp

the most appropriate electrode configuration and the optimal measuring frequency, we calculated the frequency dependence of impedance amplitude by a numerical technique, which is based on a three-dimensional finite difference method (3D-FDM) for a composite system consisting of a human body submerged in bath water. The experimental results were then compared with the results of 3D-FDM calculation.

2. Model and calculation

2.1 Model

Respiratory waveform is obtained as the difference in amplitude of BEI measured at the inspiration stage and the expiration stage [14]. Current passing between electrodes is expected to change according to electrode configuration. A model simulation is employed to select the most appropriate electrode configuration and the measuring frequency that yields large amplitude in the respiratory waveform. To examine the effects of changes in lung volume and chest circumference due to respiration, we used a simplified three-dimensional model (3D-model) as shown in **Fig. 1**. This 3D-model consists of concentric circular cylinders: the inner circular cylinder with radius R_{lung} represents the lung; the region between R_{lung} and R_{chest}-T_{cm} represents the body fluid including blood, tissue, bone; and the region between R_{chest}-T_{cm} and R_{chest} represents an electrically insulating layer equivalent to the cell membrane at the stratum corneum (SC) [16], where R_{chest} and T_{cm} denote the chest radius and the membrane thickness, respectively. The permittivity ε and conductivity κ for the lung and the body fluid are considered to be dependent on frequency due to their complicated structures [17]. However, to simplify the problem in the present study consisting of the whole body and bath water, we assumed that these regions are uniform and are characterized by ε and κ independent of frequency, as follows: $\varepsilon_{\text{lung}} = 1$ and $\kappa_{\text{lung}} = 0$ S/m equivalent to air representing the lung region, and $\varepsilon_{\text{bf}} = 78$ and $\kappa_{\text{bf}} = 1.6$ S/m equivalent to physiological saline solution representing the body fluid. For the cell membrane, $T_{\text{cm}} = 3$ nm, $\varepsilon_{\text{cm}} = 3$ and $\kappa_{\text{cm}} = 0$ S/m. The values of ε_{cm} and κ_{cm} are consistent with typical hydrophobic liquids [18]. Using the values of ε_{cm} and T_{cm}, we obtained the value of specific membrane capacitance, C_{M}, as $\varepsilon_0 \varepsilon_{\text{cm}}/T_{\text{cm}} = 8.9 \times 10^{-3}$ F/m^2, which is of the same order as that reported previously [19]. The values of R_{chest} at the maximum end inspiration stage (MaxIn) and maximum end expiration stage (MaxEx) were set at 160 and 140 mm, respectively, based on the results of chest measurements of the subjects. The values of R_{lung} at MaxIn and MaxEx were estimated to be 80 and 40 mm, respectively, from the vital capacity of the subjects. For the bath water, $\varepsilon_{\text{bw}} = 78$ and $\kappa_{\text{bw}}/\kappa_{\text{bf}} = 1/1000$, which are typical of water used for bathing. The size of the electrode was 120 × 120 mm. The 3D-model calculation regarding the frequency dependence of impedance amplitude was conducted for four different electrode configurations to choose the most suitable configuration (**Fig. 2**).

2.2 Calculation

2.2.1 Calculation method

We calculated the conductance G, the capacitance C, and the impedance amplitude $|Z|$ by the 3D-FDM at MaxIn and at MaxEx for the frequency range of 1 μHz to 10 THz. Potential distributions were calculated by the 3D-FDM at the condition of a voltage difference of 1 V between the electrodes. Boundary condition of zero vertical component in electric field was given. Calculation was carried out for the region enclosed by the bold line in **Fig. 3**, because of the symmetric property of the 3D-model. The unit lengths and numbers of units in 3D-FDM are also shown in **Fig. 3**. Electrical current that flows between the electrodes was calculated based on the 3D-FDM, and capacitance C, conductance G, and impedance-amplitude Z were obtained from the electrical current. Discrete equations about complex permittivity and discrete method of region in the 3D-FDM followed Asami's method [20]. Frequency dependence of the impedance amplitude

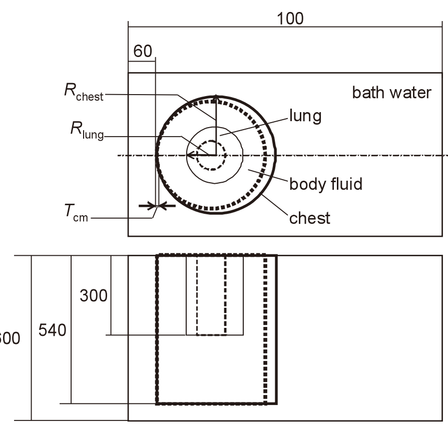

Fig. 1 A simplified 3-dimensional model of the inner structure of the upper body. Solid line and dotted line indicate MaxIn and MaxEx stages, respectively.

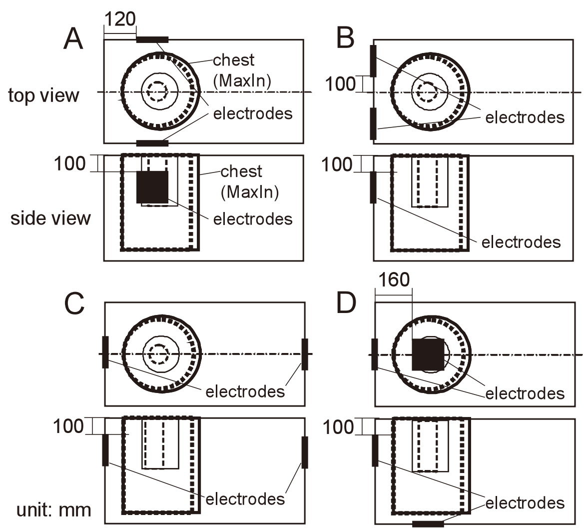

Fig. 2 Electrode configurations. Bold lines and closed squares indicate electrodes. Solid line and dotted line indicate MaxIn and MaxEx stages, respectively.

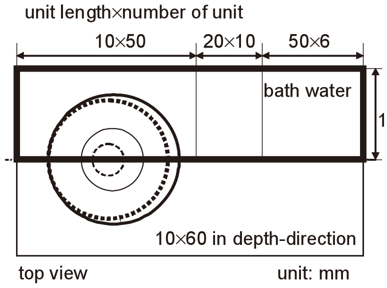

Fig. 3 The area enclosed by the bold line indicates the calculation region.

of water was also calculated by the 3D-FDM.

2.2.2 Calculation results

For the electrode configuration A, the simulation results of the capacitance, the conductance, and the impedance-amplitude Z are shown in **Fig. 4**. Frequency dependence of water showed a single relaxation time, which is the typical dielectric dispersion of water. This finding shows that the 3D-model is adequate for our analysis. Frequency dependence of the 3D-model showed dielectric dispersion including two relaxation times. The impedance amplitude at MaxIn was larger than that at MaxEx between 1 μHz and 0.1 Hz. The impedance amplitude at MaxIn was smaller than that at MaxEx between 0.1 Hz and 1 MHz. The impedance amplitudes of both water and the 3D-models showed almost the same values from the frequency of 1 MHz.

Figure 5 shows the normalized impedance amplitude difference between MaxEx and MaxIn at 1 MHz for the four electrode configurations. We chose electrode configuration A for subsequent experiments because the amplitude difference was the largest.

3. Experimental methods

3.1 Bathtub

In this study, a bathtub made from fiber-reinforced plastic (FRP) with internal dimensions of 960 × 580 × 560 mm was used in a condominium. Dimensions of the bathtub are shown in **Fig. 6**. From the calculation results, electrode configuration A was employed. The size of the electrode used in the experiment was the same as that used in calculation. The stainless steel electrode (SUS304) 100 × 100 mm in size and 0.1 mm in thickness was attached to a polyvinyl-chloride plate 2 mm in thickness, with double-faced adhesive tape. Electrode configuration A was employed according to the result of the 3D-FDM calculation. The bathtub was filled with bath water at 39 ± 1°C, and the bathroom temperature was kept between 23 and 28°C.

3.2 Subjects

The study protocol was approved by the University of Toyama Ethics Committee, and written informed consent was obtained from the each participant. Eighteen healthy volunteers had medical check-up conducted by a medical doctor, and the medical doctor permitted sixteen males

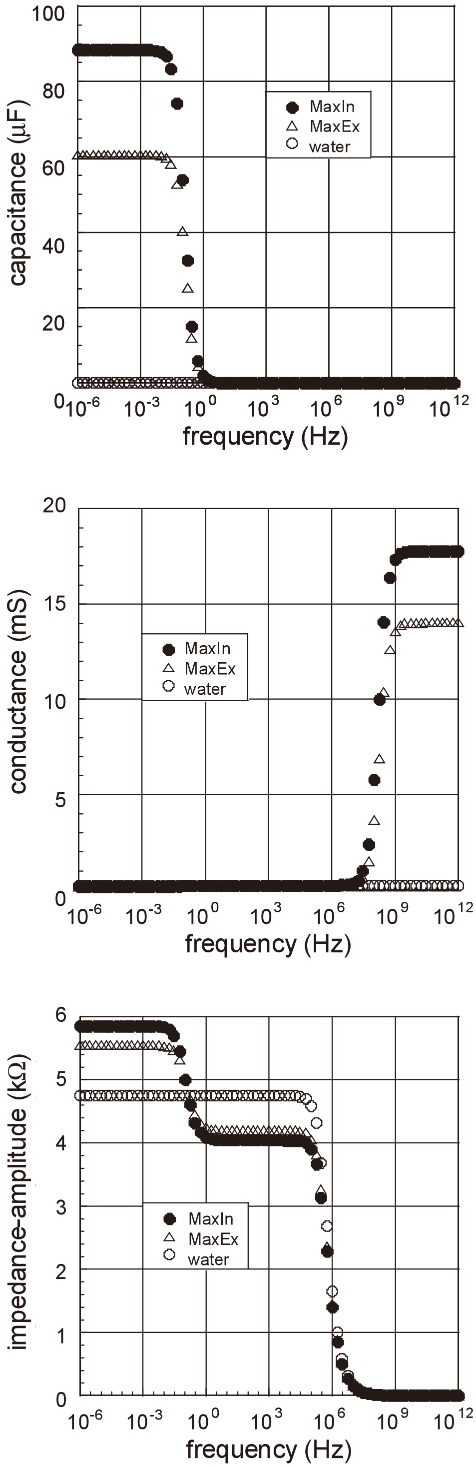

Fig. 4 Calculation results for frequency dependence of impedance amplitude using electrode configuration A.

aged 22–45 (mean ± SD: 24.2 ± 5.8) years to participate in the experimental protocol. They averaged 63.6 ± 9.1 kg in weight and 171.3 ± 5.8 cm in height.

3.3 BEI spectrum

An electrode was connected to a wire 300 mm in length with a metal binder-clip; the other end of the wire was soldered to a BNC panel jack on an aluminum board 0.5 mm in thickness, as a guard electrode (**Fig. 6**). The BNC panel jack was connected to two co-axial cables (3D2 V, Fujikura, Japan) 4 m in length by a BNC T-shape adapter; the other ends of the co-axial cables were connected to a precision impedance analyzer (4294A, Agilent Technologies, USA) with BNC connectors. Frequency dependence of impedance amplitude was measured at MaxIn and at MaxEx over a frequency range of 4 kHz to 4 MHz (8 points/decade) with an applied current of 141 μA rms using the "averaging mode" of the impedance analyzer. At this stage, the impedance of the bath water was measured without the subject.

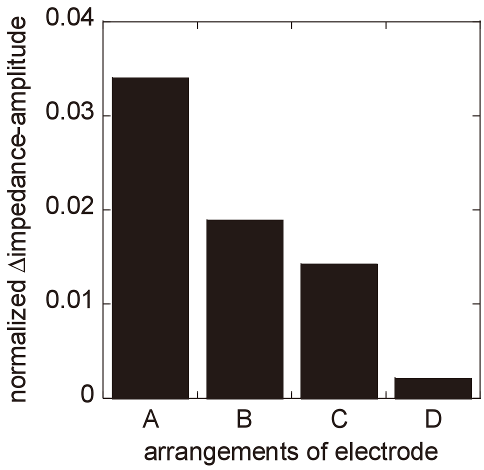

Fig. 5 Comparison of the difference in impedance amplitude among four electrode configurations.

3.4 Respiratory waveform

Figure 7 shows the electrical circuit for detecting respiratory waveform during bathing. The circuit comprised mainly three sections: constant current section, non-inverting amplifier section, and half-wave rectifier section. A constant current (1 MHz, 70 μA) was applied to the electrodes placed on the inner wall of the bathtub. Voltage between the electrodes was amplified. AC signals from the amplifier output were rectified and smoothed, and an envelope of AC signals as a respiratory waveform was obtained. Electric power for this circuit was supplied by an alkaline battery of 9 V (6LF22, Toshiba) to prevent electric shock.

Figure 8 shows a schematic diagram of the whole monitoring system to obtain respiratory waveform. For

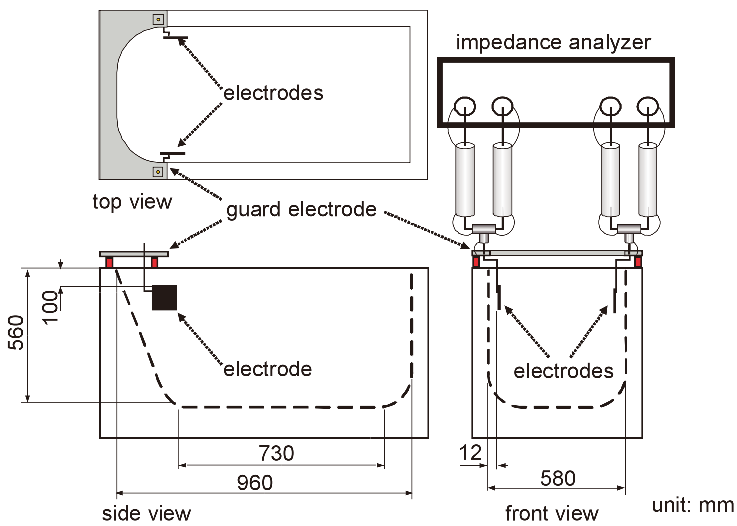

Fig. 6 Schematic representation of the system for measuring frequency dependence of impedance amplitude.

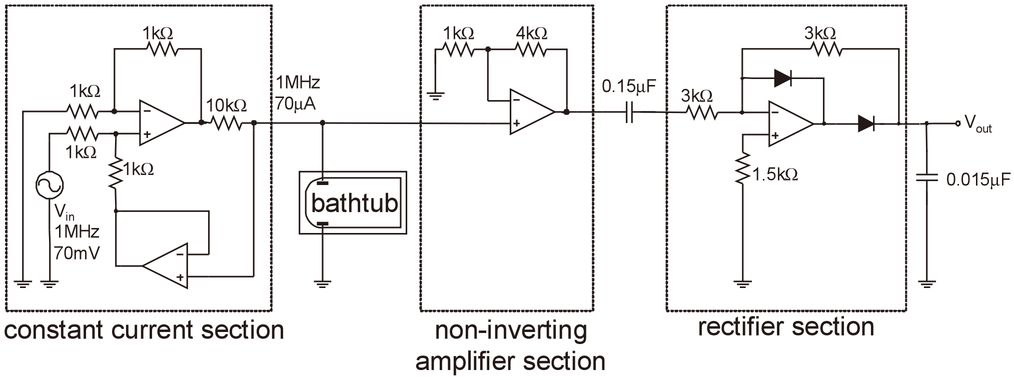

Fig. 7 Respiratory waveform detection circuits.

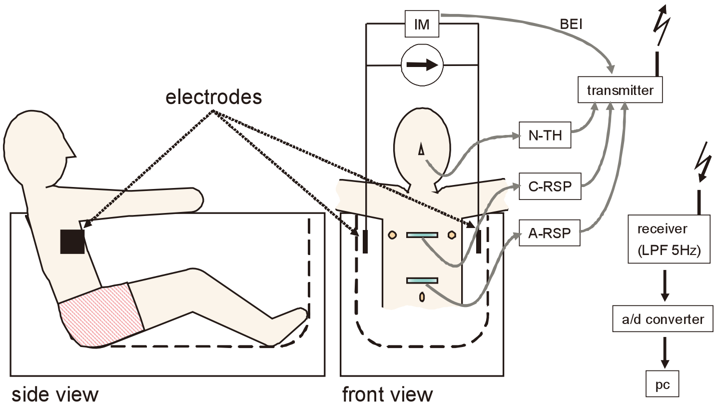

Fig. 8 Diagrammatic representation of the measuring system for respiratory waveforms. IM: amplifier for BEI measurement, N-TH: nasal thermistor, RSP: rubber strain gauge plethysmograph.

BEI measurements, the subject was sandwiched by a pair of electrodes. In all the instruments, electric power was supplied by a medical isolation transformer (SM-204 V, Nihon Koden, Japan). The electrode was connected to a wire 300 mm in length using a metal binder-clip; the other end of the wire was soldered to the electrical circuit.

Three respiratory waveforms were simultaneously measured as reference signals with two rubber strain gauge plethysmographs (RSPs) and a nasal thermistor (NTH). According to the literature [21], one RSP (C-RSP) was placed on the midline of the chest at a point that gave maximal deflection during quiet breathing. The other RSP (R-RSP) for abdominal motion was placed on the midline 50 mm cephalad to the umbilicus. These respiratory sensors were attached with medical adhesive tape (25 mm in width, Transpore Surgical Tape, 3M, Japan) to the subject. These signals were monitored by a multi-telemeter system (WEB-5000, Nihon Koden) and were acquired by a data acquisition system (Powerlab 8/30, Chart Ver. 5.4.2, AD Instruments, USA). All the sensors including BEI were recorded at sampling intervals of 0.1 s.

3.5 Protocol

The subject sat in a chair for 10 minutes and rested before starting the examination. No sensor was attached to the subject in spectrum measurement, and the respiratory sensors as reference signals were attached to the subject during respiratory waveform measurement, as shown in the previous section. After that, the subject took a shower outside of the bathtub to wash away his sweat before soaking his body in the bathtub. The subject gently soaked in the bathtub, and excess bath water was allowed to overflow. The subject was asked to put his elbows on the edges of the bathtub to maintain his posture. After the measurements, which took a total of 10 minutes, the subject slowly got out of the bathtub. The examiner wiped the subject's body with a towel and removed all the reference sensors.

3.6 Statistics

Data were analyzed for frequency dependence of impedance amplitude with respect to each study variable by Kruskal-Wallis and Steel-Dwass multiple comparison method. A 95% confidence level was considered significant.

4. Experimental results

Figure 9 shows the typical spectra of frequency dependence of impedance amplitude for water only and a subject in water. The relative shape of the frequency dependence of impedance amplitude supports the result of the 3D-model calculation by 3D-FDM (see **Fig. 4**), although the impedance amplitude increases at around the frequency of 1 MHz in the experiment.

The averaged differences in impedance amplitude between MaxIn and MaxEx for various frequencies were compared to obtain the optimal frequency for respiratory monitoring (**Fig. 10**). The averaged difference in impedance amplitude was significantly and markedly larger at the frequency of 1 MHz. Therefore, the frequency of 1 MHz is suitable for involuntary measurement of respiratory waveform during bathing.

Typical respiratory waveforms during bathing are shown in **Fig. 11**. The BEI waveform increased with expiration and decreased with inspiration. These results are consistent with those of the spectrum as shown in **Fig. 9**. The BEI waveform at the frequency of 1 MHz synchronized with that of C-RSP. In this subject, the A-RSP has inverse phase compared to C-RSP and BEI. When the subject held his breath, all the waveforms including that of BEI flattened. During the breath holding, the reference signals were smooth but the BEI signals showed a little fluctuation. The amplitude of the small fluctuations was less than 10% of the respiratory waveforms. After breathing was resumed, the synchronous waveforms were restored. The respiratory waveforms of BEI in all subjects synchronized with those of the references (197 breaths). The BEI waveforms were flattened during breath holdings in all subjects (25 breath holdings).

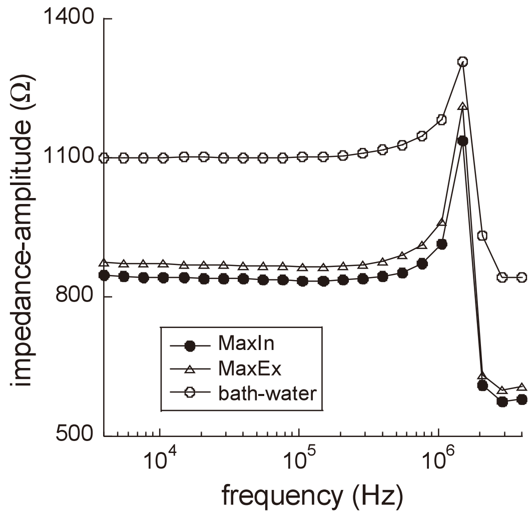

Fig. 9 Typical spectra of frequency dependence of impedance amplitude for water and a subject in water.

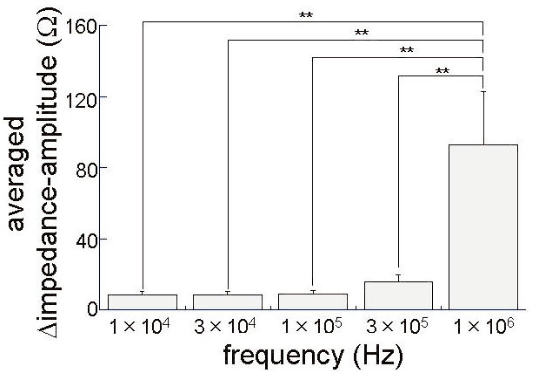

Fig. 10 Comparison of averaged difference in impedance amplitude between MaxIn and MaxEx. (No. of subject = 11, n = 24)

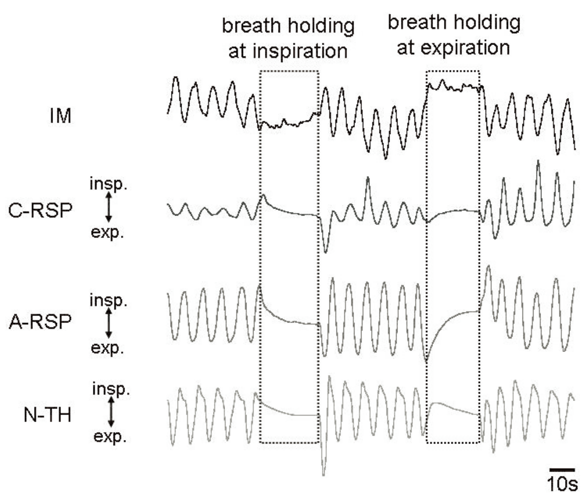

Fig. 11 Typical results of respiratory waveforms during bathing.

5. Discussion

The frequency dependence of impedance amplitude was calculated by the 3D-FDM model, and measured by the BEI method via bath water. The results of the model calculation agreed with the experimental results. In other word, the 3D-FDM model reflects the BEI measurement via bath water and the result of electrode configuration shows an adequate relationship between the frequency and the impedance amplitude. The frequency dependence of impedance amplitude was compared among four electrode configurations in this study. We found that the best configuration is the face-to-face position of the electrodes and at the closest distance between the electrodes. We chose the measuring frequency at 1 MHz because the averaged difference in impedance amplitude was significantly and markedly larger at the frequency of 1 MHz than at other frequencies. Although there are no calculation results, we experimentally confirmed that the increased impedance amplitude at around 1 MHz would be caused by the grounding method of the measuring instrument.

In this study, we showed that the BEI technique evaluates the breath holding phase as well as the respiratory waveforms. This technique evaluates the movement of the chest region. The small fluctuations of BEI signals during breath holding may reflect the movement of the chest region caused by discomfort of breath holding. This indicates that self-care motions such as arm movement during bathing influence the impedance amplitude. Therefore, the respiratory waveform obtained by the BEI technique would be affected by the self-care motions, because arm movements during bathing influence the impedance amplitude. However, the changes in BEI waveform as a result of self-care motions can be considered to be a vital sign that the person is alive.

In this study, we developed a system for involuntary measurement of respiratory waveform. The respiratory waveform obtained from the BEI system synchronized with those obtained from conventional sensors. The instruments for recording bathtub ECG use similar electrode configuration as in the present study [8–10]. However, the respiratory waveform obtained from the bathtub ECG is measured passively, whereas that from the BEI is measured actively. In other words, the signals obtained from the BEI generate the respiratory waveform and may be used to measure body fat via tap water in the future.

Other passive monitoring methods using image processing have been reported. In one method, respiratory signals during bathing are measured by an image sequence. Wavelet at the water surface in the bathtub is analyzed, and respiratory rate is estimated [22]. In another method, body surface motions including respiratory motion are evaluated using a fiber grating vision sensor in the bathroom [23]. These methods use a video camera as the image sensor, and does not disturb the subjects' movements during bathing. However, many people object to the use of a video camera in the bathroom because it is private space. On the other hand, the BEI technique that we developed does not invade the subject's private space.

The BEI technique is an active monitoring method. The body fat meter is a popular instrument used at home for body weight control and preventive medicine. Body fat is estimated by a BEI technique, and a person has to touch the electrodes of the BEI monitor to measure body fat. The BEI technique that we developed may be able to monitor body fat during bathing. As a next step, we plan to develop a BEI technique for involuntary monitoring of not only respiratory waveform but also body fat during bathing.

When bathing, Japanese customarily immerse the whole body in deep, hot water. The whole body is thus subjected to heating and hydrostatic pressure. This style of bathing may result in accidents in the bath. Abnormal breathing and breath holding during bathing can be detected by our BEI technique. Acute cardiac event also can be detected by the bathtub ECG[7, 8]. By installing these instruments in a home bathtub, the vital signs of the bather are monitored automatically and unobtrusively without the subject being aware of the monitoring.

References

1. Nowers MP. Suicide by drowning in the bath. Med Sci Law. **39**(4), pp. 349–353, 1999.
2. Budnick LD, Ross DA. Bathtub-related drownings in the United States, 1979–81. Am J Public Health. **75**(6), pp. 630–633, 1985.
3. O'Carroll PW, Alkon E, Weiss B. Drowning mortality in Los Angeles County, 1976 to 1984. JAMA. **260**(3), pp. 380–383, 1988.
4. Japanese Ministry of Health, Labour and Welfare, "Vital statistics in 2006" (*in Japanese*) [Online] Available: http://www.mhlw.go.jp/toukei/saikin/hw/jinkou/suii07/deth18. html[accessed December 25, 2008].
5. Chiba T, Yamauchi M, Nishida N, Kaneko T, Yoshizaki K, Yoshioka N. Risk factors of sudden death in the Japanese hot bath in the senior population. Forensic Sci Int. **149**(2–3), pp. 151–158, 2005.
6. Yoshioka N, Chiba T, Yamauchi M, Monma T, Yoshizaki K. Forensic consideration of death in the bathtub. Legal Med. **5**, pp. S375–81, 2003.
7. Kwatra SC, Jain VK. A new technique for monitoring heart signals--Part I: Instrumentation design. IEEE Trans Biomed Eng. **33**(1), pp. 35–41, 1986.
8. Ishijima M. Unobtrusive approaches to monitoring vital signs at home. Med Biol Eng Comput. **45**, pp. 1137–1141, 2007.
9. Motoi K, Ogawa M, Ueno H, Kuwae Y, Ikarashi A, Yuji T, Higashi Y, Tanaka S, Fujimoto T, Asanoi H, Yamakoshi K. A fully automated health-care monitoring at home without attachment of any biological sensors and its clinical evaluation. Conf Proc IEEE Eng Med Biol Soc. **2009**, pp. 4323–4326, 2009.
10. Yamakoshi K. Current status of noninvasive bioinstrumentation for healthcare. Sens Mater. **23**(1), pp. 1–20, 2011.
11. Patterson R. Bioelectric impedance measurements. In: Bronzino JD, editor. The Biomedical engineering handbook 2nd ed. Boca Raton: CRC press; 2000. 73–1–8.
12. Biomedical applications of physical sensors. In: Bronzino JD, editor. The biomedical engineering handbook 2nd ed. Boca Raton: CRC press; 2000. 47–15–16.
13. Ellett ML, Woodruff KA, Stewart DL. The use of carbon dioxide monitoring to determine orogastric tube placement in premature infants: a pilot study. Gastroenterol Nurs. **30**(6), pp. 414–417, 2007.
14. Nakajima K, Sekine K, Yamazaki K, Sakai Y, Tampo A, Fukunaga H, Yagi Y, Ishizu K, Nakajima M, Tobe K, Kobayashi M, Sasaki K. Non-contact respiratory monitoring with a bioelectric impedance technique to detect abnormal respiration during bathing. Jpn J Appl Phys. **48**, pp. 107001–1–5, 2009.
15. Nakajima K, Sekine K, Yamazaki K, Tampo A, Omote Y, Fukunaga H, Yagi Y, Ishizu K, Nakajima M, Tobe K, Kobayashi M, Sasaki K. Detection of respiratory waveforms using non-contact electrodes during bathing. Conf Proc IEEE Eng Med Biol Soc. **2010**, pp. 911–914, 2010.
16. Weavera JC, Vaughana TE, Chizmadzhevb Y. Theory of electrical creation of aqueous pathways across skin transport barriers. Adv Drug Delivery Rev. **35**, pp. 21–39, 1999.
17. Brown BH. Electrical impedance tomography (EIT): a review. J Med Eng Technol. **27**, pp. 97–108, 2003.
18. Krylovich VI, Mikhal'kov VV. Raising the accuracy of measuring the dielectric permittivity of fluids. J Eng Phys Thermophys. **56**(1), pp. 68–72, 2004.
19. Takashima S. Electrical properties of biopolymers and Membranes. Adam Hilger, Bristol, Philadelphia; 1989.
20. Asami K. Dielectric dispersion in biological cells of complex geometry simulated by three-dimensional finite difference method. J Phys D: Appl Phys. **39**, pp. 492–499, 2006.
21. Gilbert R, Ashutosh K, Auchincloss JH Jr, Rana S, Peppi D. Prospective study of controlled oxygen therapy. Poor prognosis of patients with asynchronous breathing. Chest. **71**, pp. 456–462, 1977.
22. Nakajima K. An evaluation method for water surface fluctuation during bathing, Proc. Ann. Fall Meeting of the Biomedical Engineering Society and 21st Ann. Int. Conf. IEEE EMBS. 1999; 695.
23. Nishiura T, Nakajima M. Development of a bathroom watching system using fiber grating vision sensor with a built-in color pickup device. IEICE 2006. **J89-D**(5), pp. 1001–1010, 2006 (in Japanese).

Kazuki NAKAJIMA

Kazuki NAKAJIMA is a professor of Faculty of Engineering, University of Toyama. He received his Ph.D. degree in 1996 from Yamaguchi University. He has served as a research associate in Yamaguchi University from 1988. From 1999, he served as a chief, Laboratory for Development of Equipment for Nursing and Personal Care, Department of Gerontechnology, National Institute for Longevity Sciences. His research and teaching activities are focused on biomedical transducers involving noninvasive apparatus and bio-signal analysis.

Katsuhisa SEKINE

Katsuhisa SEKINE is an associate professor of Faculty of Health Sciences, Institute of Medical, Pharmaceutical and Health Sciences, Kanazawa University. He received Ph.D. from Kyoto University in 1988. His research interests are noninvasive structural analyses of biological cells, tissues and bodies with ac electric measurements, and biological effects of electromagnetic fields.

Katsuya YAMAZAKI

Katsuya YAMAZAKI is an assistant director of Kawai Clinic, Tsukuba.

Atsushi TANPO

Atsushi TANPO received the B.E. and the M.E. degree in Engineering from University of Toyama, Japan in 2007 and 2009, respectively.

Yasushi TSUBOSAKA

Yasushi TSUBOSAKA received the B.E. and the M.E. degree in Engineering from University of Toyama, Japan in 2008 and 2010, respectively.

Juhyon KIM

Juhyon KIM received the B.E., the M.E. and Ph.D. degree in Engineering from University of Toyama, Japan in 2004, 2006 and 2009, respectively. He was a Postdoctoral Researcher in University of Tennessee Health Science Center, USA during 2009–2010. Since 2011, he is a lecturer in Faculty of Engineering, University of Toyama. His research interests are electrophysiology. He is a member of Society for Neuroscience.

Kazuyuki TOBE

Kazuyuki TOBE is a professor of First Department of Internal Medicine, University of Toyama.

Masashi KOBAYASHI

Masashi KOBAYASHI is an emeritus professor of First Department of Internal Medicine, University of Toyama.

Kazuo SASAKI

Kazuo SASAKI is an emeritus professor of University of Toyama. He received his Ph.D. degree in 1983 from Kyusyu University. He had served as an associate professor, Toyama Medical and Pharmaceutical University from 1988. He was a professor of Faculty of Engineering, University of Toyama from 1994. His research interests are neurophysiology, behavioral physiology, and biomedical engineering.