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Performance Evaluation of Five Types of Ag/AgCl Bio-Electrodes for Cerebral Electrical Impedance Tomography

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Abstract—Electrical impedance tomography (EIT) is an emerging medical imaging technique, which has already been investigated in several clinical applications due to its low-cost, non-invasiveness, non-radioactivity, high temporal resolution, and great sensitivity to impedance changes. One potential use of EIT is to perform long-term continuous imaging monitoring of brain for patients who suffer from severe cerebral diseases. However, this application requires a demanding performance of electrodes because of the characteristics of cerebral EIT measurements. Although Ag/AgCl bio-electrodes are widely used for clinical practices or EIT research at the moment, influences of different types of Ag/AgCl electrodes on cerebral EIT measurements have not been investigated. In this study, five common types of Ag/AgCl bio-electrodes were put into comparison by measuring the forearm and the brain of 10 healthy adult volunteers and evaluating those data in frequency or time domain in terms of contact impedance, uniformity, signal-to-noise ratio, and stability. Results show that Ag/AgCl powder electrode has an overall best performance with as low contact impedance as commercial ECG electrodes ($p > 0.05$), high SNR (60.3 ± 4.5 dB), better uniformity (coefficient of correlation 0.95 ± 0.03), and greater stability (slope 0.68 ± 0.03). After further improvement in design and instrumentation, Ag/AgCl powder electrode is likely to become the optimal choice for cerebral EIT measurements and provide feasible technical support for further research or application in cerebral EIT.

Keywords—Electrical impedance tomography, Electrode system, Contact impedance, Uniformity, SNR, Stability.

INTRODUCTION

Electrical impedance tomography (EIT) is an emerging medical imaging technique. As different tissues have different bio-electrical impedance, images of biological impedance distribution could be reconstructed by injecting safe currents into body and measuring electrical potentials on the body surface.[1,3,22] Although there are some existing problems such as relatively low spatial resolution when compared with other imaging modalities like X-ray and MRI, EIT has many advantages including low-cost, non-invasiveness, non-radioactivity, high temporal resolution, and great sensitivity to impedance variations caused by early physiological or pathological changes, etc.[1,2,6] To date, some research groups have already applied the technique into imaging study of thorax, abdomen, and other clinical trials.[13,14,17,24,39,44] In the aspect of cerebral EIT, the research mainly includes two directions: cerebral function imaging and dynamic monitoring of brain diseases. As to cerebral function imaging, the group directed by Holder systematically conducted a series of experiments to detect acute stroke, epileptic seizures, and visual evoked potentials by EIT method and had obtained some encouraging conclusions.[11,12,15,26,37,38]

Besides, EIT has the potential of use in dynamic monitoring of the progression of severe cerebral diseases, such as intracerebral hemorrhage (ICH), epidural hematoma (EDH), subdural hematoma (SDH), and subarachnoid hemorrhage (SAH). Researches about bedside dynamic monitoring using cerebral EIT have been conducted by our group, with major focus on intracranial hemorrhage.[8,30,32–34,41,42] Real-time monitoring on the changes of cerebral impedance can be realized by EIT method, through which the extent of brain damage due to the hemorrhage may be evaluated.[40] Hence, this can provide invaluable and timely information for the diagnosis and treatments of those patients.

Electrode system, as a key component of cerebral EIT technique, is employed to emit currents and receive voltage measurements. The performance of

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electrode system may directly influence the accuracy of the measurements and eventually affect the quality of EIT images. Besides, due to the relatively strong electrical resistivity of cranium that impedes the flow of current and the shunting effects of cerebrospinal fluid and scalp on current, the electrode system of cerebral EIT needs to have relatively high signal-to-noise rate (SNR) and low contact impedance. In addition, the electrode system should also have good uniformity and stability in that the time for monitoring of patients with cerebral trauma is often quite long.

At present, studies on electrode properties applying to EIT imaging have been carried out by a number of research groups, mainly concerning the effects of electrode form, material, working frequency, temperature, humidity on electrode properties.[18,19,22,24,28,29,35,36,40,43] Hydrogel electrode was investigated to be superior to Ag/AgCl electrode to some extent,[36] but the size of it is too huge to be suitable for long-term cerebral monitoring in clinical application. In general, Ag/AgCl electrode is preferable in biosignal recording applications like EIT measurements because it features low and stable offset potentials, low and matched interface impedance, and low polarization,[20] but up to now, there have been many types of Ag/AgCl electrodes by the different techniques, such as powder pressing, filming, hot dip coating and electrolyzing, etc., and the influences on the data measurements of cerebral EIT with different types of Ag/AgCl electrodes have not been explored. Because it has been investigated by Duan that Ag/AgCl powder electrode has an outstanding performance in EEG diagnosis and it has already been widely applied in studies about brain–computer interface and EEG-based biofeedback medical devices,[9,10] we presume that this type of electrode may also exhibit better performance in cerebral EIT. Therefore, this electrode type will be evaluated against a number of others for the cerebral EIT measurements in this study.

Considering the characteristics of cerebral EIT measurements, five common types of Ag/AgCl electrodes were compared in this paper, for the purpose of determining one optimal electrode type suitable for cerebral EIT. Specifically, five types of Ag/AgCl bioelectrode system were divided into four experiment groups (four different types of Ag/AgCl electrodes) and one control group (commercial self-adhesive ECG electrode system). By conducting impedance measurements of the left forearm and the head of 10 adult volunteers using the 2-electrode method and the 16-electrode method, respectively, comprehensive evaluation about the performance of each type electrode system can be obtained with respect to four criterions: contact impedance, SNR, uniformity, and stability.

FIGURE 1. Five common types of Ag/AgCl bio-electrodes tested in the experiment.

MATERIALS AND METHODS

Experimental Electrodes

Figure 1 shows the physical appearance of five tested types of Ag/AgCl electrodes. Electrode A is mainly made by pressing Ag/AgCl powder (Oxford Instruments, Woking, UK), of which the concrete manufacturing procedure is not clear. Electrode B is sintered and parched after being coated with the prepared Ag/AgCl paste of certain viscosity on a copper sheet (Hongxing Electronic Paste Technology Co. Ltd, Xi'an, China). Electrode C is an Ag/AgCl-coated electrode though the method of electrochemistry at room temperature, whose sheet of pure silver is cut to the same size of electrode (BX-150-02, College of Chemistry and Molecular Sciences of Wuhan University, Wuhan, China). Electrode D is also an Ag/AgCl powder electrode made without natural light and it is encapsulated with epoxide resins after compression and parches (CX-150-02, College of Chemistry and Molecular Sciences of Wuhan University, Wuhan, China). Electrode E is a commercial ECG electrode (CONMED Inc., New York, USA).

Experimental Subjects

Ten healthy adult volunteers were selected, of whom half were men and half women (age 30.5 ± 4.8 years; weight 65.7 ± 8.5 kg). The temperature for the measurements was maintained at 25 ± 1 °C, the humidity at 55 ± 5%. The left forearm of these volunteers was measured by the 2-electrode method, the head by the 16-electrode method. The subjects were at supine position and asked to hold still. We also tried to let them sleep as possible as they can to maintain a stable level of brain activities. If the subjects disobey these rules, we would stop the experiment and restart it. The experiment was carried out after being approved by Ethics Committee of the Fourth Military Medical University. All volunteers signed the informed consent form.

Experimental Setups

Precision impedance analyzer (Agilent 4294, Agilent Technologies, Palo Alto, USA) was selected to conduct impedance measurements through the 2-electrode method. Three fixed resistors (3 Ω, 810 Ω, and 1.22 kΩ) were used to calibrate the impedance analyzer before the experiment. The 16-electrode measurements were performed using the cerebral EIT monitoring system developed by our group,[31,39] of which working frequency ranges from 1 to 190 kHz, with measuring accuracy standing at $\pm 0.01\%$ and CMRR over 80 dB. In the experiment, opposite driving/adjacent measuring mode was determined for 16-electrode measurements with the driving current of 1250 μA at 50 kHz, which is proved to be safe for the human.[22] In total data of 3,000 frames were collected at acquisition speed of 1 frame/s.

In the experiment, except that evaluation of contact impedance was done by the 2-electrode method, other performance evaluations were accomplished by the 16-electrode method at the frequency of 50 kHz. The electrode placements on the forearm and the head are shown in Fig. 2. When using the 2-electrode method, two electrodes were placed on the medial line of each volunteer's left forearm with an interval of 5 cm. The distance between the right electrode and wrist is also 5 cm. When using the 16-electrode method, 16 electrodes were equally spaced placed on the head, 2 cm above supraorbital ridge in a circle. Prior to the placement of electrodes, the skin of volunteers was prepared and abraded.[20] Afterwards, conducting gel (Elefix, Z-410CE, NIHON KOHDEN, Tokyo, Japan) was used to enhance adhesion and reduce contact impedance between electrodes and skin.[17]

Performance Criterions

Four performance criterions, including contact impedance, uniformity, SNR, and stability, were used to evaluate the performance of electrodes in the research and each electrode system was tested under the same conditions, respectively. Because ECG electrode has been extensively used in clinical research and practice and presents relatively good performance in general, it was considered as the reference in comparison with other tested electrodes.

Contact Impedance

Bio-impedance measured by electrodes generally consists of contact impedance and tissue impedance. Especially at low frequency, contact impedance plays a predominant role in measured values. Although contact impedance decreases with the increasing frequency, it could still reach as high as hundreds of ohms at 100 kHz, which can significantly affect the bio-impedance measurements.[17] It is thus desirable for electrode system to have magnitude of contact impedance between bio-electrode and skin surface as small as possible, which is the key requirement for EIT measurements.[20] However, contact impedance generally cannot be measured directly in experimental practice. According to the principle of the 2-electrode measuring method, these two electrodes were simultaneously used as measuring electrode pair and driving electrode pair. Therefore, the measured impedance value partially contains contact impedance.[20] In this study, contact impedance was indirectly measured by the 2-electrode method and then normalized to reduce the variability in subjects. The difference at 50 kHz between each electrode type was also compared. In addition, variance of measured impedance over 30 min was also calculated in order to assess the changes against time.

Uniformity

The electrode system used in cerebral EIT impedance measurements are generally composed of 16 or more independent cup-shaped electrodes. As a result, the extent of uniformity of these 16 same cup-shaped electrodes would directly affect the accuracy of data collection. Because the 16-electrode method was used for measurements in the mode of opposite driving/adjacent measuring, there were totally 192 independent data, or exactly 12 independent data under individual

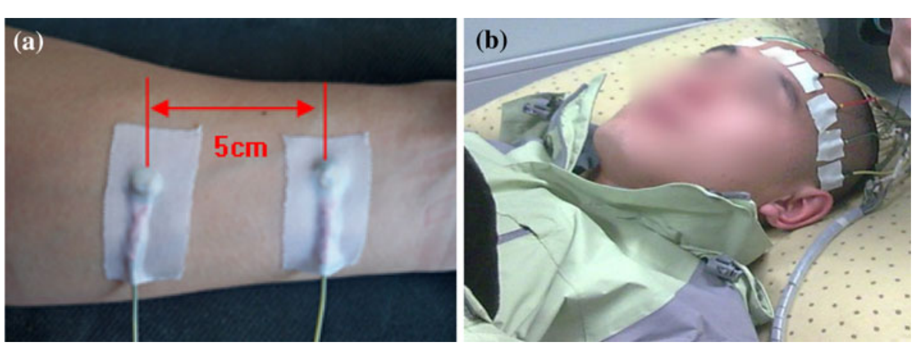

FIGURE 2. Electrode placement on forearm by the 2-electrode method (a) and on head by the 16-electrode method (b).

measuring electrode pair, in each frame.[7] A diagram of driving and measuring modes is shown in Fig. 3. The number of collected frames is $M = 3000$, and the measured data for each measuring electrode pair is a vector $\mathbf{v}_j^i = (v_1^i, v_2^i, \ldots, v_{12}^i)_j$, of which $i = 1, \ldots, 16$ stands for the number of measuring electrode pairs, $j = 1, \ldots, M$ represents index of collected frames and the average against time is $\bar{\mathbf{v}}^i$,

$$\bar{\mathbf{V}}^i = (\bar{v}_1^i, \bar{v}_2^i, \ldots, \bar{v}_{12}^i) = \frac{1}{M}\sum_{j=1}^{M} \mathbf{V}_j^i,$$

and then the data $\bar{\mathbf{v}}^1$ under the 1st measuring electrode pair is put into correlation analysis with data $\bar{\mathbf{v}}^k$ collected under other measuring electrode pairs; and correlation co-efficient r is calculated

$$r_{k-1} = \mathrm{corrcoef}(\bar{\mathbf{v}}^1, \bar{\mathbf{v}}^k),$$

in which $k = 2, \ldots, 16$. By comparing correlation co-efficient r, the uniformity of 16 cup-shaped electrodes could be evaluated. Besides, well-performed measurement should have high reciprocity. In this study, we also compared reciprocity voltages by calculating their deviation in order to further evaluate the uniformity.

SNR

We obtained $\mathbf{v}_j^i$ by the same method mentioned above at first. Afterwards, the average SNR of each measuring electrode pair under 12 different driving angles can be calculated

$$\mathrm{SNR}_m = \frac{1}{12}\sum_{k=1}^{12} 20\log\frac{|E(v_k^i)|}{\sqrt{\mathrm{Var}(v_k^i)}}$$

and then the average SNR of 16 measuring electrode pairs served as the overall SNR of certain electrode system:

$$\mathrm{SNR} = \frac{1}{16}\sum_{m=1}^{16} \mathrm{SNR}_m$$

in which m stands for the number of measuring electrode pairs.

Stability

It is inevitable for medical electrodes to be interfered with by the systematic noise and the complex electromagnetic environment in clinical settings, causing the drifts of measured signals' baseline against time and it would hide the actual measurements results. Thus, we calculated the magnitude of slope for the baseline in each type of electrode to estimate its corresponding stability. The baseline is defined as a linear function obtained by fitting the average of 192 data per frame over time.

In statistical analysis, paired t test was conducted (SPSS 8.0, SPSS, Inc., Chicago, USA) to evaluate the significance of any difference for the all measures. The level of statistical significance is $p < 0.05$.

RESULTS

The process of experiment was conducted in accordance with previous setups and no volunteers showed any discomfort or side-effect. Conditions for measurements were well-maintained throughout the experiments.

As shown in Fig. 4, the normalized impedance values reduced with increasing frequency. At low frequency range (40 Hz–10 kHz), the values of electrode E and B were relatively low and values of C and A were relatively high. Within high frequency range (10 kHz–1 MHz), it was found that the impedance values were almost similar. Furthermore, the values at 50 kHz reflected that the discrepancies of contact impedance among the electrode systems were not significant ($p > 0.05$). In addition, the variance of normalized impedance values over 30 min at 50 kHz was shown in Fig. 5a, indicating that there were no significant difference ($p > 0.05$) over time between them.

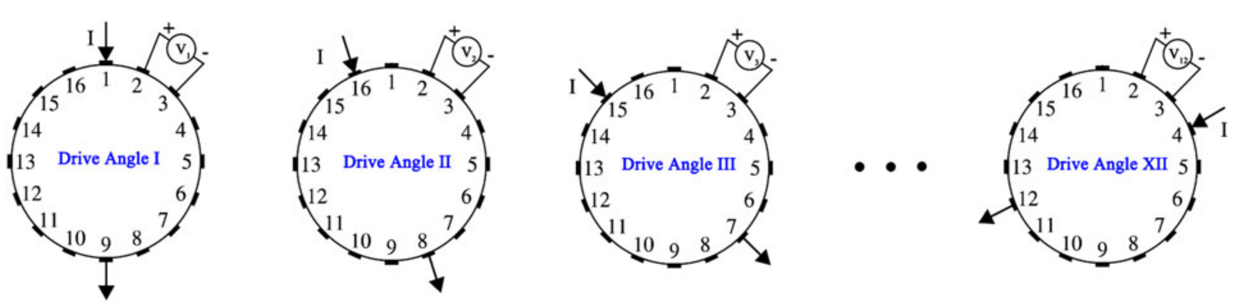

FIGURE 3. Twelve measurements under one certain measuring electrode pair. The measured data were ignored when one certain measuring electrode turned to be part of driving electrode pairs.

As for the uniformity, the correlation coefficients among 16 electrode cups in each electrode system were shown in Fig. 6. The results exhibited that correlation coefficients for electrode D had the smallest average and the greatest standard deviation, reflecting the worst uniformity ($p < 0.05$); electrode A showed relatively high average and low standard deviation, representing as good uniformity as E ($p = 0.48$). In analysis of deviation of reciprocity (Fig. 5b), electrode A also had the smallest deviation, indicating the best uniformity ($p < 0.05$) when switching electrode pairs.

In Fig. 5c, it was shown that electrode B had the lowest SNR and C had the largest variance, suggesting a worse performance in anti-nose ($p < 0.05$), whereas A and D had as great SNR as E ($p = 0.84$, $p = 0.73$).

Figure 5d compared the slopes of baseline for each electrode system. The results demonstrated that the magnitude of slope from baseline on electrode B was

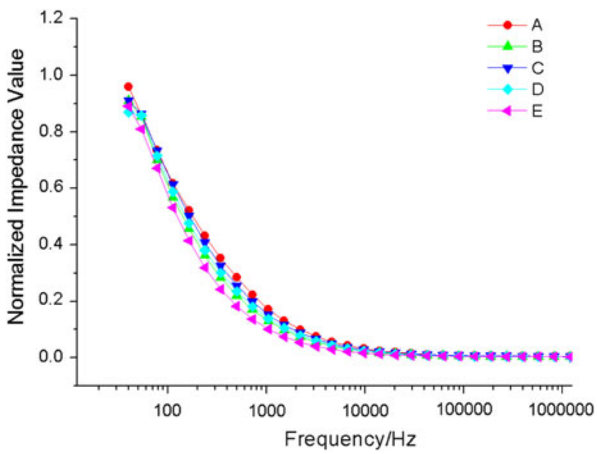

FIGURE 4. Normalized impedance values of five types of electrodes with frequency range from 40 Hz to 1 MHz by the 2-electrode method.

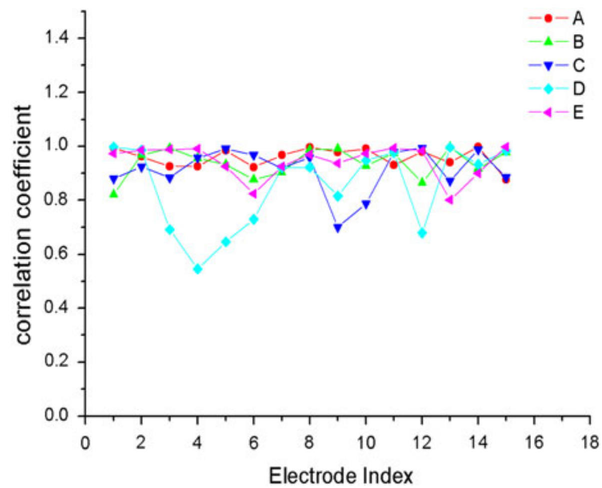

FIGURE 6. Inter-electrode correlation coefficients of each electrode system.

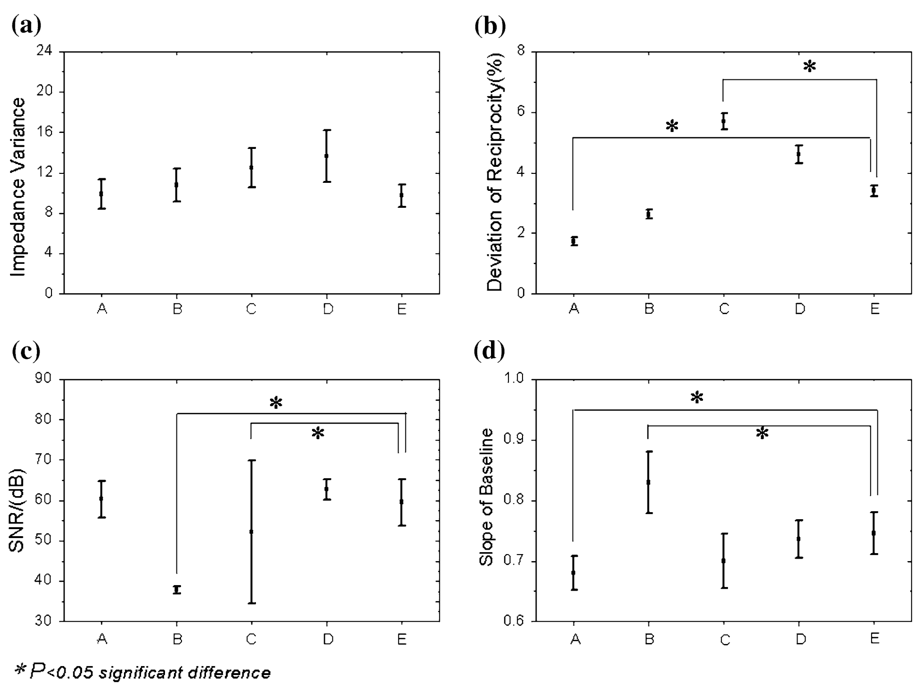

FIGURE 5. Results from EIT measurements using tested electrodes on human body. (a) The variation of impedance measured on the foream of subjects. (b) The deviation of reciprocity for each type of electrode. (c) The performance of SNR among the electrodes. (d) The slope of baseline calculated against time for the tested electrodes. Asterisk represents the significant difference at level of p<0.05.

TABLE 1. Results of five electrodes' overall performance (mean ± SD).

Type	Impedance variance	Correlation co-efficient	Deviation of reciprocity (%)	SNR (dB)	Slope of baseline
A	8.87 ± 1.43	0.95 ± 0.03	1.73 ± 0.13	60.3 ± 4.50	0.68 ± 0.03
B	10.7 ± 1.66	0.93 ± 0.05	2.62 ± 0.14	37.9 ± 0.90	0.83 ± 0.05
C	14.4 ± 1.92	0.91 ± 0.08	5.69 ± 0.27	52.1 ± 17.60	0.70 ± 0.04
D	15.6 ± 2.58	0.85 ± 0.15	4.60 ± 0.30	62.8 ± 2.50	0.74 ± 0.03
E	9.73 ± 1.11	0.94 ± 0.06	3.40 ± 0.17	59.5 ± 5.70	0.75 ± 0.04

the biggest, indicating that it had the worst stability ($p < 0.05$). In contrast, A had relatively small slope, representing relatively stable over time ($p < 0.05$).

The overall experimental results (Table 1) showed that electrode A had generally excellent performance in comparison with other types of Ag/AgCl electrodes—as low contact impedance and high SNR as electrode E, better uniformity and greater stability.

DISCUSSION

EIT has great potential in prolonged continuous monitoring of brain for patients of severe cerebral diseases. However, due to the fact that these bio-impedance changes within head always will be attenuated by the impedance presented by the cerebrospinal fluid, skull, and scalp and by partial-volume effects, it is highly desirable for cerebral EIT to have a well-performed electrode system.[16] Although Ag/AgCl electrode has been widely employed in medical application like EIT experiments, influences of different types of Ag/AgCl electrodes are not investigated before. In this study, based on the conclusion by Duan that Ag/AgCl powder electrode has an overall best performance in EEG application,[9,10] we compared five common types of Ag/AgCl bio-electrodes for cerebral EIT, by measuring the left forearm and the brain of 10 healthy adult volunteers and evaluating the performance of each type of electrodes in terms of contact impedance, uniformity, SNR, and stability.

Contact Impedance

When carrying out EIT measurements, the accurate measurements of impedance change within body are often affected difficult by the existence of significant contact impedance. In general, large contact impedances can give rise to large potential drops at the electrode–skin interface, which may hide voltage changes of the underlying tissues. The presence of reactive contact impedances also generates phase shift which can cause misleading measurements.[35] Although contact impedance declines with increasing frequency, it can still reach several hundred ohms at frequencies as high as 100 kHz. On the other hand, tissue impedance is observable between a few kHz to tens of MHz.[20] As for EIT instrumentation, large contact impedances induces large common-mode signals at the input of the measuring circuitry which has limited common-mode rejection capability in particular at high frequencies (>100 Hz).[20] Therefore, it is essential to assess contact impedance of certain electrode system so that one with low contact impedance may be determined in use of EIT. However, contact impedance is always hard to be directly measured. One alternative is the so-called 2-electrode technique, which incorporates contact impedances into the measured values,[23] so we took this indirect approach to evaluate the contact impedance of each electrode system. As a matter of fact, there are many factors affecting contact impedance, such as electrode metal, electrolyte composition and concentration, skin site, skin abrasion, temperature of conducting paste, electrode pressure, and so forth.[4] Great temperature difference between conductive paste and skin, low pressure between electrode and skin or untreated skin surface would all increase contact impedance. In order to avoid those other factors except electrode itself, we employed each electrode system as follows: the skin of volunteers was cleaned and abraded at first; conducting gel was then applied on skin before placing individual electrode; the medical adhesive elastic bandage (McDavid-4575, Bellwood, USA) was used at last to fasten the electrodes. And each electrode cup was to the same place on the skin for each type of electrode system. The measurements were performed 10 min after completing the installation of electrodes so as to maintain the same temperature of electrode–gel–skin interface. Moreover, it was also found by Rosell that repeatedly sticking and removing electrodes on skin would also have certain influences on contact impedance measurements.[28] Considering that bio-electrodes have protective effects on skin and it is not common for repeatedly sticking and removing electrodes, the influence was not taken into account in this study. Interestingly, it is shown that there was no significant difference in impedance measured at 50 kHz, indicating that there is no obvious difference in contact impedance among these electrode types. This is also proven by investigating the variation of impedance values over time. It may be related to the electrode materials. Because all these electrode systems are practically made of Ag/AgCl, the activity of the ions within them tends to be similar under the same measuring conditions.[20]

Uniformity

Given that each electrode in EIT serves as either a driving electrode or a measuring electrode, the uniformity of electrode system has an important impact on the driving/measuring procedure of EIT. For the driving circuit, one of the most important characteristics is the matching output impedance between sources as any imbalanced current that flows to ground through the input impedance of the amplifier will cause an unwanted common-mode voltage. Similarly, high input impedance for the measuring circuit is desired to prevent the decrease in current through the amplifier which would cause an unwanted voltage drop across the contact impedance and hence common-mode to differential-mode conversion. Mismatch of the input impedance, often caused by imbalance in the electrodes, leads to further degrade the common-mode rejection of the EIT system.[21] In this study, we for the first time attempted to obtain an overall evaluation in uniformity by calculating the correlation coefficient and the results show that there exist obvious differences in uniformity for these five electrode types, suggesting that electrode A has the smallest imbalance and is the most suitable for cerebral EIT measurements. Another evidence for the suggestion is that A also has the smallest deviation of reciprocity voltages. The reasons for these differences in uniformity may include the inhomogeneous distribution of Ag/AgCl films, unstable temperature for sintering electrodes and incomplete encapsulation of epoxide resin leading to insufficient oxidation of silver base in certain parts. It means that, because of the errors in manufacturing process, 16 electrode cups may not be completely consistent with respect to shape, weight, topological structure, and electrochemical characteristics, consequently resulting in the imbalance in electrode system. Hence, it is very important to demand a rigid manufacturing process for EIT electrodes.

SNR

In clinical practice, the EIT system would be affected by the presence of complicated electromagnetic environment, such as defects in electric circuit installation, interference noise from the circuit boards and other sources, and stray capacitances from electrical cables.[25] They would consequently pollute the measured data and influence their precision. Therefore, an EIT electrode system with high SNR is preferable. In this paper, the SNR was calculated after long-period monitoring of volunteers' head with EIT. According to the results, it demonstrates that there are substantial differences in SNR for these five electrode types, of which electrode A has relatively high and stable SNR for cerebral EIT measurements. In the experiment, it is hard to distinguish various sources of noises or interferences for calculation. Therefore, the SNR here is used to evaluate the overall anti-noise feature. Besides the electrodes itself, the causes of affecting the SNR probably may also arise from magnitude of signals or systematic errors.

Stability

As cerebral EIT normally requires a long-period of monitoring, the drifts are parts of EIT system errors which are inevitable. Therefore, we expect the drifts as small as possible.[21] If the baseline of the measured data in the process of monitoring drifted dramatically, the accuracy of clinical diagnosis will be undermined. In the paper, we compared the stability of each electrode system by calculating the slopes of the baselines based on the measured data. The electrode type with a greater slope tends to be more unstable. It is found that electrode A has relatively low slope and hence high stability, indicating that it may be appropriate for prolonged EIT measurements. Generally, drifts may be reduced by leaving applied electrodes for 10 min to allow gel to reach body temperature and for sweat ducts to stabilize, as we did in the study.[4,28]

Before installing an EIT electrode system on body for measurements, conducting gel is often applied to ensure a good contact of electrode with skin and to decrease the high epidermal impedance.[20,23] However, the safety issue of conducting gel should also be taken into account, especially in case of long-term EIT monitoring for the neonatal patients, as certain conducting gel is too aggressive to employ, giving rise to skin irritation or impairment. In this study, no candidates showed any discomfort or side-effect during the period of experiments.

One of limitation in this study is that we have to place each electrode one by one on the volunteers. This is absolutely time-consuming and not applicable for clinical practice. Therefore, our future work should include developing a package of electrode supporting components with high convenience and reliability for cerebral EIT, such as the electrode belt or the electrode cap which is easy to install. On the other hand, the variation of current waveform across frequencies may influence the measurements of the human head and hence electrode performance,[27] which is not considered here. The optimum current pattern may be specific to one certain emergency of the human head (acute stroke, hemorrhage, etc.) and the further investigation is required.

CONCLUSION

For the requirements of electrode system in cerebral EIT, five common types of Ag/AgCl bio-electrodes have been compared in term of contact impedance, uniformity, SNR, and stability, based on the measured impedance data for the humans by the 2-electrode method and the 16-electrode, respectively. Results show that electrode A which is Ag/AgCl powder electrode has low contact impedance, high SNR, better uniformity, and greater stability, suggesting that it is an outstanding candidate for cerebral EIT. Further studies may be required in developing the specific conducting gel to improve its electrical properties and designing a reasonable peripheral supporting structure to improve its convenience and reliability of use in clinical practice. After these improvements, Ag/AgCl powder electrode is likely to become the optimal choice for further research or application in cerebral EIT.

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REFERENCES

[1]Barber, D. C. Electrical impedance tomography. In: The Biomedical Engineering Handbook (2nd ed.), edited by D. C. Barber. Boca Raton: CRC Press, 2000, pp. 1151–1164.
[2]Bayford, R. Bio-impedance tomography (electrical impedance tomography). *Annu. Rev. Biomed. Eng*. 8:63–91, 2006.
[3]Binnie, C., A. Rowan, and T. Gutter. A Manual of Electroencephalographic Technology. Cambridge: Cambridge University Press, 1982.
[4]Boone, K., and D. S. Holder. Assessment of noise and drift artefacts in electrical impedance tomography measurements using the Sheffield Mark-I system. *Innov. Technol. Biol. Med*. 16:49–60, 1995.
[5]Boone, K. G., and D. S. Holder. Effect of skin impedance on image quality and variability in electrical impedance tomography: a model study. *Med. Biol. Eng. Comput*. 34:351–354, 1996.
[6]Brown, B., D. Barber, and A. Seagar. Applied potential tomography: possible clinical application. *Clin. Phys. Physiol. Meas*. 6:109–121, 1989.
[7]Cherepenin, V., A. Karpov, A. Korjenevsky, V. Kornienko, Y. Kultiasov, A. Mazaletskaya, and D. Mazourov. Preliminary static EIT images of the thorax in health and disease. *Physiol. Meas*. 23:33–41, 2002.
[8]Dai, M., L. Wang, C. Xu, L. Li, G. Gao, and X. Dong. Real-time imaging of subarachnoid hemorrhage in piglets with electrical impedance tomography. *Physiol. Meas*. 31:1229–1239, 2010.
[9]Duan, Y., G. M. Clark, and R. Cowan. Factors determining and limiting the impedance behavior of implanted bio-electrodes. In: Proceedings of SPIE-The International Society for Optical Engineering, 2001, pp. 498–508.
[10]Duan, Y., G. M. Clark, and R. Cowan. A study of intracochlear electrodes and tissue interface by electrochemical impedance methods in vivo. *Biomaterials* 25:3813–3828, 2004.
[11]Fabrizi, L., A. McEwan, D. Oh, and D. S. Holder. Analysis of resting noise characteristics of three EIT systems in order to optimize time difference imaging with scalp electrodes during epileptic seizures. *Proc. IFMBE* 17:568–571, 2007.
[12]Fabrizi, L., R. Yerworth, A. McEwan, O. Gilad, R. Bayford, and D. S. Holder. A method for removing artefacts from continuous EEG recordings during functional electrical impedance tomography for the detection of epileptic seizures. *Physiol. Meas*. 31:57–72, 2010.
[13]Gagnon, H., M. Cousineau, A. Adler, and A. Hartinger. A resistive mesh phantom for assessing the performance of EIT systems. *IEEE Trans. Biomed. Eng*. 57:2257–2266, 2010.
[14]Hahn, G., J. Dittmar, A. Just, M. Quintel, and G. Hellige. Different approaches for quantifying ventilation distribution and lung tissue properties by functional EIT. *Physiol. Meas*. 31:S73–S84, 2010.
[15]Holder, D. S., A. Rao, and Y. Hanquan. Imaging of physiologically evoked responses by electrical impedance tomography with cortical electrodes in the anaesthetized rabbit. *Physiol. Meas*. 4A:A179–A186, 1996.
[16]Horesh, L., S. R. Arridge, and D. S. Holder. Some novel approaches large scale algorithms for multi-frequency electrical impedance tomography of the human head. PhD Thesis University College London. 2006.
[17]Kuen, J., E. J. Woo, and J. K. Seo. Multi-frequency time difference complex conductivity imaging of canine and human lungs using the KHU Mark1 EIT system. *Physiol. Meas*. 30:149–164, 2009.
[18]Marquez, J. C., F. Seoane, E. Välimäki, and K. Lindecrantz. Textile electrodes in electrical bioimpedance measurements: a comparison with conventional Ag/AgCl electrodes. *Conf. Proc. IEEE Eng. Med. Biol. Soc*. 48:16–19, 2009.
[19]Mayotte, M., J. Webster, and W. Tompkins. A comparison of electrodes for potential use in paediatric/infant apnoea monitoring. *Physiol. Meas*. 15:459–467, 1994.
[20]McAdams, E. T., J. Jossinet, A. Lackermeier, and F. Risacher. Factors affecting electrode–gel–skin interface impedance in electrical impedance tomography. *Med. Biol. Eng. Comput*. 34:397–408, 1996.
[21]McEwan, A., G. Cusick, and D. S. Holder. A review of errors in multi-frequency EIT instrumentation. *Physiol. Meas*. 28:197–215, 2007.

[22]Medrano, G., A. Ub, N. Zimmermann, T. Gries, and S. Leonhardt. Skin electrode impedance of textile electrodes for bio-impedance spectroscopy. *Proc. IFMBE* 17:260–263, 2007.
[23]Mohamad, R., M. K. Joo, D. Andreas, T. Andrew, and B. Richard. A comparison study of electrodes for neonate electrical impedance tomography. *Physiol. Meas.* 30:73–84, 2009.
[24]Prasad, S. N., D. Houserkova, and J. Campbell. Breast imaging using 3D electrical impedence tomography. *Biomed. Pap. Med. Fac. Univ. Palacky. Olomouc. Czech. Repub.* 152:151–154, 2008.
[25]Puurtinen, M. M., S. M. Komulainen, P. K. Kauppien, J. A. Malmivuo, and J. A. Hyttinen. Measurements of noise and impedance of dry and wet textile electrodes and textile electrodes with hydrogel. *Conf. Proc. IEEE Eng. Med. Biol. Soc.* 1:6012–6015, 2006.
[26]Romsauerova, A., A. McEwan, L. Fabrizi, and D. S. Holder. Evaluation of the performance of the multifrequency electrical impedance tomography (MFEIT) intended for imaging acute stroke. *Proc. IFMBE* 17:543–547, 2007.
[27]Romsauerova, A., A. McEwan, and D. S. Holder. Identification of a suitable current waveform for acute stroke imaging. *Physiol. Meas.* 27:211–219, 2006.
[28]Rosell, J., J. Colominas, P. Riu, R. Pallas-Areny, and J. G. Wester. Skin Impedance from 1 Hz to 1 MHz. *IEEE Trans. Biomed. Eng.* 35:649–651, 1988.
[29]Searle, A., and L. Kirkup. A direct comparison of wet, dry and insulating bioelectric recording electrodes. *Physiol. Meas.* 21:271–283, 2000.
[30]Shi, X., and X. Dong. Pseudo-polar drive patterns for brain electrical impedance tomography. *Physiol. Meas.* 27:1071–1080, 2006.
[31]Shi, X., X. Dong, F. You, F. Fu, and R. Liu. High precision multi-frequency electrical impedance tomography system and preliminary imaging results on saline tank. In: IEEE-EMBS 27th Annual International Conference the IEEE, 2005, pp. 1492–1495.
[32]Shi, X., X. Dong, F. You, F. Fu, R. Liu, and M. Dai. Preliminary research on monitoring of cerebral ischemia using electrical impedance tomography technique. *Conf. Proc. IEEE Eng. Med. Biol. Soc.* 2008:1188–1191, 2008.
[33]Shuai, W., F. You, H. Zhang, W. Zhang, F. Fu, X. Shi, R. Liu, T. Bao, and X. Dong. Application of electrical impedance tomography for continuous monitoring of retroperitoneal bleeding after blunt trauma. *Ann. Biomed. Eng.* 37:2373–2379, 2009.
[34]Shuai, W., F. You, H. Zhang, W. Zhang, F. Fu, X. Shi, R. Liu, C. Xu, X. Dong, and T. Bao. Image monitoring of an intraperitoneal bleeding model of pigs using electrical impedance tomography. *Physiol. Meas.* 29:217–225, 2008.
[35]Taktak, A., P. Record, R. Gadd, and P. Rolfe. Practical factors in neonatal lung imaging using electrical impedance tomography. *Med. Biol. Eng. Comput.* 33:202–205, 1995.
[36]Tidswell, A., P. Bagshaw, and D. S. Holder. A comparison of headnet electrode arrays for electrical impedance tomography of the human head. *Physiol. Meas.* 24:527–544, 2003.
[37]Tidswell, A., A. Gibson, R. H. Bayford, and D. S. Holder. Validation of a 3D reconstruction algorithm for EIT of human brain function in a realistic head-shaped tank. *Physiol. Meas.* 22:177–185, 2001.
[38]Tidswell, A., A. Gibson, R. H. Bayford, and D. S. Holder. Three-dimensional electrical impedance tomography of human brain activity. *Neuroimage* 13:283–294, 2001.
[39]Wang, P., M. Tang, M. McCormick, and X. Dong. Preliminary results from an EIT breast imaging simulation system. *Physiol. Meas.* 22:39–48, 2001.
[40]Woo, E., P. Hua, J. Webster, W. Tompkins, and R. PallasAreny. Skin impedance measurements using simple and compound electrodes. *Med. Biol. Eng. Comput.* 30:97–102, 1992.
[41]Xu, C., X. Dong, F. Fu, W. Shuai, X. Liu, and C. Zhang. A novel image monitoring software system of electrical impedance tomography for internal hemorrhage. In: World Congress on Medical Physics and Biomedical Engineering, 2006, pp. 3882–3885.
[42]Xu, C., L. Wang, X. Shi, F. You, F. Fu, R. Liu, M. Dai, Z. Zhao, G. Gao, and X. Dong. Real-time imaging and detection of intracranial haemorrhage by electrical impedance tomography in a piglet model. *J. Int. Med. Res.* 38:1596–1604, 2010.
[43]Yin, Y., Z. Ji, W. Zhang, N. Wang, F. Fu, R. Liu, F. You, X. Shi, and X. Dong. Comparison of three kinds of electrode–skin interface for electrical impedance scanning. *Ann. Biomed. Eng.* 38:2032–2039, 2010.
[44]Zhang, J., and R. P. Patterson. Analysis on the influence of tissues/organs' movements in EIT images of lung ventilation using finite difference thorax models. *Conf. Proc. IEEE Eng. Med. Biol. Soc.* 2:1557–1560, 2004.