Analysis of resting noise characteristics of three EIT systems in order to compare suitability for time difference imaging with scalp electrodes during epilepsy

L. Fabrizi[1], A. McEwan[1], D. Oh[1] and D.S. Holder[1]

[1] Department of Medical Physics and Bioengineering, UCL, London, UK

Abstract— **We have investigated the noise in three systems suitable for imaging epileptic seizures, the UCH Mark1b, UCH Mark2.5 and KHU Mark1 16 channel systems, at applied frequencies in three steps from 1 to 100 kHz, by varying load impedance, single terminal or multiplexed measurements, and in test objects of increasing complexity from a resistor to a saline filled tank and scalp electrodes in a human volunteer. The noise was white, and increased from about 0.03% on the resistor to 0.08% on the human; it increased with load but was independent of use of the multiplexer. The KHU Mark1 system delivered the best performance with noise spectra of about 0.02%, which could be further reduced by averaging to a level where reliable imaging of changes of about 0.1% estimated during epileptic seizures appears plausible. This appears to be because the KHU Mark1 records in parallel and employs screened triaxial cables, unlike the other two systems. Work in progress is to evaluate the KHU Mark1 with longer 5m leads which are needed for clinical studies in ambulant patients and optimisation of current levels in the UCH systems. Following this, the two best systems will be evaluated in clinical studies in patients during epileptic seizures.**

Keywords— **Noise characterization, EIT system performance, UCH Mark1b, UCH Mark2.5, KHU Mark1.**

I. Introduction

In practice, clinical images acquired with biomedical Electrical Impedance Tomography appear noisy and usually suffer from limited spatial resolution. This is due to the underdetermined nature of the image reconstruction, systematic instrumentation errors, and random noise. The bottleneck is not usually clear and will depend on the practical circumstances of the clinical application. Our group at University College London has been interested in imaging brain function in general, and, specifically, electrical impedance changes during epileptic seizures. Conductivity locally in the brain changes during seizures by about 3-12% in the frequency range 1-50kHz [1][2], probably due to movement of extracellular water into the intracellular space. Such changes may be modelled to translate into very small scalp voltage changes of the order of 0.1% due to partial volume and volume conduction effects. In a preliminary clinical study of EIT recorded during seizures in 7 subjects, no reproducible images could be obtained, and this appeared to be due to a poor signal-to-noise ratio [3]. It appears likely that the bottleneck is random noise which may obscure the small boundary voltage changes of about 0.1%, which correspond to changes of about 10μV on a standing potential of about 1 mV over tens of seconds. A feasibility study showed that images of epileptic conductivity changes in the temporal lobe could be reconstructed when the random noise level was 0.04%, while, in deeper regions, a noise level of 0.002% was required [4].

Noise sources may be divided into: 1) thermal noise; 2) quantization noise; 3) electronic noise of the amplifiers; 4) external electromagnetic interference; 5) voltage fluctuations of the electrode interface; 6) physiological noise. Only the total effect of the first 3 causes is defined and can be estimate to be 0.38 μV in RMS after demodulation.

In the authors' groups at University College London and Kyung Hee University, Seoul, we have developed three EIT systems potentially suitable for imaging these small changes: the UCH Mark1b, UCH Mark2.5, KHU Mark1. This work is an examination of the characteristics of baseline noise, with the intention that this could suggest the system likely to yield the best signal quality, and any modifications which could be employed to optimise the signal-to-noise ratio. Where possible, we attempted to divide the noise into its components due to electronics and thermal noise, to metal-ion interaction and environment and to ion-skin interaction and physiological processes, by recording in resistors, saline filled tanks and human subjects. We also explored the noise dependence on measuring frequency and on load impedance and the difference between 4-terminal and multiplexed measurements.

The measurements were collected at the maximal acquisition rate and current injection allowed by each frequency, safety standard and system performance. Three representative frequencies were selected for each system. The range was limited downward by the acquisition speed of the UCH Mark2.5 at about 1 kHz and upward by the maximal operating frequency of the UCH Mark1b at about 100 kHz. The UCH Mark2.5 and KHU Mark1 could function at higher frequencies, but the sampling rate limited the acquisition and the clinical signal may be expected to decrease [4], so the frequencies recorded were between 1kHz and 100 kHz.

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In order to ascertain the sources of noise, experiments of increasing complexity were performed. The simplest measurement comprised single channel 4-terminal measurements on resistors of increasing values. It was used to examine the effect of increasing load over the whole range of transfer impedances likely to be encountered in different subjects in the human head of 8-70 Ω [5].This characterized the noise due to the electronics and the influence of increasing load on the thermal noise of a purely resistive object. This was then repeated on a resistor network with use of the multiplexer in order to assess if additional noise was introduced by the series resistance and stray capacitance of the multiplexer. Similar tests were conducted on a saline phantom to assess the noise derived from the metal-electrolyte interaction and from electromagnetic and mechanical effects, such as saline movement and temperature fluctuations. Finally, recordings were made in a resting human subject with scalp electrodes in order to determine the effects of the electrode-skin contact impedance variations and physiological noise, such as autonomic activity of muscle and blood vessels and body temperature variations.

II. Methods

A. Recording arrangement

Systems settings. Each system was connected to the test objects with cables which were 30 cm-long unscreened cables for the Mark 1b, 55cm-long 37-way unscreened ribbon cable for the Mark 2.5 and 1.2m-long triaxial cable with driven and shielded screens for the KHU Mark1. Each acquisition was executed at three frequencies, representing the low, medium and high frequency ranges, that were chosen to be as similar as possible between the 3 systems to allow comparison (Table 1). All measurements except for the single electrode combination on one resistor were made in a ring with 16 electrodes. Single channel measurements were made with current injected into electrodes 4 and 12 and recording from 13-14, 14-15 or 15-16, so that three different transfer impedances were addressed. The same set-up was used for all the systems; therefore the same loads were measured. Multiplexed measurements were made with a protocol with which current was injected through diametrically opposed electrodes in turn and voltage recordings were made serially from remaining adjacent pairs in each case. Noise was calculated from 1024 consecutive measurements from each of the three above combinations in turn; the remaining data were discarded.

Resistor. Single channel recording was on 3 loads (14, 44 and 77 Ω), with 1 kΩ as contact impedance. Multiplexed recording was on a symmetrical resistor network with 16

Table 1 Settings of the systems for single and multiple channel measurement

System	Frequency (kHz)	Acquisition rate (ms/meas)		Current (mA)
		Single ch.	Mult. ch.	
UCH Mark 1b	1.2	32	1250	0.14
	9.6	23	246	1
	65.5	21	123	2.6
UCH Mark 2.5	2	22.5	2200	0.138
	10.08	22.5	2200	0.138
	101.6	22.5	2200	0.138
KHU Mark 1 16 ch.	1	65	118	0.14
	10	60	84	1
	100	60	84	1.12

terminals (300 Ω as spokes and contact resistances and 75 Ω as rim resistances), the transfer resistances from the 3 selected channels were 18, 13 and 10 Ω.

Saline tank. A cylindrical tank (9cm diameter, 5cm deep) with 16 equally spaced, recessed stainless steel electrodes, 5mm in diameter, in a single ring, was filled with 200 ml of 0.3% NaCl solution at room temperature of 20 °C. The ratio of recorded voltages for the three channels selected to represent different loads at the middle carrier frequency was 1, 1.18 and 2.05.

Human. A single ring of 16 Ag-AgCl standard EEG cup electrodes of 10 mm diameter was placed just above the level of the ears of a healthy 27 year-old male volunteer. The ratio of recorded voltages for the three channels selected to represent different loads at the middle carrier frequency was 1, 1.24 and 5.46.

B. Data analysis

1024 boundary voltage measurements were collected during each recording, irrespective of the time taken. These were divided into 4 segments of 256 each and multiplied by a Hanning window. The DC component was subtracted from each segment and four 256-points spectra of the amplitude of the real part of the voltages (R) were analyzed with FFT (MATLAB function) and then averaged together. The spectra appeared to be white but there were occasional outlying values, presumably due to instrumentation non-idealities. For ease of analysis, each spectrum was reduced to the mean noise frequency content after exclusion of data from any outlying frequency with noise content 2 standard deviations (SD) greater than the mean. This was justified on the basis that those values were above 7 Hz, while impedance changes in epilepsy do not exceed 2 Hz. These were then assessed for statistical significance by five way analysis of variance using the statistical Toolbox of Matlab v 6.5 (www.mathworks.com); the variables were experimental preparation, carrier frequency, load, multiplexer presence and EIT system.

III. Results

Difference across variables. There was a significant difference ($p<0.05$) between all the variables except for the presence of the multiplexer for proportional noise.

Variation with frequency. The noise appeared to be largely uniform across frequency.

Carrier frequency. Overall, noise was least in the middle carrier frequency band and greatest in the lower band. Mean noise was greater in the low band (1 – 2 kHz) by 251, 71 or 771 % compared to the medium band (9.6 – 10.1 kHz) for the Mk1b, Mk2.5 and KHU Mark1 respectively. Noise in the higher band (65 – 101 kHz) was significantly greater than in the medium band only for the Mk2.5 system – by 121%.

Test object complexity. Noise increased with recording on the saline filled tank and human by 27±7 and 167±47% compared to the resistor measurements.

Load. Overall, the noise increased with load. When the load was increased 5.5 times in the single channel experiment on the resistor, the mean noise increased by 263%. The dependence on the load was more evident for the UCH Mark1b and UCH Mark2.5 than for the KHU Mark1; when the load was increased 5.5 times the mean noise increased 326%, 242% and 12% respectively.

Presence of multiplexer. The noise did not increase significantly with multiplexing.

EIT system. The KHU Mark1 had the lowest noise of the three systems. The mean proportional noise of the UCH Mark2.5 was 70% and 216% higher than that of the UCH Mark1b and KHU Mark1. It also had the least absolute noise. The mean absolute noise of the UCH Mark1b was 162% and 313% higher that that of the UCH Mark2.5 and KHU Mark1.

IV. Discussion

The noise for all the three systems was generally uniform over frequency. It was lower at medium frequencies (9.6 – 10.08 kHz) and increased with experimental complexity. It increased variably with load and was more proportional for the UCH Mk1b than for the KHU Mark1. Multiplexing did not increase the noise significantly; the KHU Mark1 had the lowest noise overall (0.02%, 0.8 μV).

A. Technical issues

The current amplitude injected at each frequency was limited by safety standards and by patient perception. At low frequencies, current injected by all systems was about 140 μA. Above 100 kHz, 5 mA could be injected according to safety limits, but none of the systems had this capability, so the current was the highest attainable by the instrumentation. The current that could be injected with the UCH Mark2.5 was limited by the injection of a composite waveform. As a result, the peak current injected was 0.14 mA compared to 2.6 and 1.12 mA for the UCH Mark 1b and KHU Mark1. The UCH Mark1b has an automatic gain controller, which may be an advantage if the noise arises after the amplifier stage. However, the number of gain levels is finite and hence may also introduce quantization effects and uncertainty due to the discrete components used to set the gain level.

In order to keep the experiment as similar as possible between systems and practically feasible, we had to use 30cm-long extensions of unscreened wires to test the KHU Mark1 on the resistor network and electrodes on human had 5cm-long unscreened wires, on which the system cables were clipped. This may have introduced extra stray capacitance.

All measurements except for the single electrode combination on one resistor were made in a ring of 16 electrodes and the transfer impedances with and without multiplexing were equivalent and changed only by selecting different combinations. The single channel measurements on the resistor were done using 3 physically different loads of 14, 44 and 77 Ω, while the transfer impedance when multiplexing was of 10, 13 and 18 Ω so the load multiples were not directly comparable. However, this appears not to have affected the conclusion that there was not a significant effect of multiplexing on the noise.

B. Possible explanation for the observed noise

The estimated noise due to thermal noise, quantization noise and amplifier noise is about 0.54 μV when integrating the noise at all frequencies. The noise measured from the KHU Mark1 was white and of mean amplitude of 0.8 μV at each frequency sample, which has a bandwidth of 0.046 Hz. If we consider 6 Hz as the entire signal bandwidth, the overall noise is 9 μV, 20 times larger than the estimated noise. It therefore appears that thermal, quantization and amplifier noise are too small to explain the observed noise.

The sources of noise are not clear; it is probably due to a combination of other factors, such as electromagnetic interferences, stray capacitance and noise deriving from other system components, such as the current source. Nevertheless, it is possible to analyze the noise empirically in relation to the different measurement variables. The increase in noise with test object complexity was as expected and is probably due to voltage fluctuations at the electrodes, present in the saline filled tanks and human subjects, and the additional noise in the latter because of physiological variability. On top of a component independent from the signal, the increase in noise with load suggests the presence of a multiplicative contribution, such as noise coming from the current drive or transimpedance fluctuations. This and reduction with recording in the middle carrier bands suggests that clinical recording should take place in the middle frequency bands for which the systems are optimised, and electrode impedance should be reduced as much as possible. Unexpectedly, multiplexing did not introduce any additional noise; so, serial recording does not appear to introduce a disadvantage in this respect, although parallel recording would permit more averaging.

C. Implications for clinical experiment

Construction and optimization of these systems has taken many man years, and the purpose of this study was to determine if it appeared that a radical re-design was needed, or whether these noise levels appeared compatible to EIT imaging of the small boundary voltage changes expected during seizures. Overall, the best noise was observed with the KHU Mark1, and was about 0.02% at each frequency and 0.16% (RMS) in total. A precise analysis of whether this might be expected to yield robust clinical images is outside the scope of this study, but the largest boundary voltage changes predicted during seizures were about 1% for peripheral lateral temporal lobe seizures, and 0.1% for deeper mesial temporal ones [4].Although these figures represent just the largest changes, there will be an averaging effect in image production, as signals from many channels will be used in reconstruction. Overall, the noise is therefore of the same or one order of magnitude less than the expected signal, and it seems plausible that it might be possible to reconstruct reliable images with the existing systems.

Without undertaking a major system re-design, the next issue is whether these findings suggest any simple modifications to the recording protocol which might improve the signal to noise ratio. The UCH Mark1b would benefit from signal post-processing tools like time averaging and band-pass filtering, which could reduce the noise by about a factor of 2 as it collects about 4 frames per second and the bandwidth of changes during epileptic seizures is near DC to 1 Hz. At the middle frequencies, this would reduce noise to about 0.02%. For the UCH Mark2.5, use of screened cables may reduce the noise at 101 kHz to that at 10.08 kHz, from 0.11% to 0.05%. Injection of current at a single frequency would allow injection of 750 μA peak-to-peak current at 10 and 101 kHz. This could reduce average noise from 0.05% to 0.01%. Noise of the KHU Mark1 could be reduced by $\sqrt{12} \approx 70\%$ by averaging the 12 frames it can collect each second, as it has parallel recording. Overall, therefore, the KHU Mark1 has the lowest noise but it suffers from the practical disadvantage that all the leads will need to be connected to the subject's head. As the next lowest noise may be expected to be obtained after averaging with the UCH Mark 1b system, which has been designed for ambulant recording with a small headbox which can be worn by the patient. We therefore plan to compare recording with both these systems during epileptic seizures.

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Address of the corresponding author:

Author: Lorenzo Fabrizi
Institute: University College London
Street: Gower Street
City: London
Country: UK
Email: l.fabrizi@ucl.ac.uk