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Effect of electrode contact impedance mismatch on 4-electrode measurements of small body segments using commercial BIA devices

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***Abstract* – Segmental and regional bioimpedance measurements have become of great clinical interest in the monitoring of many pathologies. However, the use of commercial bioimpedance analyzers to carry out measurements on small segments of the body, with their associated low inter-electrode tissue impedances, leads to some unexpected problems. Low inter-electrode tissue impedances are very prone to adverse effects resulting from the relatively large electrode contact impedances and, especially, from electrode contact impedance mismatches. The authors highlight these problems, first with measurements on electrical models of the tissue and electrode/skin contact impedances and then with measurements carried out on human calves. It is concluded that commercial equipment must be used carefully, especially when carrying out novel, localised bioimpedance measurements for which the devices were not originally designed.**

I. INTRODUCTION

BioImpedance Analysis (BIA) has been used as a monitoring tool in many medical applications [1] and traditionally has involved whole-body measurements. However, more recently there has been a growing interest in more localised segmental or regional measurements [2]. In the present study the authors are especially interested in measurements on the human calf [3] to assess patients undergoing hemodialysis therapy

Measuring segmental/regional BIA has not only the same problems encountered in whole-body studies but has additional problems associated with electrode design and placement.

Generally, standard ECG "spot" electrodes are used to perform BIA measurements, and they can be made of wet gel or hydrogel which have very different properties. As there is no standardization on the electrodes to be used, there has been a wide variety of electrodes used of differing type, size and material composition. These variations unfortunately affect the results. Band electrodes should arguably be the most appropriate for BIA as they provide better current distribution over tissue and lower electrode/skin contact impedance than spot electrodes due to their larger areas [4].

The other problem regarding segmental measurements in particular is the lower tissue impedances involved compared to whole-body measurements. In this case the relatively large electrode/skin contact impedance has more effect and can distort the measured tissue impedance. Contact impedances are not constant and vary between patients, the body site measured, electrode design, time, temperature etc. and therefore need to be assessed [5]. The effects of contact impedance on BIA measurements using commercial devices have been studied by a few authors.

For example, Bolton et al [6], when carrying out measurements on patients and on electrical models, found significant differences in the measured "tissue" impedance due to contact impedance and lead capacitance.

The effects of electrode mismatch and ground coupling were studied by Bogonez-Franco et al [7]. The authors noted that electrode contact impedance mismatch has a larger effect on the tissue impedance values than ground coupling. Electrode contact impedance mismatch on the voltage detecting electrodes has been reported to cause significant changes in the tissue impedance measured [8].

This paper will demonstrate the effect of electrode-skin contact impedance during the use of commercial impedance devices for segmental/regional measurements.

II. MATERIALS AND METHODS

The sensitivities of commercial BIA devices to contact impedance were assessed first of all using equivalent electrical models of the tissue and electrode/skin contact impedances and then with measurements carried out on human calves.

A. Electrode/skin contact impedance measurement

Electrode/skin contact impedance was measured with a

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Solartron 1255 and a biological impedance interface 1294A (Solartron Analytical Ltd. Hampshire, England) using a three-electrode configuration on the calf of a healthy person. In the three-electrode technique current is applied through the electrode-skin contact of interest and an indifferent electrode located at some remote site. A reference electrode is applied to the skin close to this "Measured" electrode-skin site. The voltage between the Measured electrode and the nearby Reference electrode is measured using a very high input impedance voltmeter. In this case, current cannot flow through the Reference electrode-skin interface, or through the intervening dermal tissue. No voltage is therefore dropped across these impedances and hence the measured voltage dropped between the Measuring electrode and the nearby Reference electrode is due to solely to the applied current flowing through the impedance of the Measurement Electrode-Skin interface. Knowing the applied current and the measured voltage, one can readily deduce the impedance of the Measurement Electrode-Skin interface.

The frequency range used in characterising the electrode/skin contact impedance was from 1 Hz to 10 kHz, as in this range the electrode contact impedance is generally most significant. Current injected was 100 μA_{RMS}. Electrodes used were 3M 2660 (3M, Minneapolis, USA), Impedimed Tab electrodes (Impedimed Ltd, Brisbane, Australia) and IMMED 2350 (Immed Europe, Ennevelin, France), all are hydrogel electrodes. No skin preparation was performed before the placement of the electrodes.

B. Tissue impedance measurement

Tissue impedance was measured using the four-electrode configuration. The frequency range of measurement was from 1 kHz to 1 MHz, as is typically used. The current injected and the electrodes used were the same as those described above.

C. Commercial impedance devices

Two commercial impedance devices were assessed; BioparHom Z-Métrix (BioparHom, Bourget du Lac, France) and Impedimed SFB7 (Impedimed Ltd., Brisbane, Australia). Data was acquired using the software provided by the respective manufacturers. Calibration of both devices were checked before measurements were performed.

D. Electrode/skin contact impedance electrical model

Equivalent electrical circuit models of electrode/skin contact impedances were built based on our previous measurements. The individual contact electrical model consisted of a resistor in series with the parallel combination of a capacitance and a resistor. Table 1 shows the component values for the equivalent electrical model.

Table 1. Component values for the equivalent electrical model.

ELECTRODE	RS (Ω)	Rp (Ω)	C (nF)
3M 2660	2,200	150,000	150
IMPEDIMED	1,495	73,881	107
IMMED 2350	759	81,189	86

Fig. 1 shows the equivalent electrical model and the parameter values used. Resistor and capacitor values were 1% and 5% in tolerance, respectively.

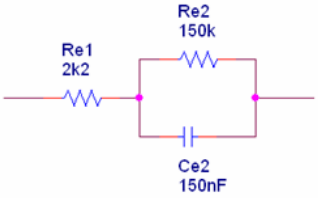

Fig. 1. Equivalent electrical model for electrode/skin contact impedance for 3M 2660 electrode.

E. Tissue electrical model

The equivalent electrical model of the tissue impedance has the same structure as the equivalent circuit for electrode/skin contact impedance. Fig. 2 shows the tissue equivalent circuit and the parameter values used.

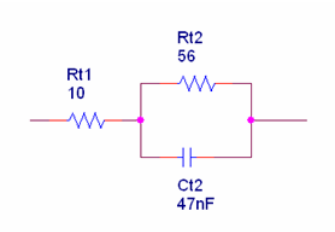

Fig. 2. Equivalent circuit model for tissue impedance.

F. Contact impedance-induced measurement errors

Absolute errors were calculated between the actual equivalent circuit parameter values used for the "tissue" impedance and those measured by the devices in the presence of mismatched contact impedance circuits.

G. Measurements on healthy people

In order to check the effect of electrode contact impedance mismatch, we also carried out measurements on the human calf. Electrode mismatch was produced by connecting two electrodes together at a given contact to cause a theoretical reduction in electrode/skin contact

impedance of 50%. Fig. 3 shows the four-electrode configuration and the possible pairing of contact electrodes to give rise to electrode impedance mismatch.

III. EXPERIMENTAL RESULTS

A. Effect of contact impedance

Using the equivalent electrical circuits described above, the effect of contact impedance on measured "tissue" (model) impedance was evaluated. Over the frequency range of interest, 1 kHz to 1 MHz, the contact impedance is dominated mainly by resistor Re1 of Fig. 1. Resistor Re1 was therefore varied in the tests from 0.5 kΩ to 22 kΩ for each lead and absolute errors were calculated with respect to actual values of the "tissue" circuit of Fig. 2.

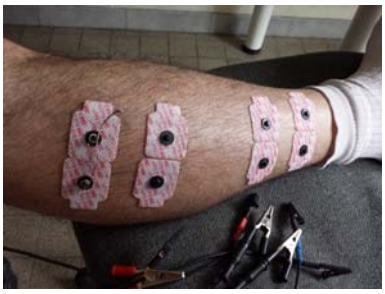

Fig. 3. Four-electrode configuration with electrode pairs used to provoke electrode mismatch.

Both devices were relatively unaffected by the presence of large contact impedances at both of the detection leads (see effects of impedance mismatch below). However, they were both affected in a similar manner by the presence of large contact impedances at the injecting leads, the BioparHom device being more sensitive to changes in the applied value of Re1. The Impedimed device was able to tolerate impedances as low as 10 kΩ while the limit for the BioparHom device was 4.7 kΩ.

B. Effect of electrode impedance mismatch

The effect of electrode/skin contact impedance mismatch was modelled by connecting the circuit of Fig. 1 to one of the four leads at a time while the remaining leads were connected directly to circuit of Fig. 2 (i.e. no contact impedance). Resistor Re1 was then varied from 0.5 kΩ to 22 kΩ. Absolute errors were calculated, as described above.

A mismatch between the injecting leads' contact impedances caused different behaviours in the devices. The Impedimed device had lower errors in the computed tissue impedance parameters, being more sensitive when the mismatch occurred in the negative injecting lead.

In the case of the BioparHom device, the calculated tissue parameter values were very sensitive to mismatched contact impedances in both the injection and detection leads, especially in the latter.

If the contact impedance mismatch occurs in the detecting leads, both devices are very sensitive and provide proper tissue impedance values only when Re1 is lower than 1 kΩ.

C. Effect of electrode impedance mismatch during in vivo measurements

Electrode contact impedance mismatches during *in vivo* measurements were augmented by connecting pairs of electrodes at a given lead in order to double the contact area and thus decrease the contact impedance by approximately 50 %. Resultant complex impedance plots measured using the BioparHom and Impedimed devices are presented in Fig. 4 and 5, respectively.

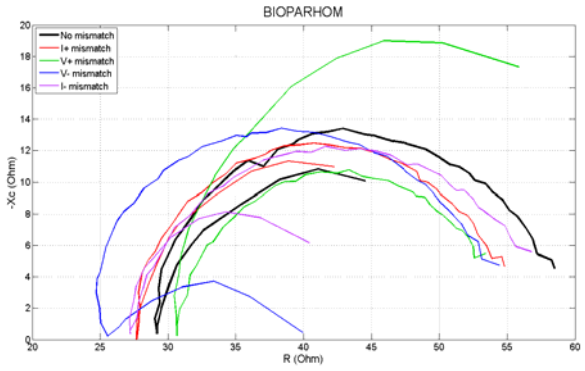

Fig. 4. In vivo complex impedance plots obtained with the BioparHom device due to a variation in contact impedance in each of the four leads.

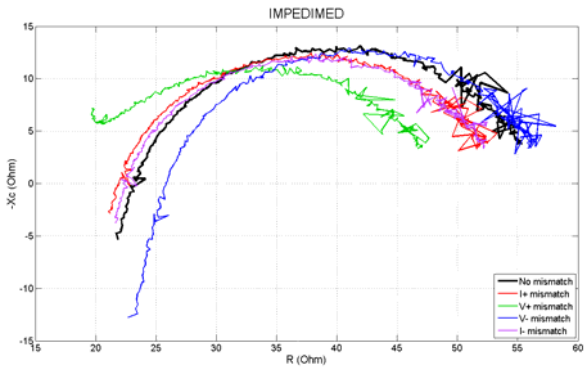

Fig. 5. In vivo complex impedance plots obtained with the Impedimed device due to a variation in contact impedance in each of the four leads.

Relative errors in measured values were calculated for the following frequencies, 5, 50 and 100 kHz, with respect to "zero" mismatch condition (i.e. unaltered contact

impedances), Table 2 and Table 3 show the relative errors for the BioparHom and Impedimed devices respectively.

IV. DISCUSSION

While contact impedance has a significant effect on the measurements of tissue impedance with both commercial devices, especially with values higher than 4.7 kΩ in the injecting leads, electrode impedance mismatch on detecting leads has the greatest effect with both devices and it must be kept under 1 kΩ to avoid negative effects on measurements.

Results from *in vivo* impedance measurements with electrode mismatch confirmed the results obtained with electrical models. Electrode impedance mismatch with the BioparHom device has a great impact on the detection of the imaginary part of the impedance, especially in the negative detecting and injecting leads. With the Impedimed device it affects mostly the real part. Both devices have significant errors at 50 kHz, a frequency that is commonly used in monofrequency BIA.

Table 2. BioparHom: relative errors in series resistance (R) and reactance (Xc) for 5, 50, 100 due to a variation in contact impedance in each of the four leads

		FREQUENCY (kHz)		
lead		5	50	100
I+	R (%)	-6.30	-8.73	-6.95
	Xc (%)	3.09	2.09	4.41
V+	R (%)	-5.06	-4.99	-11.09
	Xc (%)	-5.88	-23.30	0.30
V-	R (%)	-4.55	-2.23	-12.97
	Xc (%)	-6.95	-38.33	6.65
I-	R (%)	-4.99	1.93	-13.94
	Xc (%)	-6.94	-87.02	23.40

Table 3. Impedimed: relative errors in series resistance (R) and reactance (Xc) for 5, 50, 100 and 200 kHz due to a variation in contact impedance in each of the four leads

		FREQUENCY (kHz)		
lead		5	50	100
I+	R (%)	-2.88	-14.87	-2.31
	Xc (%)	-11.46	-5.73	17.37
V+	R (%)	-4.52	-15.54	5.78
	Xc (%)	-3.74	-14.01	-2.68
V-	R (%)	-5.54	-13.68	5.64
	Xc (%)	-5.28	-5.80	-3.52
I-	R (%)	-4.69	-13.19	6.00
	Xc (%)	0.52	21.91	21.56

V. CONCLUSION

Both commercial devices can be adversely affected by the electrode/skin contact impedances, especially when performing localized impedance measurements, for which they were not designed.

With localised impedimetric measurements, the tissue impedance under study is small and, as a result, more sensitive to contact impedances and contact impedance mismatch. This adverse effect was largest when the mismatch involves the detecting leads rather than the injecting leads.

The contact impedances must be minimized or at least be small with respect to the magnitude of the expected changes in the monitored tissue impedance. According to the results of the present study, the magnitudes of interface impedances should be no larger than 0.5 kΩ to 1 kΩ for the segment measured and the devices used.

Although it is tempting to use commercial impedance monitoring devices in novel ways to develop new and exciting applications of bioimpedimetric analysis, one must therefore be very careful to check that the new applications do not involve impedance ranges outside those for which the devices were designed.

In order to achieve this, for example, better electrodes should be designed, possibly with larger areas, such as band electrodes which may also help to optimise the current distribution throughout the segment under investigation. The authors are presently developing novel

electrode systems for their particular application.
The authors are also working with a manufacturer of BIA systems to develop a system specifically designed for localised impedimetric measurements, thus overcoming the problems associated with this challenging application.

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