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Problems encountered during inappropriate use of commercial bioimpedance devices in novel applications

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Abstract: It is often tempting to apply commercially available impedance monitoring devices to novel applications which involve impedances outside the range for which the device was designed. This problem may be more common than expected and will result in distorted impedance loci and in the inaccurate calculation of model parameters. The authors illustrate this problem by using commercial devices, designed for whole body bioimpedance spectroscopy (BIS) measurement, in a more demanding localized impedimetric study. The lower tissue impedances involved make the measurements more prone to the adverse effects of large contact impedances and, in particular, large contact impedance mismatches. One must therefore develop novel, low contact impedance electrodes or new devices specifically designed for more demanding localized applications.

Keywords: Commercial impedance device, electrode-skin impedance, segmental impedance, electrode mismatch

1. Introduction

A range of Commercial impedance devices exist for whole-body BioImpedance Analysis (BIA) in a range of clinical applications [1,2]. Recently, there has been interest in the use of such measurements on more localized regions or segments of the body, possibly leading to wearable or, at least, to the more convenient home use of systems by patients [3]. The present authors are interested in, among other things, the use of regional/segmental BIA measurements as an objective, non-invasive method of detecting and evaluating changes in the hydrational and nutritional status of patients with renal disease, especially for those undergoing dialysis at home.

Moving from whole body to regional/segmental BIA measurement is not without difficulty, however. Some of the problems encountered have already been noticed and reported in whole body BIA measurements, some are however particular to

regional/segmental BIA measurements and are reported in this paper to warn others of the risks involved when using existing commercial devices, designed for whole body measurement, in novel applications such as ours.

In Segmental-BIA, it has been identified that the design of the electrodes used and the standardization of their placement are areas which limit the value of the approach. Slight differences in electrode placement can significantly alter the encompassed tissue area and dramatically affect the derived parameter values. It has also been pointed out that there is a need to investigate and standardize electrode design in order to minimize contact impedance interferences and to optimize targeted current density and distribution. Although a variety of commercial spot electrodes are used in the literature, there is a lack of information on which type yields the best reproducibility in prediction models. Another issue is the value of spot as compared to band electrodes. Theoretically, band or circumferential electrodes should provide a more homogenous electrical field within the targeted tissues, resulting in more accurate masurements [4].

In regional/segmental BIA measurement there is the additional problem that the targeted tissue impedance is much smaller than in whole body measurements, making the contact impedances, due to electrode/skin interfaces, much more significant. In fact, researchers must be very careful that the tissue impedances encountered in such regional/segmental (or other) measurement are within the operating range of the commercial device used. The authors have found that that was not the case in some of their localized impedimetric studies, even with "gold standard" laboratory devices such as the Solartron 1255B with 1294 interface and, as a result, the data obtained in these cases were grossly distorted and largely useless. This is a potentially serious problem in the novel use of devices designed for whole body BIA measurement or for impedances larger than those observed in localized measurements. Note that contact impedance varies between individuals and, on same person, with body site, electrode design/area, time, temperature, pressure, etc [5]. One must therefore carefully assess the individual contact impedances to ensure the validity of measurements being made.

In the present study, the authors will concentrate on the problems associated with contact impedances during the use of commercial whole body systems for the study of regional/segmental impedances.

The effects of contact impedance and lead capacitance were studied by Bolton et al [6] on 3 commercial devices designed for whole-body BIA. They carried out measurements on patients as well as on electrical models and found significant differences in the "tissue" impedance recorded as a result of variations in the contact impedance.

Ward [7] performed body composition measurements using commercial impedance devices on volunteers and on equivalent electrical circuits. He observed that the reproducibility of results obtained on volunteers was significantly lower than those on electrical circuits. This result is most probably due to the variability of the invivo electrode-skin contact impedances within and between subjects.

The effect of electrode impedance mismatch in the estimation of Cole parameters were investigated by Buendia et al. [8]. They found that a mismatch of impedance in the detecting electrodes led to a change in the measuremed tissue impedance

compared to that obtained using matching electrodes.

Bogónez-Franco et al. [9] carried out a study of the effect of electrode impedance mismatch and of ground coupling on commercial impedance devices and reported that electrode impedance mismatch has a noticeable effect on measured impedance values but that ground coupling had little effect.

One of the problems caused by electrode mismatch is the conversion of differential-mode voltages to common-mode. Some researchers [10] have proposed the design of new measurement circuits in order to increase the common-mode rejection ratio (CMRR) of the measuring system and to thus reduce the effect of common-mode voltages caused by electrode mismatch.

In the present work we will investigate the effect of electrode-skin contact impedance on measured tissue impedances and the associated derived impedimetric parameters when two commercial whole-body BIA devices are used to make localized impedance measurements on the calf. The impedance properties of the human calf are of interest to studies in a range of potential clinical applications such as monitoring during hemodialysis therapy [11].

2. Materials

2.1 Commercial impedance analyzers

Two commercial impedance analyzers were used in this study; BioparHom Z-Métrix (BioparHom, Bourget du Lac, France) and Impedimed SFB7 (Impedimed Ltd, Brisbane, Australia). All data were acquired using the software provided by the manufacturer. Calibration of devices was checked before measurement using the calibration networks provided by the manufacturers.

2.2 Electrode-skin contact impedance measurement

In order to build a representative equivalent circuit model, contact impedances due to the electrode/skin interface were first measured on the calf of a test subject using a 3-electrode configuration over the frequency range of 1 Hz to 10 kHz.

3M 2660 (3M, Minneapolis, USA) hydro-gel electrodes, recommended by BioparHom, were used for the measurements and no skin preparation was performed before placement of electrodes.

All impedances at this stage were measured using a Solartron 1255 (Solartron Analytical Ltd, Hampshire, England) and a 1294A biological impedance interface. No distortion was observed in this case due to the high output impedance of current source.

2.3 Tissue impedance measurement

In like manner, typical tissue impedance values for this body segment were measured invivo over the range 1 kHz to 1 MHz using the tetra-polar electrode configuration.

2.4 Electrode-skin contact impedance electrical model

An equivalent electrical circuit model consisting of a resistor in series with the parallel combination of a capacitor and a resistor was built. The resistor and capacitor component values are shown in figure 1 and were 1% and 5% in tolerance, respectively.

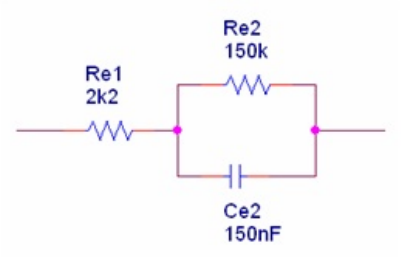

Figure 1: Equivalent circuit model for electrode/skin contact impedance.

2.5 Tissue electrical model

An equivalent electrical circuit model, similar to that used for the electrode/skin contact impedance was built for tissue. Component values are shown in figure 2. Again, resistor and capacitor values used in this circuit were 1% and 5% in tolerance, respectively.

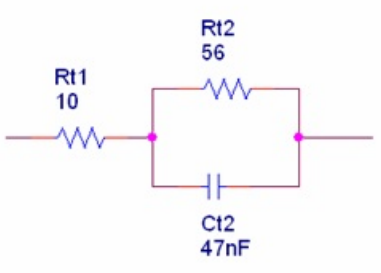

Figure 2: Equivalent circuit model for tissue impedance.

2.6 Fitting impedance measurements

The frequency-dependent capacitances measured for the actual contact and tissue impedances, normally represented by the term α in empirical Constant Phase

Elements (CPEs) [12], had to be approximated by standard, frequency-independent capacitors to enable the construction of an electrical circuit model.

We used Z-View software (Scribner Associates Incorporated, North Carolina, USA) to help in this work. As the α values obtained from the fitting process were very close to 1, 0.992 for electrode-skin and 0.996 for tissue impedance, the values of the magnitudes of the CPEs were simply used for the purely capacitive components, (Ce2 and Ct2 respectively), in the physical electrical models.

Combinations of these equivalent electrical circuits were connected to the Impedimed SFB7 and BioparHom Z-Métrix devices to assess their sensitivities to variations in contact impedances representative of localized BIA measurements

2.7 Errors

Absolute errors were calculated by comparing the actual equivalent circuit parameter values used to represent the tissue impedance and those measured by the devices in the presence of large and/or mismatched contact impedance circuits.

2.8 Measurements on healthy people

In order to check the effect of electrode contact impedance mismatch, we also carried out measurements invivo on the human calf. Electrode mismatch was reproduced by connecting two electrodes together at a given location to cause a theoretical reduction in electrode/skin contact impedance of 50%. Figure 3 shows the four-electrode configuration and the possible pairing of contact electrodes to increase the electrode impedance mismatch.

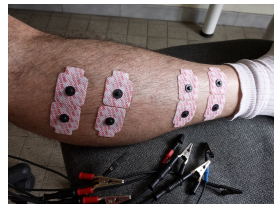

Figure 3: Four-electrode configuration with electrode pairs used to cause electrode mismatch.

3. Experimental results

3.1 Effect of contact impedance

The effect of contact impedance was evaluated using the equivalent electrical circuits described above. Over the frequencies of interest, the magnitude of contact impedance is mainly determined by series resistor Re1 of figure 1. In order to study

the effect of contact impedance on measured tissue impedance, resistor Re1 was therefore varied from 0 kΩ to 10 kΩ.

Each lead was connected to the tissue impedance via contact impedances. The parameter values used to model contact and tissue impedances were those shown in figure 1 and figure 2, respectively. Tables 1-2 show the absolute errors observed in the calculation of each of the parameters in the tissue equivalent electrical circuit model when measured with the BioparHom and Impedimed devices, respectively.

Cells filled with * in tables 1 and 2, indicate that, under the given circumstances, the measured tissue response was found to behave inductively and hence it was not possible to derive the tissue circuit's parameter values.

3.2 Effect of contact impedance mismatch

Electrode/skin contact impedance mismatch was modelled by connecting a contact circuit to one lead at a time and by varying the series resistance, Re1, from 1 kΩ to 10 kΩ.

a) Effect of contact impedance mismatch on positive injection leads

The positive injecting lead was connected via a contact impedance circuit to the tissue circuit, whereas the remaining 3 leads were connected directly to the tissue circuit. Tables 3 and 4 show the absolute errors in the measured tissue impedance parameters as derived by the BioparHom and Impedimed devices, respectively.

b) Effect of contact impedance mismatch on negative injection leads

The negative injecting lead was connected via a contact impedance to the tissue impedance circuit, whereas the remaining 3 leads were connected directly to the tissue circuit. Tables 5 and 6 show the absolute errors in the measured tissue parameters as derived by the Impedimed and BioparHom devices, respectively.

c) Effect of contact impedance mismatch on positive detection leads

The positive detecting lead was connected via a contact impedance circuit to the tissue impedance circuit, whereas the remaining 3 leads were connected directly to the tissue circuit. Tables 7 and 8 show the absolute errors in the measured tissue parameters as derived by the Impedimed and BioparHom devices, respectively.

d) Effect of contact impedance mismatch on negative detection leads

The negative detecting lead impedance was connected via a contact impedance circuit to the tissue impedance circuit whereas the remaining 3 leads were connected directly to the tissue circuit. Tables 11 and 12 show the absolute errors in the measured tissue parameters as derived by the Impedimed and BioparHom devices, respectively.

Table 1: Absolute errors in the estimation of tissue parameters for BioparHom Z-Métrix with contact impedances on each lead

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
0	1.54	-1.02	-4.40
1	26.80	-26.17	86.00
2	*	*	*
3	*	*	*
4	*	*	*
5	*	*	*
6	*	*	*
7	*	*	*
8	*	*	*
9	*	*	*
10	*	*	*

Table 2: Absolute errors in the estimation of tissue parameters for Impedimed SFB7 with contact impedances on each lead

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
0	0.15	-0.49	-2.97
1	0.62	-0.84	-1.26
2	4.50	-4.56	5.36
3	9.63	-9.69	15.89
4	14.10	-13.97	28.70
5	18.57	-18.25	41.50
6	28.98	-29.02	109.10
7	36.30	-36.36	205.6
8	44.44	-44.51	510.3
9	*	*	*
10	*	*	*

Table 3: Absolute errors in the estimation of tissue parameters for BioparHom Z-Métrix with contact impedances mismatch on positive injection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	0.10	0.90	-7.08
2	-0.47	1.20	-7.61
3	-2.01	2.41	-7.70
4	-3.40	3.96	-7.59
5	-6.10	6.94	–8.23
6	-7.65	8.19	-8.11
7	-8.25	9.01	-6.98
8	-10.28	10.59	-7.47
9	-10.46	10.84	-6.02
10	-10.65	10.97	-4.33

Table 4: Absolute errors in the estimation of tissue parameters for Impedimed SFB7 with contact impedances mismatch on positive injection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	0.54	-0.75	-2.37
2	0.64	-0.79	-2.51
3	0.62	-0.81	-2.97
4	0.54	-0.72	-3.48
5	0.64	-0.80	-3.37
6	0.73	-0.87	-3.26
7	0.70	-0.86	-3.68
8	0.82	-0.89	-3.50
9	0.75	-0.87	-4.01
10	0.96	-1.06	-3.56

Table 5: Absolute errors in the estimation of tissue parameters for BioparHom Z-Métrix with contact impedance mismatch on negative injection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	5.63	-5.08	8.47
2	9.46	-8.98	20.70
3	12.61	-12.12	28.80
4	15.00	-14.54	37.12
5	17.47	-17.11	47.98
6	19.74	-19.31	59.70
7	21.80	-21.43	73.60
8	23.80	-23.39	89.40
9	25.36	-25.05	103.90
10	26.78	-26.5	118.90

Table 6: Absolute errors in the estimation of tissue parameters for Impedimed SFB7 with contact impedance mismatch on negative injection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	0.21	-0.51	-2.86
2	0.30	-0.49	-2.78
3	0.28	-0.50	-2.68
4	0.21	-0.41	-2.84
5	0.19	-0.37	-2.83
6	0.14	-0.32	-2.83
7	0.09	-0.28	-2.87
8	0.06	-0.24	-2.84
9	0.00	-0.15	-3.03
10	0.01	-0.14	-2.95

Table 7: Absolute errors in the estimation of tissue parameters for BioparHom Z-Métrix with contact impedance mismatch on positive detection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	-5.75	6.14	-12.55
2	-12.73	12.90	-17.75
3	-20.26	20.20	-21.35
4	-26.33	26.08	-23.62
5	-31.05	30.78	-26.61
6	-36.76	36.28	-26.08
7	-41.25	40.42	-27.58
8	-44.12	43.24	-27.45
9	-47.01	46.10	-27.18
10	-48.89	48.00	-26.57

Table 8: Absolute errors in the estimation of tissue parameters for Impedimed SFB7 with contact impedance mismatch on positive detection leads

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	5.37	-5.43	7.00
2	9.95	-9.92	15.91
3	4.24	-14.52	25.84
4	7.69	-17.59	36.56
5	21.37	-21.20	51.32
6	24.89	-24.25	67.40
7	27.40	-27.31	84.40
8	30.59	-30.37	110.50
9	33.63	-33.35	147.70
10	36.32	-36.03	179.70

Table 9: Absolute errors in the estimation of tissue parameters for BioparHom Z-Métrix with contact impedance mismatch on negative detecting lead

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	4.46	5.13	1.05
2	7.69	-7.03	7.33
3	11.08	-10.35	15.11
4	13.74	-12.90	21.34
5	16.57	-15.53	20.62
6	19.78	-18.66	39.90
7	21.08	-19.91	42.40
8	23.74	-22.50	53.70
9	25.88	-24.51	63.00
10	27.50	-26.09	70.80

Table 10: Absolute errors in the estimation of tissue parameters for Impedimed SFB7 with contact impedance mismatch on negative detecting lead

Re1 (kΩ)	Error Rt1 (Ω)	Error Rt2 (Ω)	Error Ct2 (nF)
1	-3.65	3.59	-7.12
2	-9.84	9.60	-13.50
3	-17.02	16.67	-18.44
4	-23.22	22.81	-21.59
5	-28.39	28.09	-23.41
6	-35.32	34.91	-26.08
7	-42.48	41.94	-28.39
8	-49.64	49.00	-30.19
9	-55.99	55.40	-31.55
10	-63.61	62.9	-32.95

Table 11: Relative errors for BioparHom Z-métrix in resistance (R) and reactance (Xc) at 5, 50, 100 and 200 kHz due to a variation in contact impedance mismatch in each lead

lead	5	50	100	200	5	50	100	200
I+	-6.30	-8.73	-6.95	-2.95	3.09	2.09	4.41	2.38
V+	-5.06	-4.99	-11.09	-5.43	-5.88	-23.30	0.30	-10.46
V-	-4.55	-2.23	-12.97	-5.84	-6.95	-38.33	6.65	-16.48
I-	-4.99	1.93	-13.94	-5.39	-6.94	-87.02	23.40	-24.49

Table 12: Relative errors for Impedimed SFB7 in resistance (R) at 5, 50, 100 and 200 kHz due to a variation in contact impedance mismatch in each lead

lead	5	50	100	200	5	50	100	200
I+	-2.88	-14.87	-2.31	-4.55	-11.46	-5.73	17.37	1.47
V+	-4.52	-15.54	5.78	-5.12	-3.74	-14.01	-2.68	-4.89
V-	-5.54	-13.68	5.64	-3.60	-5.28	-5.80	-3.52	-5.18
I-	-4.69	-13.19	6.00	-3.23	0.52	21.91	-21.56	-2.95

e) Effect of electrode impedance mismatch in in-vivo measurements

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

Relative errors in the measured series resistance (R) and reactance (Xc) values were calculated for the following coincident frequencies on both devices: 5, 50, 100 and 200 kHz with respect to "zero" mismatch conditions (i.e. unaltered contact impedances). Tables 11 and 12 show the relative errors calculated for BioparHom device and tables 13 and 14 for Impedimed.

4. Discussion

Both commercial devices are potentially adversely affected by the presence of electrode/skin contact impedances, especially under conditions representing localized impedance measurements, for which the devices were not designed.

BioparHom device can not tolerate contact impedances higher than 1 kΩ in any of its leads. Exceding this value, the measured data behaves like that of an inductive circuit, which is obviously not the case for biological tissue. Impedimed device can tolerate contact impedances up to 8 kΩ in any lead. Values higher than 8 kΩ has the same effect as on BioparHom, the circuit measured appears to behave as an inductive circuit.

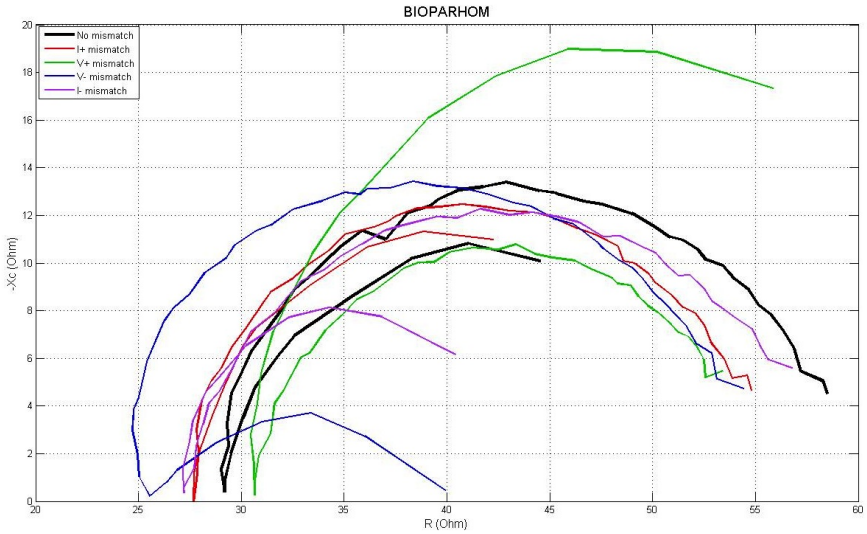

Figure 4: Complex impedance plot for Impedimed SFB7.

With the BioparHom device, causing a mismatch on the positive injection lead, gives rise to a maximum error of -10.65 Ω in the estimation of Rt1, 10.97 Ω in the estimation of Rt2 and -4.33 nF in the estimation of Ct2 when the contact impedance mismatch was 10 kΩ. The Impedimed device is less affected by contact impedance mismatch on the positive injection lead. Maximum errors were 0.96 Ω, -1.06 Ω and -3.56 nF in the estimation of Rt1, Rt2 and Ct2, respectively, when there was contact impedance mismatch.

A mismatch in the negative injection leads has a larger effect on the BioparHom device than on the Impedimed. The maximum errors for the BioparHom device when contact impedance mismatch was 10 kΩ were 26.78 Ω, -26.50 Ω and 118.90 nF for the estimation of Rt1, Rt2 and Ct2, respectively. In the case of the Impedimed device, the maximum error in the estimation of Rt1 was 0.30 Ω and this occurred when the contact impedance mismatch was 2 kΩ. For the estimation of Rt2, the maximum error was -0.51 Ω and was caused by a contact impedance mismatch of 1 kΩ. For the estimation of Ct2, the maximum error was -3.03 nF when the contact impedance mismatch was 9 kΩ.

When contact impedance mismatch was caused on positive detection leads, both devices were strongly affected. The maximum error for BioparHom in the estimation of Rt1 and Rt2 were -48.89 Ω and 48.00 Ω, respectively, when contact impedance mismatch was 10 kΩ. The maximum error in the estimation of Ct2 was -28.58 nF with a contact impedance mismatch of 7 kΩ. Maximum errors for Impedimed were 36.32 Ω, -36.03 Ω and 179.70 nF in the estimation of Rt1, Rt2 and Ct2, respectively, with a contact impedance mismatch of 10 kΩ.

Again, the effect of contact impedance mismatch on negative injection leads has

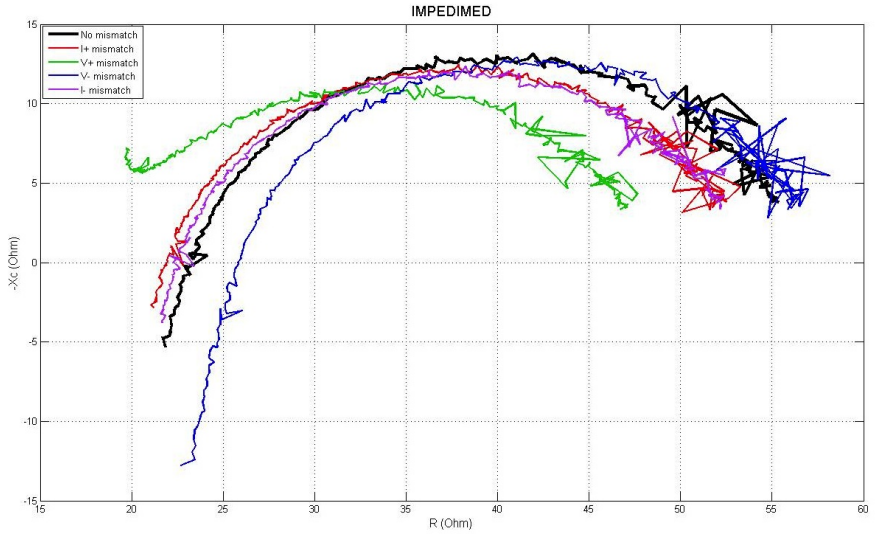

Figure 5: Complex impedance plot for Impedimed SFB7.

a larger effect than on detection leads. Maximum errors occurred for both devices, when contact impedance mismatch was 10 kΩ. For BioparHom, the maximum errors were 27.50 Ω, -26.90 Ω and 70.80 nF in the estimation of Rt1, Rt2 and Ct2, respectively. For Impedimed, these values were -63.61 Ω, 62.90 Ω and -32.95 nF, respectively.

When contact impedance mismatch was caused in in-vivo measurements we anticipate a larger adverse effect on the accuracy of the measurement when the mismatch is caused on negative injection and detection leads. For BioparHom, we obtained errors larger than 10% in the measured series resistance (R) at 50 kHz for changes in the injection lead and at 100 kHz for changes in both of the injection and detection leads. For the series reactance (Xc), errors larger than 10% were obtained on the negative injection lead at 50, 100 and 200 kHz and in both detection leads at 50 and 200 kHz. For Impedimed, errors larger than 10% in the measurement of the series resistance were obtained for changes in the injection and detection leads at 50 kHz. For the reactance, these errors were obtained in the positive injection lead at 5 and 100 kHz, and in the positive detection and negative injection leads at 50 kHz.

5. Conclusion

With localized impedimetric measurements, the tissue impedance under study is small and, as a result, more sensitive to contact impedances and contact impedance mismatch. In the present study, it was observed that the commercial, whole body bioimpedance devices were relatively less sensitive to contact impedance but very

sensitive to contact impedance mismatch.

This adverse effect was largest when the mismatch involves the detecting leads rather than the injecting leads. This effect could be reduced by increasing the input impedance and common mode rejection ratio (CMRR) of the input stage.

The influence of contact impedances must be minimized or must minimized with respect to the magnitude of the expected changes in the monitored tissue impedance. According to the results of the present study, for the body segment assessed, the magnitudes of interface impedances should be no larger than 0.5 kΩ to 1 kΩ.

In order to achieve this, better electrodes must be designed, possibly with larger areas, such as band electrodes which may also help optimize the current distribution throughout the segment under investigation. The authors are presently developing novel electrode systems for their particular application.

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

In the authors' project, involving the development of regional/segmental BIA measurements as a method of detecting and evaluating changes in the hydration and nutritional status of home-based patients, this is unfortunately the case.

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

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