Effect of stray capacitances to ground in auto-balancing impedance bridges

Most commercial impedance analyzers are nowadays based on auto-balancing bridges [11]. Figure 1 shows the equivalent circuit when impedance $Z_0$ is measured by connecting it between terminals H and L. The instrument applies a voltage to terminal H and measures the current at node L, which is held at 0 V (virtual ground) by a feedback loop. This current measurement method is analogous to a transimpedance amplifier based on an ideal op amp [11, 13]. The impedance at a given frequency is calculated by dividing the vector voltage (phasor) $V_{in}$ measured between H and ground by the vector current (phasor) $I_L$ measured at that frequency.

Measurement uncertainty depends on [14]: (1) contact impedance $Z_{e}$ between the measured material and instrument terminals H and L due to cables and electrodes;
(2) leakage (“air”) capacitance $C_{hl}$ between H and L; and (3) parasitic capacitances $C_{hs}$, $C_{lg}$ between each electrode and ground and $C_g$ between the measured material and ground.

This measurement method is immune to $C_{hl}$ and $C_{lg}$ [15], which has contributed to its popularity, but cannot measure grounded impedances [11]. Moreover, the “air” capacitance $C_{hl}$ between electrodes and the capacitance $C_g$ from the material to ground affect the measurement results [13]. $C_{hl}$ mostly depends on interelectrode distance and can be minimized by placing a small grounded electric shield between the electrodes [11]. $C_g$, however, cannot be easily minimized because it is a distributed capacitance, hence it will depend on the volume of the material and its distance to ground. Furthermore, $C_g$ increases when a grounded electric shield is placed between electrodes to reduce $C_{hl}$ [13].

Displacement current $I_g$ in figure 1 reduces the measured current $I_l$ hence the calculated impedance will be larger than the expected $2Z_c + Z_o$. When the human body impedance is measured, $C_g$ can be sizeable because of the large volume of the body and because the distance to ground can be as short as the thickness of footwear soles. The relevant effects of $C_g$ in impedance measurements, either bipolar with direct contact [13] or capacitive electrodes [16], or tetrapolar [10] warrant the interest of its measurement.

**Measurement method**

In a scenario where bipolar impedances are measured on the human body, we propose to estimate the capacitance from the body to ground by connecting a known capacitor between each electrode and the impedance analyzer. Figure 2 shows the resulting equivalent circuit. If the capacitance of the added capacitors $C$ is small enough for its impedance to be much larger than that of the body and the electrodes at the measurement frequency,

$$\frac{1}{j\omega C / 2} \gg 2Z_c + Z_o,$$

then the equivalent circuit can be simplified into that in figure 3.

![Fig.2: Equivalent circuit when a capacitor $C$ is connected in series with each electrode in bipolar body impedance measurements.](image)

By applying the star-delta impedance transformation, the impedance between H and L in figure 3, as calculated by the impedance analyzer, will be

$$Z_{hl}(j\omega) = \frac{V_{HG}}{I_l} = \frac{1}{j\omega C} + \frac{1}{1 + j\omega C} = \frac{1}{j\omega C} \frac{1}{2 + C_g / C}$$

and the equivalent capacitance,

$$C_{il} = \frac{1}{j\omega Z_{il}} = \frac{C}{2 + C_g / C}.$$  

When $C_g = 0$, we have $C_{il}$ = $C/2$ but otherwise the measured capacitance will be smaller than $C/2$. Solving (3) for $C_g$ we obtain

$$C_g = C \left( \frac{C}{C_{il}} - 2 \right).$$

Therefore, selecting $C$ such that at the desired measurement frequency the condition (1) is fulfilled, $C_g$ can be estimated from (4). Usually, selecting $C$ within the range of the expected value for $C_g$ is a good choice.

**Experimental design**

The proposed method to estimate $C_g$ has been applied in two different instruments based on auto-balancing bridges: an impedance analyzer (Agilent 4294A) and a handheld LCR meter (Motech MT4080). The relative uncertainty of the 4294A when measuring capacitances between 10 pF to 100 pF at 10 kHz ranges from ±1 % to ±0.3 % of the reading [17], whereas that of the MT4080 when measuring capacitances between 15.91 pF and 159.1 pF at 10 kHz is ±0.5 % of the reading plus ±1 [18]. The 4294A is a top-range impedance analyzer that cannot measure grounded impedances [11] whereas the MT4080 is a low-cost LCR meter that can measure grounded impedances because it is supplied by batteries, hence has floating inputs. Nevertheless, it can measure only up to 10 kHz. In order for the electromagnetic environment to be similar for both
instruments, the guard terminal of the MT4080, which is connected to its (floating) signal ground [18], has been connected to earth ground.

Four volunteers (two men, subjects #1 and #2, and two women, subjects #3 and #4) have been measured. Their basic anthropometric data and shoes’ outsole and type are given in Table 1. BMI is the body-mass index calculated by dividing the mass (kg) by the squared height (m²).

Subjects were successively connected to the impedance meters by two 4.9 cm² brass plates and 12 cm long braided wire to minimize the serial inductance of the connection. A respective ceramic 68 pF capacitor (±10 % tolerance) was soldered to the end of each cable. The actual values of the capacitors at 10 kHz, measured with the 4294A, were 74.1 pF for the one connected to terminal H and 74.0 pF for the one connected to terminal L. These values guarantee a high-enough impedance for C at 10 kHz to fulfill equation (1) as the body impedance at this frequency is about 499 Ω in series with 62 nF and the electrodes have yet smaller impedance [19].

The measurement protocol was as follows: the subject was seated on an office chair with plastic caster wheels, and held a (dry) electrode between the thumb and index finger of each hand. The hands were about 20 cm apart and rested on the wooden surface of the table that held the impedance meters. The guard terminal of the MT4080, which is connected to its (floating) signal ground [18], has been connected with the shoes in direct contact with the ground. These values are very close to the corresponding ones at 10 kHz in Table 2 and corroborate the validity of the method. Further, whereas figure 8 in [12] showed some erratic tendencies below 50 kHz, no reading fluctuations were observed here. This is because in complex impedance measurements it is easier to have good resolution in the predominant component (real or imaginary) than in the smaller quadrature component. Both here and in [12], the method relies on the “inductive component” contributed by C_HL, which increases the inductance. Each subject was measured twice: first with the feet at about 10 cm apart and rested on the wooden surface of the table that held the impedance meter us.

### Discussion

The differences between C_HL values measured in the same situation by each instrument range from 0.2 pF to 0.5 pF. However, the corresponding difference between calculated C_g values is a bit larger than 5 pF. This is a consequence of the nonlinear relationship between both capacitances, as shown in (4), but in any case, the deviation is just around 4 %, which is quite acceptable.

Calculated C_g values at 100 kHz for subject #1 in [12] were 96.6 pF with raised feet and 158.1 pF with feet on ground. These values are very close to the corresponding ones at 10 kHz in Table 2 and corroborate the validity of the method. Further, whereas figure 8 in [12] showed some erratic tendencies below 50 kHz, no reading fluctuations were observed here. This is because in complex impedance measurements it is easier to have good resolution in the predominant component (real or imaginary) than in the smaller quadrature component. Both here and in [12], the method relies on the “inductive component” contributed by

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**Table 1. Data from the measured subjects. BMI is the body mass index (kg/m²).**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Height (m)</th>
<th>Weight (kg)</th>
<th>BMI</th>
<th>Shoes’ outsole and type</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.90</td>
<td>130</td>
<td>36</td>
<td>Rubber, sports shoes</td>
</tr>
<tr>
<td>2</td>
<td>1.80</td>
<td>90</td>
<td>28</td>
<td>Leather dress shoes</td>
</tr>
<tr>
<td>3</td>
<td>1.66</td>
<td>65</td>
<td>24</td>
<td>Rubber, boots with 4 cm heel</td>
</tr>
<tr>
<td>4</td>
<td>1.50</td>
<td>42</td>
<td>19</td>
<td>Leather, boot with 7 cm heel</td>
</tr>
</tbody>
</table>

**Table 2. Capacitance C_HL measured with the 4294A impedance analyzer and calculated body capacitance to ground C_g for the four subjects and two feet heights above ground: 10 cm (u) and direct outsole contact (d).**

<table>
<thead>
<tr>
<th>Subject</th>
<th>C_HL (pF)</th>
<th>C_L (pF)</th>
<th>C_HLtot (pF)</th>
<th>C_Ltot (pF)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>22.4</td>
<td>96.5</td>
<td>18.2</td>
<td>153.0</td>
</tr>
<tr>
<td>2</td>
<td>23.4</td>
<td>86.1</td>
<td>17.8</td>
<td>159.7</td>
</tr>
<tr>
<td>3</td>
<td>25.0</td>
<td>71.1</td>
<td>20.2</td>
<td>123.2</td>
</tr>
<tr>
<td>4</td>
<td>25.8</td>
<td>64.3</td>
<td>19.8</td>
<td>128.6</td>
</tr>
</tbody>
</table>

**Table 3. Capacitance C_HL measured with the MT4080 LCR meter and calculated body capacitance to ground C_g for the four subjects and two feet heights above ground: 10 cm (u) and direct outsole contact (d).**

<table>
<thead>
<tr>
<th>Subject</th>
<th>C_HL (pF)</th>
<th>C_L (pF)</th>
<th>C_HLtot (pF)</th>
<th>C_Ltot (pF)</th>
</tr>
</thead>
<tbody>
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<tr>
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<td>25.8</td>
<td>64.3</td>
<td>19.8</td>
<td>128.6</td>
</tr>
</tbody>
</table>
C\textsubscript{g} but whereas in [12] that component added to a large real component, due to the two series-connected resistors, here it adds to the reactance of the two capacitors C and the series resistance (from the electrodes and the body) is negligible. Therefore, this novel method is clearly advantageous because it is fast and allows power-line supplied impedance analyzers to measure grounded impedances.

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References