

A fast method to estimate body capacitance to ground at mid frequencies

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Abstract

Impedance measurements that involve the human body are affected by the capacitance between the body and earth ground. This paper describes a fast method to estimate that capacitance at 10 kHz, which is valid for impedance analyzers intended to measure ungrounded impedances. The method does not require any external component other than two common capacitors and two conductive electrodes in contact with the body.

Keywords: Human body capacitance, stray capacitance, impedance measurement error

Introduction

The stray capacitance between the human body and earth ground can play a significant role in bioimpedance measurement systems where it may yield, for example, apparent inductive components not associated to any magnetic phenomena [1, 2]. It is also an intervening factor in the assessment of human body impedance for risk analysis in case of electric contacts [3] or electromagnetic radiation [4] and in electrostatic discharge studies [5], and has strong influence in power-line interference in biopotential measurements [6–8]. Finally, changes due to body capacitance are the sensing principle of capacitive tactile, proximity [9] and seat occupancy sensors [10].

Body capacitance to ground can be measured by using a grounded oscilloscope and a passive 10:1 voltage divider probe [6, 7], or a variable resistor, an amplifier and a voltmeter with grounded reference terminal [8], that estimate the power line (50/60 Hz) current through the body. These methods, however, heavily depend on the strength of existing power line fields and on the resistance of the contact with the body, hence on contact pressure. Some floating (battery-supplied) LCR meters can also measure that capacitance by connecting one of their input terminals to earth ground, but some of them can measure only up to 10 kHz, yet other cannot measure capacitances below, say, 100 pF.

Impedance meters/analyzers that have their signal ground at earth potential and are not battery-supplied cannot directly measure grounded impedances such as body-to-earth capacitance unless provided with special probes [11]. This shortcoming can be overcome by applying the method described in [12] wherein a network is built that comprises two resistors connected in series and the human body is connected to the node common to both resistors. The other end of each resistor is connected to a

respective terminal of the impedance analyzer. The measured impedance comprises a real part, due to the resistances, and a positive (hence inductive) imaginary part that is proportional to the product of the two resistances, the measurement frequency and the capacitance from the body to ground. Since the resistances of the two resistors and the measurement frequency are known, it is straightforward to calculate the capacitance to ground from the measured imaginary part. This method works fine from 50 kHz to 1 MHz but below 50 kHz that imaginary impedance component is too small compared to the real component and the limited resolution of the meter yields erratic readings.

In this paper, we propose to estimate body capacitance to ground by evaluating the effect of that capacitance when measuring two known capacitances connected in series. First, that effect is qualitatively analyzed and then the proposed method is described and its performance assessed.

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_b 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_{HG} measured between H and ground by the vector current (phasor) I_L measured at that frequency.

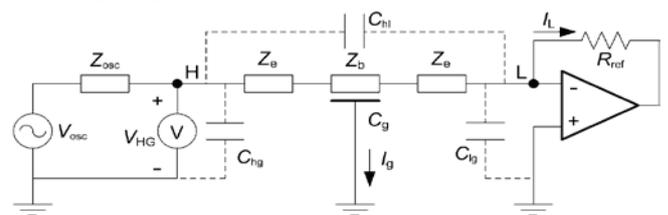


Fig.1: Equivalent circuit for impedance measurements based on an auto-balancing bridge.

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_{hg} , C_{lg} between each electrode and ground and C_g between the measured material and ground.

This measurement method is immune to C_{hg} 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 $2Z_e + Z_b$. 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 this 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,

$$\frac{1}{j\omega C/2} \gg 2Z_e + Z_b, \quad (1)$$

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

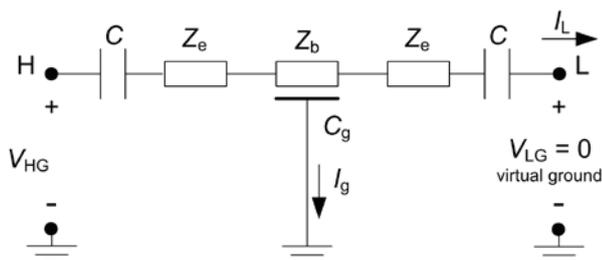


Fig.2: Equivalent circuit when a capacitor C is connected in series with each electrode in bipolar body impedance measurements.

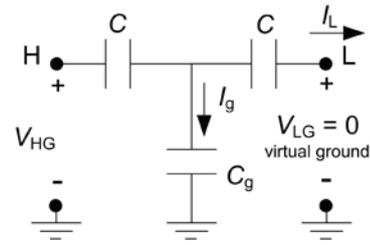


Fig.3: Simplified equivalent circuit when C is small enough for its impedance to predominate over that of the electrodes and the body at the measurement frequency.

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

$$Z_{HL}(j\omega) = \frac{V_{HG}}{I_L} = \frac{1}{j\omega C} + \frac{1}{j\omega C} + \frac{\frac{1}{j\omega C} \frac{1}{j\omega C}}{\frac{1}{j\omega C_g}} = \frac{1}{j\omega \frac{C}{2 + C_g/C}} \quad (2)$$

and the equivalent capacitance,

$$C_{HL} = \frac{1}{j\omega Z_{HL}} = \frac{C}{2 + C_g/C}. \quad (3)$$

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

$$C_g = C \left(\frac{C}{C_{HL}} - 2 \right). \quad (4)$$

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 $\pm 1\%$ to $\pm 0.3\%$ of the reading [17], whereas that of the MT4080 when measuring capacitances between 15.91 pF and 159.1 pF at 10 kHz is $\pm 0.5\%$ 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).

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

Subject	Height (m)	Weight (kg)	BMI	Shoes' outsole and type
1	1.90	130	36	Rubber, sports shoes
2	1.80	90	28	Leather dress shoes
3	1.66	65	24	Rubber, boots with 4 cm heel
4	1.50	42	19	Leather, boot with 7 cm heel

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 ($\pm 10\%$ 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 meter. This separation guaranteed a small-enough leakage (air) capacitance between electrodes. Increasing that separation could be counterproductive as the cables to connect the electrodes to the meter should then be longer, which would increase their inductance. Each subject was measured twice: first with the feet at about 10 cm from ground and later with the shoes in direct contact with the tiled floor.

Results

Table 2 shows the results obtained with the 4294A and Table 3 those obtained with the MT4080.

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 to ground: 10 cm (u) and direct outsole contact (d).

Subject	$C_{HL,u}$ (pF)	C_{gu} (pF)	$C_{HL,d}$ (pF)	C_{gd} (pF)
1	22.5	95.4	18.2	153.0
2	23.5	85.1	17.8	159.7
3	24.8	72.9	20.5	119.2
4	25.5	66.8	20.2	123.2

C_{HL} is the capacitance measured by each instrument at 10 kHz and C_g is the capacitance calculated from (4) for the two feet positions (u: up, d: down). No effect of the contact pressure with the electrodes was noticed.

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).

Subject	$C_{HL,u}$ (pF)	C_{gu} (pF)	$C_{HL,d}$ (pF)	C_{gd} (pF)
1	22.4	96.5	18.0	156.3
2	23.4	86.1	17.8	159.7
3	25.0	71.1	20.2	123.2
4	25.8	64.3	19.8	128.6

Discussion

C_g values display the same trends for all subjects and both instruments. For any given subject, the body capacitance to ground increases when the feet rest on ground, *i.e.* the distance from the body to ground decreases, regardless of the impedance meter used. That increase is larger for subjects #2 and #4 as expected from the thinner leather outsoles they wore with respect to subjects #1 and #3. This is in agreement with the results obtained with the more involved method to estimate C_g described in [12] that used a circuit similar to that in figure 2 but with two resistors instead of the two capacitors C.

For a given feet position, C_g values increase with BMI as it could be expected, but this dependence on BMI is far smaller than that on the distance from feet to ground. In Table 2, for example, C_g for subject #2 with feet on ground is larger than that for subject #1 in spite of the larger BMI of the first, and C_g for subject #4 with feet on ground is about 30 % larger than that for subject #1 with feet about 10 cm above ground, in spite that BMI for this one is 36 as compared to only 19 for subject #4. Therefore, closeness to grounded objects is more relevant than body size, and makes footwear a factor to account for in body impedance measurement and in electromagnetic hazard analysis.

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

C_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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