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Extracting the parameters of the single-dispersion Cole bioimpedance model using a magnitude-only method

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ABSTRACT

Affordable and portable indirect impedance measurement techniques with minimum hardware and post-processing requirements are necessary for various applications, and in particular for bioimpedance measurements. In this paper, a new method to extract the single-dispersion Cole bioimpedance model parameters is presented using a simple circuit consisting of a operational amplifier and a set of resistors. We examine the theory behind this magnitude-only method and demonstrate its application for the extraction of impedance data from a number of different fruit samples. Three means to compute the single-dispersion Cole model parameters are given and discussed under different measurement scenarios, and compared with experimental results.

1. Introduction

In the field of bioimpedance measurements, the single-dispersion and double-dispersion Cole electric models are widely used to characterize the electrochemical properties of biological tissues as well as to monitor physiological changes (Freeborn, 2013). Despite the higher accuracy offered by the double-dispersion model (Freeborn et al., 2014), the single-dispersion model remains the one widely used since it offers acceptable accuracy while requiring less number of unknown parameters to be computed. In particular, this model (shown within the dotted box of Fig. 1) consists of three hypothetical circuit elements: a low frequency resistor R_o in series with parallel association of a high frequency resistor R_∞ and a Constant Phase Element (CPE). The CPE is also known as the fractional capacitor and its impedance is given as $Z_{CPE} = 1/C_\alpha(j\omega)^\alpha$ where C_α is referred to as the pseudo-capacitance having units $\mathrm{F s}^{(\alpha-1)}$ while α is the dispersion coefficient $(0 < \alpha \leqslant 1)$. For simplicity, C_α commonly has the units of Farad which we will adopt here. The overall Cole–Cole impedance is described by

$$Z = R_\infty + \frac{R_o - R_\infty}{1 + \tau s^\alpha} = Z_r + jZ_i = |Z|\angle\theta \quad (1)$$

where the characteristic time constant of the biological tissue is defined as $\tau = C_\alpha(R_o - R_\infty)$ and $s^\alpha = (j\omega)^\alpha = \omega^\alpha[\cos(\theta) + j\sin(\theta)]$; $\theta = \alpha\pi/2$. Therefore, this model is a four-parameter model that requires knowing (extracting) the parameter set $(R_\infty, R_o, \alpha, C_\alpha)$ to fully characterize the impedance of the sample.

Classical bioimpedance spectroscopy relies on actual measurement of the impedance either by injecting a current into the tissue sample under consideration and subsequently measuring the developed voltage across this sample (current-excitation technique known as galvanostatic electrochemical impedance spectroscopy), or by applying a voltage across the sample and measuring the current flow (potentiostatic electrochemical impedance spectroscopy) (Grimnes and Martinsen, 2014). Both techniques are frequency-domain techniques; where the applied excitation is frequency swept to cover a range of interest (e.g. 1 Hz to 1 MHz) while Z_r and Z_i (or equivalently $|Z|$ and θ) are simultaneously extracted. This "direct" impedance measurement technique obviously requires both magnitude and phase to be accurately measured, making the hardware complex and expensive. It should be mentioned here that there are several recent attempts to propose cheap and portable impedance measurement devices such as those reported in Uvanesh et al. (2015), Hoja and Lentka (2013), Chabowski et al. (2015), Margo et al. (2013) and Melwin and Rajasekaran (2013). However, all "direct" impedance measurement techniques suffer from a major draw-back which is the necessity of having data post-processing using a suitable software optimization (data fitting) technique Trainito et al. (2015). In

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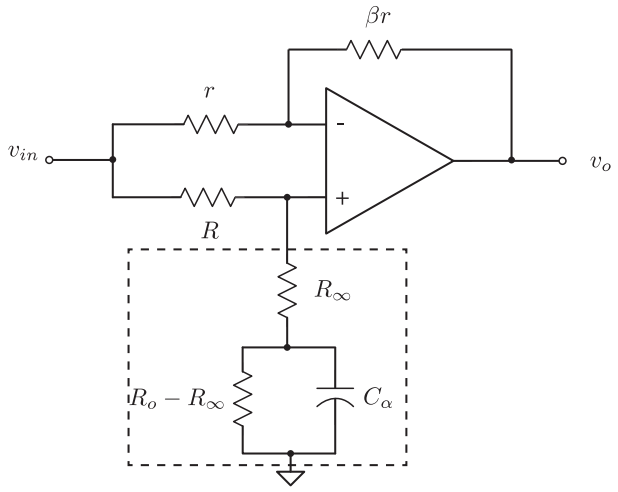

Fig. 1. Circuit used to determine the Cole parameter set $(R_\infty, R_o, \alpha, C_\alpha)$.

particular, recalling (1) and after measuring $|Z|$ and θ (or Z_r and Z_i), it remains to estimate the 4-variable parameter set $(R_\infty, R_o, \alpha, C_\alpha)$ which best fits the measured data. Several software packages have been developed to perform this task and some are freely available such as EIS Spectrum Analyzer (Bondarenko and Ragoisha, 2005).

Meanwhile, there has been continuous ongoing research efforts to propose "in-direct" impedance measurement techniques which significantly simplify the hardware requirements making it both very cheap and portable. In so doing, "in-direct" techniques rely on measuring the magnitude-only and follow a sequence of steps that systematically enable finding the parameter set $(R_\infty, R_o, \alpha, C_\alpha)$ one by one. This means that the need for data post-processing and data-fitting algorithms is also not necessary; although it can still be employed to improve accuracy. The first in-direct impedance measurement techniques were reported in Elwakil and Maundy (2010) relying on a high-pass filter setup and in Maundy and Elwakil (2012) relying on an integrator setup. Both setups require applying a voltage excitation v_{in} and measuring the magnitude of the obtained output voltage v_{out} at very low frequency $(\omega \to 0)$, very high frequency $(\omega \to \infty)$ and at half-power frequency where v_{out} drops in magnitude by -3 dB. In addition, the frequency at which the phase angle of v_{out} exhibits a minimum was analytically derived and shown to be detectable at a certain value of $|v_{out}|$, hence avoiding the need for actual phase measuring. Through these four measurements, the parameter set $(R_\infty, R_o, \alpha, C_\alpha)$ can be found. However, the two setups in Elwakil and Maundy (2010) and Maundy and Elwakil (2012) employ the sample tissue as a floating impedance which is not directly connected to the circuit ground (reference potential). Furthermore, in the integrator setup, the current flowing in the tissue sample is supplied through an operational amplifier which must have a sufficient current driving capability.

In this paper we improve upon our earlier work by proposing a new magnitude-only based technique where the tissue sample is employed as a grounded impedance with one terminal directly connected to ground potential. A single or double frequency sweep process to collect all the necessary data can be applied. In the single-sweep process, four data points are needed albeit it requires numerically solving a system of equations using floating point arithmetic to compute $(R_\infty, R_o, \alpha, C_\alpha)$. In the double sweep process, numerical solving of the equations can be avoided. It is worth noting that in-direct measurement techniques can also benefit from using optimization algorithms, such as Nonlinear Least Square Fitting to improve the accuracy of the extracted parameter values (Freeborn et al., 2012). It is also worth noting that in-direct magnitude-only time-domain (rather than frequency-domain) impedance extraction techniques have been proposed (Freeborn et al., 2013) and show excellent results. However, a detailed comparison between frequency-domain and time-domain extraction methods has not yet been done. The development of cheap and simple impedance measurement techniques is expected to have an impact on the food industry which is gradually adopting these techniques (Wu et al., 2008; Rehman et al., 2011; Damez et al., 2008).

2. Proposed technique

2.1. Circuit theory

Consider the circuit shown in Fig. 1. It consists of an operational amplifier (op amp) having a large bandwidth and several resistors with v_{in} being an applied sinusoidal voltage and v_o the measured output voltage. The tissue or biological sample to be tested is represented by the Cole model shown within the dashed lines. Straightforward analysis of this circuit yields a transfer function of the form:

$$H(s) = \frac{V_o(s)}{V_{in}(s)} = G_1\left[\frac{1-\tau_2 s^\alpha}{1+\tau_1 s^\alpha}\right] \tag{2}$$

where

$$G_1 = \frac{R_o - \beta R}{R_o + R} \tag{3}$$

is the DC $(\omega = 0)$ gain which is always less than unity and dependent on β, R and R_o. The time constants τ_1 and τ_2 are given respectively by:

$$\tau_1 = \frac{C_\alpha (R + R_\infty)(R_o - R_\infty)}{(R_o + R)} \tag{4}$$

and

$$\tau_2 = \frac{C_\alpha (\beta R - R_\infty)(R_o - R_\infty)}{(R_o - \beta R)} \tag{5}$$

Depending on the value of β $(\beta_i \geqslant 1,\ i = 1,2,3)$ it is possible to observe three types of magnitude responses corresponding to $\tau_2 < \tau_1$, $\tau_2 = \tau_1$ or $\tau_2 > \tau_1$, respectively as shown in Fig. 2 for

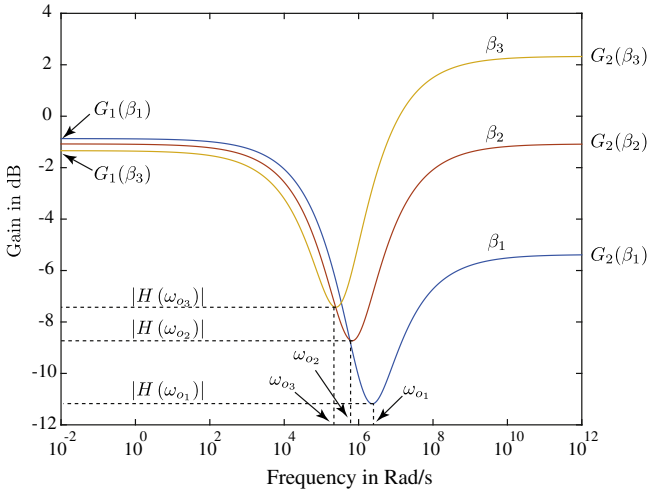

Fig. 2. Sample plot of Gain in dB versus ω with three various values of β (i.e. $\beta_1 = 1$, $\beta_2 = 1.45$ and $\beta_3 = 2$), with $C_\alpha = 1\ \mu$F, $R_\infty = 200\ \Omega$, $R_o = 20\ k\Omega$ and $R = 1\ k\Omega$. Note that with $\beta_2 = \beta_x$, we have $G_1 = |G_2|$.