As published in Innov. Techn. Biol. Med., Vol. 19, n°5, 395-399, 1998

ON USING SINGLE FREQUENCY ELECTRICAL MEASUREMENTS FOR SKIN HYDRATION ASSESSMENT

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Abstract:
A reliable method for skin moisture assessment is an important tool both in dermatology and in the field of cosmetics. The aim of this paper is to demonstrate that low frequency susceptance measurements carried out in the right manner, provide information about stratum corneum hydration with the same precision as multi frequency measurements may.

Previous research has shown that the capacitive component of the polarisation admittance is the proper electrical component to monitor for skin hydration assessment. By using low measuring frequency and a three electrode system with a small measuring electrode, this component can easily be measured with a well below 1% influence only from the other electrical components of the skin. With the right measuring technique, and with the present state of knowledge about the electrical properties of the skin, multi frequency measurements do not provide any extra precision or information about skin hydration, compared to low frequency susceptance measurements made at one single frequency.

Key words:
Skin hydration – electrical susceptance – single frequency – three electrode system.

Introduction

Skin function is closely related to skin moisture, and the monitoring of stratum corneum hydration is a central topic in bioimpedance research. Although many details are still unrevealed, we probably know enough about the major electrical characteristics of the skin, to choose the best approach for an electrical measurement of skin hydration changes. Multi frequency measurements provide in general more information about the skin or any other object, than single frequency measurements do, and multi frequency measurements are hence often regarded to be the only reliable method, in order to reduce the influence of unwanted effects, such as temperature,

sweat duct filling and the composition of applied products (1). With the present state of knowledge, however, the extra data from multi frequency measurements do not give any further information on skin moisture, and we will show that by applying the right technique, this information can be achieved with the same accuracy, by using single frequency measurements.

The Cole Equation

The Cole equation has been found able to model most measurements on biological tissue, including skin (2):

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + (iwt)^a} \qquad (1)$$

where R_0 and R_∞ are the low and high frequency resistance, respectively. τ is a time constant and α is constant such that

$$\varphi = \alpha \frac{\pi}{2} \qquad (2)$$

is a constant phase angle exhibited by the tissue. By introducing

$$G_{DC} = \frac{1}{R_0 - R_\infty} \qquad (3)$$

eq.1 is transformed into

$$Z = R_\infty + \frac{1}{G_{DC}(1 + (iwt)^a)}$$

$$or \quad Y = \frac{G_{DC}(1 + (iwt)^a)}{R_\infty G_{DC}(1 + (iwt)^a) + 1} \qquad (4)$$

The symbol G_{DC} will then represent the DC conductance in a system with no high frequency resistance. This electrical behaviour can be modelled with components in one of two equivalent ways, where the model in fig. 1 is the most commonly used (3, 4).

It follows from eq. 4 that the polarisation admittance, Y_{POL}, can be described by:

$$G_{DC}(iwt)^a \qquad (5)$$

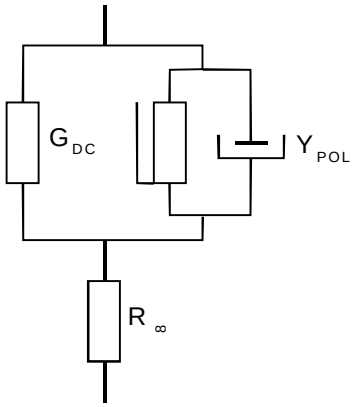

Figure 1.
Electrical model of skin

The behaviour of this polarisation admittance can be modelled by means of a frequency dependent capacitive reactance in parallel with a frequency dependent quasi-resistor. The properties of these components are basically two effects of the same polarisation mechanism, as the conductance G_{POL} can be regarded as the loss of C_{POL}. Within a limited frequency range, the constant phase behaviour can be modelled by applying the following frequency dependence:

$$C_{POL} = C_1 w^{k-1} \quad and \quad G_{POL} = G_1 w^k \qquad (6)$$

where k is a constant determining the slope of the frequency dependence, and C_1 and G_1 are the values of the frequency dependent parameters at ω=1. The following condition must then be satisfied

$$G_{DC}(iwt)^a = (G_1 + iC_1)w^k \qquad (7)$$

which means that k = α in order to provide the same frequency dependence in the two models, and that

$$G_1 = G_{DC} t^a \cos\left(\frac{ap}{2}\right)$$

$$and \quad C_1 = G_{DC} t^a \sin\left(\frac{ap}{2}\right) \qquad (8)$$

Skin hydration measurements

In a practical measurement, the components of the electrical model in fig.1 can be treated as traditional electronic components, provided that the frequency dependence of Y_{POL} is kept in

mind. The total measured admittance, i.e. the overall conductance and susceptance will then be (if $G = G_{POL} + G_{DC}$):

$$Y = \frac{G^2 R_\infty + w^2 C_{POL}{}^2 R_\infty + G}{(1 + GR_\infty)^2 + (wC_{POL} R_\infty)^2} + i \frac{wC_{POL}}{(1 + GR_\infty)^2 + (wC_{POL} R_\infty)^2} \quad (9)$$

Hence, all the components in fig.1 influence on both the in-phase and quadrature part of the measured admittance. This influence is frequency dependent, however, and isolated measurements on some of the components can be achieved at a single frequency, by the right choice of measuring technique. Isolated measurements in this context means that the influence from other components is significantly less than the inevitable error in the measurements caused by the electronic circuitry.

The first step should be to determine the proper component to be measured, i.e. the component in the model which is influenced by stratum corneum hydration, but not influenced from e.g. sweat duct filling. The series resistance R_∞ is usually taken to represent the deeper skin layers (5). It should obviously also be modelled with a polarisation admittance, since the viable skin layers also have frequency dispersions, but since the conductance is totally dominating at low frequencies, a resistor should be a good approximation (6). The G_{DC} has been found to be influenced by both the sweat ducts and the stratum corneum itself (7, 8). There is presumably not any significant polarisation admittance in the sweat ducts, however, and this complex is thus the correct objective for the electrical measurements (9). Furthermore, since G_{POL} can not be separated from G_{DC} in a single frequency measurement, ωC_{POL} is the parameter to choose. The imaginary part of eq. 9 can easily be measured separately, by means of a synchronous rectifier or lock in amplifier. The measured susceptance is modulated by all three other components in fig. 1, however, but if

$$GR_\infty << 1 \qquad and \qquad wC_{POL} R_\infty << 1 \quad (10)$$

then the measured susceptance is certainly ωC_{POL}.

Salter also finds that the polarisation admittance is the proper object to be measured for hydration assessment (10). His “hydration index” is based on an empirically derived equation which includes the Cole parameter alpha. A measurement of alpha requires of course the use of multiple frequencies. This is redundant, however, when only the quadrature component of the polarisation admittance is measured.

Typical values

The resistivity of deeper tissues has been reported to be about 5 Ωm and furthermore frequency independent up to about 10 kHz, after which it slightly decreases (6). Yamamoto and Yamamoto (6) calculate the spreading resistance for a circular electrode with the well known formula

$$R_\infty = \frac{r}{2d} \quad (11)$$

which for e.g. d = 5 mm yields 500 Ω. This equation applies only for semi-infinite, homogeneous media, however, and calculations for human skin can hence only be considered as approximations. Mørkrid and Qiao (11) report R_∞ to normally be below 100 Ω with an electrode area of 0.6 cm^2, Rosell et al. (12) found R_∞ to be about 180 Ω with 0.55 cm^2, Grimnes (13) found segment impedance values with a four-electrode system between 50 Ω and 500 Ω, and Tregear report the values of R_∞ to lie between 100 Ω and 1 kΩ (14). Tregear furthermore denotes that these values are not inversely proportional to contact area, and hence cannot be given in terms of specific resistance.

The value of R_∞ can be dramatically reduced in a practical measuring system, when a three-electrode system is used (13, 15). Finite element simulations show that by placing the reference electrode in a three-electrode system in the vicinity of the measuring electrode, the measured volume of deeper tissue is almost reduced to zero at low frequencies (unpublished results). This means that using e.g. 1 kΩ for R_∞ would represent the absolutely worst case, and that the value in a properly chosen practical situation will be far less than that.

We measured the 88 Hz conductance and susceptance on the ventral side of the forearm on 8 volunteers with a three-electrode system, and found the mean conductance to be 0.22 ± 0.19 μS/cm^2 on non-treated skin and 6.69 ± 2.75 μS/cm^2 directly after soaking the skin with water (unpublished results). The corresponding susceptance values were 0.43 ± 0.30 μS/cm^2 on non-treated skin and 8.58 ± 1.81 μS/cm^2 after soaking (16). The results from 88 Hz susceptance measurements on 11 volunteers in a different study, were 0.95 ± 0.60 μS/cm^2 before treatment and a maximum of 5.13 ± 1.92 μS/cm^2 after treatment with a 150 mg/ml liposome formulation (17). All results were obtained with a concentric electrode system for the measuring- and reference electrodes, and a gel electrode as counter electrode (16, 18). The contact area of the measuring electrode was 0.2 cm^2.

Yamamoto et al. (18) report the admittance of non-treated skin on the forearm to be about 1 μS/cm^2 at 100 Hz, 5 μS/cm^2 at 1 kHz and 30 μS/cm^2 at 10 kHz. The values after tap water treatment were about 30 μS/cm^2 at 100 Hz, 100 μS/cm^2 at 1 kHz and 500 μS/cm^2 at 10 kHz. The measurements were made with an electrode area of 0.2 cm^2, and the given values are scaled accordingly. Using the mean measured phase angles from our experiment, i.e. 62.9° before treatment and 52.1° after soaking in water, on the 100 Hz data from Yamamoto et al., gives about 0.46 μS/cm^2 for the conductance and 0.89 μS/cm^2 for the susceptance of non-treated skin. The values for hydrated skin become 18.4 μS/cm^2 for the conductance and 23.7 μS/cm^2 for the susceptance.

The values obtained by Yamamoto et al. are somewhat higher than our data, and it should be noted that their data result from measurements on one person only.

Calculating measurement error

Let us take a closer look at worst case at 100 Hz, i.e. R_∞ = 1 kΩ, G = 18.4 μS/cm^2 and B = 23.7 μS/cm^2. Both our measurements and the Yamamoto et al. measurements were done with an electrode area of 0.2 cm^2, which gives G = 3.7 μS and B = 4.7 μS. Let us furthermore assume that the worst case value of R_∞ applies also for this contact area, taking into account that the series resistance is dramatically reduced when using low frequency and a three electrode system. The denominator of the imaginary part of eq. 9 then becomes

$$\left(1 + GR_\infty\right)^2 + \left(wC_{POL}R_\infty\right)^2 = \underline{1.0074} \qquad (12)$$

This means that even in the case of what Yamamoto et al. refer to as 100 % moisturised stratum corneum, and a relatively large series resistance, the deviation or error in the susceptance measurement is only 0.7 % if an electrode area of 0.2 cm^2 is used. Using e.g. 1 cm^2 would produce an error of 3.6 %, which is in the same range as what would be obtained by increasing the measuring frequency to a few kilohertz instead.

The conductance of tissue has a temperature dependence of about 2 % /°C (19). At low frequencies, the temperature dependence of the susceptance is small in comparison (20). Schwan and Takashima (21) assert that the dominant charge carriers in tissues are electrolyte ions, and that the temperature coefficient of the permittivity of tissue electrolytes is about -0.5 % /°C. This means that the susceptance measurements are less influenced by temperature drift than conductance measurements. Electrode polarisation has also been found not to influence on these measurements (13, 16)

Conclusion

Skin hydration measurements can be made at one single frequency under the following conditions:

1. The electrode area should be kept relatively small, e.g. below 0.5 cm^2.
2. The measuring frequency should be kept low, e.g. below 100 Hz.
3. A three-electrode system should be used, with the reference electrode close to the measuring electrode.
4. Only the susceptance should be measured, in order to avoid influence from sweat ducts.

When these conditions are met, the error in the measurements will be insignificant, and no extra precision in the assessment of skin hydration can with the present state of knowledge, be achieved by using multiple frequencies.

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