Electro-Optical Synergy (ELOSä) Technology for Aesthetic Medicine The significance of skin temperature monitoring during skin treatment

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Introduction

The use of pulsed electromagnetic energy, in the red and near infra-red (NIR) ranges of the electromagnetic spectrum, has become very common in recent years. The light sources that are used for these application are either monochromatic lasers in the wavelength range of 694 to 1064nm or filtered flash lamp sources that operate in the range of 600 to 1300nm using a wide filtered spectrum. The principle of operation of both types of pulsed light sources is very similar and it is based on applying the pulsed light to up to a few cm^2 of the skin. Light photons that interact with skin can be either scattered or absorbed. Since the scattering coefficient is typically 10 to 100 times larger than the absorption coefficient the photons quickly lose their original direction of propagation and penetrate through the skin in a diffusion-like penetration process. Photons are eliminated only after they undergo an absorption event in which all of the energy of the photon is absorbed in a chromophore of the skin (such as melanin, hemoglobin or oxy-hemoglobin) and all of its energy is transformed into heat. If enough heat is generated in the target tissue (hair follicle, shaft or bulge in the case of hair removal applications), and the target reaches a high enough temperature, it can be permanently damaged, thus enabling hair removal.

The understanding of light interaction with tissue is quite difficult and our ability to model it is somewhat limited for two main reasons:

1. The exact optical and thermodynamic properties of human skin are not well understood and it is quite complex to consider them in a realistic mathematical model.
2. Even if a good optical and thermodynamic model exists, one has to take into account the variation in optical and other properties from one patient to another and even variations of properties from one area of the skin to another on the same patient.

This situation becomes quite a challenge for the physicians treating patients for hair removal by using light. This is because they are in a very narrow operational regime that on the one hand ensures patient safety, and on the other hand uses a high enough energy to achieve a long clinical effect.

The only real form of energy that is being used is thermal, and the most important parameter for both safety and efficacy is temperature. Hence, the key question is: is their any reliable way by which we can measure temperature in real time (during the pulse)?

Monitoring skin surface temperature with infrared sensor appears to be the most logical way to measure skin heating.. This method is based on measurements of thermal infrared radiation from the skin surface in the range of 4µ to 12µ. Because of strong absorption of this radiation by tissue, the depth of skin temperature monitoring is limited by 10-20µ. This upper part of skin usually consists of a layer of dead cells, called *stratum corneum*. The thermal and optical properties of this layer differ from the living tissue and its temperature does not provide enough information for skin safety. An additional problem is that using skin moistening for effective thermal and optical coupling decreases the reliability of measured data.

Therefore, a new effective method of skin heating monitoring should be developed.

In spite of the disadvantages of the method, it helps to understand the influence of different treatment parameters on skin heating.

The connection between the impedance of the skin and its temperature

Skin impedance characterizes skin conductivity, which is electrolytic. That means that electrical current is conducted by the ions of salts contained in the tissue. As for all electrolytes, skin impedance is a function of temperature [1]. The ability to measure skin impedance during treatment creates the unique potential for monitoring of skin heating during the treatment. The dependence of skin impedance on temperature is described by the equation:

$$R = \frac{R_0}{1 + aT} \qquad \textbf{(1)}$$

Where a is the temperature coefficient of impedance, which is equal to 2% per °C [2] and R_0 is initial impedance.

Skin impedance can be calculated through the measurements of electrical current and voltage using Ohm law.

$$R = \frac{V}{I} \qquad \textbf{(2)}$$

Experimental evaluation of skin impedance behavior was conducted using an RF generator that produced 1MHz radio-frequency current with maximal power (P) up to 100W. RF current was used for both measurement and skin heating. RF energy was applied to an area (S) of 2cm^2. RF penetration depth (d) is 4mm. Volumetric RF energy density applied to the skin can be estimated as

$$E = \frac{P t_p}{S d} \qquad \textbf{(3)}$$

for pulse duration (t_p) of 150ms. Energy density absorbed by the skin is about 20J/cm^3.

Temperature behavior is described by the heat conductivity equation

$$c\, r \frac{dT}{dt} = \frac{P}{S d} \qquad \textbf{(4)}$$

Where t is time, c is the specific heat of the skin, which is a little bit lower than that of water and can be estimated as 3.6 J/g K, and r is skin mass density that can be assumed as 1g/cm^3. Solving Eq. 4 and inserting Eq.1 into it, the impedance decrease during the RF pulse is calculated by

$$\frac{R}{R_0} = \frac{1}{1 + \dfrac{a P t}{d S c\, r}} \qquad \textbf{(5)}$$

Skin impedance behavior measured in vivo and calculated with Eq. is presented in Figure 1.

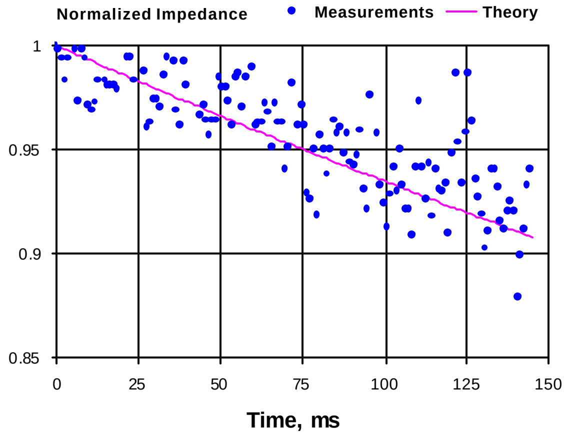

Figure 1. ***Impedance behavior during RF pulse.***

Measured data corresponds to eq. 2 predicting an impedance decrease of 10%. Thus, RF energy can be used for skin heating and for heating control.

Monitoring of skin heating by light using impedance measurements

Another very important question relates to the possibility of using skin impedance monitoring for control of skin heating by lasers or incoherent light sources broadly used in aesthetic medicine.

For the experiment, a filtered light produced byflash lamp was used. The light spectrum was 680-980mn. Output power was 2KJ and pulse duration 30ms. The irradiated area was 2cm^2 and penetration depth was about 2.5mm.

Pulses of light and RF energy were applied in parallel as shown **Figure 2**.

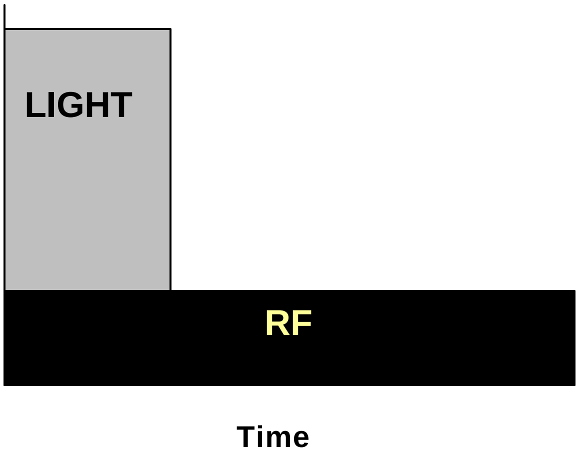

Figure 2. ***Schematic arrangement of Light and RF pulses***

Light pulse duration is 30ms with power of 2000W, while RF power is 100W, with a pulse duration of 150ms.

Skin impedance was measured during light and RF pulses. The results for dark skin (type IV according to Fitzpatrick) are presented in **Figure 3**.

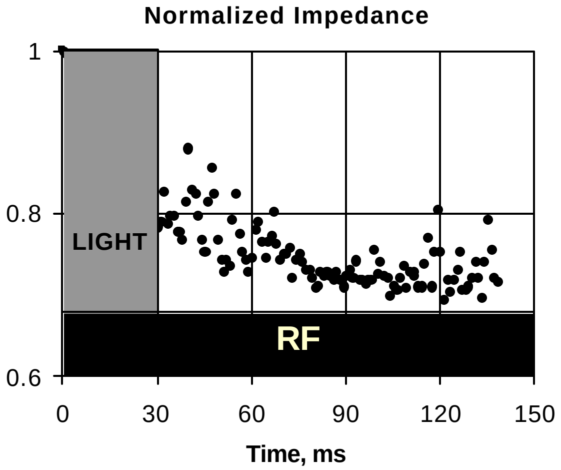

Figure 3. ***Dark skin impedance behavior under combination of light and RF pulses***

One can see that during the light pulse, impedance drops dramatically by 20% over the course of 30ms, because of the strong skin heating by light. Then, impedance decrease slows down. During the subsequent 120ms, impedance drops down by 8% due to soft heating by the RF energy.

It is a logical assumption that light skin should be heated less by light while the effect of RF energy stays the same. Therefore, impedance drop during the light pulse will be less than during the RF pulse impedance decrease will be about 8%.

Figure 4 presents impedance behavior under the same combination of light and RF pulses.

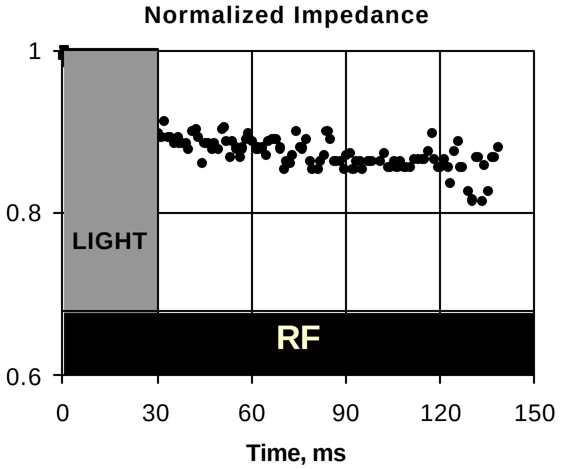

Figure 4. ***Light skin impedance behavior under a combination of light and RF pulses.***

Impedance decrease during light pulse is twice less for light skin (type I according to Fitzpatrick) than for dark skin. Impedance decrease during RF pulse is in the same range.

The results of measuring skin heating and impedence show a correlation between skin heating and impedance behavior. Higher heating causes a stronger impedance drop and vice versa.

Conclusion

Measurement of skin impedance provides information about skin heating that can be used for cooling and heating control.

The main advantage of the method is the ability it affords to measure skin impedance with sub-millisecond resolution.

Skin impedance gives integrated information from the entire skin zone where RF current flows. Penetration of the RF current can be optimized by choosing the geometry of electrodes. Penetration depth of the RF current can be estimated as the half distance between the two electrodes of a bipolar system.

References

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2. Francis A. Duck, Physical properties of tissue. Academic press limited, 1990, p. 173