Determination of the influence of dialysis on the human skin water content by means of THz spectroscopy

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***Abstract*—We report what is to our knowledge the first use of THz spectroscopy with the aim to determine the influence of haemodialysis on the human skin water contents. THz reflectance of the skin before and after dialysis was measured for ten patients and compared with other relevant parameters. It appears that the overall body hydration is not the main parameter determining the skin properties in the THz range.**

I. Introduction

In haemodialysis (HD), a reliable determination of the patients' body hydration is still an unresolved problem. During HD, potassium, urea, free water and other waste metabolites are removed from the blood, usually three times a week. Determining the optimal amount of water to be removed is important in order to both reduce unwanted side-effects and achieve the most efficient treatment. Bioimpedance spectroscopy (BIS) measurements can provide useful data, however the transformation of the raw bioimpedance data into volumes of different compartments of body fluid relies on a number of simplifying assumptions which may substantially decrease its accuracy. Consequently, up to now, the amount of fluid to be removed during the HD process is strongly dependent on the physicians' experience. Developing new methods providing at least an estimation of the body hydration state appears as highly desirable.

The THz radiation interacts strongly with water, and owing to its suitable wavelength, it can be used for imaging objects with a variable water contents including biological tissues. Several applications of this kind have been proposed and tested, the most of the effort being concentrated onto the diagnosis of malignant tissue [1], [2]. To our knowledge, no attempt to use THz radiation in conjunction with dialyzed persons has been reported so far.

II. Experimental details

A time-domain THz spectroscopy (TDTS) experimental setup used for imaging in reflection geometry under normal incidence was used for measuring the temporal waveforms of THz waves after interaction with the skin. Linearly polarized THz pulses were generated using femtosecond laser pulses and a biased photoconductive interdigitated emitter [3]. A 0.5 mm thick quartz window with a circular clear aperture of 10 mm was fixed horizontally in the focal plane of a convergent THz beam passing through a picarin lens. The lateral width of the THz beam at the window did not exceed 3 mm for the low frequency of 0.25 THz. Electro-optic sampling with a 1 mm thick ZnTe crystal was used for a phase-sensitive detection.

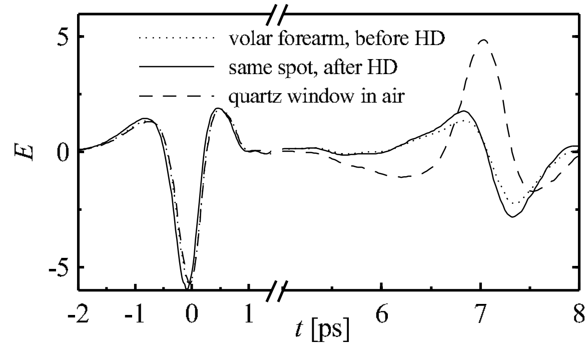

Fig. 1. Example of waveforms measured on the forearm. The first reflection comes from the air-quartz interface and does not depend on the presence of skin; its THz response is evaluated from the second one coming ≈7 ps later.

In the experiments, time-domain waveforms were recorded; the (top) air—quartz interface served as a reference, and the reflection off the bottom surface in contact with the skin was used for determining the THz properties of the skin. Its outermost layer, stratum corneum (SC), has a refractive index very close to quartz, and induced no reflection; the measured data are therefore linked to the properties of epidermis.

The measurements were performed with a group of ten stable HD patients. Both BIS and TDTS measurements were performed prior to and after two HD sessions for each patient, one after a 3 days' and one after a 2 days' interdialytic interval. Three locations were chosen for the TDTS measurements—two near the center of the volar forearm which had been indicated as convenient by previous studies [4], and the tip of the ring-finger of the non-access arm as a locus which can be easily located. This enabled us to study the influence of liquid removal from the body on the THz response of the skin and to examine the potential of the TDTS for assessing the body hydration state.

III. Data evaluation

An example of experimental data is shown in Fig. 1. The time-domain waveforms contain two distinct peaks; the first one, at $t = 0$, comes from the air—quartz interface and it can serve as a reference while the second one near $t = 7$ ps is due to the THz radiation reflected at the quartz—skin interface. By applying the Fourier transform on these two parts, one

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can calculate the reflectance spectrum. In this way, we found that within 0.3–1.3 THz, which covers the majority of the useful spectral range, the reflectance is essentially frequency-independent; as a result, three numbers are sufficient to parameterize each waveform: a time delay with respect to the reference Δt, a frequency-independent phase shift ϕ leading to the pulse reshaping, and an amplitude scaling factor r. From these values, the complex refractive index of skin can be calculated.

We model the matter involved in the interaction with THz radiation as a sequence of four layers: air, quartz, SC and epidermis separated by plane boundaries; their refractive indices are respectively 1, $n_{\mathrm{q}} = 2.12$, n_{sc} and N_{ep}. Since $n_{\mathrm{sc}} \approx n_{\mathrm{q}}$, the results do not provide access to n_{sc}; we can only roughly estimate the SC thickness from Δt if we presume that $n_{\mathrm{sc}} = n_{\mathrm{q}}$. Only epidermis is supposed to contain a significant amount of water, therefore only N_{ep} is assumed to have a non-zero imaginary part.

Denoting E_0 the incident electrical field, the reflectance of the air—quartz interface is a real number given by

$$r_1 = E_1/E_0 = (1 - n_{\mathrm{q}})/(1 + n_{\mathrm{q}}) \tag{1}$$

while that of the second interface is complex and equals to

$$r_2 = \frac{E_2}{E_0} = \frac{n_{\mathrm{q}} - N_{\mathrm{ep}}}{n_{\mathrm{q}} + N_{\mathrm{ep}}} \frac{4n_{\mathrm{q}}}{(1+n_{\mathrm{q}})^2} \mathrm{e}^{2\mathrm{i}\omega n_{\mathrm{q}}(d_{\mathrm{q}} + d_{\mathrm{SC}})/c} \tag{2}$$

where ω is the THz probing frequency and d_{q}, d_{SC} is the thickness of quartz and SC, respectively. By dividing these two reflectances we obtain the ratio between the two peaks:

$$\tilde{r} = \frac{r_2}{r_1} = \frac{n_{\mathrm{q}} - N_{\mathrm{ep}}}{n_{\mathrm{q}} + N_{\mathrm{ep}}} \frac{1 + n_{\mathrm{q}}}{1 - n_{\mathrm{q}}} \frac{4n_{\mathrm{q}}}{(1+n_{\mathrm{q}})^2} \mathrm{e}^{2\mathrm{i}\omega n_{\mathrm{q}}(d_{\mathrm{q}} + d_{\mathrm{SC}})/c}. \tag{3}$$

As the reflectance is almost constant, we can write $\tilde{r} = r\mathrm{e}^{\mathrm{i}(\phi + \omega\Delta t)}$ where r and ϕ are frequency-independent. The parameter Δt describes the time delay due to propagation of waves through the quartz and SC described by the phase term in (3). Finally, the refractive index of epidermis is equal to

$$N_{\mathrm{ep}} = n_{\mathrm{q}} \frac{A - r\mathrm{e}^{\mathrm{i}\phi}}{A + r\mathrm{e}^{\mathrm{i}\phi}} \quad \text{where} \quad A = \frac{1 + n_{\mathrm{q}}}{1 - n_{\mathrm{q}}} \frac{4n_{\mathrm{q}}}{(1+n_{\mathrm{q}})^2}. \tag{4}$$

IV. Results and discussion

The calculated values of the complex refractive index N_{ep}, measured at the volar forearms and fingertips, are shown in Fig. 2. Each of the vectors links two measurements of one patient at the same spot, pointing from the value before HD to that after HD. For the forearms, one can see a spread of values for the real part from 1.6 to 2.75 and from 0.4 to 1.7 for the imaginary part. However, the most distant data were obtained in a few measurements, and the bulk of the values are centred in a close vicinity of the value of $1.75 + 0.7\,\mathrm{i}$. The real part is in a good agreement with the value 1.7 reported for skin measured *in vitro* [5]. By contrast, our values of the imaginary part κ_{ep} are in all cases somewhat higher than those reported earlier, 0.25 ([5], [6]) and 0.4 [4].

The vectors corresponding to measurements on fingers are located clearly in a different portion of the complex plane, with the real part mostly higher—its mean value being close to 2—and the imaginary part lower than that at forearms. The values are markedly less scattered, and also the vectors are shorter owing to smaller differences occurring during the HD. The obtained real part is similar to the value stated in previous work [4] at 1 THz.

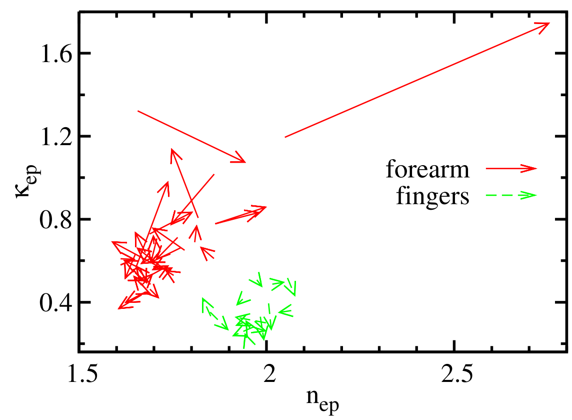

Fig. 2. Vectorial representation of changes in the complex refractive index of epidermis before and after dialysis for all patients.

The interpretation of the measurements, as shown in Fig. 2, in terms of body hydration status is not clear. Therefore, we have tried to find a relationship between changes of the refractive index during HD and the volume of fluid removed by ultrafiltration. One would expect that subjects from whom a low amount of fluid was removed would exhibit a small absolute value of changes in N_{ep}; however, the data obtained experimentally, either at the forearms or at fingertips, do not seem to follow this simple rule. Instead, they fall into all four quadrants of the complex plane, i. e., both increase and decrease in the real and imaginary part of the refractive index during HD was observed.

V. Conclusion

This small-scale study has revealed marked changes of the refractive index at the volar forearms during HD while the changes at fingertips are much smaller. The data collected in our measurements, including BIS and other patient-dependent parameters, did not allow us to identify the underlying mechanism which determines the THz refractive index N_{ep}.

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