

## Use of Quantitative Ultrasound to Measure Acoustic Properties of Human Skin

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The scattering of ultrasonic waves depends on the size, shape, acoustical properties and concentration of scatterers in the tissue. The spectrum of the ultrasonic backscatter can be used to characterize non-invasively the structural and mechanical properties of tissue. We intend to apply the custom-designed high-frequency ultrasonic scanner for the skin and cutaneous lesions characterization by evaluating their attenuating and scattering properties. In this pilot study, we have explored the possibility of extracting the human skin backscattering coefficient (BC) from the ultrasonic B-scans obtained *in vivo* at 20–30 MHz. The measured BC values of normal skin (dermis) agree well with the published data. We have found also that the spatial resolution of the BC determination using our scanner is sufficient (aprox. 1 mm<sup>2</sup>) to characterize small skin lesions and assess their penetration depth.

**Keywords:** attenuation coefficient, backscattering coefficient, dermis.

### 1. Introduction

Quantitative ultrasonic tissue parameters have been widely studied because of their potential to classify normal vs. abnormal conditions. Studies in the literature have shown that *in vitro*, both the backscatter and attenuation are sensitive to the strain [6]. Furthermore, several studies have shown that the backscatter coefficient and its frequency-dependence are related to the collagen distribution [5, 9]. Collagen is the major constituent of the dermis, therefore it can be anticipated that backscatter parameters should be equally sensitive to dermal modifications.

The detection of such modifications could be useful for tumor diagnosis, as well as for evaluation of the skin aging and cosmetic therapies.

Additional information related to skin composition and structure could be obtained using quantitative analysis of radiofrequency (RF) ultrasonic signals to calculate the frequency-dependent attenuation and backscattering coefficient. However, in most conventional applications of clinical ultrasonic scanners, only the peak-amplitude echogenicity is used to create the image. Moreover, the signal envelope detection destroys potentially useful information about the frequency dependence of acoustic properties of tissue comprised in the RF backscattered echoes.

In this pilot study, we explored the possibility of extracting information on the tissue backscattering coefficient from the RF ultrasonic B-scans, obtained by high-frequency skin scanner.

## 2. Materials and methods

### 2.1. Data acquisition

The skin scanner was developed in our laboratory. It performed a sector scan with the image frame rate up to 10 Hz. The transmitted signal and scattered echoes were sampled at 200 MHz frequency with 12 bits resolution. In this study we have used a 20  $\mu\text{m}$  thick spherical transducer (with 3 mm diameter, 8.6 mm focal length), made of the modified PZT 37 deposited on the PZT substratum using the thick-film technology (Ferroperm, Denmark). The received sequences were envelope detected and displayed. Simultaneously, the RF data were stored separately.

The measurements were performed in the dermis (thickness 0.5–3.5 mm) at a nape of a neck. Four healthy volunteers (3 women, 1 man; age about 27) participated in this study. Measurements of the slope of the attenuation coefficients were performed for the 8 cases (two measurements of the neck skin for each volunteer). Besides, one case of the basal cell carcinoma was investigated and the properties of the tumor-transformed tissue were determined. The last examination was performed in the Dermatology Clinic.

### 2.2. Backscattering coefficient

Analysis of the measured backscattered signals was performed to calculate the value and the frequency-dependence for the backscattering coefficient of the skin *in vivo*.

The ultrasound backscatter depends on the scattering and attenuating properties of the medium. The acoustic pressure amplitude of the wave propagating in the attenuating medium can be described as:

$$A(z) = A_0 \cdot \alpha_s(f) \cdot e^{-2\alpha_a(f) \cdot z}, \quad (1)$$

where  $\alpha_s$  represents the scattering coefficient and  $\alpha_a$  denotes the attenuation,  $f$  is the frequency,  $z$  – the penetration depth,  $A_0$  – the initial pressure amplitude.

Each of these parameters ( $\alpha_s$  and  $\alpha_a$ ) can be determined separately only if the remaining parameter is omitted or compensated. Experimental determination of the backscattering coefficient requires compensation of the attenuation coefficient value assumed or experimentally determined.

Both parameters, the backscattering and attenuation coefficient, were determined experimentally. The procedures were based on the calculations of the power spectrum of the ultrasonic waves scattered in the region of interest (ROI). To that end, each recorded RF echo was truncated by the Hanning window centered symmetrically within the ROI. The windowed signals were transformed by the FFT algorithm to calculate the power spectrum. The power spectra of the adjacent echoes were summed up to get the average power spectrum.

The backscatter coefficient was estimated using the expression proposed by INSANA *et al.* [3]:

$$\mu(f) = \frac{0.36R^2}{DL}W(f)E(f), \quad (2)$$

where  $D$  is the area of the transducer,  $L$  is the length of the gated segment,  $R$  is the distance between the transducer and the surface of the ROI,  $E(f)$  is the attenuation compensation function and  $W(f)$  is the normalized power spectrum, and the 0.36 coefficient results from the Hanning window application.

The normalized power spectrum  $W(f)$  was calculated from the measured spectra of echoes which were scattered from the tissue and from the plane reflector (reference spectrum). The planar surface was placed in water at the distance from the transducer corresponding to the locations of the ROI. Next, the values of backscattering coefficient were fitted by the least-squares method to power-law function  $\mu(f) = \mu_0 \cdot f^n$  over the 20–45 MHz bandwidth.

### 2.3. Compensation for attenuation

The wave propagating through the tissue is attenuated. The frequency-dependent attenuation losses come from the wave propagation between the gated region and the source as well as from propagation over the length of the gated region. If the effects of the attenuation are not properly taken into account, estimations of the scatterer properties will be incorrect.

The attenuation compensation function  $E(f)$  given in formula (2) can be described as:

$$E(f) = e^{4\alpha(f)(R+L/2)}, \quad (3)$$

where  $\alpha(f)$  is the overall frequency-dependent attenuation coefficient.

The attenuation coefficient  $\alpha(f)$  of the skin was determined using the spectral difference technique based on a comparison of the power spectra of the backscat-

tered signals after propagation through the medium. We used the following relation to determine  $\alpha(f)$ :

$$\alpha(f) = -\frac{-1}{4(x_2 - x_1)} \ln \frac{S_2(f)}{S_1(f)}, \quad (4)$$

where  $S_1(f)$ ,  $S_2(f)$  are the power spectra of pulses scattered along the penetration path and separated by the distance  $(x_2 - x_1)$ .

It is important to compensate for the effects of focusing when using the spectral difference technique, to estimate the attenuation coefficient from the backscattered signals emitted from the focused sources. Otherwise, in regions in front of the focus the attenuation is underestimated and in regions beyond the focus it is overestimated. The depth of the focus depends on the frequency and is shorter for the high-frequency waves and longer for the low-frequency waves. Thus when the acoustic pulse is drawing near the focus, the amplitude of the pulse high-frequency components increases faster than the amplitude of the lower frequencies and consequently, shifting up of the pulse spectrum occurs. For the diverging beam the opposite effect takes place. In our study we compensated the effect of diffraction or focusing using amplitude spectrum of echoes obtained from a rigid plane reflector located in water at various axial distances from the transducer.

The pressure amplitude variation of the selected frequency components for the ultrasonic pulse passing through the focus are clearly visible in Fig. 1a. For each spectral component, the correction curve that compensated the amplitude changes induced by focusing was calculated (see Fig. 1b). The attenuation coefficient was then computed using the corrected power spectra.

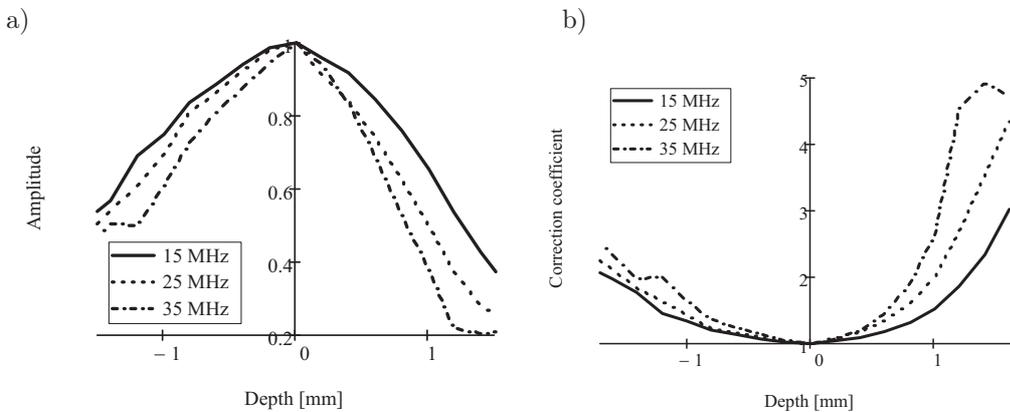


Fig. 1. The amplitude of the selected frequency components of the emitted pulse spectrum measured at different depth (a) (0 indicates focal plane position) and the corresponding correction curves (b).

### 3. Resolution

The spatial resolution of the backscattering coefficient determination technique is defined by the area of the region of interest. To find the size of the ROIs necessary to give the ROI size-independent backscattering and attenuation coefficients, we have used a homogenous part of a skin B-scan (skin scanner image). A set of concentric ROI squares was selected with the sides determining the window length (L) and the number of averaged power spectra calculated from the adjacent B-scan RF lines. All squares were analyzed for their respective average attenuation and backscattering coefficient.

### 4. Results

Figure 2 presents variation of both the attenuation and backscattering coefficients along with the increase of the ROI size. We have assumed that the coefficient evaluation is the ROI size-independent when the variation of its value is less than 5%.

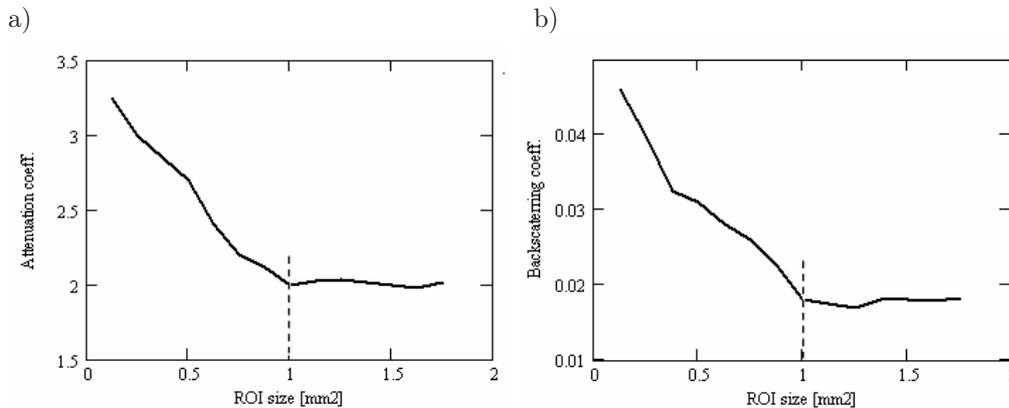


Fig. 2. Measured backscattering ( $\text{Sr}^{-1} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ ) and attenuation coefficients ( $\text{dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ ) versus the size of ROI.

The dotted line shows the minimal area assumed to give the coefficient independent of the ROI size. For the skin scanner it was found that the area equal approximately to  $1 \text{ mm} \times 1 \text{ mm}$  is the minimum area required to obtain correct values of the BC. It corresponds to  $1.5 \mu\text{s}$  time window and the averaging over 30 scan lines. For the attenuation coefficient the corresponding values were  $1.25 \mu\text{s}$  and 35 lines.

Figure 3 shows the attenuation coefficient for the neck dermis at the nape. To quantify the frequency-dependence of the attenuation, the linear fit was applied. The mean coefficient for the slope of the attenuation frequency-dependence was equal to  $2.2 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$  and ranged from  $1.73$  to  $2.38 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ .

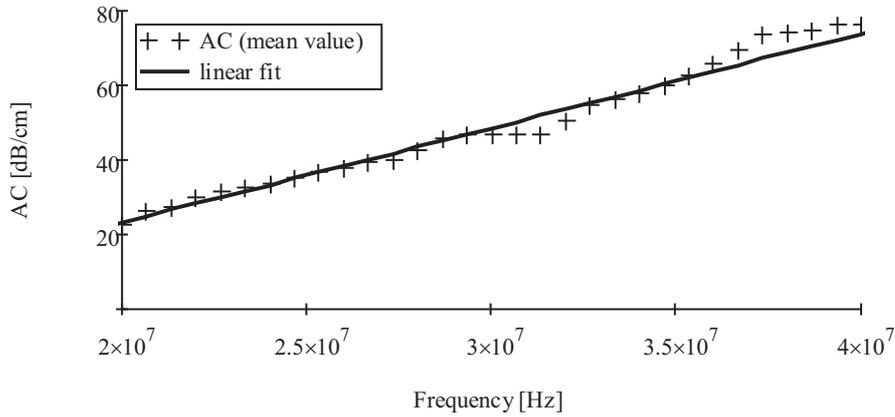


Fig. 3. Frequency dependence of the mean value of the attenuation coefficient for dermis (all eight cases were considered). Also the linear fit is shown.

The backscattering coefficient was estimated after compensation of the backscattering for attenuation. Figure 4 shows the backscattering coefficient of the neck dermis. The least-squares fit to the power-law function ( $\mu(f) = \mu_0 \cdot f^n$ ) over the 20 MHz–45 MHz bandwidth was applied. The  $\mu(f)$  varied between  $1.7 \cdot 10^{-4} f^{2.1} \text{Sr}^{-1} \cdot \text{cm}^{-1}$  and  $2.5 \cdot 10^{-4} f^{1.5} \text{Sr}^{-1} \cdot \text{cm}^{-1}$  and the averaged coefficient calculated using results of the eight cases was equal to  $2.1 \cdot 10^{-4} f^{1.66} \text{Sr}^{-1} \cdot \text{cm}^{-1}$ .

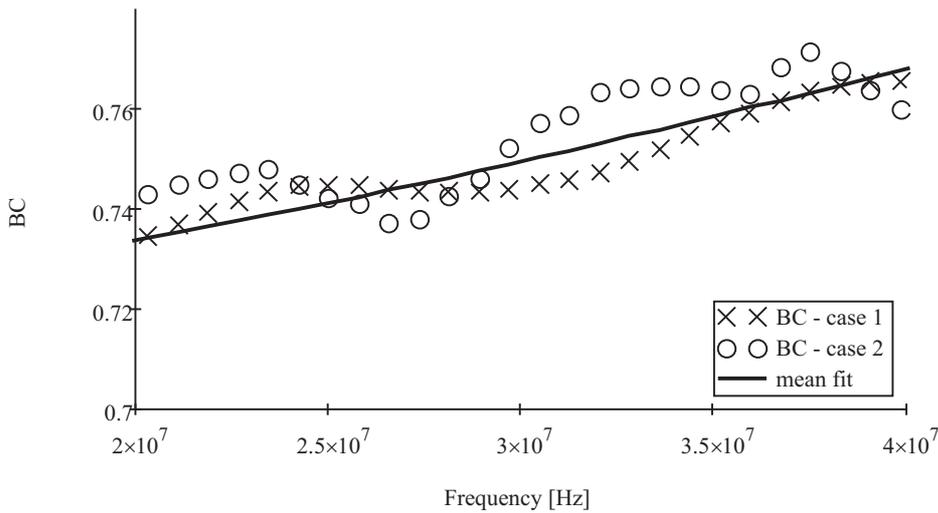


Fig. 4. Backscatter coefficient ( $\text{Sr}^{-1} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ ) of a dermis measured for two selected cases and the mean fit curve calculated using all eight cases (solid line).

Besides, the initial studies of the pathological tissue were performed. The RF-image of basal cell carcinoma located in the dermis was recorded and processed

along with the normal skin image. The basal cell carcinoma is the most common type of the skin cancer. Pathological modifications of the tissue structure and cells resulting from the tumour modifications alter the tissue-ultrasound interaction and can be potentially detected by evaluation of the attenuating and scattering properties of tissues.

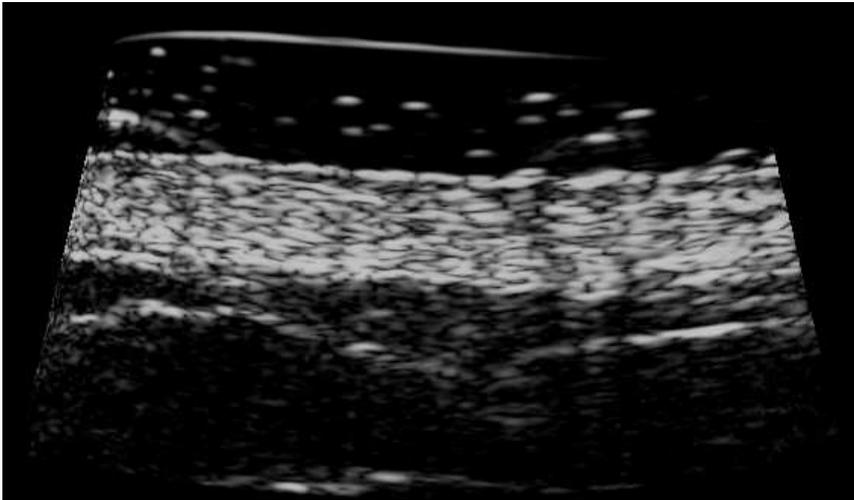


Fig. 5. B-scan image of the healthy skin for the patient with the basal cell carcinoma.

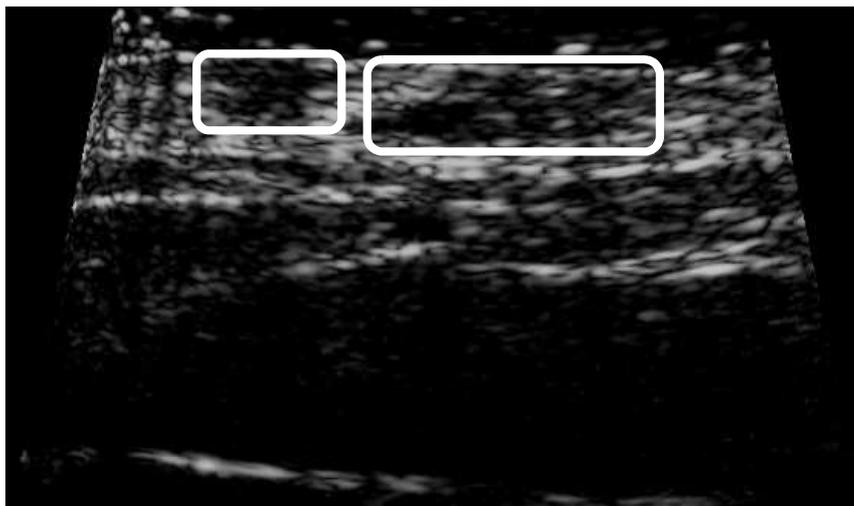


Fig. 6. Basal cell carcinoma located in dermis. Attenuation and backscattering coefficients were averaged over the marked areas (rectangles).

The linear fit of the frequency-dependent attenuation coefficient measured for the patient of the dermatology clinic in the healthy skin area shows the slope of

$1.98 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ . Figures 7a, 7b present the attenuation coefficients obtained for two dermis area where the basal cell carcinoma was identified (see Fig. 6). The linear fit with the slope of  $2.95 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$  and of  $4.34 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$  was obtained for these lesions' areas, respectively.

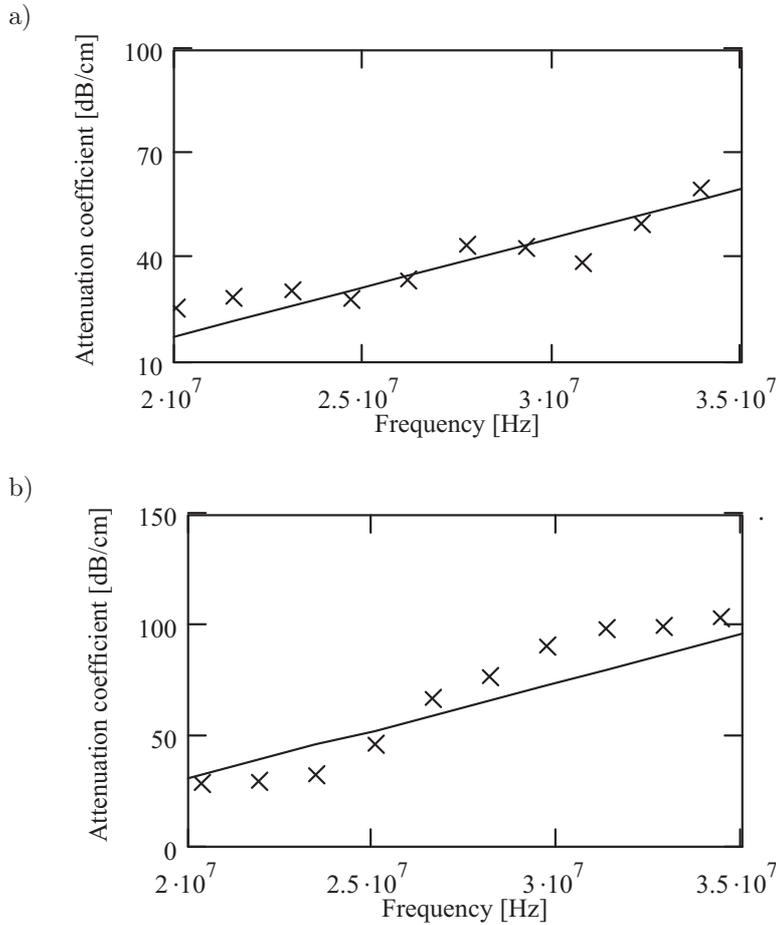


Fig. 7. Attenuation coefficient (crosses) together with the linear fit (solid lines) for the basal cell carcinoma, measured in (a) small and (b) big lesion presented in Fig. 6.

At the next step the BC was calculated for the area of healthy skin and for the skin where the pathological tissues were located.

The averaged backscattering coefficient determined from the area of healthy skin was similar to the value obtained from the dermis of healthy volunteers and equal to  $2.9 \cdot 10^{-4} f^{1.7} \text{ Sr}^{-1} \cdot \text{cm}^{-1}$ . Figures 8a and 8b show the frequency-dependence of the backscattering coefficient measured for the basal cell carcinoma. The BC were equal to  $3.1 \cdot 10^{-3} f^{1.9} \text{ Sr}^{-1} \cdot \text{cm}^{-1}$  for the smaller lesion and  $2.7 \cdot 10^{-3} f^{2.1} \text{ Sr}^{-1} \cdot \text{cm}^{-1}$  for the bigger one.

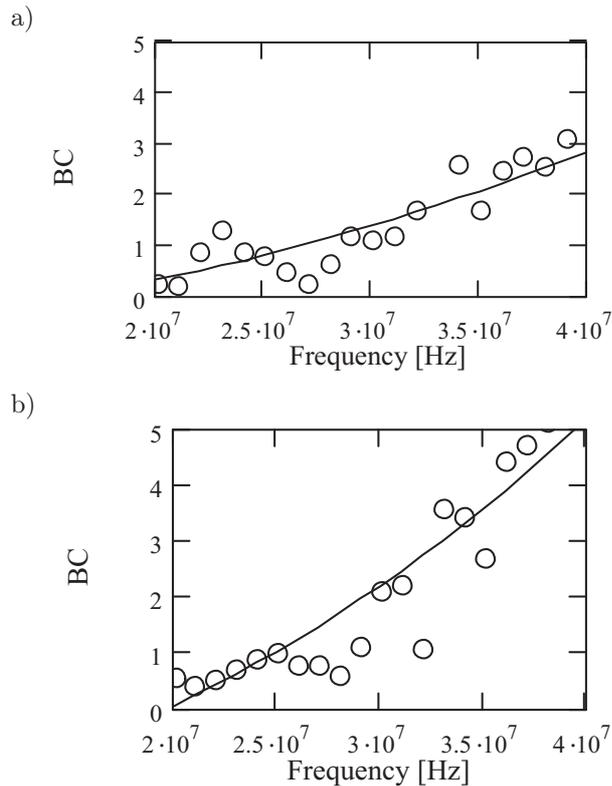


Fig. 8. Backscattering coefficient ( $\text{Sr}^{-1} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ ) of the dermis with the basal cell carcinoma (open circles) and power fit (solid lines), measured in the area corresponding to the (a) small and (b) big lesion.

## 5. Conclusion

This study reports the estimation of both the attenuation and backscattering coefficients in the human dermis *in vivo*. We have found that the RF B-scans of skin obtained with our high-frequency scanner can be used to the tissue characterization by evaluating its scattering and attenuation properties. The spatial resolution, defined by the size of the experimentally selected ROI (aprox.  $1 \text{ mm}^2$ ) is sufficient to characterize the small skin lesion and asses the lesion's penetration depth.

The measured BC and AC values determined for the healthy skin agree well with the published data [1, 2, 4, 7, 8]. The mean value of attenuation coefficient was found to be equal to  $2.2 \text{ dB} \cdot \text{cm}^{-1} \cdot \text{MHz}^{-1}$ . Attenuation was compensated when the backscattering coefficient was determined. The mean backscattering coefficient was found to be equal to  $2.1 \cdot 10^{-4} f^{1.6} \text{ Sr}^{-1} \cdot \text{cm}^{-1}$ . The frequency-dependence of the BC is similar to those published in other studies which analyzed the scattering in dermis.

Values of AC and BC obtained for the basal cell carcinoma differs considerably from those obtained for the healthy skin. Further ultrasonic study of skin lesions must be performed to evaluate the usefulness of the presented approach. Then the scanner and the proposed evaluation technique will be introduced into the medical practice for characterization of the specific skin lesions.

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