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Improvement of High Frequency Ultrasound Images by Correcting the Point Spread Function in a Coaxial Measurement with Optical Coherence Tomography
光干渉断層法との同軸計測における点広がり関数補正による高周波数超音波画像の改善

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1. Introduction

In recent years, much attention is paid to skin aging and precise evaluation method of skin becomes more important. Skin consists of epidermis (~ 0.3 mm) and dermis (~ 2.2 mm) [1]. There is few non-invasive methods for imaging the detailed structures of the skin such as collagen and elastin quality.

There are two major imaging modalities based on light and ultrasound for diagnosis of the skin. As representative examples, the former includes optical coherence tomography (OCT) [2], and the latter is high frequency ultrasound imaging (HFUS) [3]. The OCT is mainly used for the diagnosis of tissues with higher optical permeability such as eyeballs and has a spatial resolution of approximately 10 µm. Meanwhile, a penetration depth of the OCT is less than 1 mm. Therefore, there is an issue about the lower penetration. On the other hand, the HFUS has a lower resolution than that of the OCT, but the penetration depth is deeper.

In order to solve this issue, we propose a hybrid imaging by combining ultrasound and OCT in the present study. By coaxial measurment of the OCT and HFUS, the point spread function (PSF) of the HFUS image is corrected with respect to the OCT image. Thus the resolution of the ultrasound image is improved. The objective of the present study is to acquire skin images with higher spatial resolution (~ 10 µm) and deeper penetration (~ 2.5 mm).

2. Materials and Methods

2.1 Theory of Swept-source-OCT

Swept-source-OCT (SS-OCT) was an imaging modality by temporally dispersing the light with a wavelength tunable laser as the light source. As a feature, it was possible to secure a deep imaging area even for a highly scattering sample, and to decrease the measurement time. The light input was divideded into a sample side and a mirror side of the interferometer, and its reflection (scattering) was recombined and reached to a spectroscopic mirror. An interference phenomenon occured due to the optical path length difference between the sample side and the mirror side. And the tomographic image in the depth direction was acquired by performing frequency decomposition by Fourier transformation.

2.2 Coaxial Measurement of OCT and HFUS

The SS-OCT system (IVS-2000, santec Co.) and a laser light source (HSL-210-WR1, santec Co.) were used for the measurement. A PVDF-TrFE concave transduser with a central frequency of 75 MHz (Toray Engineering Co., Ltd.) was used for the HFUS. The transducer had a hole (Φ = 260 µm) at the center to arrange OCT and HFUS coaxially. **Table 1** shows imaging parameters.

Table 1 Parameters of OCT and FHUS

Parameters	OCT	HFUS
Center wavelength / frequency	1307 nm	75 MHz
Lateral resolution	13.4 µm	19 µm
Axial resolution	4.8 µm	26 µm
Focal length	3.8 mm	3.8 mm

Two signals from SS-OCT and Pulser-receiver were acquired by an A/D converter (ATS9350, AlazarTec) in a PC controlled by a Labview software as shown in **Fig. 1**. The scanning pitch of the stage controller was set to 2 µm. An artificial skin (Gunze Inc.) consisted of a three-layered structure such as silicone film, sponge layer, and gauze.

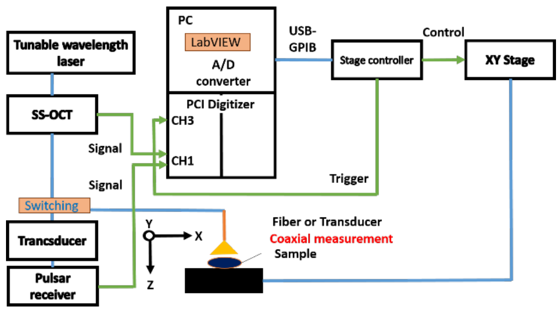

Fig. 1 Block diagram of the experimental systems

2.3 Correction Method of PSF of HFUS

First, edge detection was performed on the OCT image $f(x,z)$ and the HFUS image $g(x,z)$ reconstructed from the acquired data. A similar pattern was scanned from the array of detected high amplitudes, and the two images were aligned by the intensity based method.

Next, the PSF of the OCT image for the HFUS image was generated by a following Eq. (1).

$$h(x,z) = \frac{1}{2\pi r^2}\exp(-\frac{x^2+z^2}{2r^2}) \quad (1)$$

Variables x, z , were the position of the pixel in the ultrasound image, and the distance to the pixel position having the amplitude equal to or larger than the threshold value determined from that of the OCT image at an arbitrary position was defined as **Fig. 2**. The threshold values were decided from a difference of the OCT and HFUS edge's brightness close to the position.

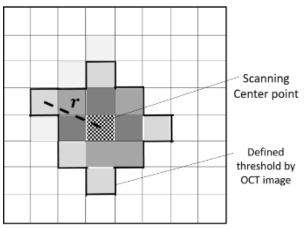

Fig. 2 Definition of r included $h(x,z)$

The relational expression of the two images was expressed by the Eq. (2).

$$g(x,z) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} f(x-\xi, z-\eta)h(\xi,\eta)d\xi d\eta$$
$$= f(x,z) * h(x,z) \quad (2)$$

The following Eq. (3) was deduced by applying Fourier transformation to the both sides of the equations.

$$\hat{F}(u,v) = G(u,v)\cdot\frac{1}{H(u,v)} \quad (3)$$

By calculating the inverse function that canceled the point spread as described above, it was possible to generate the ultrasound image with the improved spatial resolution similar to that of the OCT image. The ultrasound image generated by calculating the inverse Fourier transformed (IFT) of the Eq. (3) was utilized as the input of both Eq. (1) and Eq. (2), and then the corrected PSF was calculated iteratively until the error ε became enough minimum as expressed by the Eq. (4) . In particular, the cycle of calculation (Eq. (1) ~ Eq. (3)) was repeated in multiple times. Finally, the improved HFUS image was obtained in the spatial domain by the IFT.

$$\epsilon = \{|g(x,z) - f(x,z) * h(x,z)\ |^2\} \quad (4)$$

3. Results and discussion

Based on the reconstructed OCT image (**Fig. 3 (a)**) and the HFUS image (**Fig. 3(b)**), the ultrasound image was improved by using the proposed method (**Fig. 3(c)**). **Fig. 3(d)** shows the estimated PSF. The fiber structure of the artificial skin was observed more clearly so that blurs of the images were suppressed. While the structure of the atelocollagen layer containing much moisture was observed, the structure of the subsurface part such as silicone and gauze layer was not observed clearly.

The newly found issue was that the blurs existed and that couldn't be minimized even if the number of trials were changed. There are two solutions for the issue. The first one is to precisely calibrate the wavelength of the light during the reconstruction of the original image of the OCT so that the influence of the speckle due to scattering is suppressed to make the resolution closer to the theoretical value. As a result, PSF with higher accuracy could be generated. The other one is to weight the high luminance and the tissue parts to be focused on when generating the PSF, and narrow down the target.

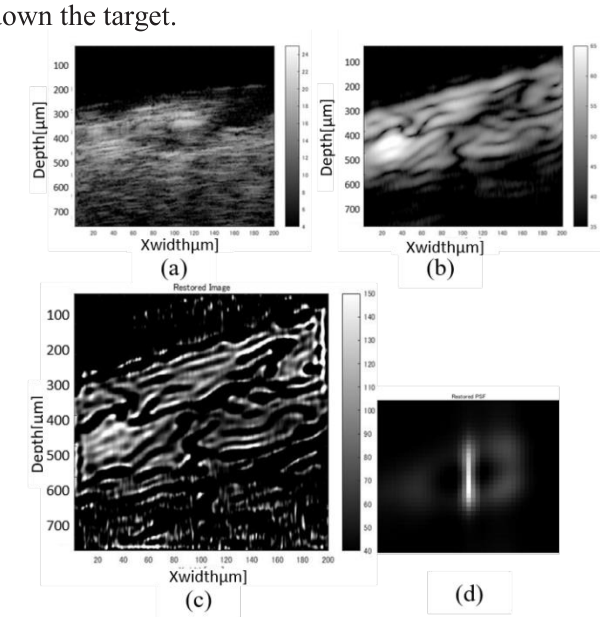

Fig. 3 (a) Reconstructed OCT image (b) Reconstructed ultrasound image (c) Ultrasound image generated using the proposed method (d) PSF mapping for 100 trials

4. Conclusion

A novel imaging method with high spatial resolution and deep penetration depth was achieved by correcting the PSF of the HFUS based on the OCT measurement. The proposed method was validated by the experimental result on the artificial skin. *In vivo* skin measurement is planned as the future works. The proposed method would contribute to the precise imaging of the skin.

References

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