Supplementary information for

Ultra-deep three-dimensional optical coherence tomography in turbid media

I. Data processing for matrix SS-OCT based 3D imaging

The overall workflow of data processing in matrix SS-OCT is illustrated schematically in Fig. S1. First, the inverse Fourier transform (IFT) is performed on the acquired interferometric spectral data to obtain depth-resolved signals with complex amplitudes. The unwrapped Hilbert phase difference $\Delta\varphi_{MZI}$ of the MZI signal and the phase difference $\Delta\varphi_R$ of the reference field are compensated respectively to A-lines and en-face images of all depths in two steps to reduce phase errors of the detected 3D complex fields. Depth-resolved reflection matrices are then formed from 3D backscattered fields based on both the sample scanning trajectory and the field scanning trajectory. The following matrix post-processing mainly consists of singular value decomposition and reconstruction for each depth-resolved matrix. Before the SVD processing applied to reflection matrices, a digital pinhole is used to filter out the most of the multiple-scattered contribution. Noted that the image quality heavily depends on the number of singular values considered for image reconstruction. To this end, correlation analysis of images pairs reconstructed from different number of singular values and corresponding remainders proposed in our previous study[1] is adopted to determine the optimal reconstructions. The reconstructed en-face images with the optimized number of singular values are then reshaped into a well-recovered 3D target.

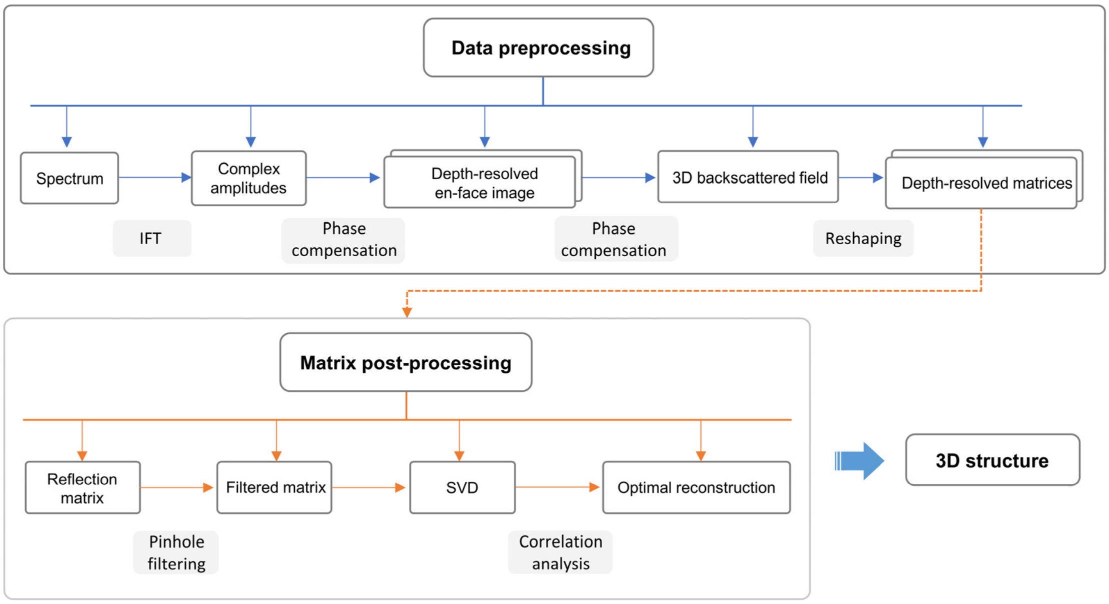

Fig. S1 Workflow of data processing for matrix SS-OCT based 3D ultra-deep imaging. Data preprocessing section: the process from the interference spectrum of the target signal to the construction of depth-resolved reflection matrices. Matrix post-processing section: SVD processing and the optimal reconstruction for each depth-resolved matrix to recover 3D structures.

II. Reconstructions of cell images below a paper tissue

The human osteosarcoma U2OS cells immobilized in the Petri dish below a paper tissue were imaged by both the matrix SS-OCT and the traditional SS-OCT. Due to slight tilting of the sample, it is necessary to perform the processing of projection summation of the en-face images (versus the increased depth) of partial imaging of the field of view (Fig. S2d) to obtain the overall distribution (Fig. S2c). The multiple-scattered light induced by the paper layer blurred the structures of cells imaged by the traditional SS-OCT (Fig. S2c). In comparison, satisfactory reconstructions similar to that acquired without the paper layer (Fig. S2b) were obtained from the matrix SS-OCT. Noted that, for samples like cells with low reflectivity and low contrast to the background, a thin scattering layer can result in a low SNR corresponding to the reconstruction threshold, while a thicker scattering layer will result in failed reconstructions.

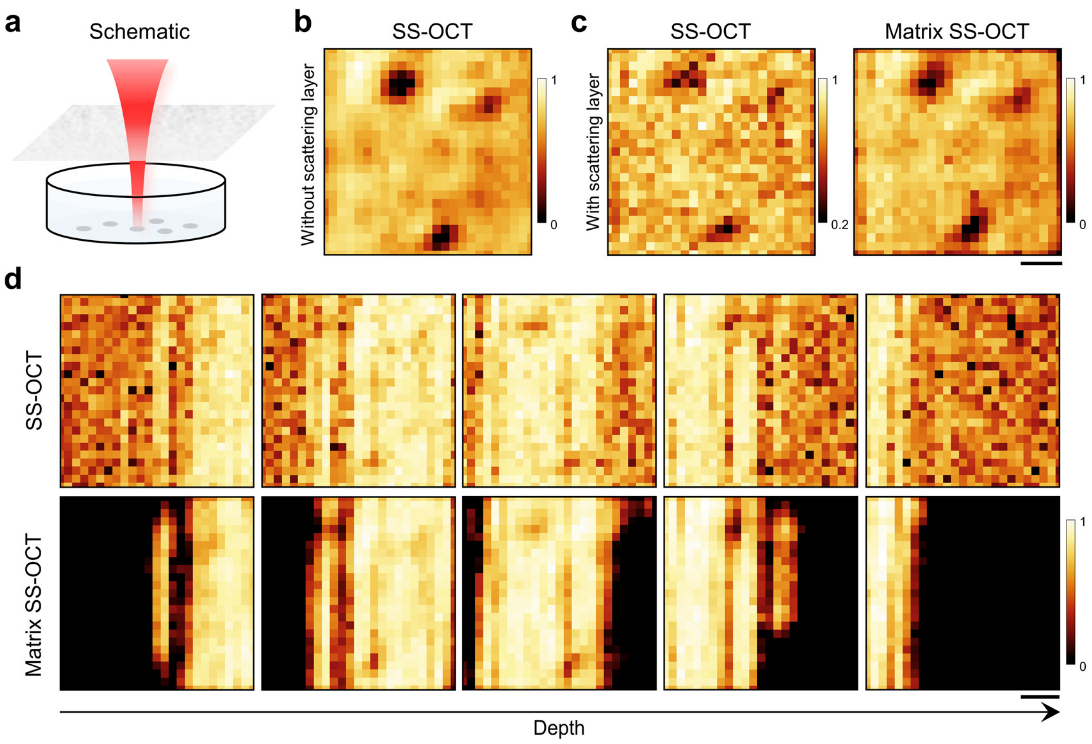

Fig. S2 Reconstructions of the en-face image of cells below a paper tissue. a Schematic of cell imaging. **b** En-face image of cells without a scattering layer obtained by the traditional SS-OCT. **c** En-face image of cells below a scattering layer, reconstructed from the projection summation of images versus the increased depth in **d** obtained by SS-OCT and matrix SS-OCT. Scale bar, 25 μm.

III. Details about the experiment of 3D imaging through a tissue phantom

Signal-to-noise ratio

The 3D distributed microbeads embedded in a tissue phantom are successfully reconstructed in Fig. 4. The signal-to-noise ratios (SNRs) of all the microbeads obtained by the traditional SS-OCT are plotted in Fig. S3a. It is noticed that only the first five microbeads are resolved with high SNRs (above 19 dB), while the remainders are far below the distinguishable SNR threshold (10 dB).

Scattering mean free path

The intensity attenuation of the ballistic signal I versus the penetration depth S can be expressed as:[2]

$$I = I_0 e^{-\frac{S}{l_s}}, \tag{1}$$

where, I_0 is the intensity of incident light, and l_s is the scattering mean free path. It is noticed that the intensity attenuation of the reconstructed microbeads can be expressed in exponential form, and thus l_s

can be estimated by fitting the exponential curve (Fig. S3b). Following curve fitting, the curve $I = I_0 e^{-0.0046 \cdot S}$ can be obtained, and the mean-square error of the fitting is 0.0016. Furthermore, the scattering characteristic of the tissue phantom can be indicated by $l_s = 1/0.0046 = 217.39\mu m$, and $L \sim$ 2.5mm ~ $11.5 l_s$, far exceeding the theoretical detection threshold of traditional OCT $\sim 7 l_s$ in previous study[3].

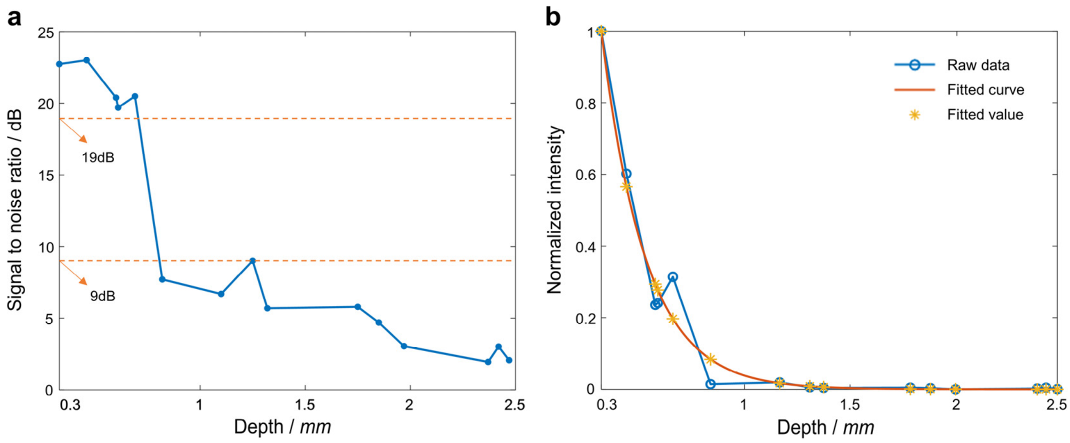

Fig. S3 Attenuation of the signal-to-noise ratio and the normalized intensity (raw data and the fitted curve) of all the microbeads versus the depth.

References

1 Yang, L. *et al.* Optimized number of the primary singular values for image reconstruction in reflection matrix based optical coherence tomography. *Optics Express* **30**, 2680-2692 (2022).

2 Akkermans, E. & Montambaux, G. *Mesoscopic physics of electrons and photons*. (Cambridge university press, 2007).

3 Badon, A. *et al.* Smart optical coherence tomography for ultra-deep imaging through highly scattering media. *Science advances* **2**, e1600370 (2016).