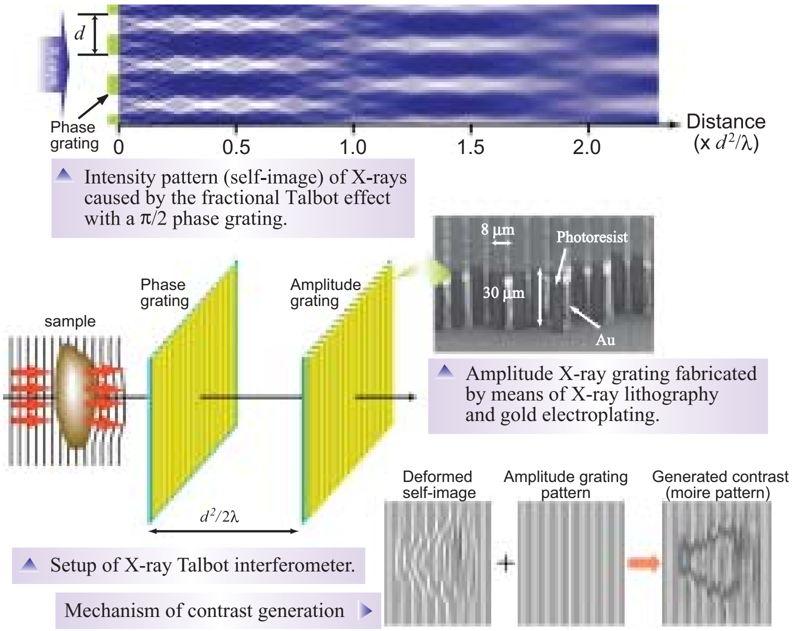
PHASE TOMOGRAPHY BY X-RAY TALBOT INTERFEROMETRY

X-ray phase imaging is attractive for the observation of weakly absorbing materials, such as soft materials and biological tissues. In the past decade, several techniques of X-ray phase imaging were reported and demonstrated [1]. Recently, we have developed an X-ray Talbot interferometer for novel X-ray phase imaging, including phase tomography. This method is unique in that transmission gratings are used to generate differential phase contrast (Fig. 1). Two gratings are aligned along the optical axis with a specific distance determined by the X-ray wavelength λ and the grating pitch d. The influence of refraction on the sample placed in front of the first grating is observed just behind the second grating. The detailed principle of the contrast generation is described in Refs. [2,3].

A critical key for realizing X-ray Talbot interferometry is the fabrication of an amplitude grating that is used as the second grating. However, it is not straightforward because a tall stripe pattern must be formed so that X-rays are fully blocked by the pattern. Gold gratings were therefore fabricated by X-ray lithography, which is advantageous in forming tall patterns, and gold electroplating, using the LIGA beamline of NewSUBARU, SPring-8. A SEM (scanning electron microscope) image of the grating is shown in Fig. 1. The pitch and pattern height were 8 µm and 30 µm, respectively.

Through the quantitative measurement of the differential phase shift by the sample, extremely high-sensitive three-dimensional imaging, that is, X-ray phase tomography, is attainable by X-ray Talbot interferometry [3]. We demonstrated this with biological samples using monochromatic undulator X-rays at beamline **BL20XU**. Figure 2 shows a rabbit liver tissue with VX2 cancer observed by X-ray phase tomography using a CCD-based X-ray image detector whose effective pixel size was 3.14 µm [4]. The cancerous lesion is depicted with a darker gray level than the surrounding normal liver tissue. The bright islands in the cancerous lesion indicate necrosis. The spatial resolution of this image was evaluated to be 15 µm. The phase tomogram approximately maps the difference in mass density, and the detection limit of the density deviation was estimated to be 1.3 mg/cm^3 from noise in the tomogram. Another observation result that was obtained for a mouse tail [4] is shown in Fig. 3. Bones and soft tissues, such as skin,



▲ Intensity pattern (self-image) of X-rays caused by the fractional Talbot effect with a $\pi/2$ phase grating.

▲ Amplitude X-ray grating fabricated by means of X-ray lithography and gold electroplating.

▲ Setup of X-ray Talbot interferometer.

Mechanism of contrast generation ▶

Fig. 1. Illustration showing principle of X-ray Talbot interferometry. The pattern of a phase grating is converted into an intensity pattern (self-image) by the fractional Talbot effect at a specific position from the grating. The self-image is deformed correspondingly to the refraction at the sample. A moiré pattern that involves information of the refraction is observed by the superposition of the deformed self-image and an amplitude grating.

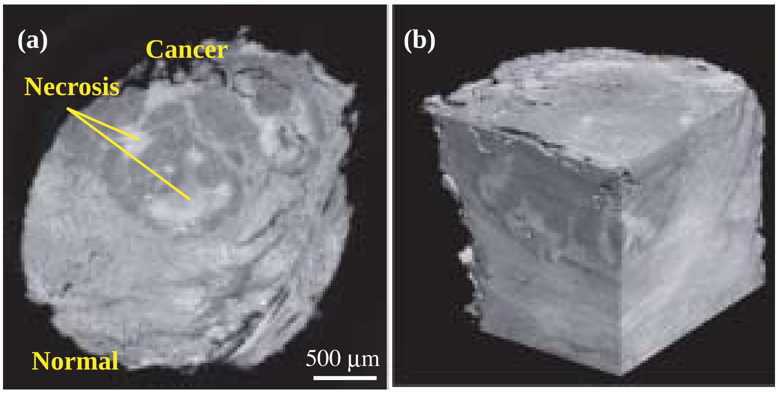


Fig. 2. Phase tomogram of rabbit liver tissue with VX2 cancer. The cancerous lesion is clearly differentiated from the surrounding normal tissue, and necrosis is depicted as bright islands in the cancerous lesion.

muscle, ligament, and intervertebral disc (cartilage), are depicted in the same view although some artifacts particularly caused by the bones remain. The images in Fig. 2 and Fig. 3 were measured with 12.4-keV and 17.7-keV X-rays, respectively, and the Talbot interferometer functioned up to 31 keV, which was confirmed by observing the sample identical to that in Fig. 3. The samples were courtesy of Drs. T. Takeda and J. Wu of the University of Tsukuba.

Three-dimensional observation by X-ray phase tomography is thus feasible for biological samples. Here, it should be emphasized that the implementation of X-ray Talbot interferometry outside synchrotron radiation facilities may be possible in the future. This is because a cone beam with a broad energy band width is available in principle owing to the use of gratings. It is also pointed out that this interferometer is much less sensitive to vibration than the Bonse-Hart crystal X-ray interferometer, and therefore it is easy to operate. Thus, imaging applications not only with synchrotron radiation as presented here but also with a compact X-ray source are expected for future practical application such as mammography, provided that a grating with a thicker pattern for higher-energy X-rays than the grating in the present study is fabricated with a wider effective area.

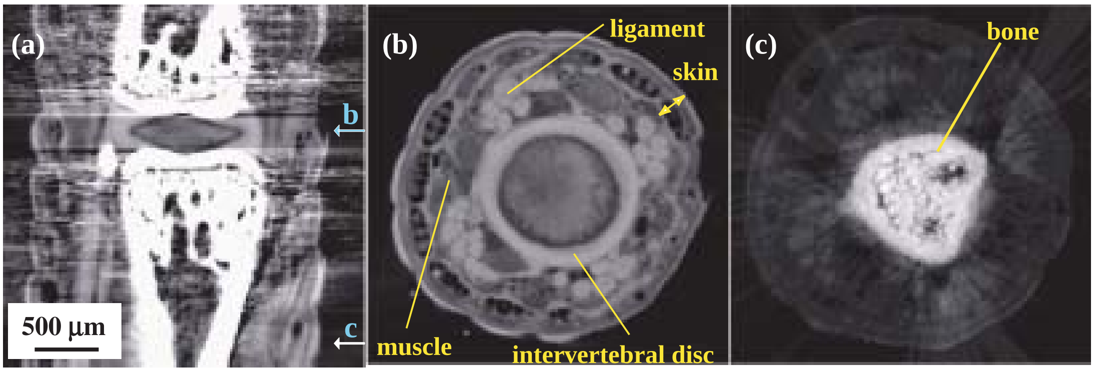


Fig. 3. Phase tomogram of mouse tail. The positions of axial images **(b)** and **(c)** are indicated by the arrows in the sagittal image **(a)**. Bones and soft tissues, such as skin, muscle, ligament, and intervertebral disc (cartilage), are depicted. The grayscale range of **(c)** is 3.5 times wider than that of **(b)**.

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