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Analysis of surface absorbed dose in X-ray grating interferometry

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ABSTRACT

X-ray phase contrast imaging using grating interferometry has shown increased contrast over conventional absorption imaging, and therefore the great potential of dose reduction. The extent of the dose reduction depends on the geometry of grating interferometry, the photon energy, the properties of the sample under investigation and the utilized detector. These factors also determine the capability of grating interferometry to distinguish between different tissues with a specified statistical certainty in a single raw image. In this contribution, the required photon number for imaging and the resulting surface absorbed dose are determined in X-ray grating interferometry, using a two-component imaging object model. The presented results confirm that compared to conventional radiography, phase contrast imaging using grating interferometry indeed has the potential of dose reduction. And the extent of dose reduction is strongly dependent on the imaging conditions. Those results provide a theoretical framework for dose estimation under given imaging conditions before experimental trials, and general guidelines for optimization of grating interferometry for those dose-sensitive applications.

1. Introduction

Since its feasibility firstly demonstrated about one decade ago [1,2], X-ray grating interferometry (GI) has become a widely used phase contrast imaging technique. The development of GI has been performed mainly using synchrotron radiation sources [3,4]. Later, Pfeiffer et al. demonstrated the feasibility of hard X-ray Talbot-Lau interferometry [5]. This is for the first time that quantitative phase images were produced with conventional X-ray tubes. Since that, X-ray GI has been applied to imaging a variety of soft tissues and organs [6–10]. The first scan of a human hand with an X-ray grating interferometer has shown significantly enhanced soft tissue contrast in the phase images, when compared with conventional attenuation data [7]. Investigations on native breast tissue have also shown that GI provides additional information to complement and improve the diagnostic process in the clinical setting with a radiation dose comparable to that of conventional mammography [9]. The feasibility of a Talbot interferometry system for the small joints of the hand has also been investigated with the potential for lower dose, simplicity and ease of clinical implementation [10]. These pioneering results demonstrate that compared to conventional absorption imaging, X-ray GI provides an increased contrast and a higher sensitivity to density changes in biomedical imaging applications, with a great potential of dose reduction. The extent of contrast improvement and dose reduction is dependent on the GI geometry, the photon energy, properties of the imaging detector and the sample. In the present work, we present a theoretical framework for a quantitative estimation of the surface absorbed dose in X-ray grating interferometry when applied to imaging biomedical samples or future patients. The framework reveals how the experimental conditions affects the surface absorbed dose, and can be of great use in optimization of X-ray grating interferometry for those dose-sensitive applications.

2. Method

Consider the object model shown in Fig. 1, where a feature tissue is embedded in its background material. The required photon number and the resulting surface absorbed dose will be calculated to statistically resolve the given feature with a specified certainty in a single raw image. In X-ray grating interferometry, an average angle of incidence at the interface between two different materials is assumed to be 45°. The following theoretical analysis assumes

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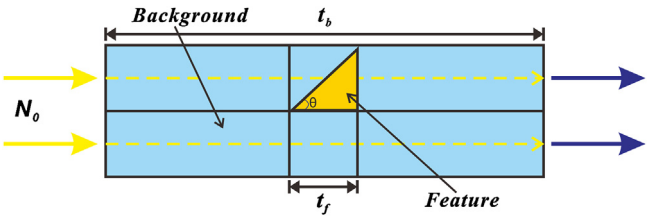

Fig. 1. Object model geometry. A feature tissue is embedded in the background material.

the use of Poisson noise characteristics, i.e., the standard deviation in each detector pixel is the square root of its photon count.

2.1. The required number of incident photons

In the following, we will find the required minimum number of incident photons to distinguish an embedded feature from its background with a $n\sigma$ statistical certainty at the thickest part using X-ray grating interferometry. Start with the condition that the refracted beam photon count is equal to the non-refracted beam photon count plus n times the square root of the non-refracted beam photon count [11],

$$N|_{x_g=\Delta x+z_m\theta} = N|_{x_g=\Delta x} + n \times (N|_{x_g=\Delta x})^{1/2} \tag{1}$$

For the object model shown in Fig. 1, the refracted beam photon count in X-ray grating interferometry is given by [2],

$$N|_{x_g=\Delta x+z_m\theta} = N_0 \exp\{-[\mu_b(t_b - t_f) + \mu_f t_f]\} \times \left[a_0 + a_1 \cos\left(2\pi \frac{\Delta x + z_m\theta}{p_2}\right)\right] \tag{2}$$

and the non-refracted beam photon count

$$N|_{x_g=\Delta x} = N_0 \exp\{-[\mu_b(t_b - t_f) + \mu_f t_f]\} \left[a_0 + a_1 \cos\left(2\pi \frac{\Delta x}{p_2}\right)\right] \tag{3}$$

where N_0 is the number of incident photons on the sample, $\mu_{b|f}$ is the linear attenuation coefficient of the background and the feature, respectively, $t_{b|f}$ is the thickness of the background and the feature, respectively, $a_{0/1}$ is the amplitude coefficient of the Fourier series, Δx is the relative transverse shift, p_2 is the analyzer grating period, z_m is the mth fractional Talbot distance, and θ is the refraction angle induced by the investigated feature.

On substitutions of Eqs. (2) and (3) into Eq. (1), we can obtain the expression of the required minimum number of incident photons to resolve the embedded feature with a $n\sigma$ statistical certainty,

$$N_0^{\min} = \frac{n^2}{\exp[-\mu_b(t_b - t_f) - \mu_f t_f]} \times \frac{[1 + V\cos(2\pi\Delta x/p_2)]}{4a_0V^2\{\sin[(\pi/p_2)(2\Delta x + z_m\theta)]\sin(\pi(z_m\theta/p_2))\}^2} \tag{4}$$

with the visibility $V = a_1/a_0$.

2.2. Surface absorbed dose

In the X-ray imaging regime, the surface dose is usually quoted when talking about doses to tissues in imaging. The surface absorbed dose is calculated by [11,12],

$$D = \frac{N_0 E_{ph}}{A} \frac{\mu_{en}}{\rho} \tag{5}$$

where E_{ph} is the photon energy, A is the cross-section area and usually defined by the pixel area, and μ_{en}/ρ is the mass energy-absorption coefficient.

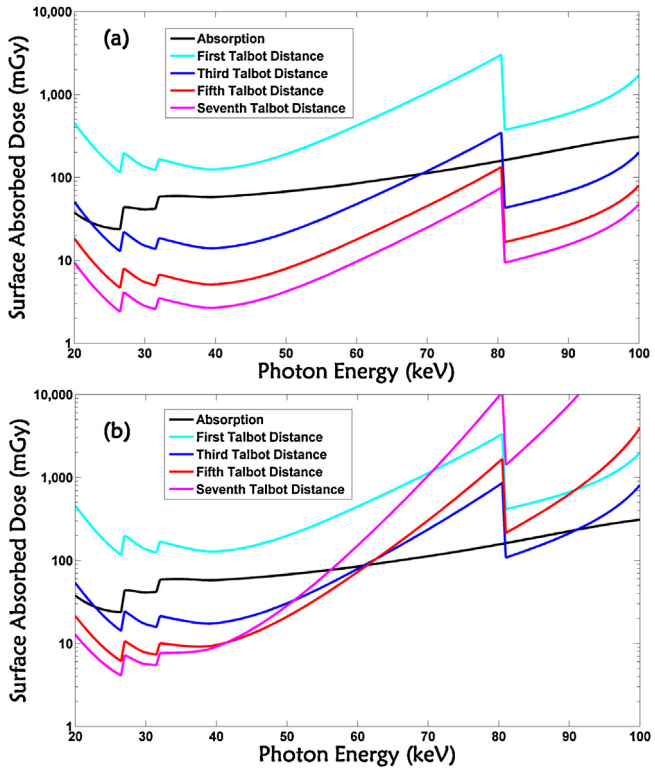

Fig. 2. Surface absorbed dose to resolve an embedded feature with a 1σ statistical certainty by using grating interferometry under monochromatic illumination. (a) Results corresponding to 1 μrad beam angular divergence. (b) Results corresponding to 5 μrad beam angular divergence.

On substitution of Eq. (4) into Eq. (5), we obtain the required minimum surface absorbed dose to resolve an embedded feature with a $n\sigma$ statistical certainty in grating interferometry,

$$D_{GI} = n^2 \frac{\mu_{en}}{\rho} \frac{E_{ph}}{4A\exp[-\mu_b(t_b - t_f) - \mu_f t_f]} \times \frac{[1 + V\cos(2\pi\Delta x/p_2)]}{a_0V^2\{\sin[\pi(2\Delta x + z_m\theta)/p_2]\sin(\pi z_m\theta/p_2)\}^2} \tag{6}$$

As shown in Eq. (6), the required minimum surface absorbed dose is inversely proportional to the square of the visibility, which emphasizes the importance of optimizing the interferometric visibility under given experimental conditions.

3. Numerical results

Numerical calculations are performed to provide a quantitative insight into the problem of surface absorbed dose in grating interferometry. We consider the case where a 1 mm-thick piece of adipose tissue embedded in the middle of a 5 cm-thick breast. The energy-dependent linear attenuation coefficients and mass energy-absorption coefficients of those tissues are taken from the National Institute of Standards and Technology [13]. The energy-dependent refractive indexes of the tissues are computed from knowledge of the tissue compositions and densities [14]. For the calculations, we used a grating period of 2 μm, and a detector pixel size of 50 μm, representative of typical experimental values. For the two gratings, we assume the use of a $\pi/2$-shifting phase grating at all photon energies, and an absorption grating with a 100 μm gold height. With respect to the detector, we assume the use of a photon-counting detector, which features a CdZnTe sensor of 1 mm thickness. The detective quantum efficiency (DQE) is taken into consideration in the dose calculations [11]. For a comparison, the

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