Sampling grating approach for X-ray differential phase contrast imaging

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Abstract: Grating-based X-ray differential phase contrast imaging (GDPCI) typically employs the phase-stepping technique to extract an object’s phase information. This method requires heavy radiation dosage and is time consuming. Another potential approach is the reverse projection (RP) method, which, however, relies on a synchrotron radiation source to obtain highly sensitive differential phase contrast (DPC) signal. Here, we present an alternative approach that enables the RP method to be used with a conventional X-ray source and substantially improves the sensitivity of the DPC signal by replacing the analyzer grating of the GDPCI with a sampling grating. This development represents a significant step towards obtaining fast and low-dosage DPC images in medical, biological, and industrial applications.

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1. Introduction

Conventional X-ray imaging relies on absorption as the contrast mechanism and has been applied in various fields, such as medical diagnostics, industrial non-destructive testing, and security checks. However, this technique cannot provide clear contrast images for soft tissues and other weak absorption materials. X-ray phase contrast imaging substantially overcomes this limitation and provides good contrast for such objects. Differential phase contrast imaging (DPCI) is a promising approach that records the phase gradient signals or refraction angles caused by the objects under study. Several types of DPCI techniques, such as diffraction enhanced imaging (DEI) [1, 2], GDPCI [3–7], code aperture imaging [8–10], and grating projection imaging [11] have been developed in the recent years. Although the configurations of these methods are all different, the physical principles involving the determination of refraction angles have some common characteristics [12, 13].

It has been shown that GDPCI holds great promise for practical applications owing to its compatibility with conventional low-brilliance X-ray tubes and capability to obtain quantitative absorption, DPC, and dark field of multi-contrast images [5]. A GDPCI, as shown schematically in Fig. 1, mainly consists of a source grating $G_0$, a phase grating $G_1$, and an analyzer grating $G_2$. $G_0$ divides a conventional X-ray source into an array of multiple line emitters. These emitters, being individually coherent, provide a sufficient spatial coherence for an interference fringe at the first order fractional Talbot distance of $d_z$ this distance is created downstream by $G_1$ even though the emitters are mutually incoherent sources in space. Passing through a phase object, the X-ray beam encounters a slight refraction. The refraction angle $\alpha$ is proportional to the gradient of the phase shift of the object, and is expressed as follows [14]:

$$\alpha = \frac{\lambda_c}{2\pi} \frac{\partial \varphi(x, y)}{\partial x},$$  \hspace{1cm} (1)$$

where $\lambda_c$ is the central wavelength of the X-ray, $x$ and $y$ represent two directions perpendicular to the optical axis $z$ in the Cartesian coordinates, and $\varphi(x, y)$ is the phase shift of the wavefront. Although the refraction angle is very small, especially for hard X-rays (typically in the order of micro-radians), it leads to a distortion in the interference pattern. A standard detector has insufficient resolution to resolve the fringe because the fringe has a period of only a few microns. Accordingly, an analyzer grating $G_2$ with period, duty cycle, and orientation similar to those of the fringe is introduced immediately in front of the detection plane. This grating acts as a mask to form resolvable moiré patterns. The intensity of the moiré patterns in each pixel $(m, n)$ can be expressed as a first order Fourier expansion [15]:

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\[ I(m,n) = a_0(m,n) + a_1(m,n) \cos \left( \frac{2\pi}{p_m} (m \cos \theta + n \sin \theta) + \phi \right), \quad (2) \]

and

\[ \phi = \frac{\lambda \cdot d_i \partial \varphi(x,y)}{p_2 \partial x}, \quad (3) \]

where \( \theta \) is a rotational misalignment angle between \( G_1 \) and \( G_2 \), and \( p_m \) is the period of the moiré fringes. \( a_0(m,n) \) is the DC term and represents the absorption contrast information, \( a_1(m,n) \) is the first Fourier coefficient that is related to the dark field information, and \( \phi \) represents the DPC information. Each of these quantities can be determined by using the phase-stepping technique.

The phase-stepping technique has an important limitation: its procedure for extracting the phase information leads to multiple exposures and excessive run times. This, in turn, leads to unacceptably high levels of radiation dosage, low mechanical stability, and slow imaging. To overcome this limitation, several interesting fast phase retrieval methods have been developed [15–20]. In 2010, Zhu P. P. et al. first introduced the idea of incorporating DEI to DPCI based on a synchrotron radiation source known as the reverse projection (RP) method. This method has great advantages in fast and low-dose applications, especially for phase contrast computed tomography [6]. However, it typically relies on a synchrotron radiation source to obtain a highly sensitive DPC signal. In this letter, we propose an alternative solution, whereby the analyzer grating in DPCI is replaced with a sampling grating. This approach uses a conventional X-ray source and substantially improves the sensitivity of the DPC signal.

2. Method

In DEI, information regarding the absorption and refraction of an object to be tested can be extracted from the two linear regions of its Bragg analyzer rocking curve. Furthermore, the intensity at the half-power points of the two side slopes can be accurately measured. The steeper the slope, the higher the sensitivity to the refraction signals. Similar to the case of DEI, the extraction of refraction information in the RP method relies on the “grating shifting curve” of the GDPCI with analyzer grating. The grating shifting curve can be acquired by scanning \( G_1 \) or \( G_2 \), without the sample, perpendicular to the grating orientation. As shown in Fig. 2(c), the intensity \( I \) versus the relative displacement \( x_g \) between \( G_1 \) and \( G_2 \), as recorded by a detector, can be expressed as:
\[
I_{xg}/d_t = I_0 S \left( \frac{x_g}{d_t} \right),
\]

where \( d_t \) is the first order fraction of the Talbot distance between \( G_1 \) and \( G_2 \), \( S(x_g/d_t) \) is the grating shifting curve, and \( I_0 \) is the intensity of the X-ray incoming in front of \( G_2 \). In addition, we define \( p_2 \) as the period of \( G_2 \), and \(-p_2/4d_t \) and \( p_2/4d_t \) as the two points on the steepest slopes of the grating shifting curve. With a sample placed in front of \( G_1 \), the incident X-ray beam refracts by a small angle \( \theta_x(x, y) \) on the plane of \( G_2 \), and the intensity of the sample image is given by:

\[
I_{xg}/d_t = I_0 \exp \left[ -\mu(x, y) \right] S \left( \frac{x_g}{d_t} + \theta_x(x, y) \right),
\]

where \( \mu(x, y) \) is the absorption of the sample. For a small value of \( \theta_x(x, y) \), \( S[x_g/d_t + \theta_x(x, y)] \) can be approximated to its first order Taylor expansion:

\[
S \left( \frac{x_g}{d_t} + \theta_x(x, y) \right) = S \left( \frac{x_g}{d_t} \right) + \frac{dS}{d(x_g/d_t)} \left( x_g/d_t \right) \theta_x(x, y).
\]

When \( x_g/d_t \) is set at the position of \(-p_2/4d_t \) and \( p_2/4d_t \), the intensity of the sample can be written as:

\[
I_{up}(x, y) = I_0 \exp \left[ -\mu(x, y) \right] \left[ S \left( -\frac{p_2}{4d_t} \right) \right. + \left. \frac{dS}{d(x_g/d_t)} \left( \frac{p_2}{4d_t} \right) \theta_x(x, y) \right],
\]

and

\[
I_{down}(x, y) = I_0 \exp \left[ -\mu(x, y) \right] \left[ S \left( \frac{p_2}{4d_t} \right) \right. + \left. \frac{dS}{d(x_g/d_t)} \left( \frac{p_2}{4d_t} \right) \theta_x(x, y) \right].
\]

Owing to the symmetry of the grating shifting curve, the intensities at \( \pm p_2/4d_t \) are equivalent and the derivative of \( dS/d(x_g/d_t) \) is opposite in sign. Under these conditions, the following equations hold:

\[
S \left( -\frac{p_2}{4d_t} \right) = S \left( \frac{p_2}{4d_t} \right),
\]

and

\[
\frac{dS}{d(x_g/d_t)} \left( -\frac{p_2}{4d_t} \right) = -\frac{dS}{d(x_g/d_t)} \left( \frac{p_2}{4d_t} \right).
\]

By combining Eqs. (7) and (8) and substituting Eqs. (9) and (10), the absorption image \( I_a(x, y) \) and refraction angle image \( \theta_x(x, y) \) can be resolved as:

\[
I_a(x, y) = \ln \left[ \frac{2I_0 S \left( \frac{p_2}{4d_t} \right)}{I_{up}(x, y) + I_{down}(x, y)} \right],
\]
and

$$ \theta_s (x, y) = C_g \frac{I_{\text{down}}(x, y) - I_{\text{up}}(x, y)}{I_{\text{down}}(x, y) + I_{\text{up}}(x, y)} $$

(12)

where $C_g = S(p_2/4d)/[dS(p_2/4d)/d(x_g/d)]$ is a constant.

Fig. 2. (a) Procedure for sampling fringe at the up-slope region. (b) Procedure for sampling fringe at the down-slope region. (c) Grating shifting curve.

However, a relatively large duty cycle of the analyzer grating, typically $\gamma_2 = 0.5$, cannot provide sufficient resolution at the two half-power points of the grating shifting curve. This reduces the steepness of the grating shifting curve at $\pm p_2/4d$ and thus decreases the sensitivity of the refraction signal. Consequently, an analyzer grating with a smaller duty cycle, called a sampling grating, is an alternative method for improving the resolution at the slope region of the grating shifting curve.

To verify the feasibility of the proposed approach, we present a numerical simulation of the procedure involved in the RP method for GDPCI. The simulation is performed by assuming that an X-ray beam passes through a protein pellet as a phase specimen. A CCD camera, with pixel size of 12 $\mu$m, is used as the image detector. All parameters are set in accordance with our previously reported experimental parameters [21, 22]. The protein pellet is assumed to have a radius of 350 $\mu$m and complex refractive index parameters of $\beta = 1.26495384 \times 10^{-10}$ and $\delta = 3.34127037 \times 10^{-7}$. Note that the phase factor $\delta$ is approximately three orders of magnitude larger than the absorption factor $\beta$. The effect on the DPCI of a sampling grating with duty cycle of $r_s = 0.25$ and an analyzer grating with duty cycle of $r_2 = 0.5$ are compared in the simulation. Considering the effect of the noise in the system [23], we choose the same noise distribution of the incident X-rays for the methods with analyzer grating and sampling grating, which means that the simulations process launch at the same anode current and exposure time for a fixed tube voltage.
Fig. 3. Simulation results of a protein pellet. (a) DPC image by a sampling grating. (b) DPC image by an analyzer grating. (c) Line profiles of (a) and (b). The rear sight of the picture width is 200 μm.

Figures 2(a) and 2(b) show the procedure for obtaining signals at ±p/4d, of the two slopes of the grating shifting curve [in Fig. 2(c)]. Figure 3 shows the simulation results according to Eq. (12). Figures 3(a) and 3(b) show the protein pellet phantoms acquired by a sampling grating and by an analyzer grating, respectively. Figure 3(c) demonstrates the quantitative difference between the two DPC images. A comparison between the results obtained by the analyzer grating and by the sampling grating reveals that the latter exhibits higher sensitivity compared to the former.

3. Experimental

Furthermore, we performed an experiment to demonstrate the proposed principle. Shown in Fig. 4 is a scanning electron microscope (SEM) image of a part of a sampling grating designed in our lab. The sampling grating was made of single crystal Si-based grating by filling Bi in it, a process involving photo-assisted electrochemical etching and micro-casting methods [24].

The imaging principle is schematically shown in Fig. 1, except that the analyzer grating G₂ is replaced with the sampling grating Gₛ. The illumination source is a conventional tungsten X-ray generator with an effective focus size of 1(horizontal) × 0.8(vertical) mm², operating at 60kV/4mA, corresponding to a mean photon energy of 30 keV. The π phase
The source grating $G_0$ is fabricated by the micro-casting process. The periods of these two gratings are $p_0 = 42 \mu m$ and $p_1 = 5.6 \mu m$ with a duty cycle of $r_0 = 0.25$ and $r_1 = 0.5$, and heights of $h_0 = 150 \mu m$ and $h_1 = 40 \mu m$. The distance between $G_0$ and $G_1$ is $l = 1.47 m$, and the distance between $G_0$ and $G_s$ is $d_t = 105 mm$, respectively. The $G_s$ is directly coupled to an ANDOR 2048 $\times$ 2048 (13.5 $\mu m$/pixel) area CCD by a 2X magnification fiber optical tape, and thus, the effective pixel size is $27 \mu m$.

Figure 5 shows the experimental results obtained by using the sampling grating. As shown in Fig. 5(a), the grating shifting curve is obtained firstly by scanning the phase grating along the transverse direction $x_g$ during one period without the sample. Images of $I_{up}$ and $I_{down}$, with the samples present, are acquired at the positions, marked by two green triangles in Fig. 5(a), corresponding to the center of linear regions on the up slope ($x_g = -p_2/4$) and down slope ($x_g = p_2/4$) of the grating shifting curve. The exposure time is 10 s for each original image and 10 images are superimposed to reduce the system noise. The absorption image and DPC image...
can be extracted by using Eqs. (11) and (12), respectively. Figures 5(b) and 5(c) show the images of a polyurethane tube with two different contrast mechanisms, demonstrating that the DPC signal has a strong advantage in resolving the difference in refractive indexes. Figures 5(d) and 5(e) show the results of a biological sample of two peanuts. Compared with the conventional absorption image in Fig. 5(d), the DPC image in Fig. 5(e) clearly provides more details, such as the textured structure of the peanut shell and kernel, which are hardly demonstrated in conventional absorption images.

4. Conclusion

In conclusion, we have demonstrated an alternative and simple method to implement X-ray DPCI by replacing the analyzer grating in GDPCI with a sampling grating. This method facilitates the realization of the RP technique using a conventional X-ray source and avoids excessive run time due to the multiple exposures required during phase stepping. The simulation and experimental results show that, in a GDPCI setup, the DPC signal obtained using a sampling grating is significantly better than that obtained using an analyzer grating. Owing to the excellent linearity of the grating shifting curve in the vicinity of $\pm \pi/4$, the mechanical precision and stability are considerably improved.

It should be pointed that the signal photons on the detector will be decreased with reducing the duty cycle of analyzer grating. RP method is implemented at the linear region in the grating shifting curve at both sides of the halfway position. Therefore, a grating with its opening the size of the linear region is appropriate. In the future, we hope to perform a rigorous analysis the characteristics of the GDPCI setup with different duty cycle of sampling grating such as noise, resolution, sensitivity and signal photons utilization.

We believe that the results presented here provides an alternative concept for promoting the development of GDPCI, and, in future, would provide faster imaging with lower radiation dosage in fields such as medical diagnoses or in vivo phase contrast imaging.

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