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Mesh-based phase contrast Fourier transform imaging

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ABSTRACT

Traditional x-ray radiography is limited by low attenuation contrast in materials of low electron density. Phase contrast imaging offers the potential to improve the contrast between such materials, but due to the requirements on the spatial coherence of the x-ray beam, practical implementation of such systems with tabletop (*i.e.* non-synchrotron) sources has been limited. One phase imaging technique employs multiple fine-pitched gratings. However, the strict manufacturing tolerances and precise alignment requirements have limited the widespread adoption of grating-based techniques. In this work, we have investigated a recently developed technique that utilizes a single grid of much coarser pitch. Our system consisted of a low power 100 μm spot Mo source, a CCD with 22 μm pixel pitch, and either a focused mammography linear grid or a stainless steel woven mesh. Phase is extracted from a single image by windowing and comparing data localized about harmonics of the mesh in the Fourier domain. The effects on the diffraction phase contrast and scattering amplitude images of varying grid types and periods, and of varying the width of the window function used to separate the harmonics were investigated. Using the wire mesh, derivatives of the phase along two orthogonal directions were obtained and combined to form improved phase contrast images.

1. Introduction

Conventional x-ray imaging utilizes differential attenuation, which arises from differences in the thickness, composition, and density of the imaged objects to generate contrast. However, attenuation variations are often quite small for low atomic number (Z) materials such as soft tissue. Higher contrast can often be obtained from phase information because the relative change in phase of x rays passing through an object is much larger than the relative change in intensity due to absorption [1]. Contrast is generated due to the difference in refractive index n of the materials [2,3]

$$n(x, y, z) = 1 - \delta(x, y, z) + i\beta(x, y, z) \tag{1}$$

where (x, y, z) denotes position, β is proportional to the absorption coefficient, and the real part, 1-δ describes the relative phase delay of x rays propagating through a material. Upon exiting the object, the x rays have accrued a total phase that can be approximated as the integral along rectilinear x-ray trajectories. Assuming the x rays are propagating along the z axis, the phase accrued in the x-y plane at the exit face of the object can be represented as

$$\phi(x, y) = -k\int\delta(x, y, z)\mathrm{d}z, \tag{2}$$

where k=2π/λ is the wave number and λ is the wavelength of x rays. For an x-ray energy range of 10–100 keV, δ ranges from approximately 10^{-6}–10^{-8}, roughly a 1000-fold enhancement over β, which ranges from 10^{-9} to 10^{-11} [4]. A variety of techniques have been developed to obtain phase contrast for low Z materials, including grating-based [5–7] and propagation-based imaging [8–11].

Propagation-based imaging measures the intensity changes that result when x rays are deflected by refractive index variations within the tissue. An x ray exiting the sample at a point (x, y) will deflect in the direction of highest phase gradient $\nabla_\perp\phi$ where $\nabla_\perp$ denotes the gradient in the plane transverse to the z axis. The angle of deflection is approximately given by

$$\boldsymbol{\theta}(x, y) = \frac{1}{k}\nabla_\perp\phi(x, y). \tag{3}$$

In most samples, this produces edge enhancement since the phase varies most rapidly at the edges of features. This is illustrated in Fig. 1.

A drawback of propagation-based imaging techniques is that since enhancement tends to appear as narrow fringes at the edges of sample features, x-ray sources with high spatial coherence are required to preserve these intensity fringes. Therefore, propagation-based techniques tend to be most relevant to sources such as large synchrotron facilities, which are not widely available, and small, low power microfocus sources, which provide low intensity and therefore require long integration times to reduce noise in the image.

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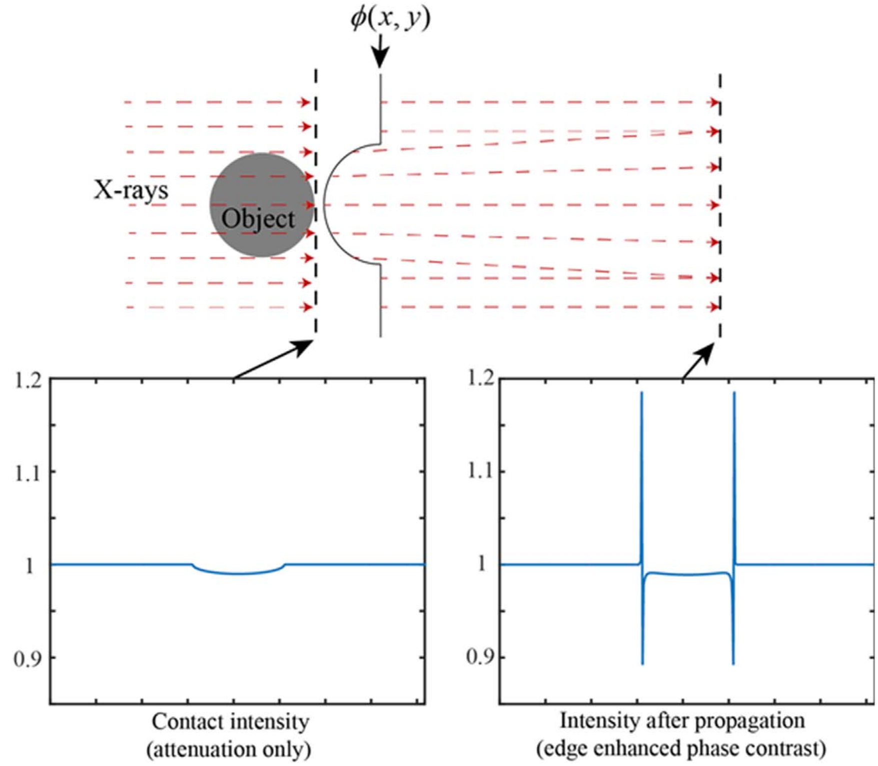

Fig. 1. Propagation based phase contrast. (top) X rays passing through an object are deflected and attenuated. The phase immediately after the object is $\phi(x, y)$. Rays strike a displaced detector at the right where, due to deflection, they accumulate in some regions producing higher intensity. (bottom) Intensity profiles (background intensity is normalized to unity) of a sphere in the contact plane and after propagation.

For grating-based phase imaging, three gratings are typically employed. First, an amplitude grating segments the source into small regions to generate the necessary coherence. Second, a phase grating is placed in the beam path to generate a Talbot image consisting of periodic fringes of the same period as the phase grating. An object placed in the x-ray beam path will impart a phase, which alters the fringe position. A grating with an extremely fine pitch d (on the order of microns) is necessary because the detector must be placed at the Talbot plane a distance $z = 2d^2/\lambda$ from the phase grating, which would be unreasonably far without small d. Since a fine grating generates fringes that are smaller than the detector pixels, an analyzer grating placed immediately before the detector is used to measure fringe displacement. The disadvantage of this method is that it requires high quality gratings, careful alignment, and multiple images taken while shifting the analyzer grating.

To overcome these practical problems with grating based techniques, an alternative, grid-based method was recently proposed by Bennett et al. [12,13], which allows grids of much coarser pitch to be used, such as those typically used in radiography, which are readily available at low cost. The experimental setup for this technique is shown in Fig. 2.

The grid-based imaging system can be understood as a modification of propagation-based phase imaging [14]. Consider x rays deflected by a sample and propagating an effective distance d_{eff} from object to detector (the physical distance is scaled to take into account off axis propagation). Due to this deflection, the x ray strikes the detector at a position displaced an amount $\Delta\mathbf{x}$ from where it would have hit with no object in place. This can be expressed, for the geometry of Fig. 2, as

$$\Delta\mathbf{x}(x, y) = d_{eff}\,\boldsymbol{\theta}(x, y) \tag{4}$$

where $\boldsymbol{\theta}(x, y)$ was given in Eq. (3). The purpose of the mesh is to introduce a high intensity contrast from which the deflection can be readily determined. Two images are recorded, the first with only the mesh in place and the second with both the mesh and patient or phantom in place (so only the second image imparts dose to the patient). The shift of the x rays will cause the image of the mesh edges to shift, and by comparing the two images the deflection can be inferred. In addition, if the object scatters x rays into a small spread of angles about the deflected direction, the edges of the mesh will blur in the resulting image. Although these concepts can be applied to general amplitude masks of any design, the image processing required to retrieve $\Delta\mathbf{x}$ and scattering amplitude is greatly simplified through the use of periodic meshes and Fourier transform-based processing.

We consider binary amplitude masks whose transmissive portions impart a uniform phase (if any) to the x ray and whose opaque portions block the x rays entirely so that the mask itself does not appreciably deflect the rays. Without a phantom in place, the image at the detector is simply a magnified version of the mesh itself. With the phantom in place, as illustrated in Fig. 2, the incident x rays are deflected by phase variations induced by the tissue, and the mesh image captured by the detector will be distorted. In addition, the tissue itself will attenuate incident x rays. The goal of the processing described in the remainder of this section is to extract separate images illustrating the x-ray attenuation and deflection by the object.

For a one dimensional periodic grid, the transmissivity can be written as

$$G(\mathbf{x}) = \sum_{n=-\infty}^{\infty} c_n \exp(\mathrm{i}\mathbf{g}_n \bullet \mathbf{x}), \tag{5}$$

where $\mathbf{x} = (x, y)$, c_n is a complex-valued constant, $\mathbf{g_n}=\frac{2\pi}{p}n\hat{\mathbf{g}}$, n is an integer, $\hat{\mathbf{g}}$ is a unit vector perpendicular to the grid lines and p is the grid period. For a mesh, *i.e.* a grid that is periodic along two orthogonal directions, the sum in Eq. (5) is extended over an additional integer index m and $\mathbf{g_{m,n}}=2\pi(\hat{\mathbf{g}}_1 n/p_1+\hat{\mathbf{g}}_2 m/p_2)$ where $\hat{\mathbf{g}}_1$ and $\hat{\mathbf{g}}_2$ are unit vectors perpendicular to the mesh lines and p_1 and p_2 are the periods. Without the phantom in place, the mesh image is captured by the detector and the Fourier transform produces delta-function-like spikes at the grid harmonics. With the phantom in place, the rays leaving the mesh strike