Phase Imaging Using Focused Polycapillary Optics

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Abstract

Contrast in conventional imaging of soft tissues is often limited due to the very similar attenuation of tissues to be distinguished. Phase contrast techniques can enable discrimination of tissues with similar attenuation. A major limitation to the widespread adoption of phase-contrast techniques is that for tabletop sources the required degree of coherence generally requires a small (10 to 50 µm) source. In this work, a polycapillary optic was employed to create a small virtual source from a large spot rotating anode. Phase contrast images obtained with two optics and several pinholes have been analyzed and preliminary results obtained for quantitative phase measurements.

Keywords: Phase imaging, propagation-based, edge enhancement, radiography, polycapillary optics, virtual source.

1. Introduction

Since the discovery of X rays by Roentgen, absorption-based images utilize the attenuation properties of the object being imaged to generate a two dimensional map. Such types of images produce high contrast for objects where the electron density differences are substantial, e.g., bone and flesh, but result in poor contrast for soft tissue imaging where the tissues exhibit similar attenuation, e.g., glandular tissue and infiltrating ductal carcinoma in breast imaging. However, X rays can accumulate significant differential phase delay even in weakly absorbing materials. If the change of phase of X rays can be imaged, tissues which would normally be indistinguishable under attenuation contrast can be distinguished. Phase contrast imaging renders different phase delays as intensity variation in the detector plane. The amount of attenuation and phase delay of a given material can be described by its complex-valued index of refraction,

\[ n = 1 - \delta - i\beta \]  \hspace{1cm} (1)

where \(\beta\) and \(\delta\) are real-valued parameters describing the absorption and phase delay of X rays, respectively.

For X-ray energies of 10 to 100 KeV, \(\delta\) of tissues is about \(10^{-6}\) to \(10^{-8}\) and \(\beta\) is about \(10^{-9}\) to \(10^{-11}\), so the phase delay term is approximately 1000 times greater than the absorption term. Because \(\delta\) remains relatively large at high energies where absorption is low, phase imaging may give the possibility of dose reduction at these energies.

Several phase contrast methods\(^2\) have been investigated to visualize phase delay. The proposed techniques all require a fairly spatially coherent source, which for typical medical X-ray sources requires a small source spot. Aside from the requirement on source coherence, many techniques utilize optical elements or gratings of fine pitch to obtain phase contrast images. Here, we investigate the simplest to implement X-ray phase contrast techniques: propagation based phase imaging, in which the physics of X-ray propagation through free space generates an image with phase contrast. In the current study, this method has been implemented to image several test objects including small diameter rods and insects.

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In order to generate a source with the required spatial coherence, polycapillary optics were used to focus the beam from a polychromatic rotating anode Mo X-ray source through a pinhole. Polycapillary optics are a collection of small hollow glass tubes through which X rays are guided along the length of the optics by total external reflection and the exiting beam is convergent to create a small intense source of X rays.\(^3\)

The differences in phase delay induced by the objects, transformed into an intensity pattern on the detector plane can be quantified using the transport of intensity equation. In the near-field region and for weakly attenuating objects, the intensity is described by\(^4\)

\[
I = \frac{I_0 T}{M^2} (1 - \frac{\lambda d}{2\pi} \varphi^2)
\]

where \(M = \frac{R_1 + R_2}{R_2}\) is the magnification, \(R_1\) is the distance of object from the X-ray source, \(R_2\) the distance of detector from the object, \(I_0\) is the incident intensity, \(T\) is the object transmission, \(d = \frac{R_1}{M}\) is the de-focusing distance and \(\varphi = k \int \delta(x, y, z) dz\) is the phase delay introduced by the object.

2. Experiment Set Up

A 5 KW Molybdenum (Mo) rotating anode X-ray source with a source size of 300 µm was used. Three types of detectors were also used for different purposes: for alignment a solid state CdZn Amptek XR-100 CR X-ray detector; and for imaging a Fuji Image plate M1040019 with an active area of 10 cm × 12 cm and 50 µm resolution or a phosphor coated CCD camera with an active area of 3 cm × 4 cm and a pixel size of 22 µm. A 1.6 mm polyethylene rod was imaged to test edge enhancement.

![Schematic diagram of the experimental set-up. The optic was translated horizontally and vertically at each point along z-axis to align it with the x-ray source.](image)

The input and output focal lengths of optic were 81 mm and 5.5 mm as given in Table 1. In this geometry, the polycapillary optic collected radiation from a divergent source and focused onto a small point.

**Table 1. Characteristics of Polycapillary Optic**

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>X-ray Optical System Inc.</th>
</tr>
</thead>
<tbody>
<tr>
<td>ID Number</td>
<td>690</td>
</tr>
<tr>
<td>Input Focal Length (mm)</td>
<td>81</td>
</tr>
<tr>
<td>Output Focal Length (mm)</td>
<td>5.5</td>
</tr>
<tr>
<td>Channel Diameter (µm)</td>
<td>9.4</td>
</tr>
</tbody>
</table>
In order to align the optic for maximum transmission, it was first placed as close to the source as possible and translated iteratively in the horizontal and then in the vertical direction to achieve the best alignment. The optic was moved away from the source and same procedure was repeated. At each distance, the full width at half maximum (FWHM) of the scan along the x and y axes was fit to a Gaussian distribution

\[ y = y_0 + A e^{-\left(\frac{x-x_c}{w}\right)^2} \]  

(3)

using Origin 6.0 software. At the input focal point, the width was minimum with a value of 0.34 ± 0.02 mm, as shown in Figure 2. The variation of beam width with the distance is shown in Figure 3.

The minimum expected value of the beam width \( w_{\text{calc}} \) at the input focal point is approximately

\[ w_{\text{calc}} = \sqrt{s^2 + (1.5 f_{\text{in}} \theta_c)^2} \]  

(4)

where \( s \) is the source size of 300 µm, \( f_{\text{in}} \) is the input focal length and \( \theta_c \) is critical angle for total reflection, which for the borosilicate glass is

\[ \theta_c = \frac{300 \text{keV}}{e} \text{ mrad} \]  

(5)

For an X-ray energy of 20 KeV, the critical angle is 1.5 mrad. The expected beam width is 0.35 mm, which is in good agreement with the measured value.

An image plate was used to estimate the output spot size. The image plate (Fuji M1040019) was placed close to the optic focal spot and the resulting image was then read in film reader (Fuji Films BAS-1800) using Image Gauge software. The smallest resolution of 50 µm was selected during reading of the image plate. Figure 4 shows the magnified view of the spot at a distance of 5.5 mm from the optic. The profile across the focal spot size drawn in Figure 4 gave a fit Gaussian distribution with a width of 114 ± 50 µm at the output focal point. The variation of spot size with the distance from the optic is shown in Figure 5.

Figure 2. Graph resulted from the scan of optic at the input focal point. The red line is the Gaussian fit to data. The width is 0.34 ± 0.02 mm.

Figure 3. Alignment of optic with the x-ray source. At the input focal point, the beam width is minimum.
Figure 4. Magnified view of focal spot at the output focal point. The profile across the spot was drawn and the resulting graph was used to estimate the FWHM.

The expected output focal spot size of the optic is

\[ s \approx c + 1.5 f_{\text{out}} \theta_c \]  

where \( c \) is the channel size of a single fiber, \( f_{\text{out}} \) is the output focal length of the optic. For the optic used here, expected value of spot size is 12.3 \( \mu \text{m} \). The difference between the measured and expected focal spot size may be due to an error in manufacturing of the optic. For short focal length optics it can be difficult to cut the optic at the position for which the channels are directed toward a single point. This optic was actually designed to be used with the short focal length side facing the source, to enable efficient source capture, in which case it was not important that all of channels be precisely aligned.

To make the spot size smaller, 25-100 \( \mu \text{m} \) diameter pinholes were placed at the output focal point of the optic and carefully aligned to maximize the output.

After aligning the optics and pinhole with the X-ray source, the output beam had high enough spatial coherence for phase imaging and was used to form phase enhanced images of rods and insects.

3. Images of Rod

3.1. 100 \( \mu \text{m} \) Pinhole

3.1.1. Image Plate

A cylindrical 1.6 mm diameter polyethylene rod was imaged using a 100 \( \mu \text{m} \) diameter pinhole. The rod was placed 17 cm from the pinhole (\( R_1 \)) and the image plate was positioned at a distance of 64 cm from the rod (\( R_2 \)). The magnification

\[ M = \frac{R_1 + R_2}{R_1} \]  

was 4.7.

Figure 5. The source size decreases with distance. It has minimum value at the output focal point.
The rod image presents a dark region with the body of the rod due to attenuation and a white band at the rod’s edge which is edge enhancement due to the phase imparted by the rod. For quantitative analysis, a profile of the rod was drawn perpendicular to the length of the rod and plotted against distance, as shown in Figure 7. This line was created by summing 5 neighboring pixels. Both the intensity dip due to attenuation and the intensity peak due to phase are evident in this plot.

The presence of noise in the image diminishes the strength of the phased enhanced signal. The conspicuity of the edge enhancement\(^6\) can be described by the edge-enhancement to noise ratio (EE/N) ratio

\[
\frac{EE}{N} = \frac{P - \bar{B}}{\sigma_B}
\]  

(8)

where \(P\) is the peak intensity, \(\bar{B}\) is the average intensity over a region of background labeled B in Figure 7, and \(\sigma_B\) is the standard deviation of intensity over region B. The edge-enhancement to noise ratio was 0.91.

The image in Figure 6 was acquired at 30 kVp and 30 mA for two and a half minutes. The product of current and exposure time is higher than the normally used for soft tissue imaging in clinical settings,\(^7\) however, the intensity loss was before the object and would not create additional patient dose. In addition, an optic which is designed for this application would produce a small spot without a pinhole\(^5\) and thus would have higher intensity and shorter exposure time.

![Figure 6. Image of 1.6 mm diameter rod taken with a 100 µm pinhole. The white lines show edge enhancement.](image1)

![Figure 7. Profile of the 1.6 mm diameter rod taken with a 100 µm pinhole. The small peaks on both sides of the valley are due to edge enhancement.](image2)
3.2. 25 µm Diameter Pinhole

3.2.1. Image Plate

The 1.6 mm diameter polyethylene rod was then imaged using a 25 µm diameter pinhole shown in Figure 8. The rod was placed at a distance of 25 cm from the output focal point of the optic ($R_1$) and the image plate was placed approximately 80-81 cm from the rod ($R_2$) giving a magnification of 4.2. The distance was increased to increase the beam coherence. The rod was exposed for 6 minutes with voltage of 30 kVp and a current of 40 mA. The exposure was increased to accommodate the loss of intensity in using the 25 µm diameter pinhole.

As with the 100 micron pinhole, the captured image shows attenuation through the rod’s body with edge enhancement due to phase at the rod’s edges. A five pixel wide profile was drawn normal to the rod in both images and a background division was performed to remove the effect of background intensity variation due to the optic structure in the image. The normalized resulting profile is shown in Figure 9. The peaks on both sides of the valley show edge enhancement. The edge-enhancement to noise ratio was 2.56.

![Figure 8. Image of a 1.6 mm rod taken with a 25 µm pinhole. The white line along the length of rod shows edge enhancement.](image1)

![Figure 9. Profile of the image of figure 8. Edge enhancement is evident from the two peaks. Background division was applied to remove nonuniformity in the beam.](image2)

3.2.2. Charged Coupled Device (CCD) Camera

The previous rod images were taken with the image plate that had a pixel size of 50 µm. A higher resolution digital detector (CCD camera) with a pixel size of 22 µm was used to test whether the edge enhancement would be improved. Flat-field and dark field images were acquired with no rod in place and were applied to the rod images.

The rod-to-pinhole distance was increased to 40 cm, and an image of the rod was taken with the object placed at a distance of 40 cm from source and a detector to rod distance of 67 cm, for a magnification of 2.6. As the exposure per frame was restricted to 30 seconds, average of 30 frames was taken to increase the exposure time to 15 minutes. A bright line along the length of the rod could be seen on both sides of the rod as shown in Figure 10.

![Figure 10. A 1.6 mm diameter rod image taken with CCD camera after flat field correction. Two bright lines along the length of rod are clearly seen.](image3)
Additional images were taken at varying object-to-detector distances. The effect of increased magnification on edge enhancement is shown in Figure 11. The edge enhancement increases approximately linearly with increasing distance of detector to object as expected from the work of other investigators using a small spot microfocus source with no optic. The virtual source provided by a polycapillary optic and pinhole performed as a microfocus source.

![Graph showing edge enhancement to noise ratio vs. distance of detector from object (cm)](image)

Figure 11. The effect of distance $R_2$ on edge enhancement. The edge enhancement to noise ratio increases with distance.

**Insect Images**

Insect images were taken to demonstrate the feasibility of phase contrast in biological tissues. A conventional absorption image, shown in Figure 12, was also acquired with the insect at a distance of $R_1 = 40$ cm from the source, with the image plate placed just behind the object at a distance of $R_2 = 2$ cm. The phase image shown in Figure 13 was taken at a distance of 67 cm from the insect keeping $R_1 = 40$ cm.

The phase image in Figure 13 shows not only the internal structures (due to absorption) but also shows clearly defined boundaries evidencing edge enhancement. Some of the improvement is due to the smaller effective pixel size, because of which there is visible edge-enhancement.

![Absorption image taken with the detector placed just behind the insect.](image)

Figure 12.

![Edge-enhanced image of the insect with the detector was placed at 67 cm from the insect. The edges of internal structures are clearly visible even though the lower resolution detector was employed.](image)

Figure 13.
4. Quantitative Phase Retrieval

Quantitative phase retrieval involves using knowledge of the physics that produces phase contrast images in order to retrieve a quantitative map of phase imparted to the X-ray beam by the object. For propagation-based imaging, one attempts to invert the transport of intensity equation, Eq. (2), to recover phase from intensity measurements. Because phase and attenuation information are both present in any pixel of the image, one typically acquires two (or more) images at different propagation distances to allow unambiguous phase reconstruction.

Capturing multiple images for processing can be difficult as the phase features are fine structures in the images (enhanced edges) which can shift by many pixels on the detector as the distances are changed in the imaging system. Even shifts of a single pixel can produce severe artifacts in a reconstructed image. To get around this problem, there are a number of phase retrieval techniques that attempt to reconstruct phase from a single image by making strong assumptions on the relationship between attenuation and phase. In this project, the phase-attenuation duality concept was employed to obtain the phase image of the insect. According to this theory the phase image and the attenuation image can both be determined by the same map of the projected electron density. The electron density was derived from the single phase-contrast image as

\[ \rho_{e,p}(r) = \frac{-1}{\sigma_{KN}} \log_e \left( \hat{F}^{-1} \left( \frac{\hat{F} \left( \frac{M^2 I}{I_{in}} \right)}{1 + 2\pi \left( \frac{\rho_x R_z}{M\sigma_{KN}} \right) u^2} \right) \right), \quad (9) \]

where \( I \) is the intensity in the image plane, \( I_{in} \) is the entrance intensity, \( \hat{F} \) is the two-dimensional Fourier transform and \( \hat{F}^{-1} \) is its inverse, \( u \) is the spatial frequency vector in the object plane, and \( \sigma_{KN} \) is the total cross-section for X-ray photon Compton scattering from a single electron derived from Klein-Nishina formula. The right image in Figure 14 is the phase at each pixel, calculated by assuming the phase is proportional to electron density.

![Measured intensity](a)

![PAD phase reconstruction](b)

Figure 14. Image (a) is the edge-enhanced image of the insect and image (b) is the pure phase image of the same insect. The phase image was obtained by employing the concept of phase-attenuation duality.

The left image of Figure 14 is the edge-enhanced image of Figure 13, except after background subtraction with a no object image. Additional features are clearly visible in the phase image.
Conclusion

Propagation based phase imaging was successfully tested with a large spot rotating anode X-ray source aligned with polycapillary optic. The polycapillary optic accepts radiation from a large X-ray source and converges it to create a small spot. A rod of 1.6 mm diameter was imaged with different geometric configurations using an image plate and CCD camera. Edge-enhancement to noise ratios up to a value of 6.5 were obtained. Conventional absorption and phase images of an insect were acquired to determine the feasibility of phase imaging in biological tissues using this technique. A pure phase object was obtained using a phase attenuation duality approach by taking only one image. Additional optimization of the optical system is required to reduce the exposure time.

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References