Full-field transmission x-ray imaging with confocal polycapillary x-ray optics

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A transmission x-ray imaging setup based on a confocal combination of a polycapillary focusing x-ray optic followed by a polycapillary collimating x-ray optic was designed and demonstrated to have good resolution, better than the unmagnified pixel size and unlimited by the x-ray tube spot size. This imaging setup has potential application in x-ray imaging for small samples, for example, for histology specimens. © 2013 American Institute of Physics. [<http://dx.doi.org/10.1063/1.4789799>]

I. INTRODUCTION

Focusing x-ray optics are widely used in x-ray imaging. The application of focusing x-ray optics in x-ray imaging can be generally divided into two types. In the first, the x-ray image of sample is magnified by a focusing x-ray optic placed between the sample and the detector.^{1–3} In the second, the sample is placed in the divergent beam after the focal spot of a focusing x-ray optic, and the enlarged image of the sample is obtained by placing an x-ray detector some distance after the sample.^{4–6} A variety of x-ray optics can be used in these systems. Very high resolution systems typically use Schwarzschild objectives⁷ or zone plates^{1,3,8} and sometimes multilayer Laue lenses⁹ or refractive optics.¹⁰ Because Schwarzschild objectives are used in normal incidence, they are essentially limited to the EUV region. Soft x-ray microscopes have also been developed for laser-plasma sources with both Wolter¹¹ and bent crystal optics.¹² Zone plates are easiest to make for soft x rays, where the thickness required to absorb the beam is small, but zone plates with high aspect ratio have been demonstrated for hard x rays.¹³ However, because the outer diameter of the zone plates or refractive optics is quite small, only a small collection angle can be achieved with a conventional source, so synchrotron sources are typically required to provide adequate flux. Similar constraints tend to limit both refractive optics and multilayer Laue lenses to synchrotron sources.

Polycapillary x-ray optics¹⁴ are attractive choices to be used as the focusing element in magnifying imaging systems for conventional sources. Such polycapillary optics, which can be used to collect the x rays in a wide energy band from a relatively large source angle, are durable and relatively simple to fabricate. A polycapillary focusing x-ray optic can focus a divergent x-ray beam onto a small sample spot for micro x-ray analysis.^{15,16} A polycapillary collimating optic can form a divergent x-ray beam into a quasi-parallel beam for x-ray diffraction.^{17–19} In addition, both focusing and collimating polycapillary optics only accept x-ray photons emitted within a finite region, and so can be used to selectively collect

divergent radiation, for example, for micro analysis.¹⁶ The principle of confocal analysis based on polycapillary x-ray optics, in which one optic focuses the incident beam onto the sample, and the second selects only emission from that region, was first proposed in the early 1990s by Gibson and Kumakhov.²⁰ In recent years, this confocal technology has been widely used in 3D micro x-ray fluorescence technology.²¹ In this geometry, the optical axes of the focusing and collection optics are typically at right angles, and the output focal spot of the focusing optic and the input focal spot of the collecting optic are adjusted in confocal configuration, so that only the x rays from the micro-volume defined by the overlap of these foci can be detected. This confocal configuration improves the signal-to-noise ratio of the experiment.

In this work, a focusing optic and collimating optic were employed in a parallel confocal geometry to magnify the image of a small sample. In this application, the collimating polycapillary x-ray optic performs much like a magnifying fiber optic taper. The optic is made of many monicapillaries. Each capillary collects the x-ray image from a different finite region of the sample and transmits it by total reflection to the detector.

II. EXPERIMENT AND RESULTS

The experimental system is shown schematically in Fig. 1. The x-ray beam from an Oxford Microfocus Mo x-ray tube was focused by the focusing optic. The object was placed beyond the focal point of the focusing optic, so that the focal spot of that optic acted as a virtual source. The image was collected by a collimating optic aligned confocally with the focusing optic. The detector was a Fuji computed radiography plate, read with a pixel size of 50 μm .

The output focal spot size was measured with a knife-edge scan to be 27 μm at 17.5 keV. This is in agreement with the expected value of $D_{\text{spot}} \approx c + 1.3 f_1 \theta_c$,¹⁴ where f_1 is output focal distance of the focusing optic, c is the channel size of the capillary holes within the optic, $c \sim 5 \mu\text{m}$, θ_c is the critical angle for total reflection, approximately 1.7 mrad at 17.5 keV, and the factor of 1.3 arises from typical values of the output divergence of each channel.²² The approximate gain in power density relative to a simple pinhole of the focusing optic is

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FIG. 1. (a) Schematic for the confocal imaging setup. A focusing optic collects radiation and focuses it to an output focal spot, a distance f_1 away. This spot is used as a source for imaging the object placed at a distance y . The collimating lens is placed at its input focal length, f_2 , from the focal point of the first optic. (b) Resolution phantom. The images were taken near the right tip.

$$\text{gain} = \eta \cdot \left(\frac{Z}{f_0}\right)^2 \cdot \left(\frac{D_{\text{inf}}}{D_{\text{spot}}}\right)^2, \quad (1)$$

where η is the measured transmission efficiency of the focusing optic,¹⁹ f_0 is the input focal distance of the focusing optic, D_{inf} and D_{spot} are the diameter of the entrance and of the output focal spot of the focusing optic, respectively, and Z is the distance from the x-ray source to the focal spot of the focusing optic, $Z = f_0 + L + f_1$, where L is the length of the focusing optic. The measured transmission, η , was 2.8% at 17.5 keV, and f_0 , D_{inf} , D_{spot} , L , and f_1 are shown in Table I. The resulting calculated gain of the focusing optic was approximately 2000 at 17.5 keV.

A pinhole with a diameter of about 1 mm located at a distance of about 1 cm from the output of the collimating optic was used to roughly determine the gain of the second optic. The measured gain of the collimating optic was approximately a factor of 13 at 17.5 keV. The high gains of the focusing optic and the collimating optic were helpful to decrease the exposure time for small samples.

The object was placed a distance $y = 9$ mm from the joint focal point, as shown in Fig. 1. The distance was chosen to illuminate a large enough region of a resolution phantom to allow image analysis. The object could be placed closer to the focal point, but not directly at the focal point where the paths from multiple channels through the collimating optic would overlap or the image would be blurred among all the output channels. The magnification M of the setup arises first from the simple propagation of the expanding beam from the

object location to the input of the collimating optic, $M_1 = f_2/y \approx 1.5$. The image is then further magnified by the collimating optic. Each channel of the optic is larger on exit than on entrance, by the ratio $M_c = D_{\text{outc}}/D_{\text{inc}} \approx 3.3$, where D_{inc} and D_{outc} are the input and output diameters of the collimating optic, respectively. The total magnification is then the product, $M = M_1 M_c \approx 5$.

A standard resolution phantom, shown in Fig. 1(b), was used to measure the resolution of the confocal imaging setup. A resulting image is shown in Fig. 2(a). These images were taken with an exposure time of 3 s with a tube voltage of 25 kV at 200 μ A. The intensity profile through the image is shown in Fig. 2(b), and the ratio to the no phantom intensity gives the expected sinusoidal variation, as shown in Fig. 2(c).

The contrast of the bar pattern was defined as $C = (I_{\text{max}} - I_{\text{min}})/I_{\text{max}}$. The intensities I_{max} and I_{min} can be taken from the original profile as in Fig. 2(b), or after dividing out the effect of the non-uniform original intensity as in Fig. 2(c). The two measures give similar results. Alternatively, the contrast can be determined by fitting to the original image profile, assuming a Gaussian intensity distribution of width σ , centered at a position x_0 , and a sinusoidal variation with a period Λ

$$I = e^{-\frac{(x-x_0)^2}{2\sigma^2}} \left(1 - \frac{C}{2} + \frac{C}{2} \cos\left(\frac{2\pi}{\Lambda}x\right)\right), \quad (2)$$

as shown in Fig. 2(d). Using the fitting parameter C for the contrast yielded similar results to the simple computation from the original image. The fitting technique allows to estimate the period of the magnified image. The images have a fitted spacing of 0.36 mm. The contrast computed from the original profiles is plotted in Fig. 3(a).

After placing a 125 μ m thick Nb filter at the source to create a more monochromatic beam, the confounding effect of absorption in the glass substrate of the phantom was reduced and the contrast was improved, as shown in Fig. 3(a), although at the expense of the overall intensity, as shown in Fig. 3(b) and the profile in Fig. 3(c). For the images with the Nb filter, the exposure time, current, and tube voltage were increased to 10 s, 300 μ A, and 30 kV, respectively.

The image line spacing of $\Lambda = 0.36$ mm corresponds, given the total magnification of a factor of 5, to an original line spacing of $\Lambda_0 = 72$ μ m, or about 14 lp/mm, which is consistent with the image having been taken near the fourth line marker from the right of the resolution phantom of Fig. 1(b).

Given an otherwise perfect imaging system, the effect of the $P = 50$ μ m pixel size of the detector readout on the image contrast is given by the Fourier transform of a square wave with period $2P$,

$$F(k) = \frac{\sin(kP/2)}{kP/2}, \quad (3)$$

evaluated at a spatial frequency of $k_0 = 2\pi/\Lambda_0$, which gives an ideal contrast of only 0.38, considerably less than the measured contrast. The Nyquist limit of the detector is 10 lp/mm, while the object was read at 14 lp/mm, due to the

TABLE I. Polycapillary optics parameters.

	Focusing	Collimating
Serial number	3159	2719
Input focal distance (mm)	50.0	13.8
Length (mm)	105.0	70.0
Output focal distance (mm)	10.0	(not applicable)
Input diameter (mm)	2.2	2.3
Output diameter (mm)	1.9	7.5

FIG. 2. (a) Image of the resolution phantom, using the polychromatic beam, with the detector 8 mm from the collimating optic, (b) intensity profile with and without phantom, (c) ratio of the intensities in (b), and (d) fit to the intensity profile through the resolution pattern of (a).

FIG. 3. (a) Contrast of a resolution pattern as a function of the distance of the detector from the final collimating optic, (b) image as for Fig. 2(a), taken with the Nb filter to produce a more monochromatic, higher energy beam, and (c) profile through (b).

successful magnification of the image using the polycapillary optics. The effect of the pixel size is reduced by the magnification, as the pixels are effectively a factor of five smaller. The contrast due to the magnified image on the pixels should be evaluated at the magnified line spacing, $k_M = 2\pi/(\Lambda_0 M)$, which gives a contrast of 97%.

Without the optics, the finite source size, d , would contribute to geometrical blur of the object, limiting the effect of the magnification, as shown in Fig. 4(a). In order to have a magnification of a factor of 5, the source to detector distance $S + D$ would have to be five times the source distance S , or $D = 4S$. The resultant blur for the no optic system would be then $blur = D\alpha = Dd/S = 4d$. The effect on the contrast in an otherwise ideal imaging system (with tiny pixels) is

$$G(k) = e^{-\frac{k^2 blur^2}{2}}. \quad (4)$$

To have a contrast of about 70% for a line spacing of 14 lp/mm, the source size would have to be around $2.5 \mu\text{m}$. The imaging was actually performed with a source size of about $150 \mu\text{m}$; the actual source size is unimportant after the focusing optic. The use of the optic allows for high resolution imaging with a conventional source.

With the polycapillary optic, the local angular divergence α is independent of source size and is given by the divergence from an individual capillary channel¹⁸

$$\alpha \sim 1.3\theta_c. \quad (5)$$

Because the pixel size effect is unimportant, the expected contrast reduction is mostly due to the propagation blur as the image is transmitted from the object to the entrance of the collimating optic. From Eqs. (4) and (5), this contrast is given by

FIG. 4. (a) Geometrical blur due to angular divergence of the beam. In this sketch, the geometry is for the no optic case, so the divergence is due to the finite source size, and (b) measured contrast as a function of distance from the collimating lens, and a fit using a post optic divergence of 0.072 mrad.

$$C = e^{-\frac{(\frac{2\pi}{\lambda_0})^2 [1.3\theta_c(f_2-y)]^2}{2}}, \quad (6)$$

where f_2-y is the distance from the object to the collimating optic in Fig. 1(a). If the average energy is taken as 17.5 keV for the filtered image, then the critical angle is approximately 1.7 mrad. The contrast is then approximately 69%, in good agreement with the data of Fig. 3(a) in the case that the detector is placed close to the output of the collimating optic.

Most of the contrast loss in the image is due to the geometrical blur, which would be improved if the object were placed farther from the focal point of the optic, but that would somewhat reduce the magnification of the system.

The computation above is valid if the detector is placed immediately following the collimating optic. If the detector is placed some distance beyond the optic, the contrast drops due to additional blurring as the image is propagated. However, it is notable that the drop is much slower than given in Eq. (6) for propagation before the optic. A fit to the contrast drop, including the magnified pixel size effect, the propagation blur of Eq. (6), and an additional blur using the effective divergence α as a fitting parameter, is shown in Fig. 4(b). The best fit for the effective divergence was 72 μ rad; the image is preserved over much larger distances from the optic that might otherwise be expected.

III. CONCLUSION

A confocal polycapillary imaging system was demonstrated to produce good resolution for magnifying small objects. The resolution of the system could be understood in terms of simple geometrical blurring due to the local divergence of the x-ray beam from each capillary channel. It should be noted that the focusing optic and collimating optic used here were not especially designed for x-ray imaging. For example, the ratio of the output and input of the collimating optic was not large, and so did not produce large magnification. In future work, a smaller resolution phantom could be placed closer to the focal point of the optic, and a system with a larger ratio of input and output optic diameters designed to give higher magnification.

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