Polycapillary optics for medical applications

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Abstract. Polycapillary optics can be designed for a wide variety of x-ray applications, including x-ray fluorescence for non-destructive materials analysis and diffraction applications such as protein crystallography. Of particular interest are medical applications, including the removal of Compton scattering with the resultant improvement in contrast and resolution in mammography, the production of monochromatic parallel beams for high-contrast imaging in clinical settings and the localization of radioactive tracers. A recent development is the use of polycapillary optics to improve beam coherence for x-ray phase imaging. Conventional radiographic techniques depend on attenuation, which provides only low contrast between soft tissues. Phase imaging can yield significantly higher contrast but requires spatially coherent beams. Conventional sources small enough to produce high coherence are necessarily low power, requiring long exposures. Polycapillary optics employed to create a small secondary source from high power sources have been shown to yield strong phase effects. Polycapillary optics can also be combined with other x-ray optics. A combination of a toroidally bent crystal with a polycapillary optic has been shown to produce enhanced resolution in a magnified image. Another x-ray imaging technique which does not rely on the small attenuation contrast is the capture of coherently scattered radiation, for which polycapillary optics could be used as angular filters to collect the beam, resulting in a map of tissue or material type.

1. Principles of polycapillary optics

Polycapillary optics are arrays of a large number of small hollow glass tubes. X rays are guided down these curved and tapered tubes by multiple reflections in a manner analogous to the way optical fibers guide light. The x rays can be transmitted down a hollow tube as long as the tube is small enough, and bent gently enough, to keep the angles of incidence less than the critical angle for total reflection, \( \theta_c \) [1,2,3,4]. The critical angle for borosilicate glass falls with photon energy \( E \) as

\[ \theta_c \approx \frac{30 \text{ keV}}{E} \text{ mrad} \]  \( \text{(1)} \)

and is about 0.1° for 17 keV photons. The angles are somewhat larger for leaded glass [5].

As shown in Figure 1, when a ray enters a curved tube on the side towards the center of curvature, the angle of incidence increases with tube diameter. The requirement that the incident angles remain less than the critical angle necessitates the use of small channel sizes, typically between 2 and 50 \( \mu \text{m} \).

Because of the mechanical limitations of producing thin-walled tubes that small, the optics are produced by pulling large diameter glass tubes to create small diameter tubes, stacking and pulling them together, and repeating, as for the example fiber shown in Figure 2.
2. Geometries and Applications

2.1. Angular Filters
Because of the small critical angle for reflection, polycapillary optics provide much higher angular resolution than, for example, anti-scatter grids for radiography, gamma camera collimators for nuclear medicine, or the soller slits for diffraction. A straight angular filter is shown in Figure 3.

2.1.1. Scatter Reduction. Compton scattering can result in substantial image degradation in radiography. In a conventional medical imaging system, scatter is partially removed by inserting an anti-scatter grid consisting of a radiotransparent material, such as graphite, embedded with lead ribbons parallel to the incoming beam. Typical anti-scatter grids have primary (unscattered beam) transmissions of 60-70% and scatter transmissions of around 20%. By comparison, the scatter transmission of a polycapillary optic, measured by moving the source off axis to an angle much larger than the critical angle, is negligible.

![Figure 1. X rays traveling in a bent capillary tube. The trajectory of the ray entering at the top at grazing incidence is projected onto the page, but in three dimensions may "toboggan" in a constant radius spiral. The x ray entering at the bottom (closest to the center of curvature) strikes at a larger angle.](image1)

![Figure 2. Cross sectional scanning electron micrograph of a polycapillary fiber with 0.55 mm outer diameter and 50 μm diameter channels.](image2)

![Figure 3. Angular filter composed of multiple polycapillary fibers. The length of the optic is 30 mm.](image3)

![Figure 4. Air gap magnification (top) showing degradation of image due to finite source size. Magnification with a long tapered polycapillary optic (bottom) showing no increased blurring after the exit plane of the patient.](image4)

![Figure 5. Image without (top) and with (bottom) a polycapillary optic in lieu of an antiscatter grid. Holes in the plastic block range from small to large (bottom to top) and shallow to deep (left to right).](image5)
than the critical angle, is less than 1% at 20 keV [6].

Polycapillary optics can also be used to magnify and demagnify images. Magnification can improve image resolution, particularly if the resolution is detector-limited, as is typically the case in digital imaging. Magnification is conventionally performed by increasing the air gap between the patient and the detector, but this will result in a loss of resolution due to the geometrical blurring from the finite source size, as shown in the top part of Figure 4. However, magnification using polycapillary optics, even with a large focal spot, would not be subject to a loss in resolution. Using a magnifying polycapillary optic increased the limiting spatial frequency of the measured modulation transfer functions (MTF) for CR plates by a factor of 1.8 [6]. The resolution was not degraded by the capillary structure, which was on a smaller scale (20 μm channel size) than the desired resolution. The optic yields an MTF increase at all spatial frequencies, which may be diagnostically more significant than the increase in limiting MTF. The simultaneous scatter rejection resulted in a contrast enhancement of more than a factor of 2, as demonstrated in Figure 5 [7]. Measurements at 40 keV showed a contrast enhancement of more than a factor of 4 [8]. Lead glass optics performed well at higher energies [5]. Magnifying capillary optics provide simultaneous contrast enhancement and resolution increase.

2.1.2. Scintigraphy

Collimators are required in nuclear medicine because without them, the rays emitted from the internal radioactive source would create an undifferentiated blur on the detector, as shown in Figure 6(a) [9]. Even with the collimator, scatter creates an obfuscating blur, as shown in Figure 6(b), so bulky and relatively low resolution energy sensitive gamma cameras are normally employed to provide discrimination against the Compton scatter. High resolution digital detectors could be employed instead, but only if the signal from the radioactive material is concentrated in small enough areas so that the intensity is larger than the background from the scatter. Scatter discrimination then occurs because the scatter intensity will be spread fairly uniformly across millions of pixels. However, maintaining a high signal requires that the collimator have a resolution on the order of the pixel size of the detector, to avoid dividing the signal over many pixels as well. Typical collimators used in nuclear medicine have resolutions on the order of a few millimeters, which is inadequate for this purpose.

Pairing a polycapillary optic such as that shown in Figure 3 with a radiographic detector provided 100-200 μm resolution [10]. Images taken with a parallel 30-mm-thick lead glass polycapillary parallel hole collimator and a computed radiography plate demonstrated the ability to image features of individual ion exchange beads within 125I brachytherapy seeds even in the presence of scattering from 30 mm of tissue-equivalent scatter material, as shown Figure 7 [11]. Calculations and measurements

![Figure 6](image6.png)  
**Figure 6.** (a) Without a collimator the image is a set of overlapping blurs, with no spatial information. The blur increases with the distance of the detector from the object plane. (b) Radiographic detectors are not normally used in nuclear medicine because their failure to discriminate against scatter results in little spatial information in the image.

![Figure 7](image7.png)  
**Figure 7.** Image of an array of 125I brachytherapy seeds with a signal-to-noise ratio of 98.
showed that for highly heterogeneous radiation distributions high signal-to-background ratios could be achieved by using the high resolution to "dilute" the scatter background without energy discrimination. In addition, the high resolution collimator/radiographic detector system is substantially more compact than typical gamma camera systems. [12]

2.1.3. Coherent Scatter

Conventional radiographic techniques depend on attenuation, which provides only low contrast between soft tissues. An alternative x-ray imaging technique which does not rely on the small attenuation contrast is the capture of the coherently scattered radiation, as cancerous and healthy tissues are known to have different coherent scatter signatures. For example, the fat scatter peak is due to the average distance between fatty acid molecules of \( d = 0.45 \) nm, which for the Mo K\( \alpha \) characteristic emission at 17.5 keV, corresponds to a peak in the fat scatter at \( 9^\circ \). The carcinoma scatter peak is at \( 13^\circ \). In addition to peaking at a different scatter angle, the scatter cross section is also higher for carcinoma than for normal tissue. [13,14,15] Differences are also seen in collagen alignment, which causes a diminishment of peaks at smaller momentum transfer values.

Rather than a system which requires raster scanning and full analysis of the coherent scatter pattern at each position, a potentially promising design employs the intensity of scatter at a fixed characteristic coherent scatter angle as a measure of the presence of that tissue type. A sketch of the selection of part of the coherent scatter ring with an angular filter (in this case an antiscatter grid) is shown in Figure 8. A schematic of the experimental system is shown in Figure 9. Coherent scatter images were obtained from a system using an antiscatter grid to select the desired peak with an exposure as low as 100-250 mAs. An example image from a checkerboard phantom is shown in Figure 10. Replacing the antiscatter grid, which has an angular acceptance of \( 6^\circ \), with a polycapillary angular filter would improve the angular resolution by an order of magnitude, and allow discrimination between fibroglandular tissue and carcinoma. The reduced intensity due to the higher

Figure 8. Coherent scatter with an angular filter tilted to the characteristic angle. The dashed black arrow is the grid normal. Only the dark part of the ring passes through the grid.

Figure 9. Schematic of the experimental setup. In the final system, a detector would be placed to capture a conventional mammogram from the primary beam. In this test system, only a single detector was employed. In a clinical slot scan system, the patient (phantom) is fixed and the source and slot are scanned past the patient, but the geometry is otherwise very similar. Replacing the grid in the test system with a polycapillary angular filter would greatly improve the angular resolution.

Figure 10. Coherent scatter image of a checkerboard phantom consisting of alternating pieces of polycrystalline graphite and fatty tissue.
angular selectivity would normally require longer exposure times, but the time (and dose) could be reduced by employing a detector with larger pixels. For example, if the coherent scatter is used to highlight regions of the original mammogram, it would be sensible to assume that no region less than 1 mm² would be useful. Increasing the pixel size in the coherent scatter image from 200 μm to 1 mm would result in a reduction in required dose of a factor of 25. The pixel size of the final coherent scatter system could be adjusted so that no additional dose is required beyond that required by the simultaneous conventional mammogram. Polycapillary optics could be used as angular filters to collect the beam, resulting in a map of tissue or material type within the image.

2.2. Collimating Optics
Collimating optics are produced by creating longer polycapillary fibers than for angular filters. The final pull is then designed to create a section with the desired shape, from which the ends are cut away, as shown in Figure 11.

2.2.1. Alignment
A typical set up for aligning a polycapillary optic is shown in Figure 12.[20] Either the source or the optic is translated perpendicular to the optic axis in small steps, producing a measurement of intensity versus relative source position, as shown in Figure 13.[21] The plot is symmetric and Gaussian, which indicates good alignment of the source, optic and detector. In order to determine the focal distance of the optic, source scans are performed at different distances from the source. At the focal distance, the ratio of the width of the scan curve to the optic-to-source distance, called the source scan angle, should be the smallest, as shown in Figure 14.[20] Transmission is the ratio of the number of photons passing through the optic to the number incident on the front face of the optic. Transmission with respect to the source-optic distance is also shown in Figure 14. The highest transmission and the lowest source

Figure 11. Sketch of the interior channels of a monolithic polycapillary collimating optic. The source is normally placed at the point where lines from the channels converge, and the output is then a nearly parallel.

Figure 12. Set up for optics measurement.

Figure 13. Source scan at 17.5 keV of a collimating optic.

Figure 14. Transmission (■) and source angle (●) with respect to the source optic. The transmission is the highest at the highest at the designed focal distance of 56 mm.
angle occurred at the focal distance.

2.2.2. Divergence
As shown in Figure 15, the output from a multifiber polycapillary collimating optic has both global divergence, \( \alpha \), and local divergence, \( \beta \). Even if the fibers are parallel (\( \alpha = 0 \)), the output divergence, \( \beta \), is not zero, but is determined by the critical angle and therefore the x-ray energy. The exit divergence from capillary optics is measured by rotating a high quality crystal in the beam and measuring the angular width of a Bragg peak. Since the Darwin width or crystal mosaicity, and the angular spread due to inherent energy width of the Bragg peak, are both typically much smaller than the exit divergence from the optic, the measurement yields the optic divergence directly.\[22\] The typical local divergence, \( \beta \), for collimating optics is ~1.3-1.5 \( \theta_c \). The factor is an experimentally determined parameter that arises from the fact that most of the beam has a divergence less than the maximum divergence of 2\( \theta_c \) produced by reflection at the critical angle. Unlike the case for pinhole collimation, the local divergence of the beam does not depend on the source size. Thus larger, higher power sources may be used without adversely affecting the resolution of the measurement. The maximum useful x-ray source spot size is limited by the acceptance area of the optic.

2.2.3. Simulations
Simulations of polycapillary optics performance are important for understanding defect structure and optimizing and predicting performance for new applications. These are typically based on ray tracing, and must take into account both the complex geometry of the optics, with hundreds of thousands of channels, and the deviation from ideality. Microscopic fluctuations of the channel walls are called roughness, and have the effect of reducing the reflectivity from the value associated with a perfectly flat surface. A typical roughness model assumes a correlation between surface deviations with a correlation length of a few microns, and a roughness height \( Z \). The effect of roughness on photon transmission is more significant when incident angles are high or when the photon bounces many times. Waviness is the midrange slope of the channel walls. The exact geometry of the surface tilt of the polycapillary channel walls is unknown, but the distribution of tilt angles has been found to be Gaussian, characterized with a width \( \sigma \).[23] The probability distribution must take into account that the photon is more likely to impact a wall tilted toward it than one tilted away. The third length scale usually considered is unintentional bending of straight channels (for example along the central axis of the lens). Results of many different optics give reasonable agreement with an unintentional

Figure 15. The output divergence from a multifiber collimating optic is characterized by global and local divergence. The minimum global divergence is proportional to the critical angle.

Figure 16: Simulation of the transmission versus energy of a collimating optic compared to experimental data.
circular curvature of 225 m, a waviness $\sigma$ of 0.125 mrad and a roughness height $Z$ of 0.5 nm with a correlation length of 6 $\mu$m.

Figure 16 shows the simulated and experimental transmission vs energy for an optic with a focal length of 56 mm. Using the above fixed parameters, the only fitting parameter was the fractional open area (the fraction of the front face which is not glass walls).[21] The best fit fractional open area was 51%, within a typical range of 40-70%. Figure 13 shows a simulated vs experimental source scan at photon energy 17.5 keV for the same optic. For this source scan, the source size was not measured when the experimental data was collected and a best-fit size of 105 $\mu$m was used by the simulation. The agreement is quite good for both independent sets of data, with only adjustable parameters. The simulation tracks the position and velocity of each exiting photon. More recent modeling includes tracking the accumulated phase.[24]

2.2.4. Monochromatic Imaging

As mentioned above, in conventional radiography, subject contrast arises from relatively low differences in absorption coefficients between different tissue types. The already low subject contrast is further reduced in a conventional system by averaging over relatively large energy bandwidths. Synchrotron measurements using monochromatic beams have demonstrated higher contrast, but synchrotrons are not clinically available. Using monochromator crystals with a conventional source without an optic is not practical because the low intensity of the diffracted beam will not allow imaging in vivo before motion blur occurs. Polycapillary collimating optics can allow sufficient diffracted beam intensity to make clinical monochromatic imaging possible without a synchrotron.[25] Measurements at 17.5 keV showed subject contrast enhancement of a factor of 2, as illustrated in Figure 17, in agreement with theoretical calculations. This contrast enhancement is in addition to that from the reduction of scattered radiation.

2.2.5. Polarization

The beam from a polycapillary optic is polychromatic and unpolarized. After diffraction from a monochromatizing crystal, the beam can be polarized, as shown in Figure 18, if the diffraction angle is approximately $2\theta=90^\circ$, so that the new propagation direction is along one of the possible directions of

![Crystal](image1.png)  
![Detector](image2.png)

Figure 17. X-ray image of plastic with 6.6 mm step, obtained with a broadband (left) and monochromatic beam (right). The phantom was thin on top and thick on the bottom, with a hole near the middle. The step is readily apparent with the monochromatic beam.

![Figure 18](image3.png)  
![Figure 19](image4.png)

Figure 18. System to measure beam polarization.  
Figure 19. The cobalt fluorescence emission is constant with $\phi$ angle, indicating good system alignment, while the scatter peak drops near $\phi=90^\circ$, indicating a good degree of polarization.
initial polarization. Since polarization along the direction of propagation is not possible, the diffracted beam is linearly polarized in the direction perpendicular to both the initial and final directions. In order to verify that the diffracted beam was polarized, a test was performed of the intensity of the fluorescence and scatter detected from a sample. The energy-sensitive detector simultaneously recorded the intensity of the beam scattered off the sample, as well as the fluorescence induced by the incident beam, which occurs at a lower, characteristic energy. Because the fluorescence is emitted isotropically, the fluorescence intensity should remain constant as the polarizing crystal, sample, and detector were rotated together. If the incident beam is polarized, the scattered beam should drop to zero when the angle nears 90°. The result of the measurement, shown in Figure 19,[26] shows a high degree of polarization.

2.3. Focusing Optics
Polycapillary optics can also be designed to focus the x-ray beam, as shown in Figure 20. The focusing effects come from the overlap of the beams from hundreds of thousands of channels, rather than from the action within a single tube.

2.3.1. Spot Size
The focal spot size of a polycapillary optic can be measured using an imaging detector. The global divergence is then the slope of the size versus distance, as shown in Figure 21.[2] Near the focal point, the spot size can be small compared to the resolution of the imaging detector. The spot can be measured using a knife edge technique as was done in Figure 21, or by scanning a small pinhole across the beam. Assuming perfect overlap, the spot size at the focal point is determined by the spot size from each individual capillary channel, which depends on channel size, c, output focal length, \( f_{\text{out}} \), and local divergence, \( \beta \), as shown in Figure 15, as

\[
d_{\text{spot}} = \sqrt{c^2 + (f_{\text{out}})^2} \beta.
\]

The critical angle, \( \theta_c \), at 20 keV is 1.5 mrad. Using \( \beta \approx 1.3\theta_c \), an optic with \( c = 3.4 \mu m \) and \( f_{\text{out}} = 9 \text{ mm} \) has a predicted spot size of 18 \( \mu m \). An intensity distribution measurement, by the pinhole technique, gave a FWHM of 21 \( \mu m \).[27] Because of the divergence from each channel, optics with smaller focal lengths have smaller spot sizes, as do measurements at higher photon energies.

2.3.2. Simulations
Figure 22 shows transmission versus energy results for a focusing optic which is 45.2 mm in length, with input focal length of 57 mm and output focal length of 128 mm. The input radius is 1.63 mm, the output radius is 2.22 mm and the channel diameter has a maximum radius of 5 \( \mu m \). The transmission
versus energy experimental results were taken using a 80 \textmu m source. For the simulation the results were fit using a fractional open area (unsupplied by the manufacturer) of 65%, so there was only a single fitting parameter.

2.3.3. Orthovoltage Therapy

Low energy x-ray beams incident on a patient have their highest intensity at the skin surface. In order to reduce the skin dose when treating the deep seated tumors, megavoltage beams were developed for tumor therapy. However, instead of using high energy beams, a focused low energy x-ray beam could spread the beam across a larger area of skin surface while increasing the beam intensity at the tumor. Such a beam can be produced with polycapillary optics, as shown in Figure 25.[28] Experimental measurements of focal spot size with PMMA blocks of varying thickness placed between the optic and focal point showed that the focal point is well maintained at least 4 cm into the plastic. MCNP5 [29] was used to model the patient therapeutic dose from a low-energy focused x-ray beam, using the measured beam parameters. Dose calculations were performed in a target area with a volume of 63 mm$^3$ inside a spherical breast tissue phantom of radius 30 mm. The optic was rotated in one plane around the sphere. The calculated dose distribution is shown in Figure 26. Skin dose was less than 10% of the tumor dose, demonstrating the potential for skin-sparing radiotherapy with a compact low energy x-ray system employing a focusing optic.

2.3.4. Microscopy

![Figure 23. (a) The focusing optic collects the 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.](image)

![Figure 24. (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.](image)
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.[30] In recent years, this confocal technology has been widely used in 3D micro x-ray fluorescence technology.[31] For most analytical applications, 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 an experiment to produce magnified imaging, a focusing optic and collimating optic were employed in a parallel confocal geometry to magnify the image of a small sample, as shown in Figure 23.[32] In this application, the collimating polycapillary x-ray optic performs much like a magnifying fiber optic taper. If the detector is placed some distance beyond the optic, the contrast drops due to additional blurring as the image is propagated, as shown in Figure 24. A fit to the contrast drop, using the effective divergence $\alpha$ as a fitting parameter, is shown in Figure 24. The best fit for the effective divergence was 72 μradians; the image is preserved over much larger distances from the optic that might otherwise be expected.

A similar, and even more striking result was obtained if the initial polycapillary optic was replaced with a toroidally bent crystal, as shown in Figure 27.[33] The contrast, shown in Figure 28 remains high as the distance is increased (in fact rises, probably due to a reduction in Compton scatter as the detector is moved farther from the resolution phantom). The result is consistent with simulation modeling.[24] The results imply that the local divergence from the focal spot of some optics is smaller and more complicated than from a point source of similar size.[34]

2.3.5. Phase

As noted, conventional radiography results in poor contrast for soft tissue imaging when 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. Phase contrast imaging renders the different phase delays as intensity variation in the detector plane. For x-ray energies of 10 to 100 KeV, the variations in phase delay are approximately 100-1000 times greater than the absorption variations. In addition to the higher inherent contrast, because the phase delay falls relatively slowly with photon energy, phase imaging could be performed at higher photon energies, which result in lower patient dose. Propagation-based phase imaging requires significant spatial coherence of the x-ray beam. This requires either synchrotron sources, unsuitable for routine clinical use, or small spot microfocus sources, which are inherently low power and thus require long exposure times. Polycapillary focusing optics can be used to produce an x-ray focal spot of the required size for propagation-based imaging from a rotating anode source.[35] Focusing polycapillary optics can produce a gain of several hundred compared to using a pinhole.
alone to reduce the spot size. The focal spot of the polycapillary optic was used as a secondary source to illuminate an object. A conventional image of a cricket taken with this illumination is shown in Figure 29. Moving the detector away from contact with the object allows for propagation-based phase effects to appear, seen as enhanced edge contrast in the image of Figure 30. Some of the improvement is due to the smaller effective pixel size due to the magnification; however there is visible edge enhancement. Because the contrast has terms due to both phase delay and attenuation, simplifying assumptions are required to reconstruct the phase delay attributable to the object. The phase-attenuation duality technique assumes that the attenuation is primarily due to Compton scatter, and hence to the electron density of the object, as is the index of refraction decrement and hence the phase delay. Assuming the two terms are proportional to each other allows generation of the image shown in Figure 31.[36]

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