HIGH CONTRAST IMAGING WITH POLYCAPILLARY OPTICS

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Magnifying polycapillary X-ray optics provide an innovative new way to control X-ray beams. The optics have measured primary transmissions greater than 50% and scatter transmission less than 1%. Contrast enhancement was nearly a factor of two compared to a conventional grid as measured using 5-cm thick Lucite phantoms.

Alternatively, using a collimating optic and diffracting crystal provides sufficient monochromatic beam intensity for medical imaging. Contrast, resolution, and intensity measurements were performed with both high and low angular acceptance crystals. At 8 keV, contrast enhancement was a factor of 5 relative to the polychromatic case, in good agreement with theoretical values. At 17.5 keV, monochromatic subject contrast was more than a factor of 2 times greater than the conventional polychromatic contrast. An additional factor of two increase in contrast is expected from the removal of scatter obtained by using an air gap which is made possible with a parallel beam. The measured angular resolution after the crystal was 0.4 mrad for a silicon crystal.

Introduction

A cross-section of a polycapillary fiber is shown in figure 1. A large diameter polycapillary fiber can be shaped into a one-piece, monolithic, optic. These are used as post-patient optics to provide rejection of Compton scattered X rays, while allowing magnification without loss of resolution, and image shaping to match with digital detectors.

Alternatively, single fibers can be strung through metal grids to form multifiber optics. This kind of optics can be used to collimate the X-ray beam to be diffracted by a monochromatizing crystal. Such a pre-patient optic could provide sufficient intensity for clinical monochromatic imaging.1,2,3,4

Magnifying Scatter Rejection

In a conventional medical imaging system, contrast can be increased by removing scattered X rays. In conventional systems, scattered X rays are removed by inserting a grid with lead ribbons parallel to the incoming beam. Alternatively, scattered X rays could be removed by inserting a capillary optic between the patient and the detector. Because capillary optics have an angular acceptance that is limited by the critical angle (1.5 mrad at 20 keV), scattered photons are not transported down the optic channels, but are mostly absorbed by the glass walls of the capillary optic. With polycapillary optics, measured transmission for scattered
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Figure 2. Tapered optics magnify without increasing the geometric blur, as usually occurs with an airgap in the projected image from conventional mammography.

photons are typically less than 1%. This leads to measured contrast enhancements of around a factor of two for energies from 20-40 keV.5,6

The tested prototype optics were also tapered and elongated to provide image magnification. Using a tapered prototype optic with a magnification factor of 1.8 the resolution was increased by the same factor. Further, the modulation transfer function (MTF) was increased at all spatial frequencies, including the diagnostically important lower frequencies.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 results were very promising, but the early optics suffered from low transmission efficiency.

Tapered Optic Defect Analysis

The transmission of a defected taper was analyzed by sectioning the optic into pieces, as shown in figure 3.7 The original optic was 32 cm long. The optic was first sectioned into two pieces, A and B. Output images of the two sections were then made, which showed that A contained a defect and B did not. In order to further locate the defect, A was cut into two 11-cm pieces; A1 and A2. In addition to X-ray images shown in figure 3, transmission measurements taken of the original tapered
Figure 4. The transmission of the defective tapered fiber. The original which was sectioned into A (22 cm) and B (10 cm). A was further sectioned into A1 and A2 of 11 cm each. And finally A2 was sectioned into A21 and A22 of 5.5 cm each.

with energy of penetration through glass inclusions blocking the channels. Optical microscope images confirm the presence of glass blockages. A similar glass inclusion in A2 was localized to A21. The output of A22 is uniform out to the hexagonal edges. The transmission of A21 has a minimum around 20 keV, not seen in A22.

**Improvements**

Partly as a result of the understanding developed from defect analysis, there have been significant advances in the manufacturing of long tapered monolithic optics. Transmission for 7 tapers from a single batch are shown in figure 5. These were initial prototypes for placement in a multioptic jig designed to increase the image area sufficiently to be of clinical use.
Applications of Multi-fiber Collimating Optics

Introduction

A somewhat different technology, multifiber collimating optics can be used as pre-patient applications for monochromatic imaging. Synchrotron measurements with monochromatic X-ray beams have demonstrated higher contrast, but synchrotrons are not practical in clinical settings. 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 result in sufficient diffracted beam intensity to make clinical monochromatic imaging possible using existing X-ray sources. Because the monochromatization is done before the patient, the patient is only subjected to those X rays which will contribute to the highest contrast image. Patient dose is reduced because of the removal of low energy photons that are heavily absorbed in the patient without contributing to contrast, and of high energy photons that cause high scattered X-ray intensity and low contrast.

Measurements

A preliminary test study using a 1 cm by 1 cm multifiber optic demonstrated the use of polycapillary optics to produce monochromatic images from phantoms. A schematic of the contrast phantom is shown in figure 6 and the resulting images are shown in figure 7. The measured contrast of the 6.6 mm step height was a factor of 5 higher for monochromatic 8 keV X rays than for the polychromatic case. Similar enhancements were seen for phantoms with compositional variation. In addition, preliminary measurements with a very low power source at 17.5 keV showed subject contrast enhancement of a factor of 2 compared to a 8 - 25 keV beam, also in agreement with theoretical calculations. This contrast enhancement is in addition to that expected from the reduction of scattered radiation.

Figure 7. Contrast images using polypropylene plastic. First column (a, b) is polychromatic and second column (c, d) is monochromatic. The step is hardly visible in the polychromatic images. First row (a, c) has step-height of 1.5 mm and second row (b, d) has 15.5 mm.

The output divergence of the optic affects the resolution and is an important parameter especially for low energy and high resolution modalities. 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. For a narrow bandwidth crystal such as silicon, the entire rocking curve width is due to the optic divergence, as shown in table 1.

Knife-edge measurements of resolution were also made. A knife-edge made of tantalum was placed after the monochromatic parallel beam. An image plate was placed 300 mm from the knife-edge to overcome the inherent resolution due to the 50-µm resolution of the image plate. The intensity profile was measured, from which the derivative of the intensity profile was calculated. The profile showed the resolved Kα doublet.

For a perfect crystal and parallel monochromatic input beam, the knife-edge image would be ideally sharp (with a perfect detector). For a monochromatic but very divergent beam, the width is limited by the acceptance bandwidth of the crystal. The 4-eV energy width of the Kα2 emission line produces an additional angular spread of

\[
\sigma_E \equiv \tan \theta_0 \frac{\sigma_{K\alpha_2}}{E_{K\alpha_2}} \approx 0.34 \text{ mrad}
\]

Combining the effects of the crystal, optic divergence, and energy spread, the calculated width is

\[
\sigma = \sqrt{4\alpha^2 \left( \frac{2\sigma_E}{E} \right)^2 + (\sigma_{\text{optic}}^2 + \sigma_{\alpha}^2)^2}
\]

where \(\alpha\) is the angular bandwidth of the crystal and \(\sigma_{\text{optic}}\) is the angular divergence of the optic output. The measured profile is additionally broadened by the detector resolution, which is 50 µm at 300 mm or 0.2 mrad. The resolution was measured with silicon, mica and graphite crystals. A summary of the results for these crystals with their theoretical values is shown in table 1. The angular resolution is quite good for either silicon or mica crystals.

<table>
<thead>
<tr>
<th>Crystals:</th>
<th>Manufacturer’s specification for (\alpha) (mrad):</th>
<th>Measured rocking curve width (mrad):</th>
<th>Angular width of knife-edge image (mrad):</th>
<th>(\sigma), Theory (mrad), with detector with 50 µm pixels</th>
<th>(\sigma), Theory (mrad), with an ideal detector</th>
</tr>
</thead>
<tbody>
<tr>
<td>Silicon</td>
<td>0.02</td>
<td>4.0 ± 0.1</td>
<td>0.5 ± 0.2</td>
<td>0.59</td>
<td>0.57</td>
</tr>
<tr>
<td>Mica</td>
<td>0.4 – 0.6</td>
<td>4.4 ± 0.2</td>
<td>0.8 ± 0.2</td>
<td>0.56 – 0.71</td>
<td>0.53 – 0.69</td>
</tr>
<tr>
<td>Graphite</td>
<td>35 – 87</td>
<td>42.5 ± 1.1</td>
<td>6.5 ± 0.5</td>
<td>4.5</td>
<td>4.5</td>
</tr>
</tbody>
</table>

Table 1. Rocking curves and resolution measurements using three different crystals. The rocking curve widths are due to the combined effects of the angular bandwidth of the crystal, the energy width of the line, and the 4-mrad output divergence of the optic. The angular resolution calculations are given by equation (2). The energy width for graphite is taken to include the whole Kα doublet.
The fraction of the input radiation within the appropriate angular and energy range for reflection from a crystal is approximately:

\[ \eta = \frac{\alpha}{\sqrt{\sigma_{\text{optic}}^2 + \alpha^2 + \sigma_E^2}}. \] (3)

The intensity reflected by the crystal is \( \eta \) multiplied by the reflectivity, \( R \), of parallel monochromatic radiation. For silicon, \( \eta \) is 0.005, \( R \) is approximately unity, and the measured efficiency was 0.003 ± 0.001. The efficiency, \( \eta \), is much higher for mica, because of its wider acceptance angle.

**Conclusion**

The use of polycapillary X-ray optics clearly provides higher contrast in imaging. Defect studies improved the quality and reproducibility of the newly manufactured monolithic tapers. The tapered monolithic optic enhances contrast by an average factor of 1.6 at 20 keV and 2.6 at 40 keV by rejecting scattered X rays.

Additionally, multi-fiber collimating optics were used for monochromatic imaging using a conventional X-ray source. The contrast was increased by a factor of 5 at 8 keV and 2 at 17.5 keV. The angular resolution was 0.6 mrad for a 50-mm thick patient when a low angular bandwidth crystal such as silicon was used.

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**References:**