Magnitude and effects of x-ray scatter in a 256-slice CT scanner

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We developed a prototype 256-slice CT scanner that employs continuous rotation of a cone-beam with a larger cone angle than conventional multidetector CTs (MDCT) to ensure a wide field of view. However, a larger cone angle may result in image deterioration due to increased x-ray scatter. Scattered radiation causes the detected signals to deviate from the true measurement of primary x-ray intensity and may result in artifacts (e.g., cupping and streak artifacts), quantitative inaccuracy in reconstructed CT number, and degradation of contrast-to-noise ratio (CNR). To reduce the effects of scatter, the 256-slice scanner incorporates an antiscatter collimator. Here, we estimated the magnitude of x-ray scatter in the prototype 256-slice CT scanner under clinical scan conditions and quantified the effects of this scatter on CT number accuracy, image noise, uniformity, and low contrast detectability. Although most experiments were performed with the antiscatter collimator, we also estimated the magnitude of x-ray scatter without the collimator to evaluate the scatter rejection efficiency of the collimator. The scatter-to-primary energy fluence ratio (SPR) without the collimator increased as cone angle increased, with estimated values of 49.7% for a 138 mm beam width with a phantom of 200 mm diameter, and 78.5% for a 320 mm diameter phantom. Estimated SPR was drastically decreased with the collimator, with an SPR reduction rate (ratio of SPR with and without the collimator) of 12.7% and 16.8% for the 200 and 320 mm diameter phantoms, respectively. The reduction in x-ray scatter by the collimator resulted in a considerable reduction in scatter effects. The measured uniformity was good and was independent of scatter amount. Although scatter still affected CT number accuracy, this could be corrected by rescaling. Further, although the CNR was decreased, in theory at least, the change was so subtle that it had no substantial effect on low-contrast detectability. © 2006 American Association of Physicists in Medicine. [DOI: 10.1118/1.2239366]

Key words: cone-beam CT, scatter, quantitative imaging, CT technology, CT image quality

I. INTRODUCTION

A prototype 256-slice CT was recently developed at the National Institute of Radiological Sciences.10 The prototype employs an x-ray cone-beam with a larger cone angle (approximately 13°) than the conventional multi-detector CT (MDCT) to ensure a wide field of view. Because the data collected by a cone-beam scan along a circular orbit is not sufficiently complete to make an exact reconstruction, artifacts occur more frequently as the cone angle widens. In general, artifacts occur when x-ray attenuation is changed rapidly along the cranio-caudal direction, and may appear as shading or streaks in soft-tissue regions near either bony structures or air pockets. We previously reported one such artifact that appeared as streaks from vertebral gaps.3 The characteristic appearance of this artifact facilitates its differentiation from anatomical structures.

The larger cone angle may also produce a second type of error arising from increased x-ray scatter. This results in deviation of the detected signals from the true measurement of primary x-ray intensity, and may result in artifacts (e.g., cupping and streak artifacts), quantitative inaccuracy in the reconstructed CT number, and degradation of the contrast-to-noise ratio (CNR).4–6

Using cone-beam CT with a flat panel detector, Siewerdsen and Jaffray7 observed significant cupping artifacts as well as reduced accuracy in CT number for larger cone angles. They also showed degradation of the CNR as a function of scatter-to-primary energy fluence ratio (SPR). Several methods to reduce the scatter effect in digital x-ray imaging have been investigated, including cone-beam CT. These include adjustment of object-to-detector gap,8,9 use of an anti-scatter grid,10–12 estimation of scatter using pixel values beyond the beam stopper13,14 or collimator leaves,15 and software correction using convolution.16–19 For the 256-slice CT prototype, however, we did not employ any of these approaches but rather an antiscatter collimator, because it was expected to provide a substantial reduction in x-ray scatter with only a small decrease in the primary x-ray.20 This collimator was similar to the conventional CT collimator but its longitudinal size was sufficiently large to cover 256 detector rows.

To our knowledge, no systematic experimental work has been done to investigate the magnitude of x-ray scatter and its effect on image quality in cone-beam CT using an antiscatter collimator. Here, we estimated the magnitude of scatter in the prototype 256-slice CT scanner under clinical scan
conditions and examined its effects on CT number accuracy, image noise, uniformity, and low contrast detectability. Most measurements were performed with the antiscatter collimator, but we also estimated the magnitude of scatter without the collimator to evaluate the scatter rejection efficiency of the collimator.

II. MATERIALS AND METHODS

A. Acquisition system

The prototype 256-slice CT scanner uses a wide-area two-dimensional (2D) detector designed on the basis of present CT technology and mounted on the gantry frame of an advanced CT scanner.¹ The number of elements is 912 channels × 256 segments, and element size is approximately 1 × 1 mm². Scanning time is 1.0 s/rotation, data sampling rate is 900 views (frames)/s, and dynamic range of the A/D converter is 16 bits. Nonuniform response of detector elements was calibrated by so-called water calibration, which uses measurement with cylindrical water phantoms of the same diameter as the fields of view (FOVs). A Feldkamp-Davis-Kress (FDK) algorithm is used for reconstruction. The 2D detector is equipped with an antiscatter collimator, an assembly of thin, equally spaced molybdenum blades that are placed in front of the detectors, and are adjusted for concentration on the longitudinal axis where the x-ray focal spot lies. Blade pitch is the same as the detector element pitch (1 mm), while thickness is 0.2 mm and collimator height is 30 mm. The detector element consists of a scintillator and a photodiode. The scintillator is Gd₂O₂S ceramic with a thickness of approximately 2 mm, as also used in the scintillator of a commercial multi-detector CT scanner (Toshiba Medical Systems Aquilion). The three types of wedge (large, small, and flat) on the 256-slice CT scanner were designed to extend the conventional wedges of the multidetector CT in the longitudinal direction. The large and small wedges were shaped to compensate for the path length of the patient across the scan field of view. The small wedge is used as an object for fields of view (FOVs) less than 240 mm, and the large wedge for those greater than 240 mm (e.g., chest and abdomen). The central part of the flat wedge is thicker than the respective part of the other two.

Definitions of the coordinate system are shown in Fig. 1. The x-y coordinate plane is parallel to the transverse direction, and the z-coordinate axis is parallel to the crano-caudal direction.

B. Estimation of scattered-to-primary energy fluence ratio (SPR)

SPR at the detector plane was measured in a manner similar to that of Johns and Yaffe⁴ or Siewerdsen and Jaffray.⁷ Briefly, we used two cylindrical water phantoms (200 and 320 mm diameter, and 250 mm height) to measure x-ray scatter by interrupting the primary with a Pb disk blocker in front of the phantom (Fig. 1).

First, x-ray intensity behind the phantom without the Pb blocker was measured by

\[ M_1 = P + P' + S \]  

where \( P \) is the transmitted primary x-ray intensity, \( P' \) is off-focal radiation, and \( S \) is scattered radiation. Detector dark current was corrected at the time of measurement by subtracting the offset data, which were measured in the absence of x-ray irradiation. To compensate for statistical fluctuation, data for 64 views were averaged.

The Pb blocker was then inserted, and the x-ray intensity behind the phantom and the Pb blocker was measured by

\[ M_2 = P' - P_B' + S - S_B, \]  

where \( P_B' \) and \( S_B \) are the reductions in off-focal and scattered radiation due to shadowing by the blocker, respectively, which decreases as the diameter of the blocker decreases.

The Pb blocker was 5 mm thick and 20, 30, 40, or 50 mm in diameter. A region of interest (ROI) with a diameter of 20 mm and approximately 300 pixels was placed in the center of the blocker shadow within the umbra. X-ray intensity was obtained as the sum of pixel values in the ROI and plotted against the diameter of the blocker. The sum of off-focal and scattered radiation \( P' + S \) was obtained as the intercept of the linear fitting function [Fig. 2(a)]. The primary x-ray intensity \( P \) was obtained by subtracting \( P' + S \) from \( M_1 \) [Eq. (1)].

To separate the scattered and off-focal radiation, further measurement was made without the phantom. X-ray intensity without the phantom or blocker was measured by

\[ M_3 = P_{air} + P'_{air}, \]  

where \( P_{air} \) and \( P'_{air} \) are the primary and off-focal x-ray intensities transmitted through air, respectively. The Pb blocker was then inserted and the x-ray intensity behind it was measured by
\[ M_4 = P'_{\text{air}} - P'_{\text{air, blocker}}, \]  
(4)

where \( P'_{\text{air, blocker}} \) is the reduction in off-focal radiation through air due to shadowing by the blocker. The same analysis as that used in the separation of \( P \) and \( P' + S \) was done to obtain \( P'_{\text{air}} \) and \( P'_{\text{air}} \) separately [Fig. 2(b)].

The off-focal x-ray intensity behind the phantom \( P' \) was obtained from \( P'_{\text{air}} \) as follows:

\[ P' = T \times P'_{\text{air}}, \]  
(5)

where \( T \) is transmittance of the off-focal radiation and was approximated by the transmittance of the primary x-rays as follows:

\[ P' = \frac{P}{P_{\text{air}}} \times \frac{P'_{\text{air}}}{P_{\text{air}}} = P \times \frac{P'_{\text{air}}}{P_{\text{air}}}. \]  
(6)

Finally, the scatter-to-primary ratio (SPR) was obtained as follows:

\[ \frac{S}{P} = \frac{P' + S}{P} - \frac{P'}{P} = \frac{P' + S}{P} - \frac{P'_{\text{air}}}{P_{\text{air}}}. \]  
(7)

These measurements were repeated five times and the results were averaged. X-ray tube voltage was 120 or 135 kV for two phantoms, with a small or flat wedge used for the 200 mm diameter phantom and a large or flat wedge used for the 320 mm diameter phantom according to clinical head and body conditions. X-ray exposure was made for 64 views without gantry rotation by keeping the x-ray tube and detector stationary, and exposure time was 0.07 s (64 views \( \times 1 \text{s/900 views} \)). X-ray tube current was selected as the maximum available value.

Beam width was set to 20, 42, 74, 106, or 138 mm, as defined at the midplane of the phantom. The beam width was defined as the full-width-at-half-maximum (FWHM) of the x-ray intensity profile at the rotation center and was approximately 10 mm larger than the nominal beam width given by the product of the slice thickness and its number.\(^{21}\) The difference between the actual and nominal beam widths is a margin established to cover the penumbra and mechanical errors. Table I lists details of scan conditions.

### C. Image noise and uniformity

Image noise was measured with the same water phantoms as the SPR estimations. Figure 3 shows the 200 mm diameter phantom. Standard deviations (SDs) of CT numbers were calculated in nine ROIs of 10 mm diameter aligned every 20 mm along the diameter of the phantom in transverse section. The same phantom and ROIs used for the noise measurement were used for the uniformity measurement, and averages of CT numbers were calculated in these ROIs. Scan conditions were 120 kV, 200 mAs, small wedge, 5 mm slice thickness (given from the average of ten 0.5 mm slices) and a 20, 42, 74, 106, or 138 mm beam width for the 200 mm

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**Table I. Scan conditions.**

<table>
<thead>
<tr>
<th>Evaluation items</th>
<th>Phantom diameter (mm)</th>
<th>Tube voltage (kV)</th>
<th>Tube current (mA)</th>
<th>Beam width (mm)</th>
<th>Scan time (s)</th>
<th>Wedge</th>
<th>Slice thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SPR</td>
<td>200</td>
<td>102</td>
<td>400</td>
<td>20–138</td>
<td>0.07</td>
<td>Small</td>
<td></td>
</tr>
<tr>
<td></td>
<td>320</td>
<td>135</td>
<td>300</td>
<td>20–138</td>
<td>0.07</td>
<td>Large</td>
<td></td>
</tr>
<tr>
<td>Image noise</td>
<td>200</td>
<td>120</td>
<td>200</td>
<td>20–138</td>
<td>1</td>
<td>Small</td>
<td>5 (0.5 mm ( \times ) 10)</td>
</tr>
<tr>
<td>Uniformity</td>
<td>320</td>
<td>120</td>
<td>200</td>
<td>20–138</td>
<td>1</td>
<td>Large</td>
<td>5 (0.5 mm ( \times ) 10)</td>
</tr>
<tr>
<td>CT no. accuracy</td>
<td>200</td>
<td>120</td>
<td>200</td>
<td>20–138</td>
<td>1</td>
<td>Small</td>
<td>5 (0.5 mm ( \times ) 10)</td>
</tr>
<tr>
<td>Low contrast</td>
<td>200</td>
<td>120</td>
<td>400</td>
<td>20,74,138</td>
<td>1</td>
<td>Small</td>
<td>20 (1 mm ( \times ) 20)</td>
</tr>
</tbody>
</table>

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**FIG. 2.** Relationship between intensity and Pb disk diameter using a 106 mm beam width (a) with and (b) without a phantom.
diameter phantom. The same conditions were applied to the 320 mm diameter phantom except for 300 mAs and a large wedge.

D. CT number accuracy

CT number accuracy was evaluated by measuring air CT numbers in an air cavity located at the center of the water phantom and outside the phantom. The air cavity was produced by inserting a vacant acrylic cylinder of 10 mm diameter (Fig. 4). Scan conditions were 120 kV, 200 mAs, small wedge, and 5 mm slice thickness (given from the average of ten 0.5-mm slices) and a 20, 42, 74, 106, or 138 mm beam width. Details are provided in Table I.

E. Low-contrast detectability

Figure 5 shows schematic drawings of the phantom for low-contrast detectability (Catphan 500 with module CTP515). The low-contrast phantom included three supra-slice and subslice sets of cylinders. The subslice targets had z-axis lengths of 3, 5, or 7 mm and diameters of 3, 5, 7, or 9 mm, while the supra-slice sets consisted of cylinders of 2, 3, 4, 5, 6, 7, 8, 9, or 15 mm diameter, and contrast with the backgrounds of three sets of cylinders was 0.3%, 0.5%, and 1.0%. A 1.0% contrast meant that the mean CT number of the object differed from its background by 10 Hounsfield units (HU). Scan conditions were 120 kV, 400 mAs, small wedge, and a 20, 74, or 138 mm beam width. Slice thickness was 20 mm (given by the average of 20 1.0 mm slices). To avoid volume-averaging errors, the results were evaluated by observing images with supra-slice targets.

F. Estimation of SPR without a collimator

The measurements described in Secs. II B–II E above were performed with the anticollimator installed. Scatter measurement without the collimator was not performed at that time because the high precision required for alignment of the collimator to the detector element would have hampered subsequent restoration of the detector assembly following removal of the collimator. However, just before upgrading the first prototype used here to the second improved model,22 we removed the center part of the collimator (120 channels × 256 segments in size) and performed scatter measurement to estimate SPR using the method in Sec. II B, with only the small and large wedge filters used for the 200 and 320 mm diameter phantoms, respectively.

III. RESULTS

A. Estimation of SPR

Estimation of the SPR without the anticollimator is shown in Fig. 6 and Table II. At a 20 mm beam width and 120 kV tube voltage, SPR was 18.2% for the 200 mm phantom and 41.8% for the 320 mm phantom. SPR increased with increasing beam width, with respective values for a 138 mm beam width of 49.7% and 78.5%. There was no significant difference in SPR between the 120 and 135 kV tube voltages. This tendency was qualitatively similar to the results obtained using flat-panel detectors.25

Figures 7 and 8 (and Table II) compare estimated SPRs with and without the collimator for the 200 and 320 mm phantoms, respectively. X-ray tube voltage was 120 kV. These figures and table also show the ratio of SPR with and without the collimator (SPR reduction rate). Estimated SPR was drastically decreased by the collimator, with reduction
rates for the 20 mm beam width of 4.9% and 6.7% for the 200 mm and the 320 mm phantoms, respectively, increasing with increased beam width (138 mm) to 12.7% and 16.8%, respectively.

Figures 9 and 10 show details of the estimated SPR with the antiscatter collimator. For the 20 mm beam width and 120 kV tube voltage, SPR was 0.9% for the 200 mm diameter phantom and 2.8% for the 320 mm phantom, also increasing with increased beam width (138 mm) to 6.3% and 13.2%, respectively. Further, SPR was larger with the flat wedge than with the small or large wedge. There was no significant difference in SPR between the 120 and 135 kV tube voltages.

B. Image noise and uniformity

The magnitude of image noise was obtained from the standard deviations of CT numbers in the nine ROIs in the transverse section (Fig. 3). Averaged standard deviations for the nine ROIs were 5.6–5.9 HU for the 200 mm phantom and 23.9–25.7 HU for the 320 mm phantom (Table III). Figure 11 shows the relationship between noise magnitude and SPR. The solid lines show the least squares fit to the following equation:

$$\sigma = \sigma_0 \left(1 - \frac{S}{2P}\right),$$

where $\sigma$ is the noise magnitude. This equation (8) is derived in the Appendix as (A15). The results suggest that this equation may provide a good estimate of the relationship between noise magnitude and SPR.

For the uniformity measurement, we calculate averages of CT numbers in ROIs in the transverse section (Fig. 3). Figure 12 shows the relationship between the $x$ position and average CT numbers. Standard deviations of the average CT numbers were 0.8–1.7 HU for the 200 mm phantom and 0.6–1.8 HU for the 320 mm phantom (Table III). From these results, uniformity was considered to be independent of SPR. This independence of SPR could be attributed to the water calibra-

### Table II. Scatter-to-primary ratio (unit: %).

<table>
<thead>
<tr>
<th>Phantom diameter (mm)</th>
<th>Wedge</th>
<th>Voltage (kV)</th>
<th>Collimator</th>
<th>With</th>
<th>Without</th>
<th>W/WO*</th>
<th>Beam width (mm)</th>
<th>With</th>
<th>Without</th>
<th>W/WO*</th>
</tr>
</thead>
<tbody>
<tr>
<td>200</td>
<td>Small</td>
<td>120</td>
<td>With</td>
<td>0.9</td>
<td>18.2</td>
<td>0.049</td>
<td>138</td>
<td>6.3</td>
<td>49.7</td>
<td>0.127</td>
</tr>
<tr>
<td></td>
<td></td>
<td>135</td>
<td>With</td>
<td>1.2</td>
<td>19.5</td>
<td>0.062</td>
<td></td>
<td>6.4</td>
<td>48.6</td>
<td>0.132</td>
</tr>
<tr>
<td></td>
<td>Flat</td>
<td>120</td>
<td>With</td>
<td>1.4</td>
<td>N/A</td>
<td>N/A</td>
<td></td>
<td>7.2</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td></td>
<td>135</td>
<td>With</td>
<td>1.5</td>
<td>N/A</td>
<td>N/A</td>
<td></td>
<td>6.7</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td>320</td>
<td>Large</td>
<td>120</td>
<td>With</td>
<td>2.8</td>
<td>41.8</td>
<td>0.067</td>
<td></td>
<td>13.2</td>
<td>78.5</td>
<td>0.168</td>
</tr>
<tr>
<td></td>
<td></td>
<td>135</td>
<td>With</td>
<td>6.2</td>
<td>42.1</td>
<td>0.147</td>
<td></td>
<td>12</td>
<td>82.5</td>
<td>0.145</td>
</tr>
<tr>
<td></td>
<td>Flat</td>
<td>120</td>
<td>With</td>
<td>4.5</td>
<td>N/A</td>
<td>N/A</td>
<td></td>
<td>13.8</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td></td>
<td>135</td>
<td>With</td>
<td>6.2</td>
<td>N/A</td>
<td>N/A</td>
<td></td>
<td>14.8</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>

*W/WO is the SPR reduction rate that is a ratio of SPR with and without the collimator.
tion process, in which scatter contributions to object measurement and calibration are similar and may in fact compensate for each other.

C. CT number accuracy

We found that CT numbers in the cavity (CT_{cavity}) were different to those for outside air (CT_{outside}) (Fig. 13). Figure 14 shows the relationship between CT number in the cavity and beam width. Although CT_{outside} is constant (~1000 HU) with increasing beam width, CT_{cavity} increases with increasing beam width. Here, although CT_{cavity} is usually adjusted to ~1000 HU, we set CT_{outside} to ~1000 HU to facilitate the observation of physical properties. Figure 15 shows the relationship between CT_{cavity} and SPR. The solid line shows the least squares fit to a straight line.

D. Effect of scattered radiation on contrast

Figure 16 shows low contrast images obtained with 20, 74, and 138 mm beam widths. Objects can be observed down to 2 mm at 1% contrast and to 7 mm at 0.3% for all three widths. Table IV lists the minimum diameters of detected cylinders. Although the circular bandlike artifacts might have affected the results, they indicate that low contrast detectability is independent of SPR over the beam width range used in this work.

IV. DISCUSSION

In this study we estimated the magnitude of x-ray scatter in a prototype 256-slice CT scanner and examined the effects of scatter on CT number accuracy, image noise, uniformity, and low contrast detectability. The prototype employed an antiscatter collimator that efficiently rejected scatter x-ray, clearly differentiating the prototype from other cone-beam CT scanners. To clarify the efficiency of the collimator, we made additional measurements to estimate SPR without the collimator just before the scanner was decommissioned.

The measurements without the collimator showed that the SPR for the 200 mm diameter phantom was 18.2% and 49.7% for the beam widths of 20 and 138 mm, respectively, while that for the 320 mm phantom was 41.8% and 78.5% (Table II). The SPR increased as beam width increased and there was no significant difference between ratios at the 120 and 135 kV tube voltages. This tendency was qualitatively similar to the results obtained using flat-panel detectors, although quantitative comparisons were difficult due to differences in experimental conditions (beam geometry, etc.).

<table>
<thead>
<tr>
<th>Phantom diameter (mm)</th>
<th>Beam width (mm)</th>
<th>Average image noise (HU)</th>
<th>SD of uniformity (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>200</td>
<td>20</td>
<td>5.9</td>
<td>0.8</td>
</tr>
<tr>
<td></td>
<td>42</td>
<td>5.8</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>74</td>
<td>5.7</td>
<td>1.7</td>
</tr>
<tr>
<td></td>
<td>106</td>
<td>5.6</td>
<td>1.0</td>
</tr>
<tr>
<td></td>
<td>138</td>
<td>5.6</td>
<td>0.8</td>
</tr>
<tr>
<td>320</td>
<td>20</td>
<td>25.7</td>
<td>0.8</td>
</tr>
<tr>
<td></td>
<td>42</td>
<td>25.0</td>
<td>1.8</td>
</tr>
<tr>
<td></td>
<td>74</td>
<td>24.8</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>106</td>
<td>24.3</td>
<td>0.9</td>
</tr>
<tr>
<td></td>
<td>138</td>
<td>23.9</td>
<td>0.6</td>
</tr>
</tbody>
</table>
Scatter rejection by the collimator was highly efficient. The ratio of SPR with and without the collimator (SPR decreasing ratio) for the 200 mm diameter phantom was 4.9% and 12.7% for the beam widths of 20 and 138 mm, respectively, while that for the 320 mm phantom was 6.7% and 16.8% (Table II). Alignment of the blades in the collimator was so precise that transmittance of the primary x-ray in the collimator would be approximated by a geometrical factor of 0.8. From these values, approximately only 10.2% =12.7%/0.8 and 13.4% =16.8%/0.8 of scatter could transmit the collimator even at the 138 mm beam width with the 200 and 320 mm phantoms, respectively.

Siewerdsen and Jaffray\(^7\) showed that elevated SPR degrades the CNR and that it can be restored by increasing dose. For example, the CNR in a cone-beam CT image decreased to 1/2 when SPR=100%, but could be fully restored by increasing the dose by a factor of 4 (\(=2^2\)). In our case, using a 200 mm phantom and 138 mm beam width, SPR was approximately 50% without the collimator, and CNR could be restored by increasing the dose by a factor of approximately 2.25 (\(=1.5^2\)). With the collimator, however, the same CNR could be achieved by first increasing the dose by a factor of 1.12 (\(=1.06^2\)) and then by compensating for the primary photon rejected by the collimator to increase x rays by a factor of 1.25 (\(=1/0.8\)). This process results in an increase in dose by a factor of 1.4 (\(=1.25\times1.12\)), a value substantially smaller value than that without the collimator. The use of a collimator will therefore help to decrease patient dose in clinical practice.

The collimator produced a dramatic decrease in SPR: that for the 200 mm diameter phantom was 0.9% and 6.3% for the beam width of 20 and 138 mm, respectively, while that for the 320 mm phantom was 2.8% to 13.2% (Table II). These small values may justify the linear approximation in the Appendix.

As shown in the Appendix, x-ray scatter degrades the measured contrast by the ratio \(S/P\) and decreases image noise by \(1/\sqrt{S/P}\), if \(S/P\) is negligible to unity. As mentioned above regarding Fig. 11, the relationship between noise magnitude and SPR may also follow this relationship.

With regard to the measured contrast and CT number accuracy, the CT number of the air cavity from Eq. (A6) in the Appendix is given by

\[
\text{CT}_{\text{cavity}} = 1000\frac{S}{P} - 1000. \tag{9}
\]

The measured result supported this finding, given that Fig. 15 shows that the relationship between \(\text{CT}_{\text{cavity}}\) and \(S/P\) is fitted to the following equation,

\[
\text{CT}_{\text{cavity}} = 902.3\frac{S}{P} - 1004.3, \tag{10}
\]

and also given that the two equations agreed well. The smaller gradient of the measured data might be attributable to the approximated derivation of Eq. (9). The agreement suggests that the CT number inaccuracy of the air cavity was mainly attributable to scatter and that the equations (A4)–(A7) are a good approximation within our experimental conditions. It is therefore likely that CT number deviation in the 256-slice CT can be corrected by rescaling the CT number using Eq. (A7).

As shown in the Appendix, the decrease in measured contrast by the ratio \(S/P\) and that in image noise by the ratio \((1/2)S/P\) caused a net decrease in the contrast-to-noise ratio (CNR) by the ratio \((1/2)S/P\), where the CNR is closely related to lesion detectability. SPR for the 200 mm phantom...
was approximately 6% for the 138 mm beam width, giving a 3% degradation of the CNR from the ideal case without scatter. This ratio of degradation had only a very subtle effect on images, however, and might not affect their visibility. Evidence for this comes from the lack of difference in low-contrast detectability among the three beam widths in Fig. 16.

In conclusion, the collimator used for the 256-slice CT was so effective that only a small portion of scatter could transmit. Image uniformity was therefore good and independent of scatter amount. Although scattered radiation affected CT number accuracy, this could be corrected by rescaling. These findings indicate that the 256-slice CT scanner is as quantitative as conventional CT scanners. Although the CNR was decreased, in theory at least, the change was so subtle that it had no substantial effect on low-contrast detectability. Artifacts caused by scatter were not investigated in the present study but will be examined in a later study.

APPENDIX

1. Contrast

Consider two cylinders of uniform materials, the first enclosed within the second, with diameters \(d_1\) and \(d_2\), where \(d_1 \gg d_2\). Linear attenuation coefficients are \(\mu_1\) and \(\mu_2\). According to Siewerdsen and Jaffray, \({}^7\) measured contrast \(\hat{C}\) is given by

\[
\hat{C} = C + \frac{1}{d_2} \ln \left( \frac{1 + (S/P)\exp(-Cd_2)}{1 + S/P} \right),
\]

where, \(\hat{C} = \hat{\mu}_1 - \hat{\mu}_2\), \(C = \mu_1 - \mu_2\), where \(\hat{\mu}_1\) and \(\hat{\mu}_2\) represent measured values of \(\mu_1\) and \(\mu_2\), respectively, and \(Cd_2 \ll 1\).

If \(S/P \ll 1\), Eq. (A1) is approximated by

\[
\hat{C} \approx C + \frac{1}{d_2} \ln \left( 1 - \frac{S}{P} [1 - \exp(-Cd_2)] \right)
\]

\[
\approx C - \frac{1}{d_2} \frac{S}{P} [1 - \exp(-Cd_2)].
\]

Since \(Cd_2 \ll 1\), Eq. (A2) is further approximated by

\[
\hat{C} \approx C - \frac{1}{d_2} \frac{S}{P} Cd_2 = C \left( 1 - \frac{S}{P} \right).
\]

The last result shows that the measured contrast \(\hat{C}\) decreases by the ratio \(S/P\) due to scattered radiation if \(S/P\) is negligible to unity.
shows the noise-to-signal ratio in x-ray intensity measures as follows:

\[
\text{var} = \text{var} + \text{var}
\]

2. CT number accuracy

From Eq. (A3), the measured CT number difference between the two materials \(\Delta CT\) (HU) is given by

\[
\Delta CT = \Delta CT_0 \left(1 - \frac{S}{P}\right),
\]

where \(\Delta CT_0\) is the difference without scatter. If the first material is water and water calibration is employed, its attenuation coefficient is approximately independent of the SPR because scatter contributions to object measurement and calibration are similar and may compensate for each other. On this basis we can set its CT number to 0. The CT number of the second material is given by

\[
CT = CT_0 \left(1 - \frac{S}{P}\right),
\]

where \(CT_0\) is the CT number of the second material without scatter. Equation (A5) shows that the CT number is deviated from \(CT_0\) by the ratio \(S/P\). If the second material is air in the cavity, its CT number \(CT_{\text{cavity}}\) is given by

\[
CT_{\text{cavity}} = -1000 \left(1 - \frac{S}{P}\right) = 1000 \frac{S}{P} - 1000,
\]

because the CT number of air without scatter equals -1000. From (A5) and (A6), we can estimate \(CT_0\) from the measured quantities as follows:

\[
CT_0 = \frac{-1000}{CT_{\text{cavity}}} \cdot CT.
\]

3. Noise

Noise in a reconstructed image is related with the noise in x-ray intensity measurement. The noise-to-signal ratio in a reconstructed image is defined as \(\sigma/I\mu\), where \(\mu\) is the linear attenuation coefficient at a point, and \(\sigma\) is the standard deviation in a set of measurements of the attenuation at that point. Endo et al.20 derived a formula that relates the noise in a reconstructed image to that in x-ray intensity measurements as follows:

\[
\left(\frac{\sigma}{I}\right)^2 = K^2 \left(\frac{\sigma_I}{I}\right)^2,
\]

where \(I\) is the intensity of transmitted x rays and \(\sigma_I\) is the standard deviation in a set of intensity measurements. \(\sigma/I\) shows the noise-to-signal ratio in x-ray intensity measurement. \(K^2\) is independent of the noise magnitude and is given by

\[
K^2 = \frac{1}{2\mu^2 a^2 n},
\]

if the Shepp and Logan correction function is used, where \(a\) is the linear sampling distance and \(n\) is the number of projection views.

The noise-to-signal ratio in x-ray intensity measurement is roughly approximated by the inverse of the x-ray photon number absorbed in the detector if additional detector noises are negligible. This condition can be applied to the present study since the CT detector we used shows a very low noise level. Therefore \((\sigma/I)^2\) is given by

\[
\left(\frac{\sigma_I}{I}\right)^2 = \frac{1}{N_X},
\]

where \(N_X\) is the x-ray photon number absorbed in the detector. It is approximated by

\[
N_X = \frac{P + S}{E_p},
\]

if \(S \ll P\), where \(E_p\) is the mean energy of the primary photon. From (A8), (A10), and (A11),

\[
\sigma^2 = \frac{K^2 \mu^2 E_p}{P + S}.
\]

If \(\sigma_0\) denotes the image noise without scattered radiation, it is given by

\[
\sigma_0^2 = \frac{\sigma_0^2}{1 + S/P},
\]

From (A11) and (A12),

\[
\sigma^2 = \frac{\sigma_0^2}{1 + S/P},
\]

and

\[
\sigma = \sqrt{\frac{\sigma_0^2}{1 + S/P}} = \sigma_0 \left(1 - \frac{S}{2P}\right),
\]

since we assume that \(S/P\) is negligible to unity.

4. Contrast-to-noise ratio

From (A3) and (A15), the contrast-to-noise ratio \(\hat{C}/\sigma\) is given by

\[
\frac{\hat{C}}{\sigma} = \frac{C}{\sigma_0} = \frac{1}{\sqrt{1 + S/P}} \approx \frac{C}{\sigma_0} \left(1 - \frac{1}{2} \frac{S}{P}\right),
\]

where \(C/\sigma_0\) is the contrast-to-noise ratio without scattered radiation.

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