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A prototype piecewise-linear dynamic attenuator

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Abstract

The piecewise-linear dynamic attenuator has been proposed as a mechanism in CT scanning for personalizing the x-ray illumination on a patient- and application-specific basis. Previous simulations have shown benefits in image quality, scatter, and dose objectives. We report on the first prototype implementation. This prototype is reduced in scale and speed and is integrated into a tabletop CT system with a smaller field of view (25 cm) and longer scan time (42 s) compared to a clinical system. Stainless steel wedges were machined and affixed to linear actuators, which were in turn held secure by a frame built using rapid prototyping technologies. The actuators were computer-controlled, with characteristic noise of about 100 microns. Simulations suggest that in a clinical setting, the impact of actuator noise could lead to artifacts of only 1 HU. Ring artifacts were minimized by careful design of the wedges. A water beam hardening correction was applied and the scan was collimated to reduce scatter. We scanned a 16 cm water cylinder phantom as well as an anthropomorphic pediatric phantom. The artifacts present in reconstructed images are comparable to artifacts normally seen with this tabletop system. Compared to a flat-field reference scan, increased detectability at reduced dose is shown and streaking is reduced. Artifacts are modest in our images and further refinement is possible. Issues of mechanical speed and stability in the challenging clinical CT environment will be addressed in a future design.

I. Introduction

Dynamic control of the x-ray flux has been proposed by several groups as a method for reducing dose or improving image quality in CT scans (1-5). The measurements in CT can have very different noise, and rays with the fewest photons can make a large contribution to the noise of the reconstructed image (6). The ability to redistribute incident flux from signal-rich rays to signal-poor rays can therefore significantly improve image quality. Simulations of this effect have shown peak variance improvements of 50% (7) to 86% (8).

For decades, beam shaping filters (also called bowtie filters) have been used for this very reason, and also because they reduce the scatter-to-primary ratio and the dynamic range required of the detector. Tube current modulation has also become standard in clinical CT as another form of flux control (9). However, the static beam shaping filter combined with tube current modulation still leave room for improvement.

More recently, several groups have explored dynamic attenuators (or dynamic beam shaping filters) as the next level of sophistication in flux control (2-5, 10-12). These dynamic attenuators provide additional degrees of freedom and can personalize the radiation exposure to the patient and clinical task. For example, one of these dynamic attenuators allows the system to select among a family of thickness functions parameterized by a single variable (11), such as a family of thickness functions adapted to patients of different sizes. When scanning an elliptical patient, with a different width along the major and minor axis, the dynamic attenuator could adjust itself on a view-by-view basis to better conform to the patient anatomy. We have compared the performance of several of these dynamic attenuators, along with algorithms which may be used to control them to optimize image quality, in Ref. (13).

We proposed the piecewise-linear attenuator as a dynamic attenuator that is able to produce any piecewise-linear thickness function in each view (1). This provides a great deal of flux control, and also produces a thickness function which is continuous. Previously, we showed in simulation that this dynamic attenuator can reduce dose by 30% without increasing the maximum variance in a scan (14), and it can reduce the impact of count rate limitations of photon-counting detectors (15). A key advantage of the piecewise-linear attenuator that has not been explored is its continuous attenuation profile, which we hypothesized would reduce artifacts compared to other designs which are discontinuous. The purpose of this work is to provide an experimental demonstration of a prototype, piecewise-linear attenuator, in order to demonstrate low imaging artifacts with improvements in dose and noise objectives.

A related concept, the digital beam attenuator, has also been proposed as a dynamic attenuator (2, 3, 12). The digital beam attenuator shares many similarities with the piecewise-linear attenuator, and can be considered an attenuator that produces a piecewise-constant thickness instead of a piecewise-linear one. The level of flux control is similar. The digital beam attenuator has been experimentally demonstrated to reduce variance at a fixed dose (12). However, significant ring artifacts were present with the digital beam attenuator, and some of these artifacts persist even after data substitution techniques were used. The work presented here is intended to be complementary to the existing experimental work with the digital beam attenuator. The primary goal of our work is to demonstrate low artifact levels (on the order of 10 HU) with the piecewise-linear attenuator. A secondary goal is to demonstrate improved image quality at reduced radiation dose, with the understanding that similar, compelling results have already been realized with the digital beam attenuator (12).

II. Materials and Methods

II. A. Piecewise-linear design

We briefly review the mathematical theory for the piecewise-linear attenuator. The reader is directed to Ref. (1) for further information.

A piecewise-linear function can be described using triangle functions as a basis set. Any piecewise-linear function with control points at $x = -N, -N+1, -N+2 \dots N-1, N$ can be decomposed as follows:

$$f(x) = \sum_{k=-N}^N \ell_k \Lambda(x - k) \quad (1)$$

$\Lambda(x)$ is a triangle function with FWHM of 1. By varying the values of ℓ_k , we allow the function to take on any desired piecewise-linear form.

These triangle functions are implemented physically using wedges with triangular cross-sections whose height varies along its length. These wedges are connected to linear actuators which move the wedges in the z-direction, parallel to the axis of rotation of the CT scanner, and as the wedge is translated the thickness of the triangle exposed to the beam changes. Within each layer the wedges are offset by the triangle base (or twice the FWHM of the triangle), so that the shadows of the wedges on the detector touch. The two layers are themselves shifted by half the triangle base (or the FWHM of the triangle). Therefore, the apex of a triangle in one layer aligns with the vanishing point of two triangles in the other layer.

Figure 1 shows a drawing of the prototype dynamic attenuator as well as an ideal dynamic attenuator. In this prototype, due to the size of the components, the wedges and actuators are split into a set above the beam and a set below the beam. However, if the actuators could be sufficiently miniaturized, the entire set could be on one side, as shown in Figure 1(c-e). The motion of the k^{th} actuator shifts a wedge upwards or downwards and directly adjusts $\ell(k)$. In Figure 1(d), the part of the attenuator that directly intersects with the collimated beam is isolated. It can be seen that wedges that have been translated upwards contribute greater thickness into the beam. Because the individual wedges taper smoothly to zero thickness, the attenuation as a function of fan angle will remain continuous. This suppresses artifacts that may arise from discontinuities in attenuation.

II. B. Fabrication

Figure 1 shows a drawing of the prototype dynamic attenuator. Figure 2 shows the constructed dynamic attenuator. For this first prototype system, we simplified the design. Because the field of view of our tabletop CT scanner is much smaller than a clinical CT scanner (about 25 cm), and because of the size limitations, six wedges and actuators were used instead of the originally proposed 15. The geometry is also simplified. In Figure 1, the wedges are arranged in two layers and placed in equiangular increments. In our experimental prototype, the wedges are still arranged in two layers, but in order to maximize the number of wedges available to the system given the size of the actuators, wedges in each layer are driven from both directions rather than only from below.

The frame consisted of acrylic sheets that were laser cut. The laser cut acrylic pieces were connected at right angles using captive bolts and nuts. An earlier version of the frame was 3D printed. We found that rapid prototyping technologies enabled rapid iteration of the frame design. The frame design was not completely rigid, and some flexion of the frame could be felt under pressure. It was observed that motion from the actuators created stress in the frame, and the nature of the stress was not totally reproducible. While we were

concerned that this lack of rigidity would result in artifacts in the reconstructed images, the final images appear acceptable.

The continuously varying thickness of the piecewise-linear attenuator as a function of fan angle is intended to reduce imaging artifacts at transitions between wedges compared to a piecewise constant attenuator. However, it is difficult to machine wedges which truly taper to zero at their edges, because such knife edges are easily damaged in the machining process. We investigated several machining techniques and found that electric discharge machining (or spark machining) produced the best results. To further improve sharpness at the edge, the edges of the wedges were honed with the same techniques that are used to sharpen knives. The wedges themselves were machined from stainless steel. The portion of the wedge inserted into the beam is 10 mm in width and 4 mm at the thickest point. The attenuating portion of the wedge was attached to a square stem holder, which can be seen in Figure 2. The square stem holder is stabilized using two layers of acrylic sheets, which is also seen in Figure 2. These acrylic sheets contained square holes which were laser cut to be a size slightly larger than the size of the square stem holder. When only a single layer of acrylic was used to stabilize the wedge motion, the resulting artifacts from uncontrolled lateral motion were unacceptable. Stiffer actuators could be used in the future to reduce lateral motion.

The wedges were tapped and connected to Firgelli L12-P actuators (Firgelli Technologies Inc., Victoria BC). Firgeli LAC control boards were used to drive these actuators. The control boards themselves were driven via USB interface by a custom computer program. The wedges could not be screwed in tightly because the orientation of the wedge would not be correct. The orientation of the wedge was held by using the square guide holes in the frame.

II. C. Artifacts

Artifacts arising from the prototype can be traced to either stochastic or deterministic causes. Stochastic errors arise from mechanical limitations and, in particular, from noise in the actuators. Stochastic errors could be improved with improved actuators or using a motion detection system, such as an optical encoder, that is able to sense the position of the wedge with higher precision and accuracy than the actuator. Deterministic errors arise from our inability to perfectly model physical processes such as beam hardening, off-focal radiation, focal spot motion, and scatter. Often, these artifacts are amplified near discontinuities of the wedges and produce ring artifacts. These deterministic errors could be alleviated with more sophisticated data models and processing.

Wedge stochastic noise—A reasonable concern for the piecewise-linear dynamic attenuator is that actuator noise would lead to prohibitively large errors in the context of a high-speed clinical CT scanner. We used simulation to assess this concern.

We assumed that in each view, the true wedge position is distributed about the expected wedge position following a Gaussian with standard deviation of $\sigma_{location}$. We estimate $\sigma_{location} = 0.1 \text{ mm}$ for our actuators. If optical linear encoders were used, the precision of the

wedge location could be $<1 \mu\text{m}$. We assume that within each view, the errors of different wedges are uncorrelated.

The error in the wedge position in the axial direction (recall that the wedges translate in the axial direction) directly produces an error in the wedge thickness. The constant of proportionality is determined by the geometry of the wedge. The triangular portion of our wedges (not including the square stem) is 40 mm in axial extent and 0 to 4 mm in thickness. Therefore, in our geometry, we assumed the standard deviation in thickness is

$$\sigma_{\text{thickness}} = \frac{4}{40} \sigma_{\text{location}} = 10 \mu\text{m}.$$

In the best case, the errors in the wedge location in a view would be uncorrelated with the errors in any other view. However, the errors could be correlated with adjacent views. We use a simple linear model to address these correlation effects, as follows. We assume that a clinical scanner uses 1000 views. For each wedge, we generate 100 Gaussian random variables with standard deviation $\sigma_{\text{thickness}}$. These random numbers are then upsampled by a factor of 10 using linear interpolation to produce the error for each view index. Next, we convert errors in thickness to errors in attenuation measurement, assuming a monoenergetic beam of 60 keV and an attenuator composed entirely of Fe. The actual measurements are polychromatic, but we believe that the error induced by the monoenergetic approximation is small in assessing these artifacts. These errors are filtered and backprojected to estimate artifacts in reconstructed images.

Beam Hardening Correction—The use of stainless steel wedges significantly hardens the beam. A beam hardening correction is therefore of high importance.

Beam hardening was compensated using an analytic model. We used the TASMIP model to estimate the 100 kVp tube spectrum (16, 17) and simulated passage of the spectrum through varying thicknesses of water for several fixed thicknesses of iron. For any given thickness of iron, a quadratic polynomial was fit and used to convert log-normalized measurements of detected counts into estimates of water length. These estimates of water length were then reconstructed. This form of water beam hardening correction with an iron dynamic attenuator was previously shown to be competitive with standard water beam hardening correction with a conventional, water-like bowtie filter (1), although the appearance of the artifacts is different from traditional beam hardening artifacts. We assumed an ideal energy-integrating detector.

Scatter—Scatter was suppressed by collimating the beam in the slice direction. Lead plates were simply taped on in front of the dynamic attenuator, and the position was not precisely controlled. In most studies, the collimated beam had a width of about 2 cm in the axial direction at isocenter. A 10:1 anti-scatter grid was further used to reduce scatter.

II. D. Experiments

The dynamic attenuator prototype was bolted directly onto the x-ray tube. A 4030CB flat-panel detector was used (Varian Medical Systems, Palo Alto CA). The object to be scanned was placed on a rotating stage. All experiments were performed at 100 kVp. We collected

625 frames over a 360 degree rotation at a rate of 15 frames per second. The total scan time was approximately 42 seconds.

Each of our experiments involved three scans. First, a reference scan was taken with the dynamic attenuator removed but with the phantom present. Next, a primary scan was taken with the wedges moving in a pattern to compensate for the phantom. The trajectory of the attenuator was determined from the reference scan using the method which will be described in Section II.E. However, in principle, the attenuator of the trajectory can be calculated using real-time feedback of the primary scan itself, without needing the reference scan first. In the primary scan, both the phantom and the dynamic attenuator are present. Finally, an “air scan” was taken with the wedges moving according to the same trajectory, but with the phantom removed. The air scan is used to determine the thickness of steel in the beam, and is used with the primary scan to produce a beam hardening corrected sinogram. Figure 3 shows an example of these three scans.

In one instance, we scanned a uniform cylinder phantom. The cylinder phantom is an ideal phantom for seeing many types of artifacts. However, for a cylinder phantom positioned near the center, one would expect that the dynamic attenuator would be stationary, defeating the purpose of the artifact detection scan. We therefore moved the wedges according to a random trajectory.

A surrogate for dose was estimated using the air scan. Rays which intersect the object were determined using the sinogram of the object scan, and the total signal in this region was integrated in the air scan to estimate the dose. Our measurement of dose more accurately captures the energy of the photons incident on the patient rather than the total energy absorbed by the patient. However, since the same method was used to estimate dose both for the dynamic attenuator scan and the reference, attenuator-free scan, we felt that the comparison was still fair.

It should be noted in Figure 3 that the thickness of the dynamic attenuator, which is seen in the air calibration scan, does not look very piecewise-linear. This is because of two imperfections. First, the wedges are not machined into perfect triangular shapes, and second, the wedges are not perfectly at their intended location. In theory, at any point along the piecewise-linear function there should be contribution from two triangle functions. A bright stripe in Figure 3 appears when three triangle functions overlap at a single fan angle, and a dark stripe appears when there is only one. This was compensated using the air scan.

II. E. Control

The trajectory of the wedges was found by solving a convex optimization problem using the reference scan of the phantom without any dynamic attenuator. We used a linear model for the thickness introduced by the wedges as a function of fan angle γ :

$$f_{attenuator}(\gamma) = f_0(\gamma) + \sum_{i=1}^6 f_i(\gamma) \ell_i \quad (2)$$

ℓ_i is the thickness of the i^{th} wedge, which is directly proportional to its location and is controlled by its corresponding actuator. $f_0(\gamma)$ is the total thickness of the dynamic attenuator (including the effect of all constituent wedges) when the positions of all the wedges are set to zero. In Equation (2), we define zero to be a baseline position, with the wedges already partly inserted into the beam, not the state when the wedges are completely retracted. $f_i(\gamma)$ is therefore the change in thickness when the i^{th} wedge is moved by one unit further into the beam, increasing the thickness of the i^{th} wedge. $f_0(\gamma)$ and $f_i(\gamma)$ are measured experimentally. If the actuator is ideal, $f_i(\gamma)$ should be a triangle function. In practice, we find $f_i(\gamma)$ to be distorted because the wedges are not perfectly machined, and they are arranged in an equilinear fashion, in multiple parallel rows, rather than in an equiangular arrangement. We observed that this linear model does not predict $f_{\text{attenuator}}(\gamma)$ well enough to eliminate the need for the air scan, but it suffices for the purpose of controlling the wedge trajectory. For a perfectly calibrated system, an air scan should not be necessary because it would be possible to predict the air scan from the wedge control trajectory.

With the linear model defined, the wedge trajectory for the attenuator was determined using power law control, with an exponent of $\beta = 0.75$. As explained in Ref. (13), power law control is a simple way of reducing mean or peak variance for the perfect attenuator with arbitrarily fine control. Power law control with exponent β implies that for any ray with

attenuation from the phantom of $\frac{I_0}{I}$, the attenuation introduced by the attenuator should be proportional to $\left(\frac{I_0}{I}\right)^{-\beta}$. An exponent of $\beta = 1$ produces uniform signal across the detector and is a good choice for reducing peak variance. An exponent of $\beta = 0.5$ is a good choice for reducing mean variance, and under certain assumptions minimizes mean variance for fixed dose (18). Intermediate exponents, such as $\beta = 0.75$, are compromises between mean and peak variance. In general, power law control is not perfectly optimal, but it is simple to implement and does not require a priori knowledge of the patient anatomy.

The desired attenuation from the dynamic attenuator, determined by power law control and after log normalization, is therefore

$$f_{\text{power}}(\theta, \gamma) = -\beta \log \left(\frac{I(\theta, \gamma)}{I_0(\theta, \gamma)} \right) + 1 \quad (3)$$

c is a constant chosen to keep the target attenuation, in log-normalized terms, non-negative.

The piecewise-linear attenuator is only an approximation to the perfect attenuator with arbitrarily fine control. We determined the wedge thicknesses ℓ_i in each view θ by solving the following equation in the least-squares sense:

$$\alpha f_{\text{attenuator}}(\gamma) = f_{\text{power}}(\theta, \gamma) \quad (4)$$

α is a proportionality constant which converts thickness of attenuator into log-normalized attenuation. In solving Equation (4), we considered only those measurements which

intersected the object. Rays which did not cross the object were ignored in the least-squares calculation.

Power law control is appealing because it is simple and uses only information present in the current view. In a real-time control loop, the detector data from any given view could be used to determine the wedge trajectory in a subsequent view. Because the sinogram data is typically slowly changing between one view and the next, the increase in variance would likely be small. This type of control would eliminate the need of a reference scan. Other control schemes are possible, some of which require knowledge of the patient anatomy and may necessitate a low dose pre-scan to determine the wedge trajectories (13).

III. Results

III. A. Actuator Noise Simulation

Figure 4 shows simulated artifacts from actuator noise, following the method described in Section II.C. Two examples are shown: one image for the current prototype system, and another image representing a possible clinical system. In both cases, the artifacts induced by actuator noise appear to be limited to 2 HU. We assumed in the clinical system that more accurate actuators or feedback mechanisms are used. While the simulation for the clinical system includes reduced actuator noise, the wedges are also spaced closer together, and hence the simulated errors are higher spatial frequency and amplified by the ramp filter. For the same field of view, increasing the number of wedges would require an increase in the actuator precision to avoid introducing artifacts from actuator noise.

III. B. Uniform Cylinder

Figure 5 shows scans of a uniform cylinder with a 16 cm diameter and various inserts. Contrast inserts were not used, but small gaps are present around the holes of the inserts. If controlled normally, the dynamic attenuator would effectively become a stationary attenuator for this scan. As explained in Section II.D, the wedges were moved randomly on this phantom. The dynamic attenuator introduces few artifacts in this example, including most prominently a broad band artifact with a magnitude of about 10 HU.

III. C. Anthropomorphic Phantom

Figure 6 shows images of the pelvic region of an anthropomorphic phantom taken with uniform illumination (no attenuator), with the dynamic attenuator at 13% reduced dose, and with uniform illumination at four times the dose of the first scan. Some of the low contrast objects are more easily seen in the dynamic attenuator image than in the slightly higher dose reference scan. Some artifacts are also introduced with the dynamic attenuator. A series of streaks are apparent and emanate from the center of the image. This artifact is not present on the other images. We believe that this artifact results from actuator motion. For much of the scan, the center actuator is kept to the minimum position. When assigned to stay at one position, the actuator will not always reach that position but will oscillate about that target at a rate of about 1 Hz. This frequency is consistent with the pattern seen in this image for a 42 s scan. Other artifacts are reduced, including beam hardening artifacts.

Figure 7 shows images from the thorax region of the same phantom. The artifacts from the reference scan and the dynamic attenuator scan are comparable. The dose with the dynamic attenuator is reduced by 55%, although the noise has also increased.

Figure 8 shows images from the shoulder region of the same phantom. In this case, the dose with the dynamic attenuator is slightly higher than that of the reference scan. The artifacts are comparable in the two images. In this image, as with Figure 6, the dynamic attenuator appears to reduce the blooming from the bones.

Figure 9 shows images from a two-object composite. To assemble this phantom, two sections from the anthropomorphic phantom were placed side-by-side. This two-object composite was chosen to demonstrate noise streaks, which are alleviated by the dynamic attenuator. Some artifacts are introduced by the dynamic attenuator. A likely cause of these artifacts is imperfect correction for beam hardening.

IV. Discussion

The preliminary images with the attenuator prototype show improvements in dose and noise objectives. These benefits would be increased for larger objects with greater inhomogeneity in photon statistics, but would be decreased if compared to a reference scanner that employed tube current modulation and the bowtie filter. Nonetheless, our previous simulation work leads us to expect dose reductions of 30% in clinically relevant applications (7, 8, 10, 13, 14).

The artifacts present in our images are generally comparable to the artifacts in the reference images with uniform illumination. Compared to the reference images, the dynamic attenuator images include a beam hardening correction and more homogeneous signal statistics. The latter is relevant for ring artifacts, which are more pronounced with our detector at low signal levels. The artifacts seen in these images represent a significant improvement over the digital beam attenuator, which is essentially the present system with the use of rectangular wedges rather than triangular ones in the fan-angle direction. The present artifact levels may still be unacceptable for clinical systems. Nonetheless, the results are very encouraging.

In simulations, the problem of actuator noise appears to lead to artifacts of only 2 HU. We note that this level is dependent on a number of assumptions. For example, if we assumed that the noise of the actuator in each view was uncorrelated with the others, the artifact levels would be reduced. However, it seems likely that if linear encoders are used with submicron precision, the effect of actuator noise would be negligible.

Some residual artifacts remain from beam hardening. Our correction for beam hardening is based on theory and not derived from empirical data. More sophisticated beam hardening corrections will need to be incorporated in the future.

Scatter is problematic for the dynamic attenuator because existing methods to estimate and subtract scatter assume a fixed bowtie filter. In this work, scatter was not estimated in the present images but was instead controlled with narrow collimation and an anti-scatter grid.

The problem of scatter will be aggravated in larger phantoms and wider cone angles. A wide variety of methods exist for estimating scatter in conventional systems (19, 20). Many of these approaches would need to be adapted to the dynamic attenuator.

In this present work, we measured both an object scan and an air scan. The requirement to measure an explicit air scan would be a severe impediment to clinical adoption. An alternative to measuring the air scan is to predict it using a model. We found that the linear model for predicting the thickness of attenuator $f_{attenuator}$ from Equation (2) was not sufficiently accurate for this purpose. The current actuators also show hysteresis. This hysteresis is amplified by the method used to stabilize the wedges, which are currently fit snugly through two square holes in acrylic, as seen in Figure 2. The square holes are only slightly larger than the wedge, and the frictional forces which hold the wedge are not always consistent. Better methods of guiding the wedges will be investigated in a future prototype.

Notwithstanding these limitations, the level of artifacts observed and the improvements in dose and noise are encouraging. Work is underway to validate these results in larger fields of view and also to experimentally demonstrate possible advantages with photon-counting detectors (15, 21).

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Figure 1a

Figure 1b**Figure 1.**

Drawings of the piecewise-linear dynamic attenuator. (a) A side view of the prototype with eight actuators. Four of the actuators are hidden behind the other actuators in this illustration. X-rays proceed along the red dashed line. In this figure, the wedges are completely retracted. (b) A perspective view of the prototype. (c) Schematic of a full dynamic attenuator covering a larger fan beam. The actuators are not shown. The wedges all lie on one side of the beam for clarity. The fan beam is denoted with gray lines. (d) Cross-section of the same image, showing the volume that intersects with the path of the x-rays. Wedges that have been translated upwards insert greater thickness into the beam (e) A separate perspective.

Figure 2.
(Left) Prototype dynamic attenuator, mounted directly to the x-ray tube. Three actuators are above and three actuators are below, with the wedges in between them. (Right) A close up of the wedges. Two layers of acrylic are held in place using brass nuts and bolts, and these acrylic layers stabilize the wedge motion.

Figure 3.

Example sinogram for the thorax dataset. Data is shown after log normalization. (Left) Reference scan with uniform illumination. (Middle) Scan with the phantom present, and the dynamic attenuator controlled to reduce dose and improve image quality. (Right) Air calibration scan with the phantom removed, but with the dynamic attenuator moving in the same path. Note that where the patient attenuation is large, the dynamic attenuator thickness is reduced, and where the patient attenuation is small, the dynamic attenuator thickness is increased. Arrows point to bright and dark streaks where there is unintended overlap or gap between adjacent wedges. The left sinogram is used to generate our reference image. The middle and right sinograms are processed using the beam hardening correction to produce the dynamic attenuator reconstruction.

Figure 4.

Simulated artifacts from stochastic noise in the actuators in HU. We assumed that the errors in actuators were correlated from view-to-view, so that 100 effective views were used. (Left) 6 wedges, $\sigma_{location} = 100 \mu m$. Field of view is 25 cm. This is a simple model for our prototype system. (Right) 15 wedges, $\sigma_{location} = 30 \mu m$. Field of view is 50 cm. This is a simple model for a larger, next-generation system.

Figure 5.

Uniform cylinder with random motion. (Left) Logarithm of raw data, prior to air scan normalization. The presence of the wedges and their motion can be seen in the horizontal striped pattern. The wedges were controlled to move randomly. (Middle) Logarithm of the air scan with the wedges moving. (Right) Reconstructed cylinder. An arrow points to the band artifact. WW = 200 HU.

Figure 6. Pelvis dataset. (Top left) Reference scan with uniform illumination (no flux control). WW = 200 HU. (Top right) Scan with the dynamic attenuator at 13% less dose. WW = 200 HU. A black arrow points to a bone insert, and a white arrow points to a low contrast object that is arguably more visible with the dynamic attenuator than in the reference scan. (Bottom left) Reference scan with uniform illumination at four times the dose of the first reference scan. WW = 200 HU. (Bottom right) Same as bottom left, but with WW = 400 HU. This shows that the bone inserts which are visible in the dynamic attenuator scan, but not visible in the reference scans, are still present but were obscured by beam hardening effects.

Figure 7.
Thorax dataset. (Left) Reference scan with uniform illumination. (Right) Dynamic attenuator scan at 55% less dose. WW = 200 HU.

Figure 8.
Shoulder dataset. (Left) Reference scan with uniform illumination. (Right) Dynamic attenuator scan at 10% increased dose.

Figure 9. Two object composite dataset. (Top) Reference scan with uniform illumination. (Bottom) Dynamic attenuator scan at 30% less dose. Noise streaking is reduced with the dynamic attenuator, although some new artifacts are also introduced. WW = 200 HU.