Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2018 February ; 10573: . doi:10.1117/12.2293696.

Dynamic beam filtering for miscentered patients

Andrew Mao, William Shyr, Grace J Gang, and J Webster Stayman

Department of Biomedical Engineering, Johns Hopkins University, Baltimore, MD 21205

Abstract

Purpose—Accurate centering of the patient within the bore of a CT scanner takes time and is often difficult to achieve precisely. Patient miscentering can result in significant dose and image noise penalties with the use of traditional bowtie filters. This work describes a system to dynamically position an x-ray beam filter during image acquisition to enable more consistent image performance and potentially lower dose needed for CT imaging.

Methods—We propose a new approach in which two orthogonal low-dose scout images are used to estimate a parametric model of the object describing its shape, size, and location within the field of view (FOV). This model is then used to compute an optimal filter motion profile by minimizing the variance of the expected detector fluence for each projection. Dynamic filtration was implemented on a cone-beam CT (CBCT) test bench using two different physical filters: 1) an aluminum bowtie and 2) a structured binary filter called a multiple aperture device (MAD). Dynamic filtration performance was compared to a static filter in studies of dose and reconstruction noise as a function of the degree of miscentering of a homogeneous water phantom.

Results—Estimated filter trajectories were found to be largely sinusoidal with an amplitude proportional to the amount of miscentering. Dynamic filtration demonstrated an improved ability to keep the spatial distribution of dose and reconstruction noise at baseline levels across varying levels of miscentering, reducing the maximum noise and dose deviation from 53% to 15% and 42% to 14% respectively for the bowtie filter, and 25% to 8% and 24% to 15% respectively for the MAD filter.

Conclusion—Dynamic positioning of beam filters during acquisition improves dose utilization and image quality over static filters for miscentered patients. Such dynamic filters relax positioning requirements and have the potential to reduce set-up time and lower dose requirements.

Keywords

Patient centering; dynamic filtration; fluence-field modulation; multiple aperture devices

1. INTRODUCTION

A common technique for dose reduction in CT is the use of bowtie filters. These filters generally shape the x-ray beam (through selective attenuation) to decrease the incident fluence for measurements expected to have lower overall attenuation. This reduces exposure

for measurements that do not require high fluence. As such, bowtie filters are often designed to flatten the fluence profile arriving at the detector. However, existing CT scanners cannot translate the bowtie, resulting in poor dose utilization, reduced image quality, and CT number variations when the patient is not centered within the bore.¹⁻³ Toth et. al. have reported that patients are routinely miscentered in elevation by an average of 2.3 cm and up to 6 cm clinically. Objects miscentered 3 cm and 6 cm can experience an 18% and 41% increase in dose and a 6% and 22% increase in noise respectively.¹ Similar concerns arise in emergency medicine where physicians often have 1) limited prior knowledge of the location of potential disease and therefore need to visualize the entire body volume with high image quality; and 2) limited time to properly position the patient. For these reasons, emergency room (ER) CT scanners often operate without a bowtie, thus forsaking the dose-saving benefits of beam filtration.

Both miscentering in ordinary diagnostic CT and bowtie-free ER scanning would benefit greatly from an automatic method to determine the patient's position in the bore and dynamically position the beam filter during acquisition. In this work, we present a general method to calculate dynamic motion profiles for any type of beam filter, and demonstrate its ability to alleviate possible concerns about dose and image quality. We demonstrate the methodology in physical experiments using two beam filtration approaches: 1) a traditional bowtie filter with variable thickness; and 2) a filter based on a multiple aperture devices (MAD) that uses a binary filter with very fine slits to achieve spatial modulation of the x-ray beam. The MAD-based filter^{4,5} is an emerging technology that can shape the x-ray beam while maintaining a very compact profile. This is important since modern CT gantries allow little room for additional hardware in front of the x-ray tube. In essence, MADs operate under the following principle: wider slits permit higher local transmissivity while thinner slits restrict local fluence. Since the slits are small relative to resolution limiting elements of the system (focal spot and detector pixels), the induced modulation patterns can be relatively smooth.

2. THEORETICAL AND EXPERIMENTAL METHODS

2.1 Patient Position Estimation

To estimate the patient's position within the FOV, we propose a parametric model for the object shape, $\alpha(\Omega)$, in the central axial plane as a rectellipse function. This model has a shape parameter, s , for a continuously variable shape between $s = 0$, an ellipse, and $s = 1$, a rectangle. Additional parameters define its center position (x_c, y_c) , width w , height h , and uniform attenuation coefficient μ . Mathematically, the parameterized object is given by the quartic Cartesian equation

$$\left(\frac{u}{w}\right)^2 + \left(\frac{v}{h}\right)^2 - s^2 \left(\frac{uv}{wh}\right)^1 \leq 1 \quad (1)$$

where $u = x - x_c$ and $v = y - y_c$ represent the translated coordinates of x and y respectively.⁶

Low-dose scouts/topograms of the patient are acquired at two orthogonal positions – the anteroposterior (AP) and lateral views – from which the central row measurements are used to form a data vector g . The six parameter object model $\Omega = \{x_c, y_c, w, h, s, \mu\}$ is then optimized by minimizing the mean squared error between the forward projected object and g according to

$$\hat{\Omega} = \arg \min_{\Omega} \|\mathbf{P}o(\Omega) - g\|_2^2 \quad (2)$$

where $\mathbf{P}$ is the fan-beam forward projector of x that generates the AP and lateral views using the Siddon approach.⁷ In this work, the optimization is performed using the Nelder-Mead simplex method.⁸

Using $\alpha(\hat{\Omega})$ the linear miscentering in the AP and lateral directions is $^{\wedge}$

$$\begin{aligned} e_{\text{ap}} &= (x_c - x_0)a_x \\ e_{\text{lat}} &= (y_0 - y_c)a_y \end{aligned} \quad (3)$$

respectively, where x_0 and y_0 are the center coordinates of the central-slice, axial image and a_x and a_y are the axial image voxel sizes.

2.2 Filter Trajectory Estimation

2.2.1 Filter Calibration—Given the source-to-filter distance (SFD) in the scanner (which is assumed to be fixed and parallel to the imaging plane) and the design specifications of the beam filter (e.g. thickness of the bowtie along its width), the expected modulation pattern $f(t)$ for a specified translation t can be calculated analytically based on ray tracing. Alternatively, when $f(t)$ is not straightforward to compute (e.g., in MAD-based filtering due to obliquity effects) a prior calibration scan can be conducted by discretely translating the filter through the range of achievable actuation positions, and then interpolating to find the modulation patterns at specific actuation positions. Note that in this work calibration scans to obtain $f(t)$ for the MAD filter are additionally low-pass filtered to avoid fitting the high-frequency spatial modulations created by the MAD.

2.2.2 Estimation of Filter Actuation—Multiplying these fluence profiles with the forward projected object model yields the expected detector fluence as a function of t and the projection angle θ . To find the optimal translation of the filter at each projection $\hat{t}(\theta)$, we consider two approaches:

1. Analytically match the ray passing through the center of the object with the center of the beam filter at the SFD, i.e.

$$\hat{t}(\theta) = SFD \frac{r \cos(\theta - \phi)}{SAD + r \sin(\theta - \phi)} \quad (4)$$

where

$$\begin{aligned} r &= \sqrt{a_x^2(x_c - x_0)^2 + a_y^2(y_0 - y_c)^2} \quad (5) \\ \sin \phi &= a_x \frac{x_c - x_0}{r} \\ \cos \phi &= a_y \frac{y_0 - y_c}{r} \end{aligned}$$

The derivation for this expression can be found in Kak and Slaney 2001.⁹

2. Promote uniform detector fluence by minimizing the spatial variance of the expected detector fluence, i.e.

$$\hat{t}(\theta) = \arg \min_t \text{Var} \left\{ f(t) e^{-\mathbf{P}(\theta) \circ \hat{\Omega}} \right\} \quad (6)$$

where $\mathbf{P}(\theta)$ is the Siddon fan-beam forward projector of $\alpha(\Omega)$ at angle θ and $\text{Var}\{\cdot\}$ computes the spatial variance of the signal across the length of the detector.

2.3 Phantom Dose and Noise Studies

To investigate the efficacy of the above method in mitigating the dose and image noise penalties of miscentering, experiments were conducted on a 16 cm CTDI phantom scanned at approximately 0, 2, and 4 cm of miscentering with and without dynamic filter positioning for two beam filters: an aluminum bowtie and a single MAD. For the aluminum bowtie, the simple estimation approach based on matching the object center (4) was applied; and for the MAD filter, the spatial variance minimization approach (6) was used. The CBCT test bench and beam filters used for the experiments are shown in Figure 1. Experiments were performed at 100 kV, 92 mAs (total), and 10°/15° vertical beam collimation for dose/noise measurements, respectively.

Dose measurements were made in the 16 cm CTDI phantom, which mimics the size and attenuation of the human head—a common subject of imaging studies in the ER. The accumulated dose over a full 360° projection scan for each dosimeter location (as labeled in Figure 1a) d_i is measured, and the weighted dose is calculated using the standard weighted average of central and peripheral dose

$$CTDI_w = \frac{1}{3}d_1 + \frac{2}{3}\frac{\sum_{i=2}^5 d_i}{4} \quad (7)$$

To analyze noise, two sequential data acquisitions were performed and reconstructed with the Feldkamp (FDK) p approach.¹⁰ Difference images were formed to create noise-only image volumes with a $1/\sqrt{2}$ correction factor applied. A $7 \times 7 \times 7$ voxel box standard deviation filter is convolved with the difference image, and the central slice of the result is used to qualitatively assess the noise distribution. Additionally, $31 \times 31 \times 7$ voxel ROIs are also defined over each of the 5 dosimeter locations in the central volume of the difference image and used to quantitatively assess the standard deviation of voxel values.

FDK reconstructions are performed with a Hamming window, 0.8 Nyquist frequency cutoff, $512 \times 512 \times 220$ voxel image volume and $0.5 \times 0.5 \times 0.5$ mm voxel size. For bowtie filter acquisitions, a simple additional gain scan of the bowtie at each position in the filter trajectory is necessary. However, for the single MAD acquisitions, the high-frequency modulations produced by the MAD are susceptible to focal spot changes (blooming and/or shifting) as well as spectral effects, necessitating a modified reconstruction pipeline. We model mean measurements $\bar{y}$ for MAD acquisitions on our flat panel detector system as

$$\bar{y}(\theta) = g_S g_D g_M(\theta) e^{-\alpha(\theta)\mu l} \quad (8)$$

where g_S is the emitted x-ray distribution from the tube, g_D is the detector sensitivity, and g_M is the spatial MAD transmissivity. The α term is a first-order spectral correction factor that was calibrated using specialized scans of variable thickness slabs. To accommodate subtle changes in g_M due to focal spot variations, fiducials in projection data are used to estimate focal spot shift/blur and appropriately modify g_M from a calibration scan. The measurements are assumed to be uncorrupted by scatter.

3. RESULTS AND BREAKTHROUGH WORK

The calculated filter trajectories for both the aluminum bowtie and the single MAD are shown in Figure 2. Consistent with intuition, these trajectories are largely sinusoidal with the amplitude equal to half the miscentering offset, which is consistent with our CBCT system geometry where the magnification of the filter, $SAD/SFD \approx 2$. For the MAD filter, trajectories are similarly sinusoidal but not as smooth, which may be attributable to non-smooth transitions between the measured MAD modulation patterns. In the MAD case, the variance minimization approach yields a larger amplitude than the center matching approach because, for a (left) miscentered object and fan-beam projection, the projection is skewed to the left of the projected object center. We note here that there is more variation in beam width and amplitude induced by the MAD filter as a function of displacement due to obliquity effects, as seen in Figure 2b. Such amplitude variations could be mitigated using tube current modulation, which was not applied in this study.

Figure 3 demonstrates the results of the phantom studies. For the static aluminum bowtie, miscentering has a large effect on both dose and noise: for 4 cm of offset (toward the left of the panel), location 3 sees a 39% dose increase and a 22% noise decrease while location 5 sees a 42% dose decrease and a 53% noise increase relative to the “centered” condition. With a dynamic bowtie, these variations are largely mitigated, with the dose and noise remaining close to baseline levels for all dosimeter locations as seen in Figure 3. Figure 4 shows that use of the dynamic filter promotes homogeneous noise distribution whereas the static filter causes a significant noise increase on the left and decrease on the right.

The static MAD data shows similar dose and noise penalties as observed with the bowtie. The overall noise levels in Figure 4 are somewhat higher, which is attributed to the overall lower fluence levels of the MAD (55% vs 90% maximum transmissivity as shown in Figure 2b). The MAD fluence profiles are also narrower than those of the bowtie and are suboptimal for promoting uniform detector fluence with a 16 cm CTDI phantom, leading to the higher observed central noise in Figure 4. Additional narrowing and decreasing fluence levels for large shifts (9% lower maximum transmissivity for a 2 cm shift corresponding to 4 cm offset as seen in Figure 2b) leads to overall dose decreases for all locations at 4 cm miscentering even with the dynamic MAD. Notably, location 3 experiences the largest and location 5 the smallest decrease in dose, which is again consistent with obliquity effects and the variance minimization approach. Despite these differences, Figure 4 demonstrates that the dynamic MAD is still able to distribute noise similarly to the baseline while the static filter cannot.

4. DISCUSSION AND CONCLUSION

A system was developed to dynamically position the beam filters during CT image acquisition, promoting consistent image performance and improved dose utilization for arbitrary patient positioning in experiments as compared to static filtration strategies. The dynamic beam filtering scheme can return the benefits of beam filtration to challenging clinical settings like ER CT, where it can both improve clinical workflow and reduce sensitivity to set-up errors, and is generally applicable to a variety of beam filters, including emerging fluence-field modulation approaches like MAD filters. Future work will further analyze different approaches for trajectory estimation, extend the method to dual MAD filters that permit modulation of both beam width and centering for generalized fluence control, and investigate the efficacy of the method for anthropomorphic objects that may be encountered in the clinical arena.

Acknowledgments

This work was supported in part by NIH Grant U01EB018758. The authors would like to thank Dr. Russell Taylor (Department of Computer Science, Johns Hopkins University) for his valuable advice and feedback on this work.

References

1. Toth T, Ge Z, Daly MP. The influence of patient centering on CT dose and image noise. *Medical Physics*. 2007; 34:3093–3101. [PubMed: 17822016]

2. Habibzadeh, MA., Ay, MR., Asl, ARK., Ghadiri, H., Zaidi, H. The influence of patient miscentering on patient dose and image noise in two commercial CT scanners. XII Mediterranean Conference on Medical and Biological Engineering and Computing 2010 IFMBE Proceedings; 2010. p. 327-330.
3. Szczykutowicz TP, DuPlissis A, Pickhardt PJ. Variation in CT number and image noise uniformity according to patient positioning in MDCT. American Journal of Roentgenology. 2017; 208(5): 1064–1072. [PubMed: 28267350]
4. Stayman JW, Mathews A, Zbijewski W, Gang G, Siewerdsen J, Kawamoto S, Blevis I, Levinson R. Fluence-Field Modulated X-ray CT using Multiple Aperture Devices. Proc of SPIE Medical Imaging 2016: Physics of Medical Imaging. 2016; 9783:97830X1–6.
5. Mathews AJ, Tilley S, Gang G, Kawamoto S, Zbijewski W, Siewerdsen JH, Levinson R, Stayman JW. Design of dual multiple aperture devices for dynamical fluence field modulated CT. Conf Proc Int Conf Image Form Xray Comput Tomogr. 2016
6. Guasti MF. Analytic geometry of some rectilinear figures. Int J Math Educ Sci Technol. 1992; 23:895–901.
7. Siddon RL. Fast calculation of the exact radiological path for a three-dimensional CT array. Medical Physics. 1984; 12(2):252–255.
8. Nelder JA, Mead R. A simplex method for function minimization. The Computer Journal. Jan.1965 7:308313.
9. Kak, AC., Slaney, M. Principles of Computerized Tomographic Imaging. Society for Industrial and Applied Mathematics; Philadelphia, PA, USA: 2001.
10. Feldkamp LA, Davis LC, Kress JW. Practical cone-beam algorithm. J Opt Soc Am A. 1984; 1(6): 612–619.

Figure 1.

(a) CBCT test bench with Varian 4343CB detector and Varian Rad-94 x-ray tube. CTDI dosimeter locations are numbered and an arrow with dotted line indicates the direction of miscentering. (b) Aluminum bowtie and (c) single MAD filter.

Figure 2.

Fluence as a function of actuation (a) calculated for the aluminum bowtie, and (b) measured for the MAD filter. (c) Estimated filter trajectories for each filter and miscentering scenario.

Figure 3.

Dose and noise results for the static/dynamic aluminum bowtie and MAD filter. Dose and noise at each dosimeter location (locations 1–5) are plotted as a function of the amount of left miscentering $-e_{ap}$.

Figure 4. Noise maps for each scenario showing the static/dynamic cases split into the top/bottom halves (due to symmetry). ROIs used to quantify noise in Fig. 3 are labeled in the top-left and match the numbering shown in Fig. 1a.