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Abstract. Traditional CT image acquisition uses bowtie filters to reduce dose, x-ray scatter, and detector dynamic range requirements. However, accurate patient centering within the bore of the CT scanner takes time and is often difficult to achieve precisely. Patient miscentering combined with a static bowtie filter can result in significant increases in dose, reconstruction noise, and CT number variations, and consequently raise overall exposure requirements. Approaches to estimate the patient position from scout scans and perform dynamic spatial beam filtration during acquisition are developed and applied in physical experiments on a CT test bench using different beam filtration strategies. While various dynamic beam modulation strategies have been developed, we focus on two approaches: (1) a simple approach using attenuation-based beam modulation using a translating bowtie filter and (2) dynamic beam modulation using multiple aperture devices (MADs)—an emerging beam filtration strategy based on binary filtration of the x-ray beam using variable width slits in a high-density beam blocker. Improved dose utilization and more consistent image performance with respect to an unmodulated baseline (static filter) are demonstrated for miscentered objects and dynamic beam filtration in physical experiments. For a homogeneous object miscentered by 4 cm, the dynamic filter reduced the maximum regional noise and dose penalties (compared with a centered object) from 173% to 16% and 42% to 14%, respectively, for a traditional bowtie, 29% to 8% and 24% to 15%, respectively, for a single MAD, and 275% to 11% and 56% to 18%, respectively, for a dual-MAD filter. The proposed methodology has the potential to relax patient centering requirements within the scanner, reduce setup time, and facilitate additional CT dose reduction.

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1 Introduction

A common technique for dose reduction in x-ray computed tomography (CT) is the use of bowtie filters. These filters typically shape the x-ray beam spatially through selective attenuation to decrease the fluence incident to the patient for measurements expected to have lower attenuation. Consequently, this reduces the exposure for measurements that do not require high fluence. More specifically, bowtie filters are often designed to flatten the fluence profile arriving at the detector.^{1,2} However, clinical CT scanners use static filtration without the ability to translate the bowtie filter. This results in poor dose utilization, reduced image quality, and CT number variations (e.g., due to artifacts) when the patient is not well centered within the bore.^{3–5} Toth et al.³ have reported that patients are routinely miscentered in elevation by an average of 2.3 cm and up to 6 cm clinically. In those studies, miscentering of 3 and 6 cm showed an 18% and 41% increase in dose and a 6% and 22% increase in noise, respectively.

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. One could eliminate the bowtie filter, but

one would forgo the dose-saving benefits of beam filtration. Although techniques have been developed for automatic centering which involve readjusting the patient in the bore,⁶ both miscentering in ordinary diagnostic CT and bowtie-free ER scanning would benefit greatly from a method to dynamically position the beam filter during acquisition, thereby improving the overall efficiency of the CT examination without disrupting the clinical workflow.

This work presents a general method to calculate actuation profiles for various types of dynamic beam filters based on two low-dose “scout” scans of the patient (i.e., topograms). Such scout images are routinely obtained for positioning to ensure that the region-of-interest is in the field-of-view (FOV),⁷ to reasonably center patients,⁸ and to determine an appropriate x-ray technique and other acquisition parameters.^{9,10}

Various dynamic beam filtering strategies for CT have been investigated including double-wedge systems,¹¹ piecewise-linear attenuators,^{12,13} and fluid-filled bowties.^{14,15} Such methods rely on variable attenuation of filter elements and moving those elements within the beam path. A related strategy using digital beam attenuators^{16,17} achieves beam modulation by combining several exposures with an essentially binary filter at different locations. An emerging technology for dynamic beam filtration based on multiple aperture devices (MADs)^{18,19} can

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shape the x-ray beam while maintaining a compact profile (i.e., the MAD filters are ~ 2 -mm thick). This is important as modern CT gantries allow little room for additional hardware in front of the x-ray tube, and finding compact beam filtration solutions and actuation hardware within the limited space of a clinical CT scanner is a challenge.

MAD filters are essentially binary and block or pass x-rays using small slit-shaped apertures in a high-density material (e.g., tungsten). MADs operate under the following principle: wider slits permit higher local fluence while thinner slits restrict local fluence. Moreover, the local size and spacing of the tungsten slits can be optimized for two MADs placed in a series in front of the x-ray beam to achieve a wide range of modulation patterns. In essence, the two binary MAD filters create low-frequency moiré patterns of varying shapes and widths, which can be varied via relative displacement of one MAD (MAD1) with respect to the other (MAD0). These modulation patterns become cyclic for translations greater than one MAD “period,” or the distance between neighboring apertures. Similarly, the center of the beam profile may be shifted by changing the absolute positioning of both filters simultaneously. As the slits are small relative to the resolution limiting elements of the system (e.g., the x-ray focal spot), the induced modulation patterns can be relatively smooth even though the filters are binary. A dual-MAD-based filtration scheme offers both centering and width control of the beam with small (mm scale) linear actuation, though a single MAD may also be used to adapt the center of a fixed fluence modulation profile.

In this paper, we describe strategies to drive dynamic beam modulation for miscentered patients based on knowledge of the set of beam patterns achievable with a given modulation strategy as well as scout scan data. We demonstrate the methodology in physical experiments using three beam filtration approaches: (1) a traditional aluminum bowtie filter with the capability of lateral translation, (2) a single-MAD beam filter (MAD0 only) with lateral translation, and (3) a dual-MAD filter with two independently actuated filters that permit additional control over the beam profile. Image quality and dose are compared for both static and dynamic modulation strategies. This paper extends our previous work in which we presented results with the bowtie and the single-MAD filters.²⁰ Specifically, this work presents a more detailed analysis of filter trajectory

design methods, demonstrates results using the two degree-of-freedom (2DOF) dual-MAD filter, applies the beam modulation to an anthropomorphic phantom, and estimates the potential for dose reduction.

2 Materials and Methods

2.1 Patient Position Estimation

To optimize beam modulation, some form of patient model is required to predict the attenuation along different measurement ray-paths. To estimate the patient position within the FOV, we propose a parametric model for the patient/object shape, $o(\Omega)$, in the central axial plane as a rectellipse function as shown in Fig. 1(a). The rectellipse is similar to the ellipse in that it has parameters describing its center position (x_c, y_c) , width w , and height h (in units of voxels). However, the rectellipse also includes a shape parameter c for a continuously variable shape between $c = 0$, an ellipse, and $c = 1$, a rectangle. Additional parameters define a uniform monoenergetic attenuation coefficient μ and a rotation angle about the center R . Mathematically, the parameterized object is given by the quartic Cartesian equation:

$$\left(\frac{i}{w}\right)^2 + \left(\frac{j}{h}\right)^2 - c^2 \left(\frac{ij}{wh}\right)^2 \leq 1, \quad (1)$$

where

$$\begin{bmatrix} i \\ j \end{bmatrix} = \begin{bmatrix} \cos(R) & -\sin(R) \\ \sin(R) & \cos(R) \end{bmatrix} \begin{bmatrix} x - x_c \\ y - y_c \end{bmatrix}, \quad (2)$$

represents the translated and rotated coordinates of x and y in the axial plane, respectively.²¹

Low-dose scouts of the patient are acquired at the two orthogonal anteroposterior (AP) and lateral (LAT) views. Measurements from the central detector row are used to form a data vector g . The seven parameter object model $\Omega = \{x_c, y_c, w, h, c, \mu, R\}$ is then optimized by minimizing the mean squared difference between the forward projected object and g , according to the following equation:

Fig. 1 (a) Geometry for object calibration showing the parametric rectellipse model in the central axial plane and the positions at which the two scout scans of the object are acquired. (b) Geometry for calculating the bowtie transmissivity. Tracing the ray through known points along the length of the bowtie, shown as black dots, yields line integral measurements at an angular sampling different than that of the detector. The translation t shifts the origin of the x coordinate, where $x = 0$ is at the center of the bowtie and assumed to be aligned with isocenter (u_0 is the central detector element) with no filter translation.

$$\hat{\Omega} = \arg \min_{\Omega} \|\mathbf{P}o(\Omega) - g\|_2^2, \quad (3)$$

where $\mathbf{P}$ is the fan-beam forward projector of o that generates the AP and LAT views using the Siddon approach²² and $\|\cdot\|_2$ denotes the l_2 -norm. The optimization in Eq. (3) is nonconvex and does not readily admit a closed-form solution. In this work, this problem is solved using the Nelder–Mead search method²³ using initializations of the height, width, and center parameters obtained by simple thresholding of the orthogonal projections and computing the centroid, where the object is initially assumed to be elliptical ($c = 0$), be minimally rotated with respect to the scanner axes, and has an attenuation coefficient close to that of soft tissue at the mean x-ray beam energy $\bar{E}$. In this work, initial $\bar{E}$ was estimated for an all-water object using Spektr,²⁴ a computational tool for x-ray spectral analysis, and the known physical settings for the x-ray test bench. The parameters $\Omega_1 = \{x_c, y_c, w, h\}$ are optimized first to obtain a good positional estimate, before optimizing $\Omega_2 = \{\mu, c, R\}$ while holding Ω_1 constant to match the magnitude, curvature, and asymmetry of the projections, respectively. Then, the entire parameter set Ω is optimized together to obtain the final object parameterization. This approach was validated in simple circular and elliptical phantoms for various projection pairs over 360 deg and found to be consistent in its ability to estimate accurate parameters.

From the calibrated object, patient miscentering in the AP and LAT views can be converted from voxel indices to physical distances as

$$e_{\text{ap}} = (x_c - x_0)a_x \quad e_{\text{lat}} = (y_0 - y_c)a_y, \quad (4)$$

respectively, where a_x, a_y are the natural voxel sizes in the axial plane and x_0, y_0 are both the center coordinates of the object image (origin at top-left) and the center of object rotation.

2.2 Filter Trajectory Calculation

2.2.1 Filter calibration

To design the filter actuations for dynamic beam filtration, the available beam profiles must be characterized. Consider the case of a traditional bowtie filter based on variable attenuation of a uniform material. 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 [i.e., the thickness of the bowtie $h(x)$ at discrete points x along its width], the expected modulation pattern at the detector $f(u; t)$ (where u is the positional index across the detector) for a specified translation t can be calculated analytically based on ray tracing. This geometry is shown in Fig. 1(b). The path length l of a ray traveling through the bowtie at the lateral position index x to reach the angular sampling position ϕ at the detector for a translation t , assuming no re-entrant crossings, is given as

$$l[\phi(x; t)] = \frac{h(x; t)}{\cos[\phi(x; t)]}, \quad (5)$$

$$\phi(x; t) = \tan^{-1} \left[\frac{x + t}{\text{SFD} + h(x; t)} \right]. \quad (6)$$

Line integrals through the filter may be estimated using a monoenergetic forward model with knowledge of the bowtie

material and attenuation coefficient, μ_B , and the mean x-ray beam energy, $\bar{E}$. For this work, these physical quantities were estimated using Spektr.²⁴ Piecewise cubic Hermite interpolating polynomials were used to find estimates on the detector for particular angular samples $\phi(u)$, followed by exponentiation to yield the expected bowtie transmissivity:

$$f_b(t) = \exp\{-\mu_B(\bar{E})l[\phi(u; t)]\}, \quad (7)$$

where we have dropped the explicit dependence of f on u for notational convenience.

However, $f(t)$ is not straightforward to compute in MAD-based filtering due to obliquity effects. That is, even though the filters are relatively thin, the MAD slits act as a focused grid and translation of the MADs away from the center will narrow the beam. Alternatively, $f(t)$ may be estimated through a calibration scan that can be conducted by discretely translating the filter through the range of achievable motion and then interpolating to find the modulation patterns at specific actuation positions. We have performed such a calibration scan for both single- and dual-MAD filtration setups on our CT test bench. The MAD calibration results are shown in Fig. 2 and include an illustration of the fluence function computed for a translating aluminum bowtie filter. The single-MAD calibration was conducted using 0.125-mm steps over ~ 7 cm. For the dual-MAD setup, MAD0 was actuated in 0.1-mm increments with relative displacements of MAD1 using 0.05-mm increments over one period ($\sim \pm 0.45$ mm¹⁸) at each MAD0 position. MAD calibration scans were smoothed with an additional low-pass filtering step (using a Butterworth infinite impulse response low-pass filter) to avoid fitting residual high-frequency spatial modulations created by the MAD. Interpolation using splines was applied to obtain $f(t)$ at arbitrary actuation positions. (Note that test bench relative positioning and encoding error is estimated to be < 5 μm for accurate positioning, calibration, and artifact-free imaging.) Note that the range and spacing of actuations that need to be sampled during this calibration is dependent on where the object and MADs are placed in the field of view. The SFD for the test bench scanner is ~ 39 cm, whereas in a commercial scanner, the MADs would be placed much closer to the source (e.g., ~ 15 cm), requiring finer sampling but less total actuation.

In general, translation of the filter controls the centering of the beam for the bowtie and single-MAD filters. For the dual MADs, the centering is controlled by the absolute translation of MAD0, but additional capability for beam width control is conferred by the relative translation of MAD1 with respect to MAD0. For both the single- and dual-MAD filters, however, there is additional variation in beam width and amplitude as a function of absolute displacement owing to the aforementioned obliquity effects. Such beam narrowing is shown in Figs. 2(b) and 2(c).

2.2.2 Calculation of filter trajectory

Combining the patient/object model with the achievable fluence functions from the previous section permits estimation of the actuation required to drive the dynamic filters. Mathematically, multiplying f with the forward projection of $o(\hat{\Omega})$ (from Sec. 2.1) yields the expected detector fluence p as a function of t :

$$p(t; \theta) = f(t)e^{-\mathbf{P}(\theta)o(\hat{\Omega})}, \quad (8)$$

Fig. 2 Fluence functions for the three beam modulation strategies: (a) Computed fluence for a translating aluminum bowtie $f_b(t)$, (b) measured fluence patterns for a single-MAD $f_s(t)$, and (c) measured fluence patterns for dual-MAD filters $f_d(t)$. For the dual-MAD setup, the absolute translation of MAD0, in millimeters, is labeled in the top left of each cell in (c). The profiles achievable within approximately one period of relative MAD1 translation for the centered MAD0 position are shown in (d). (e) Comparison between the central line profiles $f(0)$ of the three filters (labeled by the dotted lines), where the fluence levels are, in general, lower but broader for the single-MAD filter than the bowtie, and lowest for the dual-MAD filter. All images and plots are normalized based on fluence without filters.

where $\mathbf{P}(\theta)$ is the Siddon fan-beam forward projector applied to $o(\hat{\Omega})$ at the projection angle θ . We choose to estimate the optimal translation of the filter $\hat{t}$ at each individual projection θ_i , where $i \in \{0, 1, \dots, N-1\}$, which is a low-dimensional optimization of $p(t; \theta_i)$ over t . To compute $\hat{t}(\theta_i)$, we consider several possible design approaches:

1. Center matching (CM): Analytically match the ray passing through the center of the object with the center of the beam filter at the SFD, given as

$$\hat{t}(\theta_i) = \text{SFD} \frac{r \cos(\theta_i - \phi)}{\text{SAD} + r \sin(\theta_i - \phi)}, \quad (9)$$

where r and ϕ describe the distance and orientation of the object center with respect to the image origin, respectively. Mathematically,

$$\begin{aligned} r &= \sqrt{a_x^2(x_c - x_0)^2 + a_y^2(y_c - y_0)^2} \\ \sin \phi &= a_x \frac{x_c - x_0}{r} \\ \cos \phi &= a_y \frac{y_0 - y_c}{r}, \end{aligned} \quad (10)$$

which may be substituted into Eq. (9) by application of the trigonometric angle addition formulae. The right-hand term in Eq. (9) represents the tan term for the fan-beam ray γ passing through the center of the object at gantry rotation angle θ_i . The full derivation for this expression can be found in Kak and Slaney.²⁵

2. Minimum standard deviation (MSD): Minimize the spatial standard deviation of the expected detector fluence via the following optimization:

$$\hat{t}(\theta_i) = \arg \min_{t \in T(\theta_i)} \text{Std}_Y\{p(t; \theta_i)\}, \quad (11)$$

where $\text{Std}\{\cdot\}$ computes the spatial standard deviation of the detector signal. Υ is a support region for the calculation where either (a) $\Upsilon = U$, the “full-field” case where U is the entire horizontal detector extent, or (b) $\Upsilon = \Phi(\theta_i)$, where $\Phi(\theta)$ denotes measurements at angle θ containing only the object [e.g., $\Phi(\theta) = \{u | e^{-\mathbf{P}(\theta) \circ (\hat{\Omega})} < 1\}$]. This generates two distinct strategies that we refer to as MSD-U and MSD-O, respectively.

3. Normalized minimum standard deviation (NMSD): Minimize the ratio of the spatial standard deviation of the expected detector fluence to its spatial mean, according to

$$\hat{t}(\theta_i) = \arg \min_{t \in \mathcal{T}(\theta_i)} \frac{\text{Std}_{\Upsilon}\{p(t; \theta_i)\}}{\text{Mean}_{\Upsilon}\{p(t; \theta_i)\}}, \quad (12)$$

$$\mathcal{T}_2(\theta_i) = \begin{cases} \{t \in \mathbb{R}^2 | |t - t(\theta_{i-1})| \leq [\epsilon \quad \eta]^T \wedge \min_{u \in \Phi(\theta_i)} f(t) > \lambda\} & \theta_i \neq \theta_0 \\ \{t \in \mathbb{R}^2 | \min_{u \in \Phi(\theta_i)} f(t) > \lambda\} & \theta_i = \theta_0 \end{cases}, \quad (14)$$

where $t = [t^0 \quad t^1]^T$ are the absolute MAD0 and MAD1 translations, respectively, and $|\cdot|$ is the element-wise absolute value function. The quantity λ is a hard minimum transmissivity constraint within Φ to avoid imaging through the low MAD transmissivity region [$\sim 2\%$ as seen in Fig. 2(e)], where the scatter-to-primary ratio is likely to be very high and would contribute to significant image artifacts—as demonstrated in previous studies of the effect of bowtie filters on CT image quality.^{26,27} In this study, $\epsilon = 1$ mm, $\eta = 0.1$ mm, and $\lambda = 0.05(5\%)$. Due to these definitions of $\mathcal{T}_1$ and $\mathcal{T}_2$, the trajectory optimizations can be solved quickly even by brute-force search.

The CM approach Eq. (9) simply attempts to match the center of the filter and the center of the object, which reduces to the solution of a similar triangles problem. The MSD method Eq. (11) is a uniform fluence objective, which seeks to find the modulation profile that will have the lowest spatial variation in signal at the detector. Such an objective will also tend to homogenize noise as variance is dominated by primary quantum noise, which is Poisson and proportional to the signal level. The NMSD objective Eq. (12) modifies Eq. (11) to prefer higher mean fluence arriving at the detector. This is because while low spatial variation may be desired, using low overall fluence to achieve flatness is not desirable. By considering the ratio of spatial standard deviation to the mean signal levels, we may avoid trivial solutions where the fluence is simply zero. Such patterns are achievable with dual MADs due to the second degree of translation freedom.

The five candidate metrics (CM, MSD-U, MSD-O, NMSD-U, and NMSD-O) were individually analyzed and used to identify a single best design objective for each of the three beam filtration strategies used in this study. The selected trajectory design strategies were then used to carry out the physical experiments outlined in the following section.

2.3 Experiments

2.3.1 CT Benchtop with filter motion stages

Dynamic beam filtration was implemented on a CT test bench as shown in Fig. 3. The test bench includes a Varex 4343CB

where $\text{Mean}\{\cdot\}$ is the spatial mean operator. We again denote the two separate strategies referring to the choice of Υ as NMSD-U and NMSD-O, respectively.

For the optimizations in Eqs. (11) and (12), $\mathcal{T}$ is a convex set that varies depending on the filtration strategy. For the 1DOF filters (i.e., the bowtie and single-MAD filters):

$$\mathcal{T}_1(\theta_i) = \begin{cases} \{t \in \mathbb{R} | |t - t(\theta_{i-1})| \leq \epsilon\} & \theta_i \neq \theta_0 \\ \mathbb{R} & \theta_i = \theta_0 \end{cases}, \quad (13)$$

where the parameter ϵ is a hard constraint on the magnitude of the shift in absolute filter position at each angular step after the starting angular position θ_0 to promote overall trajectory smoothness. For the 2DOF filter (i.e., the dual MAD):

amorphous silicon flat-panel detector, a Varex Rad-94 x-ray tube (Palo Alto, California) with an added 2-mm Al and 0.2-mm Cu of beam filtration, a six-axis Hexapod for object rotation (ALIO Industries, Arvada, Colorado), and two Velmex XSlide linear motion stages for filter positioning (Bloomfield, New York). Actuation of the rotation and linear motion stages was synchronized with the x-ray source pulsing to perform step-and-shoot image acquisitions (i.e., the filters move into position before each frame of acquisition). The system geometry was chosen to emulate the source-to-detector distance of a clinical CT scanner and to be consistent with the geometry for which the MADs were designed,^{18,19} with an SFD of 380 mm, a source-to-axis distance of 823 mm to maximize the FOV, and source-to-detector distance of 1100 mm.

2.3.2 Image quality studies

To investigate the performance of the above methods in mitigating the dose and image noise penalties of miscentering, experiments were conducted with a 16-cm CTDI phantom—a uniform PMMA cylinder that mimics the size and attenuation of the human head. The CTDI phantom was scanned with

Fig. 3 (a) CT test bench with dual MADs indicated by the green arrows. CT dose index (CTDI) dosimeter locations are numbered and the pink dotted arrow indicates the direction of left miscentering. (b) Aluminum bowtie and (c) MAD filter.

miscentering of $\sim 0, 2$, and 4 cm toward the left side of the detector for both static (no filter actuation) and dynamic filter positioning for three beam filtration strategies: an aluminum bowtie, a single-MAD [shown as MAD0 in Fig. 3(a)], and the dual-MAD filters. For each combination of miscentering and beam filter, the static beam position was set to be the mean absolute translation of the dynamic trajectory. Mathematically, the actuation position was computed as follows:

$$t_{\text{static}} = \frac{1}{N} \sum_{\theta} \hat{t}(\theta), \quad (15)$$

where N is the number of angular projections. For the dual-MAD case, the relative MAD1 translation that minimized the design objective for the mean MAD0 position was used.

Scout scans were acquired from single cone-beam projections at the AP and LAT views using 100 kV and 0.256 mAs. Image acquisition was performed with $N = 360$ projections over a full 360-deg rotation (with a constant angular step) at 100 kV, 92 mAs (0.256 mAs/projection), and 10-deg/15-deg vertical beam collimation (cone angle) for dose/noise measurements, respectively. The design of the aluminum bowtie was optimized for the CTDI phantom at 100 kV, whereas the MADs were designed to flatten the fluence for a wide range of object diameters.^{18,19} Because fluence levels differ between filters and with translation and are not simply scaled versions of each other (as is apparent from Fig. 2), the acquisitions using different beam filters were not dose matched.

Dose measurements were made in the 16-cm CTDI phantom using a Radcal 6 cc gas ionization chamber (Model 10x6-0.6, Monrovia, California). The accumulated dose over a full 360-deg projection scan for each dosimeter location [as labeled in the CTDI phantom in Fig. 3(a)] d_j was measured, and the weighted dose was calculated using the standard weighted average of central and peripheral dose:

$$\text{CTDI}_w = \frac{1}{3} d_1 + \frac{2}{3} \frac{\sum_{j=2}^5 d_j}{4}. \quad (16)$$

The % difference in the dose measurement with respect to the centered condition was then computed to investigate the change as a function of the amount of miscentering.

To analyze noise, two sequential data acquisitions were performed and reconstructed with the Feldkamp (FDK) algorithm.²⁸ Difference images were formed to create noise-only image volumes with a $1/\sqrt{2}$ correction factor applied. Regions-of-interest (ROIs) of $31 \times 31 \times 7$ voxels were defined for each of the five dosimeter locations in the central volume of the difference image and used to quantitatively assess the standard deviation of voxel values σ_j . Additionally, a $7 \times 7 \times 7$ voxel box standard deviation filter was convolved with the difference image. The central axial slice of the result was used to qualitatively assess the noise distribution within the object. The spatial mean of the noise over this noise map was computed and defined as σ_w . The % difference (from the centered object scenario) was again used to assess the effect of beam modulation on the noise measurements.

From the noise maps, the “noise-adjusted” exposure level³—the % dose change needed to maintain the same peak variance as the centered case—was calculated assuming the dose should be increased by an amount proportional to the square of the peak noise increase. That is, presuming that the noise is dominated by

photon statistics (i.e., low electronic noise) permits a scaling of the exposure (and therefore dose). For a global image quality measure, peak variance was used, as in related studies.^{29,30} This presumes all portions of the image volume are subject to the same maximum noise constraint and are consistent with diagnostic tasks for which the location of disease/defects is unknown.

2.3.3 Anthropomorphic head phantom study

CT scans were also conducted of an anthropomorphic head phantom (CIRS ATOM Head Partial Phantom Slice #6, Model 701-HN, Norfolk, Virginia) under centered and miscentered conditions to investigate an anatomical imaging case with a miscentered, nonuniform, and noncircular object. The dual-MAD filter was chosen for this experiment as its width modulation capability is appropriate for elliptically shaped, anthropomorphic objects. Scouts were obtained at 100 kV and 0.512 mAs, and images were acquired at 100 kV and 184 mAs (0.512 mAs/projection) with 15-deg vertical beam collimation. Noise was qualitatively analyzed in a manner similar to that described before but computed only over voxels interior to the patient. A patient mask was identified using a segmentation based on Otsu’s binary threshold algorithm.³¹ Additionally, the computed axial noise maps were each normalized by the mean value over the entire interior head region for comparison on a common window level.

For this study, we focused on the NMSD-O design strategy. The static beam profile was the same across both the centered and miscentered cases. The MAD0 position t_{static}^0 was again defined according to Eq. (15) using the miscentered trajectory, and the relative MAD1 translation was selected by finding the fluence pattern that minimizes the sum of the objective in Eq. (12) for the miscentered object $o(\hat{\Omega}_M)$ over all angular projections, i.e.,

$$t_{\text{static}}^1 = \arg \min_{t^1 \in \mathcal{T}_{\text{static}}} \sum_{\theta} \frac{\text{Std}_{\Phi(\theta)} \{p(t^1; t_{\text{static}}^0, \hat{\Omega}_M, \theta)\}}{\text{Mean}_{\Phi(\theta)} \{p(t^1; t_{\text{static}}^0, \hat{\Omega}_M, \theta)\}}, \quad (17)$$

$$\mathcal{T}_{\text{static}} = \{t^1 \in \mathbb{R} \mid \min_{u \in \Phi(\theta)} f(t^1; t_{\text{static}}^0) > \lambda \forall \theta\}. \quad (18)$$

This optimization favors a beam profile wide enough to encompass the entire object at every view in the miscentered case. This approach is consistent with the traditional engineering design choice in which a larger (i.e., less “aggressive”) bowtie filter is conservatively selected to maintain image quality for an object with both width variability and miscentering.

2.4 Image Reconstruction

FDK reconstructions were performed with a Hann window, 0.8 Nyquist frequency cutoff, $512 \times 512 \times 220$ voxel image volume, and a 0.5-mm isotropic voxel size. For bowtie filter studies, a simple additional gain scan associated with the bowtie at each position in the filter trajectory is necessary. However, for the single and dual-MAD studies, the high-frequency modulations produced by the MAD are susceptible to focal spot changes (blooming and/or shifting) as well as spectral effects that may introduce deterministic artifacts (such as rings) in the reconstruction. This necessitates a modified reconstruction pipeline to avoid such artifacts.

For the MAD reconstructions, we model mean measurements $\bar{y}$ at projection θ as

$$\bar{y}(\theta) = g_D [g_{I_0} g_M(\theta) e^{-\alpha(\theta)\mu_l} + s(\theta)], \quad (19)$$

where g_D is the x-ray detector sensitivity, g_{I_0} is the emitted x-ray distribution from the tube, g_M is the MAD transmissivity, and s is a constant (shift-invariant) x-ray scatter term. A first-order spectral correction factor, α , was estimated using calibration scans of variable thickness PMMA slabs. Because the MAD filter is placed relatively close to the x-ray source, MAD structures exhibit a high degree of magnification. Although focal spot blur smooths out the x-ray profile, residual patterns are susceptible to focal spot variations. If ignored, uncorrected data yields ring artifacts that worsen with larger focal spot shifts. However, these effects may be modeled and calibrated as discussed in Gang et al.³² In short, to accommodate subtle changes in g_M over time, fiducials on the MAD filter are tracked in projection data and used to estimate focal spot shifts and size changes. The g_M from a calibration scan is then modified (e.g., shifted and/or blurred) to fit the current data and avoid ring artifacts in the reconstruction.

A method to estimate and subtract the scatter signal from the data was also employed. This is particularly important to eliminate the bias that occurs when imaging through the very low transmissivity regions of the bowtie and dual-MAD. For the CTDI phantom studies outlined in Sec. 2.3.2, $s(\theta)$ was estimated by comparing reprojections of a binarized image volume with ideal attenuation $\mu_{\text{PMMA}}(\bar{E})$ against the actual measurements at the center of the object to obtain a constant scatter correction (per projection angle). This same correction was applied

for both dynamic and static cases for each of the three filters investigated for consistency. Though more sophisticated methods for scatter correction could be applied, such as Monte Carlo methods,^{33,34} this method was chosen for its simplicity. No scatter correction was applied for the anthropomorphic head data.

3 Results

3.1 Filter Trajectories

3.1.1 Performance of trajectory metrics

The five candidate trajectory designs for each of the three beam filters are summarized in Fig. 4 for an object with 4 cm of miscentering. As observed in Fig. 4(a), all design objectives were approximately equivalent in the aluminum bowtie case. Because the bowtie was designed specifically for this object, there is little contribution to the spatial variance from the fluence outside the object, making metrics over $\Upsilon = U$ or $\Upsilon = \Phi$ almost identical. Similarly, as beam shape is approximately independent of filter translation and the beam simply shifts with translation [Fig. 2(a)], the mean fluence penalty has little effect on the result.

For the single-MAD filter, there was more variability between the objectives as shown in Fig. 4(b). We see that the MSD approach is able to accommodate trade-offs in beam centering and beam width not present in the traditional bowtie [Fig. 2(b)]. Specifically, the single-MAD beam width is narrower for off-center translations, and the MSD objective can accommodate these variations by increasing the amplitude of the trajectory slightly relative to the CM approach. More significantly, we observe that the MSD-O metric resulted in

Fig. 4 Optimizers considered for (a) the bowtie, (b) single-MAD, and (d) dual-MAD trajectories with the shared legend shown in (a). (c) Illustration of the differences between the MSD-U and MSD-O objectives in the single-MAD case for the t_U and t_Φ positions respectively (marked by *) at $\theta = 0$ deg. (d) The relative MAD1 translations (dotted lines) overlaid on the absolute MAD0 positions (solid lines). The CM approach, in blue, provides a reference for the location of the projected object center but does not represent its own 2DOF trajectory in (d). All plots (a–d) were computed for the CTDI phantom with offset $e_{\text{ap}} = -4.02$ cm.

a trajectory that was significantly different from the others. With a symmetric, perfectly designed fluence profile (e.g., the aluminum bowtie case), the only way to minimize the spatial standard deviation of the detector fluence is to center the beam on the object. For symmetric fluence profiles that cannot flatten the detector fluence and/or are not matched in width to the object (e.g., the single MAD for the CTDI phantom), it was possible to find a solution that will further lower MSD-O with an asymmetric profile. The varying symmetry of designs is shown in Fig. 4(c) for the MSD-U and MSD-O solutions for the 0-deg projection angle (indicated by arrows between subplots). Note that $f(t_U)$ is more centered on the object than $f(t_\Phi)$. {Object boundaries are evident in the plot of $\exp[-\mathbf{Po}(\hat{\Omega})]$.} The largest contributors to the MSD-U metric are the two large peaks just outside of Φ . Although both solutions gave similar uniformity within Φ , the penalty computed over $\Upsilon = U$ added an implicit penalty for asymmetry by comparing the fluence on either side of the object region, therefore tending to favor more “equalization” of the two peaks.

Dual-MAD results are shown in Fig. 4(d) with solid lines showing absolute MAD0 translation and dotted lines showing relative MAD1 translation. MSD objectives failed in this case because the second degree of translational freedom allowed solutions that minimize spatial standard deviation by simply pushing the overall fluence to zero (fluence profiles not shown). In contrast, the NMSD objectives avoid these trivial solutions and accommodate both beam centering and width design. The NMSD-O and NMSD-U solutions are very similar with sinusoidal absolute motion of the filters. Relative motion is also sinusoidal; however, it exhibits discontinuities around 50 deg and 350 deg due to a phase-wrapping effect. In particular, because dual-MAD fluence patterns are periodic over ~ 0.8 -mm relative translations [recall Fig. 2(c)], jumps in relative

translation between ± 0.4 mm produce nearly identical beam profiles. As only a single relative translation period was used for design, these phase-wrapping jumps are apparent. Moreover, the observed differences in phase wrap location between NMSD-O and NMSD-U are not significant.

Based on these studies, objectives were selected for each strategy and are summarized in Table 1. Specifically, CM was chosen for the bowtie filter as it is the simplest method and all other methods were comparable. For the single MAD, MSD-U was selected for its ability to encourage centered fluence patterns, and NMSD-O was used for dual-MAD filters.

3.1.2 Estimated filter trajectories

The calculated filter trajectories for the three beam filtration strategies and the miscentered 16-cm CTDI phantom are shown in Fig. 5. Consistent with $\sim 2\times$ system magnification, these trajectories were largely sinusoidal with an amplitude equal to approximately half of the miscentering offset. For the MAD filters, the trajectories were sinusoidal but not as smooth (likely due to irregularities in the measured MAD modulation patterns). In the single-MAD case, the MSD-U approach yielded a slightly larger amplitude than the CM approach for the bowtie because, for a (left) miscentered object and fan-beam projection, the projection center-of-mass is skewed to the left of the projected object center. This object obliquity effect becomes more pronounced for larger miscentering offsets as is shown in Fig. 5(a). The overall dual-MAD trajectory was similar to the other approaches with the smallest translation amplitude. This is likely due to the increasing cost of low fluence levels using the NMSD metric and, recalling Fig. 2(c), the general decrease in beam amplitude as a function of the dual-MAD absolute displacement. As seen in Fig. 5(b), the relative MAD1 translation followed the same general trend as MAD0, indicating that approximately the same beam width was selected at every MAD0 position. The previously described phase-wrapping of relative translations is apparent for the largest miscentering case.

3.2 Dose and Image Noise Measurements

Figures 6 and 7 show the results of the CTDI phantom studies. For the static aluminum bowtie, miscentering had a large effect on both dose and image noise: for 4 cm of offset (toward the left of the detector), location 3 saw a 39% dose increase and a 22% noise decreases, whereas location 5 saw a 42% dose decrease

Table 1 Summary of selected trajectory designs.

Beam filter	Design strategy	Design objective
Bowtie	CM	$\hat{t}_b(\theta) = \text{SFD} \frac{r \cos(\theta - \phi)}{\text{SAD} + r \sin(\theta - \phi)}$
Single MAD	MSD-U	$\hat{t}_s(\theta) = \arg \min_{t \in \mathcal{T}_1(\theta)} \text{Std}_U\{p(t; \theta)\}$
Dual MAD	NMSD-O	$\hat{t}_d(\theta) = \arg \min_{t \in \mathcal{T}_2(\theta)} \frac{\text{Std}_{\mathbf{p}(\theta)}\{p(t; \theta)\}}{\text{Mean}_{\mathbf{p}(\theta)}\{p(t; \theta)\}}$

Fig. 5 (a) Estimated filter trajectories for each filter and miscentering scenario, where the “offset” was the estimated e_{ap} . (b) Dual MAD trajectories for the same miscentering offsets as (a).

Fig. 6 Noise maps for each scenario showing the static/dynamic cases split into the top/bottom halves (exploiting symmetry for visualization) masked by $\alpha(\hat{\Omega})$ to show only the object region. The window was adjusted to reflect the distinct noise levels for each filter case. ROIs used to quantify noise in Fig. 7 are labeled in the top-left (not to scale) and match the numbering shown in Fig. 3(a).

and a 173% noise increases relative to the centered condition [Figs. 7(a) and 7(c)]. This large noise increase at location 5 and the general noise increase at all other locations other than 3 was the result of imaging through the edge of the bowtie, where the transmissivity decreased sharply down to $\sim 4\%$ (due to the “aggressive” design of this bowtie that sought perfectly flat detector fluence for a 16-cm CTDI phantom). (Note the dose measurements shown do not perfectly reflect the inverse square root proportionality with noise due to scatter correction.)

With the dynamic bowtie variations were largely mitigated. Dose and noise remained close to baseline levels for all dosimeter locations as well as the overall CTDI_w as seen in Figs. 7(b) and 7(d). Figures 6(a)–6(c) show that use of the dynamic filter promoted homogeneous noise in the reconstruction in contrast to the static filter. A slight noise asymmetry from left to right was observed in the case of 4-cm miscentering. We note that a fixed filter shape cannot account for the combination of obliquity and depth-dependent magnification effects that occur with the miscentered object, and so even the translating bowtie cannot perfectly flatten the fluence for the miscentered object.

The static single-MAD data show similar left-to-right dose and noise gradients as those observed with the aluminum bowtie. Because the low transmissivity region only approached 10%, however, the noise penalty for location 5 at 4-cm offset was not as severe as with the bowtie, only reaching 29% as seen in Fig. 7(g). The noise levels as observed in Fig. 6(d) were higher overall, which is attributed to the overall lower fluence levels of the MAD [55% versus 90% maximum transmissivity as shown

in Fig. 2(e)]. The single-MAD fluence profiles were also sub-optimal for achieving perfectly uniform detector fluence for the 16-cm CTDI phantom, leading to a bowl-shaped noise distribution with higher observed central noise in Fig. 6(d). 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 Fig. 2(b)] lead to overall dose decreases for all locations at 4-cm miscentering and overall higher noise [Figs. 7(f), 7(h), and 6(f)], even with the dynamic single MAD. Despite these differences, Fig. 6 shows that the dynamic MAD was still able to distribute noise similarly to the baseline while the static filter could not. Notably, Fig. 7(h) shows that there was less left-to-right variation in this case than for the bowtie, which may be related to the broad fluence profiles of the single MAD, which can help mitigate the inability to modulate beam width.

The trends observed for the dual-MAD case were, in general, very similar to those of the bowtie. The overall observed noise levels for the centered case shown in Fig. 6(g) were $1.9 \times 10^{-3} \text{ mm}^{-1}$ for the dual MAD compared with $1.2 \times 10^{-3} \text{ mm}^{-1}$ for the bowtie, representing approximately a 50% increase that intuitively reflects the noise proportionality $\sigma \propto \frac{1}{\sqrt{\text{Dose}}}$, where the dual-MAD fluence levels were at 40% (maximum transmissivity) compared with 90% for the bowtie [as shown in Fig. 2(e)]. Because of the transmissivity constraint applied within Φ , the static beam profile was wider than that of the bowtie and hence the noise increase in the 2-cm miscentering case was not as severe. However, due to the more “aggressive” design of the dual MAD, a more dramatic noise penalty was observed at location 5 for 4 cm of miscentering with a static filter [Fig. 7(k)].

Fig. 7 Dose and noise results for the static/dynamic aluminum bowtie, single-MAD, and dual-MAD filters. Dose and noise at each dosimeter location (locations 1 to 5) were plotted as a function of the amount of estimated linear miscentering e_{ap} . Note that the x axis has been inverted so that increased miscentering follows from left to right. Error bars on the centered condition for the dose plots indicate the inherent 5% uncertainty in dose measurements made at ambient conditions, according to the ion chamber manufacturer.³⁵

With the dynamic dual MAD, dose and image noise distribution were more homogeneous and similar to the baseline. More left-to-right variation in the observed noise distribution was apparent for the 4-cm case [Fig. 6(i)] than the bowtie, though still within 10% of the baseline as seen in Fig. 7(l). This asymmetry is believed to be due to the smaller amplitude of the filter trajectory resulting from the trade-off between fluence uniformity and

mean fluence in the NMSD metric, which has the effect of slightly overexposing the right part of the object and underexposing the left. Though it was not applied in this study, techniques such as tube-current modulation could be used to help eliminate the residual nonuniformities in both the bowtie and dual-MAD cases.

Peak noise measurements for each combination of miscentering and beam filter along with the associated noise-adjusted exposure calculations are shown in Table 2. Values were normalized to the centered case in each row. In general, the dynamic filters require significantly less dose adjustment to maintain the same image quality for a miscentered object. For the more “aggressive” dynamic bowtie and dual-MAD filters, an eight-fold and 13-fold respective decrease in noise-adjusted dose is achievable. For the more moderate single-MAD filter, 24% dose savings are possible by using the dynamic filter to maintain the same peak variance at 4 cm of miscentering.

Table 2 Dose reduction with dynamic filtration.

Beam filter/Miscentering (cm)	Peak noise			Noise-adjusted dose		
	0	2	4	0	2	4
Bowtie—static (%)	100	149	297	100	223	881
Bowtie—dynamic (%)	100	99.7	104	100	99.4	109
Single MAD—static (%)	100	107	116	100	115	135
Single MAD—dynamic (%)	100	96.2	104	100	92.5	109
Dual MAD—static (%)	100	117	423	100	138	1790
Dual MAD—dynamic (%)	100	97.4	114	100	94.9	129

3.3 Dynamic Dual-Multiple Aperture Devices Filtration for the Anthropomorphic Head Phantom

Application of dynamic beam modulation to the anthropomorphic head phantom using the dual-MAD filter is shown in Fig. 8. The calibrated object models in Figs. 8(a) and 8(c) reasonably approximate the shape and location of the real object in the FOV.

Fig. 8 Head phantom study comparing static and dynamic filter trajectories, including the calibrated object models (a) and (c), calculated dual-MAD motion trajectories (b) and (d), reconstructed axial slices (e–h) and the associated noise maps (i–l) for both centered and miscentered cases. The yellow arrows indicate the direction and amount of miscentering in centimeter.

According to the calibration, e_{ap} and e_{lat} were -0.20 and 0.14 cm, respectively, for the centered case and -2.29 and 1.71 cm, respectively, for the miscentered case. Because of the linear miscentering in both x and y directions, the anthropomorphic head had inherent width variability, projection-dependent obliquity, and centering effects that required accommodation by the dynamic filtration strategy. The dynamic filter trajectory for the miscentered case in Fig. 8(d) shows that the (relative) MAD1 trajectory had a much larger amplitude, reflecting the need for dynamic width control for an elliptical object. The relative position of MAD1 for the static trajectory indicates the selection of a wide beam profile by inspection of Fig. 2(d).

The reconstructed images demonstrate the lack of ring artifacts arising from the MAD filter; however, some beam hardening and cupping effects are observed, as no scatter or beam hardening correction was applied. Some bias is evident at the bottom of the image in the nose region that was outside of the object model and hence well within the low transmissivity region of the filters (contributing to artifact sensitivity). From the noise maps, we see that the noise distribution was largely homogeneous in the centered case [Figs. 8(i) and 8(j)] with some variations due to anatomical inhomogeneity and the lack of tube-current modulation. However, the differences between the static and dynamic filter are still apparent. Consistent with our expectation from the results observed in from Fig. 6(i), the dynamic filter was able to distribute noise similarly to baseline in the miscentered case, whereas significant increases in image noise in the top left portion and significant decreases in the bottom right portion of the image are observed in the static miscentered case [Fig. 8(k)], despite the use of a broad beam profile. A slight noise gradient from top left to

bottom right was still observed in the dynamic case, which may again be attributable to the same reasons described earlier in Sec. 3.2 for the 4-cm dual-MAD case.

4 Discussion and Conclusions

In this work, dynamic beam modulation for miscentered patients was demonstrated using three beam filtration strategies. For the bowtie, simple translation of the filter using a basic trajectory estimation approach, based on centering the beam on the center of the patient, was used. For the more complex beam modulators based on the dual-MAD filter, we considered a more sophisticated metric seeking an optimal trade-off between minimal variation in the detected signal and maximal mean detected fluence. In general, a simple monolithic filter can be actuated effectively using the simple CM design objective if well matched to the object while more complex dynamic filters require an objective that can balance field-flattening and overall signal levels like the NMSD-O metric. All three scenarios demonstrate the ability to keep the spatial distribution of dose and image noise at baseline levels (i.e., compared with static filtration with a centered object) across arbitrary levels of miscentering for the homogeneous CTDI object.

The results also strongly suggest that the dual MADs operate effectively as dynamic bowtie filters and that their capability for beam width control allows for increased uniformity of image properties. In particular, the noise gradients observed in the static filtering scenarios with miscentered objects illustrate two key effects that need to be accommodated by a dynamic filter: (1) patient centering and (2) depth-dependent magnification of projection data, which are compensated by beam centering control and beam width modulation, respectively. Moreover,

variations in patient size and anatomical site result in additional variation in the width of patient projections. Thus, while all investigated methods demonstrate improved uniformity in dose and noise, fluence-field modulation strategies like the dual-MAD filter may have additional advantages across a greater range of clinical scenarios.

The quantitative results presented in this paper differ somewhat from those of previous studies.^{3,4} This is because an “aggressive” bowtie was used here, attempting to perfectly flatten the detector fluence for the 16-cm CTDI phantom. In contrast, Toth et. al. employed bowties that taper more moderately to a minimum transmissivity of 10%.³ Habibzadeh et. al. utilized GE Healthcare’s “large-body” bowtie for their study,⁴ which would also filter less aggressively for smaller objects. The single-MAD filter used in this work was more moderate with similar transmissivity, making observed 24% dose savings for the dynamic vs static version more consistent with the results of Toth et al.

The performance of the dynamic methods in this paper relies on a good estimate of the object (Ω). While effective object estimation was found for the scenarios investigated in this work, there is likely to be increased variability and inhomogeneity in actual patient data. Studies validating the methodology in such diverse data are the subject of future work. Should the proposed estimation technique be insufficient, more robust algorithms for global non-convex optimization may be required.

Although the studies conducted here using dynamic beam modulation showed greater control of dose and noise properties, the methods did not use additional exposure control (e.g., tube current modulation) as a function of rotation angle.³⁶ Automatic exposure control is another important element of dose reduction and noise minimization, which can have a significant impact on noise correlations in reconstructed images. Ongoing studies are exploring the combination of spatial beam modulation in conjunction with angular beam modulation of the overall fluence for additional control and improvements in dose utilization. Similarly, while the studies presented here focused on a simple metric of image quality (i.e., noise in FBP reconstructions), more sophisticated image quality metrics tailored to specific clinical tasks may be more appropriate. Recent studies have suggested that such task-based optimization as well as advanced model-based reconstruction methods can lead to nontraditional modulation strategies.^{37,38} Future studies will include investigations of optimized beam modulation for miscentered patients using such task-based metrics and advanced reconstruction.

All of the investigated FFM techniques in this paper create uniform flux patterns along the z-direction. With the increasingly large cone-angles of commercial CT systems (and increased z detector coverage), the methods for estimating the patient model would need to be extended to account for anatomical variations along the z-direction. For example, the proposed methodology could be generalized using fitting to a three-dimensional (3-D) “rectelliptical” cylinder. More sophisticated approaches based on anatomical atlases or even patient-specific models from low-dose 3-D scouts³⁹ could also be used. Such extended cone-beams similarly suggest approaches that allow additional modulation in the z-direction. For example, a design involving more than one set of MADs spaced axially could help achieve the desired fluence modulation along the z-direction.

One of the trade-offs of the dual-MAD system is that small relative motions induce relatively large changes in fluence. This helps keep the system compact; however, there are increased

requirements for calibration and hardware positioning. The stricter requirement is on position encoding, as slight changes in fluence pattern do not have a large impact on noise and dose; however, gain calibration is much more sensitive. Encoding errors will, in turn, lead to gain calibration errors, which manifest as ring artifacts in the reconstruction. With the high accuracy encoding on the test bench system, we do not observe such artifacts in this work. Ongoing work seeking to develop dual-MADs for a commercial CT system must contend with these engineering challenges; however, similar encoding accuracy can be found in fast linear motors and offers one potential solution.

In conclusion, a workflow and methodology was developed to dynamically position beam filters during CT image acquisition. The techniques are generally applicable to other beam filtration strategies and promote consistent imaging and dose performance for arbitrary patient positioning. These advantages could ease patient setup requirements and reduce repeat scanning in cases of poor positioning. Such benefits are particularly applicable in challenging clinical settings like ER CT, where a simplified patient setup procedure will benefit workflow and maintain the dose advantages of spatial beam shaping.

Disclosures

No conflicts of interest, financial or otherwise, are declared by the authors.

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