Design of dual multiple aperture devices for dynamical fluence field modulated CT

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Abstract—A Multiple Aperture Device (MAD) is a novel x-ray beam modulator that uses binary filtration on a fine scale to spatially modulate an x-ray beam. Using two MADs in series enables a large variety of fluence profiles by shifting the MADS relative to each other. This work details the design and control of dual MADs for a specific class of desired fluence patterns. Specifically, models of MAD operation are integrated into a best fit objective followed by CMA-ES optimization. To illustrate this framework we demonstrate the design process for an abdominal phantom with the goal of uniform detected signal. Achievable fluence profiles show good agreement with target fluence profiles, and the ability to flatten projections when a phantom is scanned is demonstrated. Simulated data reconstruction using traditional tube current modulation (TCM) and MAD filtering with TCM are investigated with the dual MAD system demonstrating more uniformity in noise and illustrating the potential for dose reduction under a maximum noise level constraint.

Index Terms—Fluence field modulation, Radiation dose reduction, Dynamic bow-tie filter, Region-of-interest CT, X-ray beam modulation, Patient-specific CT.

I. INTRODUCTION

X-ray computed tomography has found widespread clinical utility; however, increasing concerns about the risks associated with ionizing radiation have driven the search for exposure reduction strategies. While many algorithmic strategies for producing better images at lower exposures have been developed, there has been relatively little research on innovative hardware-based dose reduction methods. Dose to an individual patient is naturally tied to the particular exposure settings of a CT scanner; however, finding minimum dose strategies is both complex due to the dependence on patient size, anatomical site, etc. and, currently, somewhat limited due to the relative inflexibility of modern CT scanners to control the distribution of x-rays used to image a patient.

Typical clinical scanners permit coarse control of the x-ray beam through exposure settings (tube current and voltage), and many systems have tube current modulation hardware that permits variation of exposure as a function of rotation angle and table position. Control of the spatial distribution of the x-ray beam is typically very limited and is achieved through the introduction of a bow-tie filter. Some systems allow selection from a small number (typically three or fewer) bow-tie filters based on patient size. Typical filters attenuate x-rays at large fan angles to achieve higher fluence levels in the center of the patient (where the attenuation is highest) and lower fluence at the edges (where attenuation is low). Unfortunately, such static beam shaping is limited and cannot account for variability in the width/size of the patient as a function of angle and table position. Similarly, static bow-tie filters can be sensitive to positioning since a well-centered patient is presumed.

As discussed in [8], the pitch (spacing between blockers) of the MAD device may be designed to minimize high-frequency patterns at the detector. For example, if the focal spot of the x-ray source is assumed to be a rectangle, the MAD pitch may be placed at the first null frequency associated with the focal spot blur MTF. In this fashion, the fine bar pattern of the MAD device is blurred out and is not visible at the detector. Desirable (lower frequency) spatial modulation associated with the variable bar width is still achievable.

A single fluence pattern can be obtained with a single MAD device. With multiple MADs in series, capable of moving with respect to each other, a range of fluence patterns can be obtained since it is the composition of two binary filters. Moreover, small relative displacement of the MADs with respect to each other can induce large changes in the fluence pattern. Because small actuations have a large effect on the x-ray distribution, speed and acceleration requirements can be reduced for device

Fig. 1: Illustration of fluence modulation using Dual MAD filters.

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B. Initial Phantom Study and System Geometry

For initial investigations, we have concentrated on fluence optimization for a single target object; however, the approach may be extended to classes of objects. Specifically, the known object in the simulation study was chosen to be an anthropomorphic phantom body of uniform material (acrylic), as illustrated in Fig. 2. This digital phantom emulates commercially available physical phantoms (QRM GmbH, Morehendorf, Germany) that will be used in subsequent studies.

The system geometry was chosen to emulate a CT scanner’s source-to-detector distance and also geometry achievable in a flat-panel-based experimental test bench that is available for subsequent experiments. For our investigations, we considered 360 degree rotation, in steps of 0.5 degree. To create projection data for MAD design and analysis, we used a polycrystalline forward model and Spektr [9], a computational tool for x-ray spectral analysis, corresponding to a tube voltage of 100 kVp with additional filtration (2 mm of Al, 0.2 mm of Cu). The model also includes fluence adjustments to accommodate divergent beam effects.

C. Optimization Procedure

1) MAD parameterization

In order to design a set of MAD filters, the location and dimensions of many MAD attributes must be specified. The elements of a dual MAD design are identified in the illustration in Figure 3. Specifically, the free design parameters include: 1) \( b_0(x) \), the thickness of each bar as a function of position in MAD0 that locally blocks X-rays; 2) \( b_1(x) \), the analogous bar function for MAD1; 3) \( \delta(x) \), a local offset function that specifies the position of individual bars in MAD1 relative to MAD0; and 4) the MAD pitch (e.g., the spacing interval between bars). As mentioned in [8], the MAD pitch may be designed independently of other parameters based on the first null frequency of the focal spot, magnified to the MAD plane. For a rectangular focal spot size, \( f_s \), the optimal MAD pitch is

\[
m = f_s \times \left(1 - \frac{SMD}{SDD}\right)
\]

We note that for nonrectangular focal spots, one can similarly find a null or minimal pass frequency to enforce smooth fluence profiles. Additionally, even though Fig. 3 shows MAD0 and MAD1 to be parallel with identical pitch, each of the flat MADs have a slightly different pitch and the bars/slots must be focused to the source due to the diverging x-ray beam.

The last parameter that is important for design is the control parameter \( \Delta \), which denotes the relative offset between MAD0 and MAD1. This is the one-dimensional actuation that controls the fluence profile enforced by the MAD filters. In general, this parameter must be part of the design process as well, and is a function of the CT rotation angle and/or table position, which we will denote as \( \Delta(\theta) \).

With MAD pitch specified, the remaining parameters: \( b_0(x) \), \( b_1(x), \delta(x), \) and \( \Delta(\theta) \) are sought. In [8], these values were determined analytically using an “endpoint” design to match two desired profiles by considering the minimum and maximum blocking conditions of a dual MAD system. While this approach is attractive due to its closed-form solution, it fails to provide best fit solutions for a wide range of desired fluence patterns. In this work, we seek that more optimal solution, which may be stated as a nonlinear, nonconvex optimization (discussed in the next section).

To facilitate optimization, we have chosen to further parameterize the dual MAD design using a low-dimensional set of basis functions. For example, rather than have a parameter for every bar width in MAD0, we presume neighboring bar widths vary smoothly as a function of position. Specifically, we chose to represent our parameters with a small set of Fourier coefficients, \( c_p(\omega) \) such that

\[
p(x) = \frac{m}{(1 + e^{-\beta(x)})} \hat{p}(x) = F^{-1}[c_p(\omega)]
\]

where \( p(x) \) is one of \( \{b_0(x), b_1(x), \delta(x), \) or \( \Delta(\theta)\} \). Thus, the optimization will focus on finding the optimal coefficients: \( c_{p1}(\omega), c_{p2}(\omega), c_{p3}(\omega), \) and \( c_{p4}(\omega) \) which are functions of the spatial (or, for \( \Delta, \) angular) frequencies selected for the basis set.

2) Objective function

To define our optimization objective, we construct a model of the fluence output which is a function of the design and actuation values and can be written in terms of the original parameters or vectors of low-dimensional Fourier coefficients:

\[
M(b_0, b_1, \delta; \Delta) \leftrightarrow M(x, \theta; c_{p1}, c_{p2}, c_{p3}, c_{p4})
\]

Note that \( M \) is a function of spatial location (e.g., a fluence profile) as well as rotation angle.

Using this model, we pose the following optimization:

\[
\{\delta, \hat{c}_{p1}, \hat{c}_{p2}, \hat{c}_{p3}, \hat{c}_{p4}\} = \text{argmin} \sum_{x \times \theta} \left[ f(\theta, x) - \frac{M(x, \theta; c_{p1}, c_{p2}, c_{p3}, c_{p4})}{M_{\Delta}(\theta)} \right]^2
\]

where \( f(\theta, x) \) denotes desired fluence patterns as a function of rotation angle. The objective is computed as the mean squared error between the desired and modeled fluence patterns over all projections that intersect the phantom (or patient). As such, X-rays passing outside the phantom (e.g. not contributing to dose) will be ignored in the optimization process.

Also note that both the modeled and desired fluence patterns are normalized by \( M_{\Delta}(\theta) = \sum_x M(x, \theta; \cdot) \) and \( t_{\Delta}(\theta) = \sum_x t(\theta, x) \), respectively. This normalization concentrates the design process on achieving the proper fluence shape. The magnitude of the profile can be adjusted post-design through exposure settings and tube current modulation. While there are many potential desired fluence patterns that one might seek including those that enforce minimum peak variance [10], combined noise and dose objectives [1], or maximize task-based detectability [11], we will focus on fluence patterns that flatten the signal and homogenize noise in projection data.
using the CMA-ES optimization is also plotted. Shown in Fig. 4. The fluence obtainable with the dual MADs designs. A subset of these target fluence profiles are symmetric basis functions were employed to enforce symmetric Fourier coefficients for each MAD feature (4 total), and only required to flatten this fluence is computed using the detector plane, and the fluence profile at the MAD plane 360 degrees in steps of 0.5 degrees. The fluence is simulated at nonconvex optimization since a population of solutions is employed to avoid local optima. We implemented the objective function and the profile modeling function in efficient C++ code including parallelized computation of objective function values (over the population) using OpenMP. The CMA-ES algorithm was initialized to the output of the end-point design process from traditional bowtie (e.g. more fluence in the center of the field range of approximately 50 µm to 800 µm. Such designs are largely within the constraints of modern tungsten sintering technology, though features <100 µm can present some challenges (such constraints can potentially be integrated into the design process). The local offset function, δ(x), is predominantly negative, meaning that the MAD1 bars are located to the left of the center position in each MAD period. The actuation control shown in Figure 5D illustrates that MAD1 is displaced between 0.15 mm to 0.4 mm as the MAD1 design is almost the opposite (when acting alone). The bar widths in both MADs span the range of control actuation (displacements of MAD1 with respect to MAD0), and less at the edges. The MAD1 design is almost the opposite (when acting alone). The bar widths in both MADs span the range of approximately 50 µm to 800 µm. Such designs are largely within the constraints of modern tungsten sintering technology, though features <100 µm can present some challenges (such constraints can potentially be integrated into the design process). The local offset function, δ(x), is predominantly negative, meaning that the MAD1 bars are located to the left of the center position in each MAD period. The actuation control shown in Figure 5D illustrates that MAD1 is displaced between 0.15 mm to 0.4 mm as the projection angle changes from 0 to 360 degrees. This minimal movement of the MADs causes the large change in the fluence patterns seen in Figure 4 and can be attributed to the relatively small MAD pitch. From an implementation standpoint, the potential mechanical advantage is the fast switching speed of the MAD fluence profiles as the CT gantry spins around the patient. The smooth profile of the displacement also reduces the acceleration requirements on the actuator. Though not done here, one could integrate specific acceleration limits as part of the optimization.

C. Achievable Fluence Patterns

It is interesting to note that the design of the previous sections only utilizes part of the actuation control range. Fig. 6 shows the full range of fluence patterns achievable as the second MAD is moved with respect to the first MAD within a single MAD pitch (e.g. one cycle). Recall, that for the selected phantom, only fluence profiles between MAD1 displacements of 0.1 to 0.4 were used.
However, from the fluence map, it is clear that much sharper fluence patterns can be obtained by changing the displacement to 0.7 mm. This potentially enables other applications such as region-of-interest fluence modulation and suggests additional design flexibility for larger classes of profiles (e.g., complex objects, multiple classes, etc.).

D. Tube Current Modulation (TCM)

Although a variety of fluence patterns have been demonstrated, practical application and fitting to the desired fluence profiles requires proper scaling. This scaling can be achieved through tube current modulation (TCM). Typical Automatic Exposure Control (AEC) seeks to provide a constant fluence at the center of the detector. We have applied this strategy for the no filter scenario. For the MAD scenario, we applied the same strategy of providing constant fluence at the central detector pixel, through the Dual MAD and phantom. For comparison between the no filter and MAD filtered scenarios we have ensured that the total fluence (i.e., the number of simulated photons) incident on the phantom is constant for the two approaches. Specifically, TCM is scaled to enforce a total of 100,000 photons incident on the phantom.

Fig. 7 shows the TCM required to convert the fluence generated by the MADs to the required target fluence. Without the MAD filter, the TCM is largest when the path length of X-rays through the phantom is largest. The dual MAD filter has maximum attenuation when the fluence profile is narrow. Therefore, more photons are required at 0 and 180 degrees to flatten the fluence with MAD than at 90 or 270 degrees. The MAD requires higher scaling and modulation to generate the same number of photons incident on the phantom.

E. Simulated Projection Data

Figure 8 and 9 show the fluence profiles with and without the phantom in the field of view for the no filter and MAD filtered scenarios (TCM is used in both cases). In Fig. 8, we see that the no filter scenario can only modulate the per view number of photons through TCM while the dual MAD filter can customize both the shape and intensity of the beam. In Fig. 9, the post-object fluence is more uniform across object projections (the design goal) than the no filter, TCM-only scenario.

F. Simulated Reconstructions

With Poisson noise added to the projection data in Fig. 9, filtered backprojection reconstructions were performed for both filtering scenarios. Results are shown in Figure 10. Both methods show approximately the same average noise level (as expected due to an equal number of incident photons). However, we see much greater noise uniformity in the MAD filtered image. This is significant if a minimum noise level is prescribed to obtain sufficient image quality. The TCM-only case will require more incident photons (hence larger dose) to obtain the same minimum noise level over the entire image.

G. Ongoing and Future Work

While these initial results suggest that dual MAD filters can successfully achieve a broad class of fluence patterns, we are working to fabricate physical MAD devices and evaluate performance in an experimental CT system.

REFERENCES