Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2019 February ; 10948: . doi:10.1117/12.2513427.

Volume-of-interest Imaging Using Multiple Aperture Devices

W. Wang, G. J. Gang, J. H. Siewerdsen, and J. W. Stayman

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

Abstract

Volume-of-interest (VOI) imaging is a promising strategy for dose reduction in computed tomography (CT) while retaining image quality. However, implementation of VOI-CT has been challenged by the lack of adequate hardware and the interior tomography reconstruction problem. Multiple aperture devices (MAD) are a novel filtration scheme that can achieve x-ray fluence field modulation in a compact design with small translations. In this work, we propose a general approach for VOI imaging using MADs. MAD trajectories are designed to dynamically tailor the fluence for prescribed VOI. A penalized-likelihood reconstruction algorithm is proposed for fully truncated projections extended with scout views. Physical experiments were conducted to verify the feasibility for non-centered elliptic VOIs. Image quality and dose were estimated and compared with standard fullfield protocols. The ability of MAD-based VOI imaging to retain high image quality while significantly decreasing the total dose is demonstrated, suggesting the potential for dose reduction in clinical CT applications.

Keywords

Fluence field modulation; volume-of-interest imaging; model-based iterative reconstruction

1. INTRODUCTION

While CT is used for a wide range of disease diagnosis, treatment monitoring, and assessment, overall radiation exposures to patient populations and individual doses to radiosensitive organs have been a growing concern. In many diagnostic applications, only a relatively small image volume needs to be assessed -e.g. cardiac and spine imaging, or follow-up studies on single organs. As such, VOI imaging using a narrow X-ray beam limited to the particular anatomical target and sparing radiation dose to surrounding tissue is a potentially promising technique. VOI imaging is challenging in two aspects. First, creating hardware capable of fluence field modulation on modern CT has been a challenge due to size and speed constraints. Prior investigations have concentrated on *static* collimators or filters to block or decrease the incident fluence outside the VOI.^{1,2} However, these methods permit only centered cylindrical VOIs. While research in flat-panel cone-beam CT has considered dynamic collimation design³ or used increased motion capabilities of a C-arm⁴ to realize a non-centered VOI, implementation in CT remains a challenge. The second difficulty for VOI imaging is the reconstruction of truncated data. The interior tomography

problem is ill-posed and is not easily handled with conventional reconstruction methods. Previous work showed that additional prior knowledge permits exact reconstruction with truncated projections (within the VOI) using Hilbert transform methods.⁵

Previously, we introduced novel binary filters called multiple aperture devices (MADs) that permit fluence field modulation in a compact space with relatively small motions.^{6,7} Briefly, dual-MAD filters, placed in series, effect a varying fluence field pattern based on changing the relative displacement between the two filters. The beam center may be shifted by moving both filters simultaneously as a whole. In this work, we extend a general approach⁸ for VOI imaging using MAD filters. We leverage a workflow that includes full-field anterior-posterior (AP) and lateral scout views for both VOI specification and for use in reconstruction to solve the interior reconstruction problem. Methods for fluence field design, scatter correction, and model-based reconstruction are presented. Image quality and dose were compared between MAD-based VOI imaging and standard full-field protocols.

2. MATERIALS AND METHODS

2.1 Filter trajectory design for VOI fluence field modulation

The proposed workflow and modulation strategy is summarized in Figures 1 and 2. The dual MAD filtering system provides the flexibility for dynamic fluence field modulation for VOI imaging with arbitrary fluence width and centering in the transverse plane. While there is potential to design a variety of VOI shapes, we focus on elliptical cylinder VOIs -wherein an ellipse circumscribes the anatomy of interest in the axial view. Two projection views (AP and lateral) are acquired in advance to identify the VOI size and location relative to the center of field of view. These two projection frames cover the entire support and are used in interior image reconstruction to maintain the accurate CT values within the VOI.

We denote the desired VOI with a 2D binary mask, m_{VOI} , where voxels within the VOI are labeled 1, and outside, 0. Projection data that is directly associated with the VOI area at projection angle θ is given by another binary mask, $g(\theta) = \mathbb{1} \{ \mathbf{A}(\theta) m_{\text{VOI}} > 0 \}$ where $\mathbf{A}(\theta)$ is the fan-beam (central slice) system matrix that projects a 2D image volume to a 1D projection profile at angle θ . The logical operator, $\mathbb{1} \{ \cdot \}$, returns 1 if the input statement is true such that $g(\theta)$ has value 1 where the VOI mask projection is non-zero. The set of masks $g(\theta)$ over all θ represents the projection support required for obtaining the desired VOI. Specifically, we may specify the detector column indices required for obtaining the desired VOI as $u_{\text{VOI}}(\theta) = \{ u | g(u, \theta) > 0 \}$.

To estimate the MAD actuation required to achieve specific fluence patterns, a calibration of all possible patterns is acquired through an exhaustive sweep. We denote $f(u; t_0, t_1)$ as the lateral transmissivity profile of the dual-MAD filters with individual MAD0 and MAD1 filter positions labeled t_0 and $(t_1 + t_0)$, respectively. Note that t_1 represents a shift of MAD1 relative to MAD0.

Fluence field design for VOI imaging using MAD-based dynamic filtration follows the following goals: 1) Seek maximum (relative) exposures within the VOI region while limiting

exposure outside the VOI. And 2) constrain minimum fluence within VOI projections to not be lower than a threshold $f_{\min}$ (i.e., to maintain data fidelity at the edge of the VOI). Mathematically, the optimization problem is posed as

$$(\hat{t}_0, \hat{t}_1)(\theta) = \arg \max_{(t_0, t_1) \in \mathcal{T}_\theta} \frac{\sum_{u \in \{u_{\text{VOI}}\}(\theta)} f(u; t_0, t_1)}{\sum_u f(u; t_0, t_1)}, \quad (1)$$

$$\mathcal{T}_\theta = \left\{ (t_0, t_1) \left| \min_{u \in \{u_{\text{VOI}}\}(\theta)} f(u; t_0, t_1) > f_{\min} \right. \right\}. \quad (2)$$

Specifically, the fraction of the exposure within VOI projections is maximized but constrained to a specific minimum transmissivity, $f_{\min}$, within the VOI projection. Because MAD fluence patterns are periodic, actuation estimates are susceptible to jitter between periods. To avoid this an initial estimation of $t_0(\theta)$ and $t_1(\theta)$ is followed by a sinusoidal fitting of $\tilde{t}_0(\theta)$ and a new estimation of $t_1(\theta)$ with fixed $\tilde{t}_0$.

2.2 Penalized-likelihood reconstruction for truncated VOI data

MAD-based VOI imaging data must be corrected to account for a number of physical effects including the incident fluence distribution and detector sensitivity as well as other sensitivities (e.g., focal spot dynamics and spectral changes) as described in previous work.⁷ Fluence levels outside the projected VOI are too low to be useful for reconstruction; therefore, the projection data can be severely truncated. To perform the interior tomography reconstruction problem we choose to leverage the full-view AP and lateral scouts that were used to specify the VOI. Providing this additional information that covers the entire projection support in two views should theoretically permit exact reconstruction in the VOI.⁹ We adapt a penalized-likelihood method that uses both the scout views and the truncated data in a single measurement vector for VOI image reconstruction.

Scattered radiation differs greatly between the full-field AP/lateral views and MAD-VOI scan due to incident fluence differences. Therefore projection data are preprocessed to eliminate/equilibrate the scatter between the full-field scouts and the MAD-VOI data. A simple constant scatter correction is applied to the full-view scout data. For MAD-VOI data, scatter correction uses a low-dimensional model based on Monte Carlo simulation. The parametric model includes a scatter distribution term related to the MAD transmissivity profile and a scatter magnitude parameter. The latter parameter is estimated by matching the MAD-truncated projections at AP/lateral view angles with the corresponding scatter-corrected full-view data. The combined scatter-corrected scout view and MAD-VOI data are used as an input to model-based reconstruction. To fit the forward model for penalized-likelihood, the incident fluence is estimated using bare-beam variance measurement scaled

by the transmissivity profile $f(u; \hat{t}_0(\theta), \hat{t}_1(\theta))$ and MAD-VOI data outside the projected VOI are set explicitly to zero.

2.3 Experimental setup for investigation of MAD-based VOI imaging

We tested the feasibility of MAD-based VOI imaging on a cone-beam CT test bench with dual MAD filters (Figure 3a). Two scenarios were studied: 1) a cylindrical VOI for spine imaging using a QRM abdomen phantom (Figure 3b), and 2) an elliptical cylinder VOI around the left lung in a CIRS Xsight phantom with 6 subsolid nodules (Kyoto Kagaku) (Figure 3c). MAD filter trajectories were designed based on AP/lateral views in each case.

In addition to the MAD-VOI scans, two full-view scans were acquired with a “high dose” reference and “low dose” scan for comparison. All scans used 100 kVp and 720 frames with equiangular steps over 360° . The bare-beam fluence, I_0 , was estimated to be 8.1×10^4 photons/pixel in the high-dose cases and 1.3×10^4 photons/pixel in the low-dose cases, and averaged 2.7×10^4 photons/pixel within the VOI in the MAD scans.

All reconstructions used a separable quadratic surrogate (SQS) algorithm with shift-invariant quadratic penalty.¹⁰ The effects of regularization were approximately matched between different protocols by setting the ratio of the (average) bare-beam fluence I_0 and the regularization parameter β to a constant 0.8. Each reconstruction used 300 iterations (20 subsets for 5 iterations, 10 subsets for the next 5, and 1 subset for the remaining). Due to differences in scatter correction and full-field versus MAD processing, the residual difference between reconstructed CT values between protocols were matched by linear scaling based on two uniform regions. Dose maps and total accumulated dose was calculated using Monte-Carlo simulation¹¹ in all scans.

3. RESULTS

3.1 MAD-VOI fluence field modulation

MAD filter actuation estimates are shown in Figure 5a for each of the two imaging scenarios. The MAD0 translation is shown in red, and MAD1 translation relative to MAD0 in blue. The MAD0 translation is ± 25 mm and sinusoidal while the relative MAD0-MAD1 motion is constrained to a single MAD period and ± 0.4 mm.

Achieved fluence field patterns are shown in Figure 5b. For spine VOI, the fluence width is nearly constant over projection angles, while the center of the beam is shifted sinusoidally. For elliptical lung VOI, the fluence width changes over projection angles. The fluence width reaches the maximum at 270° where the ellipse major axis is parallel to the detector plane with the highest magnification.

3.2 Image quality analysis and dose calculation

PL reconstructions of the VOI for full-field high dose, full-field low dose, and MAD-VOI scans are shown in Figure 5c. The dose distributions in full-field low-dose and MAD-VOI scans are shown in Figure 4. Contrast-to-noise ratios (CNRs) for the mixed nodules and spine (bone/soft tissue) were computed for each scan. The image quality and total dose

metrics are summarized in Table 1. We observe that the noise in MAD-VOI reconstructions is lower than in full-field low dose case. The dose absorbed in lateral anatomy is significantly reduced, while the dose within the ROI is increased. CNR of MAD-VOI reconstruction is higher than full-field low dose images for all testing objects (and approaching the quality of the high-dose full-field). Moreover, the total dose is reduced by 20% in abdomen phantom scan, and 10% in lung phantom scan compared with full-field low dose protocol.

4. DISCUSSION

In this work, we proposed a general method for VOI imaging with MAD-based fluence field modulation and validated the feasibility to achieve a non-centered elliptic cylindrical VOI. Scout views are used for VOI positioning, scatter correction, and truncated data reconstruction. Examples of optimized MAD actuation and achieved fluence modulation for prescribed VOI are demonstrated. Experiments show that the MAD-VOI imaging outperforms a full-view low dose protocol while reducing the total absorbed dose suggesting opportunities for patient radiation reduction. Future work will investigate additional constraints on modulation patterns (e.g., to avoid radiation dose to specific anatomical regions¹²), consider variations to the methodology for helical CT scanning, and conduct VOI scans in cadaver and/or patient studies.

ACKNOWLEDGMENTS

This work was supported, in part, by NIH grant U01EB018758.

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Figure 1.
MAD-based VOI imaging framework.

Figure 2.
MAD-based VOI imaging diagram and notations.

Figure 3.

a) Test bench experiment setup. b) Abdomen QRM phantom with a non-centered cylindrical VOI. c) CIRS Xlight lung phantom with a non-centered elliptic VOI. (VOIs identified by a red dashed line).

Figure 4. Monte-Carlo absorbed dose distribution in: a) full-field low dose scan of abdomen phantom, b) MAD-VOI scan of abdomen phantom, c) full-field low dose scan of lung phantom, d) MAD-VOI scan of lung phantom. The colormap is shown in log-scale, and the display window is lower for MAD-VOI lung to better show variations.

Figure 5. Top row: Circular VOI spine experiment. Bottom row: Elliptical VOI lung experiment. a) Trajectories of MAD filters for VOI imaging. b) MAD-based barebeam fluence field modulation patterns. c) Penalized-likelihood reconstructions of full-field high dose, full-field low dose, and MAD-VOI scans.

Table 1:

SNR, CNR and Integral Absorbed Dose

Experiment	CNR							Dose (mGy)	
	nodule 1	nodule 2	nodule 3	nodule 4	nodule 5	nodule 6	spine	abdomen	lung
High-dose full-field	32.9	73.0	22.2	16.4	10.9	16.0	22.1	50.4	47.3
Low-dose full-field	30.7	42.3	16.2	14.0	9.1	12.9	12.4	7.52	7.09
FFM-VOI	32.3	60.6	19.0	16.2	10.0	15.1	19.1	6.00	6.40