

Realistic Undersampling Model for Compressed Sensing Using a Multi-Slit Collimator

Baiyu Chen, Matthew J Muckley, Thomas O'Donnell, Aaron Sodickson, Thomas Flohr, Karl Stierstorfer, Bernhard Schmidt, Florian Knoll, Thomas Vahle, Andrew Primak, David Faul, Daniel Sodickson, Ricardo Otazo¹

Abstract— Compressed sensing methods use undersampled CT projection data to reduce radiation dose. Instead of pausing the x-ray tube, a multi-slit collimator (MSC) has been proposed to undersample the data by partially blocking the beam before it reaches the patient. One potential concern for the MSC approach is the penumbra caused by the finite size of the focal spot. Therefore, this study evaluates several MSC designs with the penumbra considered, and quantitatively demonstrates that a wider MSC slit and a smaller focal spot provide better dose efficiency and beam separation. Further work is needed to examine the data incoherence of the undersampled data and to design the MSC with an optimal tradeoff between dose efficiency and incoherence.

Index Terms— CT, compressed sensing, sparse CT, undersampling, multi-slit collimator, penumbra

I. INTRODUCTION

To minimize public health risks from radiation exposure, NIH called for new technologies to reduce the effective dose of routine CT exams to less than 1 millisievert (sub-mSv level) [1]. This is challenging for thoracic and abdominal CT, because a 5-7 fold reduction of routine dose would be needed without compromising diagnostic accuracy.

One promising technology to achieve such dose reduction is compressed sensing (CS), which uses undersampled projection data to reconstruct images without streaking artifacts. Previous studies have shown order-of-magnitude dose reductions with CS using reduced-views projection data [2, 3]. However, because the x-ray tubes of CT scanners cannot be pulsed quickly enough (the thermal inertia of the cathode can't be overcome quickly enough) to directly acquire the reduced-views projection data, the data used in prior CS studies were retrospectively undersampled from complete datasets, meaning that the actual dose of the scan was not lowered. To make CS applicable to radiation dose reduction in clinical practice, hardware modifications that directly acquire the undersampled data are needed.

Recently, a multi-slit collimator (MSC) has been proposed

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Baiyu Chen (e-mail: Baiyu.Chen@nyumc.org), Matthew J Muckley, Florian Knoll, Daniel Sodickson, and Ricardo Otazo (e-mail: Ricardo.Otazo@nyumc.org) are with NYU Langone Medical Center, New York, NY USA.

Thomas O'Donnell, Thomas Flohr, Karl Stierstorfer, Bernhard Schmidt, Thomas Vahle, Andrew Primak, and David Faul are with Siemens Healthineers.

Aaron Sodickson is with Brigham and Women's Hospital, Boston, MA USA.

[4, 5] to undersample the projection data by partially blocking the beam between the x-ray tube and the patient. However, the effect of the beam penumbra, caused by the finite size of the focal spot, has not been evaluated. We examined several MSC designs with the penumbra considered, and compared the designs from the perspective of beam separation and dose efficiency.

II. METHODS

The MSC is illustrated in Figure 1. Because the x-ray beam is partially blocked before reaching the patient, patient dose is reduced; because the detector rows are partially irradiated, the projection data are undersampled. The MSC is jittered (or the focal spot moves) as the gantry rotates to sample different rows of the detector.

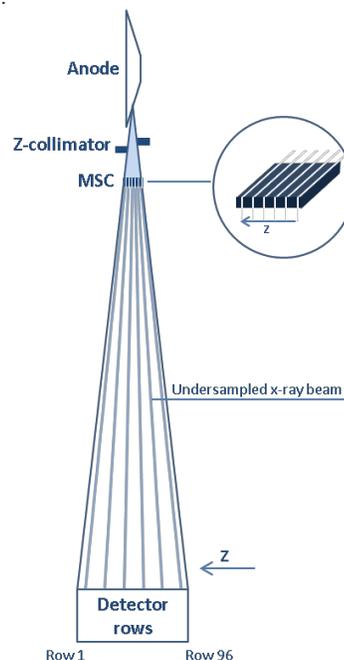


Figure 1: The multi-slit collimator (MSC) partially blocks the beam before it reaches the patient, such that undersampled projection data are acquired while reducing patient dose.

Ideally, if the focal spot were infinitely small, the photon distribution through an MSC slit would have a rectangular profile, where the flux would be either zero or the same as the flux without the MSC. However, the finite size of the focal spot creates a penumbra [Figure 2(a)], which spreads the

photon/dose from a rectangular to a bell-shaped profile [Figure 2(b)]. The penumbra raises two concerns. First, because the bell-shaped distribution covers wider region, the beams through neighboring MSC slits might not be well-separated. Second, because the penumbra effect distributes the intensity of the beam, the noise of the projection data increases, the reliability of the projection data decreases, and the dose efficiency decreases.

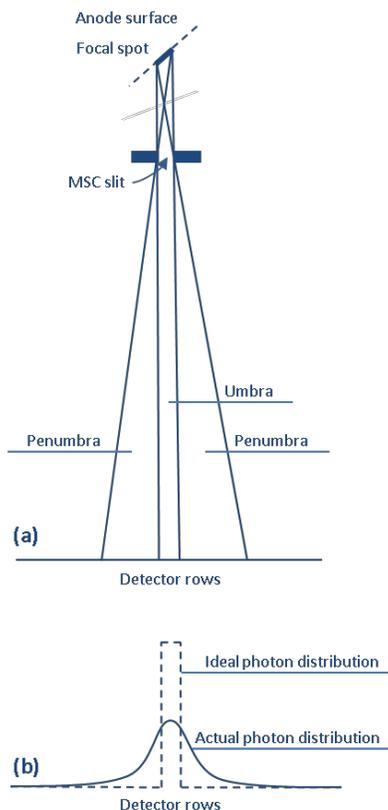


Figure 2: Due to the finite size of the focal spot, the beam through the MSC contains penumbra. (b) Because of the penumbra, the same amount of photons (same dose) is spread from a rectangular distribution into a bell-shaped distribution.

To assess the impact of these two concerns, we simulated the photon distribution of various MSC configurations and focal spot sizes, and quantified the corresponding dose efficiency.

A. A numerical simulation tool

We developed a numerical simulation tool to understand the photon distribution after the MSC. The following features were modeled in the simulation: focal spot size, focal spot intensity distribution, anode angle, CT gantry geometry, and MSC specifications. Everything other than the MSC was modeled after a 96-row commercial CT scanner (Siemens SOMATOM

Force).

To validate the simulation tool, photon distributions were also experimentally measured from a Siemens SOMATOM Force scanner. Because the scanner did not have an MSC installed, the existing z-collimator (shown in Figure 1) was brought together in close proximity to mimic a single slit of the MSC. The slit width, projected to the detector surface, was selected to be the same as the width of one detector row. This experiment was repeated at multiple slit locations (irradiating different detector rows) using two focal spot size modes (*stdHR*, large focal spot; and *superHR*, small focal spot). The experimental results are compared to the numerical simulation results.

B. MSC design

The simulation tool, once validated, was used to evaluate four MSC configurations, as listed in Table 1. The slit separation is the distance between neighboring slits. The undersampling factor is the ratio between the slit width and the slit separation. The photon distribution was simulated for each MSC configuration, from which the dose efficiency was further calculated.

The dose efficiency (DE) was calculated using the equation below, where R is the undersampling factor, $D(x)$ is the dose at detector Row X , $C(x)$ is the flux at detector Row X , $w(x)$ is the weighting factor at detector Row X , and the subscript “MSC” indicates with or without MSC. The weighting factor is inversely proportional to the variance of the post-log projection data (calculated from Poisson statistics of $C(x)$), similar to the weighting factor used in statistical iterative reconstruction methods [6]. This weighting factor can be understood as a measure of each photon’s contribution to the image quality of the reconstructed image. DE is maximized at 1 when the focal spot is infinitely small (no penumbra, rectangular flux distribution).

$$DE = \frac{R * \sum_{Row1}^{Row96} D_{MSC}(x) * w_{MSC}(x)}{\sum_{Row1}^{Row96} D(x) * w(x)} = \frac{R * \sum_{Row1}^{Row96} C_{MSC}(x) * w_{MSC}(x)}{\sum_{Row1}^{Row96} C(x) * w(x)}$$

III. RESULTS

A. Validation of simulation tool

Figure 3 compares the numerically simulated photon distributions to the experimentally measured photon distributions using two focal spot modes. The photon distributions are shown in terms of normalized flux, which is the flux with MSC divided by the flux without MSC. Overall the simulation and experimental results agree well, which validates the simulation tool. The only exception is observed towards the cathode, where the simulation results have slight larger magnitude, possibly because the thickness of MSC was not modeled in the simulation.

Table 1: MSC configurations of different slit widths and slit separations.

Configuration name	Undersampling factor	MSC slit width (projected to detector surface)	MSC slit separation (projected to detector surface)
W1S8	8 fold	1 detector row	8 detector rows
W2S16	8 fold	2 detector rows	16 detector rows
W3S24	8 fold	3 detector rows	24 detector rows
W4S32	8 fold	4 detector rows	32 detector rows

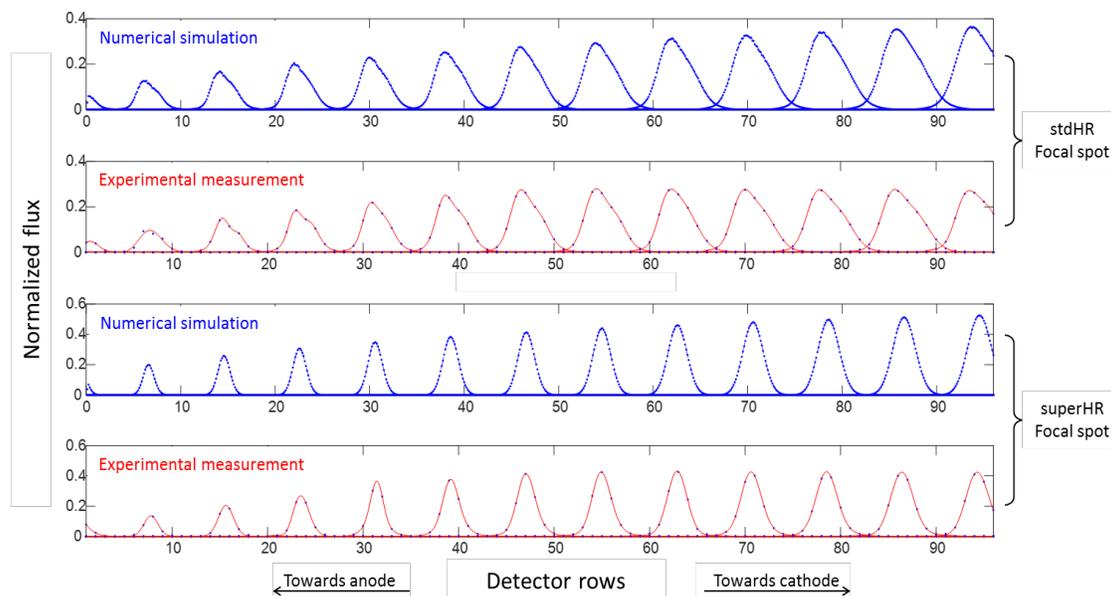


Figure 3: The numerically simulated photon distributions and experimental measured photon distributions at multiple MSC slit locations using two focal spot sizes (*stdHR* and *superHR*). Each peak corresponds to an MSC slit.

In Figure 3, the flux towards the cathode is approximately 3-time-higher than the flux towards the anode. This variation is largely artefactual, caused by the uneven heights of the z-collimator leaves used to mimic a MSC slit in experimental validations (Figure 4). Since a true MSC design would have both sides of each slit at the same height, all following simulations assumed even height. Also note that the relative flux in Figure 3 is consistently less than 1, which can be explained by Figure 2(b).

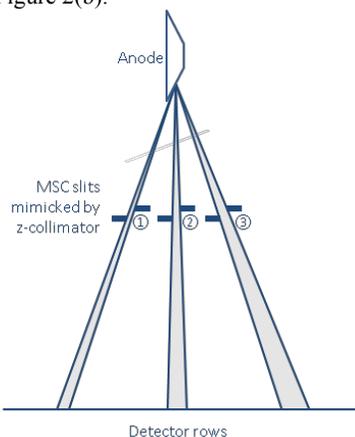


Figure 4: The z-collimator is narrowed and moved to simulate MSC slits at three different locations. Because the two leaves of the z-collimator are at uneven height, the slit towards the anode results in less flux.

B. MSC design

The photon distributions reaching the detector corresponding to 4 MSC configurations and 2 focal spot modes are shown in Figure 5. Several trends are observed. First, the smaller focal spot size produces a more concentrated photon distribution, as evidenced by comparing the *superHR* mode to *stdHR* mode or by comparing the anode side to the cathode side (since the

effective focal spot size is smaller towards the anode side). Second, the flux increases proportionally to slit width, evidenced by the area under each curve. Third, the FWHM does not increase proportionally with the slit width, because the FWHM is dominated by the size of the penumbra, which largely depends on the size of the focal spot. Finally, wider slits allow better beam separation. To achieve an 8-fold undersampling with the *stdHR* focal spot, a slit width of at least 2 detector rows (projected to detector surface) is needed to separate the beams from neighboring slits.

Based on the aforementioned observations, a wider MSC slit and a smaller focal spot has higher dose efficiency, as confirmed in Table 2.

IV. DISCUSSION

While this study demonstrates that a wider MSC slit and a smaller focal spot are preferable from the perspective of dose efficiency and beam separation, there are additional factors to be considered. For example, considering anode heating limit, a smaller focal spot might not be suitable for scans that require a large tube current; considering undersampling incoherence, a wide MSC slit will decrease the incoherence and thus be disadvantageous to compressed sensing. Further work is needed to examine the impact of penumbra on undersampling incoherence in order to design the MSC with optimal tradeoff between dose efficiency and incoherence.

V. CONCLUSION

A MSC has been proposed to undersample CT projection data for compressed sensing reconstruction with reduced patient dose. Several MSC designs were examined from the perspective of beam separation and dose efficiency. To our knowledge, this is the first exploration of a practical CT undersampling scheme, including consideration of penumbra effects.

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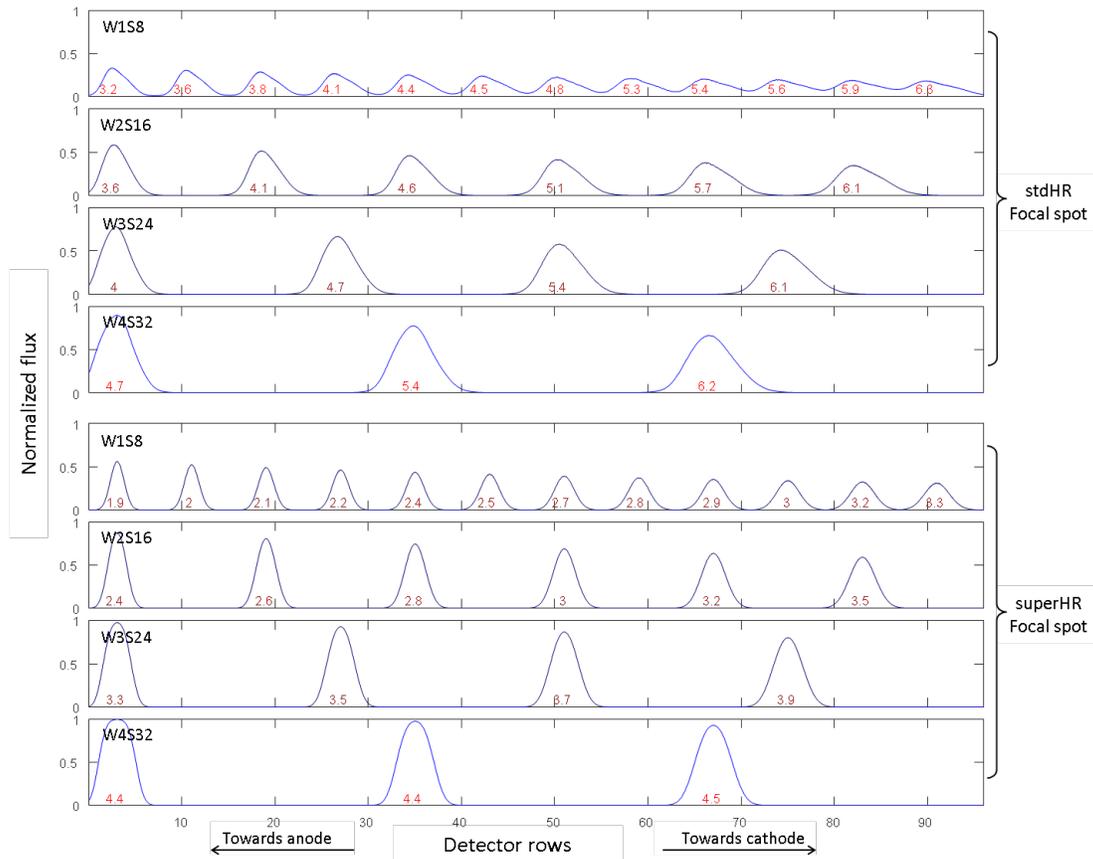


Figure 5: Photon distributions corresponding to four MSC configurations and two focal spot modes. The numbers in red are the FWHM of each peak, in units of detector rows.

Table 2: The dose efficiency corresponding to four MSC configurations and two focal spot modes.

Focal spot mode	MSC configuration	Dose efficiency
<i>stdHR</i>	W1S8	17%
<i>stdHR</i>	W2S16	30%
<i>stdHR</i>	W3S24	41%
<i>stdHR</i>	W4S32	50%
<i>superHR</i>	W1S8	30%
<i>superHR</i>	W2S16	50%
<i>superHR</i>	W3S24	62%
<i>superHR</i>	W4S32	71%