Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2016 February 27; 9788: . doi:10.1117/12.2216248.

Design and evaluation of a grid reciprocation scheme for use in digital breast tomosynthesis

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Abstract

This work describes a methodology for efficient removal of scatter radiation during digital breast tomosynthesis (DBT). The goal of this approach is to enable grid image obscuration without a large increase in radiation dose by minimizing misalignment of the grid focal point (GFP) and x-ray focal spot (XFS) during grid reciprocation. Hardware for the motion scheme was built and tested on the dual modality breast tomosynthesis (DMT) scanner, which combines DBT and molecular breast tomosynthesis (MBT) on a single gantry. The DMT scanner uses fully isocentric rotation of tube and x-ray detector for maintaining a fixed tube-detector alignment during DBT imaging. A cellular focused copper prototype grid with 80 cm focal length, 3.85 mm height, 0.1 mm thick lamellae, and 1.1 mm hole pitch was tested. Primary transmission of the grid at 28 kV tube voltage was on average 74% with the grid stationary and aligned for maximum transmission. It fell to 72% during grid reciprocation by the proposed method. Residual grid line artifacts (GLAs) in projection views and reconstructed DBT images are characterized and methods for reducing the visibility of GLAs in the reconstructed volume through projection image flat-field correction and spatial frequency-based filtering of the DBT slices are described and evaluated. The software correction methods reduce the visibility of these artifacts in the reconstructed volume, making them imperceptible both in the reconstructed DBT images and their Fourier transforms.

Keywords

anti-scatter grid; reciprocation; digital tomosynthesis; power spectrum; image filtering

1. INTRODUCTION

Loss of contrast due to scattered radiation is a common problem in full-field digital mammography (FFDM) and digital breast tomosynthesis (DBT). Scatter can be detrimental to image quality, especially for thicker, denser breasts. To remedy the foggy appearance of images that resulted from scattered rays, Gustav Bucky invented the anti-scatter grid, which is placed between the detector and the object being imaged to reduce the amount of scattered radiation that reaches the detector.¹ In modern mammography, anti-scatter grids have been

shown to improve contrast by anywhere from ~15% to over 70%, depending upon the thickness and composition of the breast.^{2,3,4} Figure 1 is a schematic of how anti-scatter grids are typically implemented. Many anti-scatter grids used in mammography are 1-dimensional (1D), or linear, comprised of narrow high-attenuation strips, aligned in the posterior-anterior direction. The strips (lamellae) are made from high atomic number materials that remove secondary photons whose trajectories have appreciable components perpendicular to the direction of the strips [2].

To minimize cutoff of primary rays originating from cone beams, most 1D grids are focused, wherein strip orientations are increasingly non-normal to the detector surface with increasing distance from the grid midline. The rate of change of the strip angle is selected so that the gaps between the strips are oriented such that their axes all intersect at a single line in space. A lower atomic number material, such as aluminum or carbon fiber, is used to separate and align grid walls. A linear grid can be characterized by a grid ratio, which is the ratio of the height of the grid to the thickness of the interspace material, and its strip density, provided as a frequency expressed in line pairs per cm or line pairs per inch (lp/cm or lp/in). The grid pitch, which is the sum of the widths of a single wall and a single space, is then the inverse of this grid frequency. The degree of scatter rejection will depend upon the strip attenuating material, the grid ratio, and the grid pitch. Larger grid ratios generally provide better scatter rejection.^{5,6} Grid ratios typically used in mammography are of the order of 5:1, and grid frequencies are typically ~30 lp/cm.⁷

With 1D grids scattered photons traveling parallel to the strips are still transmitted, so only partial elimination of scatter is possible. Focused 2D anti-scatter grids increase scatter rejection by using an array of holes rather than strips to block scatter.^{8,9} Mechanically sturdy 2D grids can be manufactured with air interspaces, thereby improving interspace primary transmission compared to grids with solid interspace material. Most cellular grids have square holes, and only the central hole at the chest wall edge has walls that are perpendicular to the surface of the grid. All other holes in both the anterior direction and the left and right directions away from the central hole have increasingly angled walls to match the paths of primary rays originating from the x-ray source. When extended, the axes of these holes intersect at a single point in space, referred to as the grid focal point (GFP). The GFP is located a fixed distance measured perpendicularly from the midpoint of the grid's posterior edge, called the grid focal distance. When a 2D grid is centered on the detector, the GFP is located in the same point in space as the x-ray focal spot (XFS).

In addition to removing a portion of the scattered photons exiting the breast, anti-scatter grids also remove primary (non-scattered) photons. The missing primary photons result in an image of the high-attenuation components of the grid, Grid line artifacts (GLAs) are the images of the grid walls, superimposed upon the image of the breast. Both linear and cellular grids with typical mammographic grid frequencies create GLAs in the image if the grid is kept stationary during exposure. Kim and Lee have proposed using stationary, high frequency cellular grids with hole matrices rotated with respect to the detector sampling matrix.¹⁰ Specific values of grid hole pitch and grid rotation are chosen to place the primary and first few harmonic grid peaks on the boundaries of the region $[-f_N < u < f_N]$; $[-f_N < v < f_N]$ where $f_N = 1/(2p)$ is the Nyquist frequency and p is the detector sampling interval. This

approach makes low pass or band pass frequency space peak filtering approaches possible without affecting any breast information in the final image.¹⁰ While this is an effective method for removing artifacts, high frequency cellular grids with selectable spatial frequency are presently not readily available.

To minimize GLAs, grids are typically reciprocated (translated) in the left-right direction during the x-ray exposure. Hollis E. Potter pointed out that elimination of GLAs requires that every position on the detector must experience the same attenuation by the grid over the duration of the exposure, which can be accomplished by translating the grid at a uniform speed by some integer multiple of grid pitches during the x-ray exposure.¹¹ The required repeat distance r_d is thus defined as the distance measured in the direction of translation over which the pattern of the grid walls repeats.¹¹ Potter explained that the most effective method of blurring grid lines was to move the grid by at least 10 to 20 repeat distances.¹¹ Equation (1) shows the relationship among grid reciprocation speed v_g , a given exposure time t_{exp} , and an integer multiple N of grid pitches, assuming a constant x-ray fluence over t_{exp} for any given point on the detector.

$$v_g = \frac{N r_d}{t_{exp}}. \quad (1)$$

Blurring by reciprocation is relatively straightforward for 1D grids where grid motion is perpendicular to the direction of the strips, enabling the posterior edge of the grid to extend beyond the posterior edge of the field of view of the detector throughout the exposure. In contrast, the 2D nature of the holes (usually square in shape) of cellular grids make removal of the grid image via reciprocation more challenging. If the grid walls were oriented to be parallel to the translation direction, dark bands would form in the image where the shadows of the parallel walls fall on the detector throughout the exposure. For this reason, the square hole matrix of 2D grids is typically rotated so that left-right translation is not parallel to the direction of any walls. Like linear grids, cellular grids must also be moved by a specific distance in order to uniformly reduce the primary fluence to every point on the detector. Unlike linear grids where the repeat distance is equal to the grid pitch, rotated cellular grids have repeat distances that depend on both the hole pitch and the rotation angle. The lamellae of square grids are periodically spaced in two orthogonal directions. In order for a repeat pattern to be created, Pellegrino et al. explain that an integer number of pitches must be translated in both grid dimensions.¹² Assuming that the pitch p is the same in both dimensions, the repeat distance can be calculated as shown in Equation (2) from integers n and m that correspond to the number of grid pitches p in the two orthogonal directions of the grid matrix.

$$r_d = p \sqrt{m^2 + n^2}. \quad (2)$$

An example of a rotated grid with $m = 4$ and $n = 5$ is shown in Figure 2, which shows a pattern of walls and spaces that repeats every r_d .

A right triangle is formed from the repeat distance and the walls with lengths mp and np as shown in Figure 2. These integers m and n are related to the grid rotation angle φ as shown in Equation (3).

$$\varphi = \tan^{-1}\left(\frac{m}{n}\right). \quad (3)$$

For the case of $m = n$, a 45° grid rotation angle is formed and the repeat distance is then simply $\sqrt{2}p$. Similar to Hollis Potter's findings, Pellegrino et al. also explain that the more grid lines that are traversed, the less precise the motion of the grid must be in order to result in acceptably low artifacts in the image.¹² However, Pellegrino et al. also explain that a 45° angle of rotation would lead to bright streaks in the image where the attenuation by the aligned intersections of the grid walls would be lower than that of the walls adjacent to these intersections because the amount of material at the intersections is approximately half of that in the sum of the adjacent walls.¹² Figure 3(a) illustrates this concept using an image of a portion of a grid where $m = n$. The image is an attenuation map of the grid where non-zero pixel values indicate attenuating material and zero pixel value indicates air interspaces. Ten repeat distances are included in the image. Results of summing the pixels through this grid image along the rows of the image matrix simulate the total attenuation that would be experienced by a given detector element beneath the grid shown in Figure 3(a) moving at a uniform speed. The total normalized attenuation by the grid can then be plotted versus the index of the row to simulate the attenuation seen by a column of detector elements as is shown in Figure 3(b). Dips in the plot of Figure 3(b) show where the detector would see more primary rays due to the lower attenuation of the intersections described by Pellegrino et al.¹²

Cha-Mei Tang describes a way of remedying these brighter bands by placing additional material at the intersections to increase the attenuation at these locations,¹³ which would remove the valleys seen in the plot of Figure 3(b). Another method of reducing these fluctuations, as presented by Pellegrino et al., is to use a rotation angle that is slightly offset from 45° such that less of the intersections are aligned along the direction of translation.¹² Values of m and n suggested by Pellegrino et al. ranged from $m = 9, n = 1$ to $m = 9, n = 8$, which would result in values of r_d ranging from approximately $9p$ to $12p$. Because the number of grid lines that covers each detector element would be sufficiently large at these values of m and n , these grids would only need to be translated by a single repeat distance.¹² Figure 4(a) is an example of a grid with $m = 9$ and $n = 8$, or $\varphi = 48.4^\circ$. Figure 4(b) is a plot of the estimated attenuation that a column of detector elements would experience as a function of the row index similar to Figure 3(b). This choice of m and n notably reduces the non-uniformity in fluence to the detector along a column of pixels, indicating a lower probability of seeing bright band artifacts in the final image.

Attenuation of primary photons by the grid degrades the statistics of the breast image, potentially leading to lower detectability of suspicious structures. X-ray exposure to the breast is typically increased in order to compensate for the lost signal at the detector, thereby increasing breast radiation dose.¹ Thus, the net change in the image signal-to-noise ratio due

to the introduction of a grid depends on the relative reduction in signal (primary) and noise (scatter) photons.

To maximize primary transmission through a focused cellular grid, the distance between the XFS and GFP should be minimized over the entire x-ray exposure, which is not the case for conventional planar grid reciprocity. For alignment of the XFS and GFP throughout grid motion, the grid trajectory would ideally be such that the grid's midpoint at its posterior edge travels along an arc of a circle whose center is located at the XFS and whose radius is equal to the grid focal distance. In this trajectory, each point along the grid's edge would trace out a circle. The largest circle would be traced out by the corners of the grid, the radius of which is equal to the distance between the corner and the GFP.

We define a coordinate system in which the x-axis is parallel to the detector surface, the y-axis is parallel to the axis of rotation of the DBT scanner, and the z-axis points from the detector towards the x-ray tube, with the coordinate system origin located at the midpoint of the grid's posterior (chest wall) edge when the grid is parallel to the detector surface. In that coordinate system the XFS is located at a distance h above the origin, and the curve traced out by a chest wall edge corner of the grid in the ideal circular trajectory can be described as in Equation (4).

$$z = -\sqrt{(h^2 + w_{grid}^2) - x^2} + h, \quad (4)$$

where w_{grid} is half the width of the grid. For travel over a single repeat distance the x-

coordinate of the grid corner would lie within the interval $\left[w_{grid} - \frac{r_d}{2}, w_{grid} + \frac{r_d}{2}\right]$ where r_d is the repeat distance of the grid.

Since grid motion repeat distances of the order of centimeters are small in comparison to the large focal spot-to-detector distances typical of x-ray breast imaging (60 to 80 cm), the ideal arc trajectory traced by the corners of the grid can be well approximated by moving the corners along straight line trajectories that are tangent to the ideal circular arc at the corner locations when the grid is centered beneath the focal spot. Those tangent lines have the following slopes:

$$m = \begin{cases} \frac{-w_{grid}}{h}, & x < 0 \\ \frac{w_{grid}}{h}, & x > 0 \end{cases}. \quad (5)$$

Since the grid is rigid, this motion can be achieved if one edge of the grid is driven along a flat surface oriented at an angle θ with respect to the detector surface, while the opposite non-driven edge of the grid is constrained to track along a second flat surface oriented at an angle $\pi - \theta$ where θ is defined as in Equation 6.

$$\theta = \tan^{-1}\left(\frac{w_{grid}}{h}\right). \quad (6)$$

The net motion of the grid is then a combination of translation (shift) and rotation (twist) about the grid midline. The shift-and-twist (ST) reciprocation scheme can be implemented using a single linear positioning stage that is mounted at angle θ to drive the ST motion of the focused grid, with the opposing edge supported by a linear bearing mounted at angle $\pi - \theta$, or vice versa.

Figure 5(a) is a schematic of the grid in its two extreme positions as viewed along the y-axis direction during ST reciprocation. The origin of the coordinate system in the plots of Figure 5 is the location of the midline of the grid surface when the grid and detector are parallel. Arrows show the directions in which the grid edges move during the exposure, with the right (non-driven) edge moving along a straight line at angle θ with respect to the detector surface and the left (driven edge) moving along a line at an angle $\pi - \theta$ with respect to the detector surface. The solid line in Figure 5(a) represents the posterior edge of the grid at its initial position, and the dashed line shows the final orientation of the grid after reciprocation along the inclined surfaces. Figure 5(b) is a graphical example of the trajectory of the right-most corner of a focused grid as viewed along the y-axis for three different motion schemes: conventional translation parallel to the detector surface, translation along the ideal circular trajectory, and translation along lines tangent to the ideal circular trajectory.

In the example of Figure 5, the anti-scatter grid has a width of 32 cm, a focal distance of 80 cm, and a 2.5 cm repeat distance. The dotted plot in Figure 5(b) shows the ideal arc trajectory traced by the grid's right corner when XFS and GFP are always kept aligned as it

is reciprocated. This arc has a radius equal to $\sqrt{h^2 + w_{grid}^2}$, the distance between the GFP and the grid's right-most corner. The solid line with diamonds shows a line that is tangent to the ideal arc at the point where the right corner lies when the grid surface and the x-ray detector surface are parallel. The angle of inclination θ of the tangent line for the given example is 11.3° . For the small 2.5 cm repeat distance, the tangent line-based trajectory approximates well the ideal arc trajectory as seen by the close overlap between these two curves. For comparison, the corner trajectory for conventional grid translation parallel to the detector surface is also shown.

The dual modality breast tomosynthesis (DMT) scanner is an investigational system that is located at the University of Virginia Breast Care Center. In DMT scans, DBT and molecular breast tomosynthesis (MBT) are performed with the breast in a single configuration. The image sets are reconstructed and used in conjunction to create a single three-dimensional (3D) image of the breast with both anatomic and functional information. A copper 2D focused cellular grid using the grid ST reciprocation mechanism described above has been built, installed, and tested on the DMT scanner. In this paper we present measured grid performance, including primary transmission under stationary and reciprocating conditions and severity of GLAs in the projection and DBT images as quantified using the 2D image power spectrum (PS).^{14,15,16} Because grid motion in the ST reciprocation scheme is not

confined to a single plane, all detector elements do not experience exactly the same coverage by the grid septa. As a result, some residual GLAs can be present in the projection images. When used in DBT acquisitions, GLAs from the projection images create artifacts in the reconstructed volume. Also included in this paper are descriptions of two methods for reducing grid line visibility from both projection and reconstructed images post image acquisition. The first method involves correction of GLAs from projection images by flat-field correction.¹⁷ The second method involves frequency space-based filtration of GLAs from the reconstructed image slices.

2. METHODS & MATERIALS

2.1 Image Acquisition

The DMT x-ray scanner consists of the 2923MAM, a CMOS flat panel x-ray detector manufactured by Dexela/Perkin-Elmer (London), and a Varian tube with a RAD-70 tungsten target and a 50 micron rhodium filter. All images were taken with the detector in high-sensitivity (HS) mode.¹⁸ The source-to-detector distance (SID) is 82 cm. The breast is compressed on a support that is a distance of 14 cm above the x-ray detector, while the axis of rotation (AOR) for both x-ray detector and x-ray source is located 4 cm above the breast support.

A focused 2D stack-laminated copper anti-scatter grid prototype manufactured by Mikro Systems, Inc. was used for these experiments. The grid holes are square. Hole size, septal thickness, and grid height are 1 mm, 0.1 mm, and 3.85 mm, respectively. The prototype grid has an active area of 17 cm $\times$ 26 cm and a focal distance of 80 cm. The grid rotation angle is 47° and the repeat distance r_d is 19.8 mm. The VCS13 voice coil motor manufactured by H2W Technologies, Inc., was used for grid reciprocation. The exposure time t_{exp} was 500 ms for all acquired images. The positioning stage was programmed to provide an approximately constant speed over the duration of the exposure, which was calculated from Equation (1).

Image reconstruction was performed on the tomosynthesis images presented here using iterative reconstruction software provided by Dexela/Perkin Elmer, the manufacturer of the x-ray detector. The resulting images have reconstructed slice thicknesses of 1 mm and in-plane voxel dimensions of 150 microns. All image analysis and filtering was done using ImageJ and Matlab.

2.2 Measuring Primary Transmission

The primary transmission of the grid was measured by comparing mean pixel values of grid-in images to images acquired using the same techniques but without the grid prototype in the beam. The stationary grid was positioned at a distance of 80 cm below the x-ray source. It was placed above the detector such that the center of the grid along the chest wall edge and the focal spot projection onto the detector were both aligned. There was no compression paddle or breast support present for these images. Five consecutive images were acquired for grid-in and grid-out conditions with tube parameters of 28 kV and 2.5 mAs. Gain correction

was not performed on the images, but they were corrected for dark current. Next, projection images were obtained with the grid in and reciprocated using ST motion.

To account for any changes in tube output from one exposure to the next, a factor α was calculated as described by Fetterly and Schueler¹⁹ and as shown in Equation (7)

$$\alpha = (\overline{R_{Bkg, grid-out}} / \overline{R_{Bkg, grid-in}}). \quad (7)$$

Since the grid is smaller than the active area of the x-ray detector, the mean pixel value in a region outside of the grid of the grid-in image $\overline{R_{Bkg, grid-out}}$ and the mean pixel value of the same region in the grid-out image $\overline{R_{Bkg, grid-in}}$ were used for the calculation of α .

Nine regions of interest (ROIs) were drawn over the images with dimensions of 300×300 pixels. For a given ROI, Equation (8) was used to calculate primary transmission T_p through the grid,

$$T_p = \alpha \overline{R_{grid-in}} / \overline{R_{grid-out}}, \quad (8)$$

where $\overline{R_{grid-in}}$ is the mean pixel value within a grid-in ROI and $\overline{R_{grid-out}}$ is the mean pixel value within the corresponding grid-out ROI.

2.3 Grid Line Artifact Processing

2.3.1 Acquisition of Phantom Images—For this study, two phantoms were imaged. The first phantom was a 2.5 cm block of acrylic. The second phantom consisted of four 1 cm slabs of the CIRS Model 020 BR3D²⁰ to create a 4 cm phantom simulating a breast composed of approximately 50% glandular and 50% adipose tissue. Features simulating calcifications, masses, and fibers were embedded within one of the four slabs. Two sets of tomosynthesis images were acquired of this phantom. The first set of images was acquired without any form of scatter rejection. The second set of projection images was acquired with the copper prototype grid in the beam. The exposure techniques were set to obtain a total average glandular dose (AGD) of approximately 1.15 mGy. This total dose was divided equally among thirteen 500 ms exposures acquired at $\pm 12^\circ$, $\pm 8^\circ$, $\pm 5^\circ$, $\pm 3^\circ$, $\pm 2^\circ$, $\pm 1^\circ$, and 0° relative to the direction of compression using a step-and-shoot gantry motion scheme. Dark subtraction and flat-field correction were performed on all breast phantom images. All corrected images are of 16-bit unsigned data type.

2.3.2 Generation of Flat-field Correction Images—The first method of removing GLAs processes grid lines out of the projections by flat-field correction. Two sets of flat-field correction images (FFCIs) are acquired, one set with the grid in and a second single one with the grid out. For both grid-in and grid-out FFCIs the FFCI is created by combining seven consecutive exposures of a 2.5 cm acrylic block placed on a 1.2 mm thick carbon fiber breast support below the compression paddle. Each pixel in the final FFCI is set equal to the median pixel value of the corresponding pixels from the seven exposures. A single grid-out

FFCI, generated from the seven consecutive exposures of the acrylic block and obtained with the gantry arm vertical, is used for processing all subsequent grid-out projection images of the breast phantom.

The rotation of the DMT gantry during DBT scanning results in changing gravitational force on the grid, producing slight variations in grid motion from projection to projection. Therefore, multiple grid-in FFCIs are acquired at several gantry arm angles for flat-fielding of the grid-in projections, depending upon the compression angle. Grid-in FFCIs were obtained at each of the following angles for simulation of a craniocaudal (CC) view: $\pm 12^\circ$, $\pm 8^\circ$, $\pm 5^\circ$, and 0° . These seven grid-in FFCIs were used for correction of projections acquired in the ranges between them. Since variations in motion between -3° to $+3^\circ$ were small, it was possible to use the 0° FFCI grid-in to correct the projections taken within this angular range at $\pm 3^\circ$, $\pm 2^\circ$, $\pm 1^\circ$, and 0° . Unlike for a CC view, the positioning stage is configured to always go either with gravity or against gravity for all projection views of a given mediolateral oblique (MLO) compression. For MLO acquisitions, a single FFCI acquired at 0° with respect to the direction of compression was found to be sufficient for correction of all projection images acquired within an angular range of $\pm 12^\circ$ away from the direction of compression.

2.3.3 Flat-field Correction—Three data sets were generated from the two sets of projection images of the breast phantom and the two sets of FFCIs. The first data set consisted of grid-out projections corrected by the grid-out FFCI. To illustrate artifact formation in grid-in images when the grid is not present in the FFCI, the second data set consisted of grid-in projections corrected by the grid-out FFCI. The third data set was created by correcting grid-in images by the grid-in FFCIs.

The presence of periodic grid lines in a projection image can be seen as peaks in its 2D image power spectrum (PS). The PS is defined here as ten times the log of the absolute value squared of the fast-Fourier transform (FFT) of a region of an image $I(x,y)$ as shown in Equation (9):

$$PS(u, v) = 10 \cdot \log_{10}(|FFT(I(x, y))|^2). \quad (9)$$

The PS is a two dimensional function of spatial frequency coordinates u and v , where u is the conjugate of the x direction or the rows of the image, and v is the conjugate of the y direction, or columns of the image. For the PS of a digital image the matrix size of the PS is identical to that of the image or ROI chosen. Here, 512×512 pixel and 256×256 pixel ROIs were chosen for PS analysis. The frequency axes of the PS matrix cover a range from +Nyquist to -Nyquist frequency with the center of the image being the origin ($u = v = 0$). Since all detector elements of the 2923MAM are square ($75 \mu\text{m} \times 75 \mu\text{m}$), the Nyquist frequencies for both axes of the 2D PS are the same and equal to $1/(2p) = 6.67 \text{ mm}^{-1}$ where p is the detector sampling pitch. The frequency bin size of the PS is given by the reciprocal of the sample length, $1/(Np)$, where N is the dimension of the ROI. Therefore, for the 512×512 ROIs, each frequency bin has a width of $1/(512 \times 0.075 \text{ mm}) = 0.026 \text{ mm}^{-1}$.

Locations of grid artifact peaks in the PS are determined by the grid frequency and rotation angle. For example, since the grid prototype has a rotation angle of 47° with respect to one axis of the detector element sampling matrix, the peaks in the PS that correspond to the GLAs are expected to be along lines that are at approximately this same angle with respect to the frequency axes.

The PS for each of the 0° projection views from the breast phantom scans was obtained and they are shown here as examples. The effectiveness of the flat-field correction method was tested by comparing 1D profiles through the 2D PS of the three data sets along the line in frequency space upon which the peaks at the grid's primary frequency and its harmonics lay. The line was determined in the spectra of the second data set, in which the peaks were very conspicuous because they were derived from grid-in projections corrected by grid-out FFCIs.

2.3.4 Filtering of Grid Line Artifacts (GLAs)—Flat-fielding the projection images using the procedure described above ensured that the reconstructed images have minimal GLAs. However, in cases where the residual grid artifacts are unacceptably visible after flat-field correction, additional artifact removal is necessary. Because of their predictable locations, the amplitude of the grid peaks in the 2D PS can be reduced through an automated custom filtering routine that is designed to adapt to the magnitude of the GLAs in each reconstruction slice. First, a 256×256 voxel ROI $R_{0,k}$ containing GLAs is selected for a given slice k of the reconstructed volume. Next, the 2D fast-Fourier transform (FFT) of $R_{0,k}$ is computed. The $PS_{0,k}$ is calculated as shown in Equation (10) for slice k .

$$PS_{0,k} = 10 * \log_{10}(|FFT(R_{0,k})|^2). \quad (10)$$

This PS is then modified by attenuating peaks from $PS_{0,k}$ that were attributed to GLAs as follows. An algorithm was written in Matlab for finding grid peaks along the line on which they were known to lie. Pixels within 8×8 pixel ROIs centered on known peak locations in $PS_{0,k}$ were identified as resulting from grid artifacts when they had values above a threshold of 60% of the 0 frequency peak. This threshold level was chosen because it was the ratio of the highest amplitude grid peak to the 0-frequency peak, on average, in spectra from reconstructed breast phantom and acrylic phantom slices with the most conspicuous GLAs. A modified PS, referred to as $PS_{mod,k}$, was created by setting the pixels that had values above the threshold to unity. $PS_{mod,k}$ was then divided by $PS_{0,k}$ as shown in Equation (11) to create a customized filter F_k for the k^{th} slice of the reconstruction volume.

$$F_k(i, j) = \frac{PS_{mod,k}(i, j)}{PS_{0,k}(i, j)}. \quad (11)$$

In the above equation, i and j are the indices of the rows and columns, respectively, of each of the 2D PS image matrices. The filter has a value of 1 in all frequency bins except those that contained grid artifact peaks in the original slice FFT, where its value is the inverse of the peak values.

Once the filter image was created, it was used to filter out the GLAs of the whole area of the k^{th} slice. In this step, the FFT of the original slice is multiplied element by element by a resized version of F_k whose dimensions match that of the FFT of the original slice. F_k was resized using the Matlab `imresize` function using bicubic interpolation. The inverse Fourier transform (iFFT) is taken of this product to create the filtered slice $I_{F,k}$ as shown in Equation (12).

$$I_{F,k} = iFFT(FFT(I_{0,k}) \times F_k). \quad (12)$$

Finally, the voxel values for the given slice k were scaled to convert the floating point values of the filtered image $I_{F,k}$ to 16-bit unsigned integers in the final processed image $I_{P,k}$ to match the mean value of the original image as seen in Equation (13).

$$I_{P,k} = I_{Filtered,k} (\overline{R_{0,k}} / \overline{R_{F,k}}). \quad (13)$$

where $\overline{R_{0,k}}$ is the mean voxel value within $R_{0,k}$, and $\overline{R_{F,k}}$ is the mean voxel value within the same corresponding ROI location in the filtered image $I_{Filtered,k}$.

This algorithm was executed on both projection data and reconstructed images. Two phantoms were tested. The first was a uniform 2.5 cm thick block of acrylic covering the full field of view (FOV). The second was the BR3D phantom, whose background contains more complicated patterns and structures. To determine whether there was any loss of information as a result of the filtering, profiles were drawn through the 2D PS of the filtered image and compared to that of the grid-out PS and the un-filtered grid-in PS.

3. RESULTS

3.1 Primary Transmission

Figure 6 illustrates the calculated distance between the XFS and the GFP as a function of the distance that the midline of the grid is moved away from its centered position above the detector midline during an x-ray exposure for the three motion schemes presented in Figure 5.

The distance between the XFS and the GFP for the ideal case is shown with circles, that for the ST motion scheme is shown with diamonds, and that of the conventional, planar scheme is shown with squares. In this plot, the center at 0 cm along the x-axis corresponds to the one point during the grid's motion where the grid midline and detector midline overlap. This plot illustrates better alignment of XFS and GFP in the ST motion scheme where the grid edge is driven along the flat surface inclined at angle θ rather than in the translation-only configuration, in which the entire grid is driven parallel to the detector surface.

Figure 7 shows the images acquired for calculation of T_p . Each figure has multiple boxes to illustrate where ROIs were drawn for determining primary transmission. The boxes with solid lines show the ROIs used for calculating mean pixel values $\overline{R_{grid-in}}$ and $\overline{R_{grid-out}}$,

and the box with dashed lines show the ROI drawn for calculating α , which accounts for any variation in tube output. Figure 7(a) is an example of a grid-out projection image. The stationary grid-in image is shown in Figure 7(b), and the reciprocated grid-in image is shown in Figure 7(c).

Table 1 contains the results of the calculations of the primary transmission. Maximum transmission was seen beneath the focal spot projection (third column, second row of each grid-in projection) and minimum transmission was seen through the central box (second row, second column) in both the stationary and reciprocating grid cases. These minima, maxima, and the average values of all nine ROIs are shown in the table.

3.2 Flat-field Corrected Projection Images

Figure 8(a) shows a 512×512 pixel ROI of a projection image of the 2.5 cm uniform acrylic phantom without any scatter rejection. Figure 8(b) shows the same phantom imaged with the reciprocating grid, but corrected with a grid-out FFCI. Obvious GLAs can be seen in the grid-in image of Figure 8(b), which present as peaks in the PS shown in Figure 9(b). These can be confirmed as peaks created by the grid as there is no evidence of these peaks in the grid-out PS of Figure 9(a).

In Figure 9(b) the distance from the center ($u = v = 0 \text{ mm}^{-1}$) to the first grid peak along the 47° angle away from the u axis is 35 pixels, which is at a frequency of approximately $35 \times 0.026 \text{ mm}^{-1} = 0.91 \text{ mm}^{-1}$, consistent with the 1.1 mm pitch of the grid prototype (0.91 mm^{-1} frequency). Peaks at higher frequencies away from the center are higher harmonics of this frequency. Figure 10(c) is the same grid-in projection image of the acrylic phantom as Figure 8(b), but after correction using the 0° grid-in FFCI. Figures 10(a) and 10(b) are the same 0° projection views as Figures 8(a) and 8(b), respectively, shown for comparison. Figures 11(a)–(c) are the spectra of the images in Figures 10(a)–(c), respectively.

Figure 12(a) is the 0° grid-out projection view of the BR3D phantom. Figure 12(b) is the 0° projection view of the grid-in image corrected by the grid-out FFCI, and 12(c) is the same grid-in projection corrected by the 0° grid-in FFCI. A close-up of the ROIs in red boxes drawn in Figures 12(a)–(c) are shown in Figures 13(a)–(c), respectively, and they illustrate the ROIs that were drawn for creating the 2D PS images shown in Figures 14(a)–(c). A 512×512 pixel region was chosen for the image spectra. Figure 14(a) shows the PS of the grid-out image, and Figures 14(b) and 14(c) are the spectra for Figures 13(b) and 13(c), respectively.

Figure 15 contains profiles through the 2D PS images along the line where peaks were found that were a result of grid artifacts seen in the projection image in Figures 12(b) and 13(b). Figure 15(a) is the same 2D PS in Figure 14(b), which shows where the profile was drawn. Figure 15(b) is the profile plot of the grid-out PS. Figures 15(c) and 15(d) are profile plots through the PS of the grid-in 0° projection image corrected by the grid-out FFCI and grid-in FFCI, respectively.

3.3 Filtered Reconstruction Slices

Figure 16 shows close-up views of the ROI that was drawn in the top-most slice (slice 40) of the reconstructed DBT images of the BR3D phantom. This slice also corresponds to the top of the phantom and AOR. Figure 16(a) is a slice from the grid-out DBT image. Figure 16(b) is a slice from the grid-in DBT image where the projection data was corrected using the grid-out FFCI. Figure 16(c) shows a slice from the grid-in reconstruction where the projection data was corrected using the grid-in FFCIs.

Figures 17(a)–(c) are the 2D PS images of Figures 16(a)–(c), respectively. The peaks in the PS that are associated with the GLAs are located along a line that is at approximately a 47° angle with respect to the u-axis of the 2D PS. This angle is consistent with the rotation angle of the cells of the grid prototype with respect to the detector sampling matrix. However, the frequencies at which the GLA peaks are found are slightly different from those found in the projection data since the grid artifacts result from GLA contributions from all projections that go into the reconstruction. Obvious grid lines can be seen in Figure 16(b). To filter out these lines from the reconstructed slice, a modified PS $PS_{mod,40}$ was created from Figure 18(a) (same PS as Figure 17(b)) by setting GLA peak pixel values in this PS to unity as seen in Figure 18(b). Figure 18(c) is the resulting filter image, which was created from $PS_{mod,40}$ (Figure 18(b)) and the original PS $PS_{0,40}$ (Figure 18(a)) using Equation (11). Figure 19(a) is the same ROI shown in Figure 16(b), and Figure 19(b) is the filtered image.

Voxel value profiles extracted along the red lines shown in Figure 20 (a) of the grid-in image with GLAs and 20(b) of the filtered image are plotted in Figure 20(c). The effectiveness of the filtering method can be seen by the fact that the difference of the two profiles in (d) contains primarily modulation at spatial frequencies corresponding approximately to the fundamental grid frequency.

Speck groups, masses, and fibers of varying sizes are embedded within the BR3D phantom. Further evidence that there was no obvious loss in information is the fact that the same features are detectable in both the grid-out and filtered grid-in reconstructions. Figure 21 contains the reconstructed slice containing all objects for the (a) grid-out and (b) filtered grid-in images. The grid-in image has slightly better visibility of the mass margins seen in the circles on the right side of the image slices.

4. DISCUSSION

When averaged over the FOV, primary transmission for the prototype cellular grid under the ST motion scheme is 72.2%. This is a reduction in primary transmission of 2.7% compared to when the grid is stationary and aligned for maximum primary transmission.

Though primary transmission is reduced by a relatively small amount and GLAs are largely eliminated when ST reciprocation is used, some residual GLAs remain in the projection images, as evidenced by peaks in the PS. The first method of grid line removal using flat-field correction of projection images effectively reduces the presence of grid lines, as can be seen by the reduction in amplitude of the grid line peaks in the PS of the projection images. Nevertheless, some GLAs do appear in the reconstructed DBT images from the grid-in flat-

field corrected projections. These artifacts are the result of small differences in grid motion between the grid-in FFCIs and the grid-in projection images.

Other methods for reducing grid line visibility have been described, such as tube output adjustment as described by Gauntt and Barnes.²¹ Much like the flat-field correction method presented here, the tube output must closely and reproducibly be synchronized with grid motion for artifact removal to be consistent. Gauntt and Barnes explain that even with their novel technique, GLA presence is never exactly zero.²¹ Because of the possibility of inconsistency from one acquisition to the next, post-acquisition image filtering was considered. Marayuma and Yamamoto proposed a median filtering method for smoothing grid lines in x-rays films.²² While this reduces image artifacts, it also has the potential for removing breast data.

The correction software described here uses an ROI that is smaller than the full image. One reason for this is that we have found that in order to enable clean automated location and removal of grid artifact peaks from the PS, the ROI size must be chosen small enough (i.e. the frequency bin size must be large enough) to confine grid peaks to a small number of frequency bins. If the entire image is used to create a filter, the grid artifact peaks are not easily detectable in the resulting high resolution PS when they are embedded in the Fourier transform of the breast image. In this work an ROI length of ~256 or 512 pixels (or voxels) permitted reliable peak thresholding. The smaller ROI was used for generation of the filter, which then was applied to the entire area of the original slice.

We found that GLAs are not equally apparent in in all DBT slices or their Fourier transforms. Using a slice-specific frequency space filter rather than a single filter applied to the entire reconstructed volume reduces the likelihood of over-correction in slices with minimal GLAs. Though for the prototype grid tested here some grid peaks in the PS are located at spatial frequencies typical of breast structures, little phantom information was found to be removed as a result of the filtering process. As shown in Figure 20 differences in profiles through pre-filtering and post-filtering images show that the filtering process removes primarily a nearly periodic pattern corresponding to the GLAs, with little evidence of removal of phantom structure.

Since the work reported here with the 17 cm × 26 cm prototype grid, a full-size copper cellular grid with an active area of 23.5 cm × 31 cm and designed for the FOV of 2923MAM detector has been fabricated and installed on the DMT scanner. Table 2 lists the physical parameters of each grid.

In addition to having a larger FOV, the full-size grid has a finer structure than the prototype. The higher grid frequency and slightly different rotation of the full-size grid permit a much smaller repeat distance, and therefore lower acceleration and velocity during ST reciprocation. The smaller pitch, hole size, and height result in a less massive grid even though its FOV is 1.65 times that of the prototype and its grid ratio is higher (4.41 versus 3.0). The lower mass, along with lower acceleration and velocity requirements result in lower load on the voice coil positioning stage and improved mechanical stability. Early evaluation suggests that fewer gantry angle-dependent FFCIs are necessary with the full-size

grid. Characterization of the new grid in terms of scatter rejection, GLA suppression, and impact on image quality and radiation dose, is ongoing.

5. CONCLUSIONS

The combined translation and rotation grid reciprocation method described here can provide efficient removal of scatter with minimal residual grid line artifacts. A combination of projection image flat-fielding and when needed, post reconstruction frequency space filtration is effective in suppressing residual GLAs. The ST motion approach to grid reciprocation is designed to minimize cutoff of primary radiation, thereby reducing the radiation dose penalty associated with simple lateral grid translation. Combined with the better scatter rejection provided by 2D grids compared to 1D grids,^{3,8} DBT image quality can potentially be improved without a large dose penalty. Analysis of the impact, under conditions of constant radiation dose, of scatter removal and primary photon loss on image quality for the prototype grid similar to the one described here using ST reciprocation are described in a separate paper [Effects on Image Quality of a 2D Anti-scatter Grid in X-ray Digital Breast Tomosynthesis (DBT): Initial Experience using the Dual Modality Breast Tomosynthesis (DMT) Scanner. *Medical Physics* 2016; 43(4):1720-1736. PMID: PMC4798991

Acknowledgments

The authors thank Mikro Systems, Inc (Charlottesville VA) for their invaluable efforts in the development of the prototype anti-scatter grid. This work was partly supported by the National Institutes of Health (NIH) (R21 EB002321, R01 CA149130). Further development and clinical evaluation of DMT is being funded by the NIH (R01 CA149130) and the Komen Foundation (KG100479).

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Figure 1.

Schematic of how anti-scatter grids are employed in FFDM. Walls of the grid block a large portion of scattered rays that are generated within the breast while primary rays are transmitted through the grid spaces. The height of the walls, the material used for building the grid, and thickness and spacing of the walls determine grid performance.

Figure 2. Schematic of a rotated cellular grid illustrating the distance in the direction of translation over which the grid pattern repeats. The repeat distance is a function of integers m and n and the grid pitch p , which are related to the rotation angle of the grid φ .

Figure 3.

Example of the attenuation by a grid with a rotation angle of $\phi = 45^\circ$ (a) mapping of the attenuation by the grid walls where non-zero (white) pixel values show presence of attenuating material and zero (black) pixel values show air interspaces. (b) plot of the sum of pixel values along the rows of the image in (a) versus the row index. The sum for each row indicates the attenuation experienced by a detector element in that row during grid reciprocation. Sums were taken over a length of ten repeat distances where $r_d = \sqrt{2}p$. The sharp valleys in (b) correspond to rows where wall intersections are aligned along the direction of reciprocation.

Figure 4.

Example of the attenuation of a grid with a rotation angle of $\phi = 48.4^\circ$ (a) mapping of the attenuation by the grid walls where non-zero pixel values show attenuating material and zero pixel values show air interspaces. (b) Plot of the sum of pixel values along the rows of the image in (a) versus the row index over a length of one repeat distance where $r_d = 12p$ ($m = 9$; $n = 8$).

Figure 5.

(a) Motion of a grid with one edge driven along an incline at angle $\pi - \theta$ with respect to the detector surface and the other edge is constrained to move along a line angled at θ , where θ is determined by Equation (6). The solid line represents the posterior edge of the grid at the beginning of the exposure. The dashed line shows the final orientation of the grid at the end of the exposure. The vertical scale of the drawing is intentionally different than that of the horizontal scale in order to better visualize the vertical displacement of the grid corners. (b) Trajectory of right-most corner of grid represented by the diamond in (a) as viewed from patient's perspective for three different motion schemes. Vertical position is with respect to the z-coordinate of the grid surface when it is parallel to the detector surface. The circular symbols show the corner coordinates for the ideal circular trajectory of the grid such that the x-ray focal spot (XFS) and grid focal spot (GFP) are always exactly aligned. The solid line with diamonds shows travel along a line that is tangent to the ideal corner trajectory when the grid surface is parallel to the detector surface. The conventional trajectory with lateral translation parallel to the detector surface only is shown with square symbols along a solid line.

Figure 6.

Calculated distance between XFS and GFP as a function of the lateral distance that the center of the grid's posterior edge is moved away from the detector's midline during an x-ray exposure. The distance between XFS and GFP for ST reciprocation is shown with diamonds, and that of the translational-only scheme is shown with squares. The ideal circular trajectory is shown with circles. For maximum transmission of primary x-rays the distance between XFS and GFP should ideally be 0 over the entire x-ray exposure.

Figure 7.

Projection images acquired with (a) nothing in the x-ray beam, (b) with the stationary grid prototype aligned with the XFS for maximum primary transmission, and (c) with the grid prototype moving with the ST motion scheme. ROIs were drawn for determining primary transmission through nine different 300×300 pixel ROIs (solid squares). The background ROI whose mean pixel value was used for calculating a in Equation (7) is drawn using dashed lines to the left of each projection.

Figure 8.

Close-up of 0° projection view of acrylic phantom (a) without the grid and (b) with the reciprocating grid, but gain corrected with a grid-out FFCI. The grid was reciprocated in a direction that goes from the bottom to the top of the image.

Figure 9.

Close-up view of power spectra of the images shown in Figure 8 for the (a) grid-out and (b) grid-in projection. PS images have the same 512×512 dimensions as the ROIs of Figure 8

Figure 10.

Close-up of 0° projection view of acrylic phantom (a) without the grid, (b) with the reciprocating grid corrected by the grid-out FFCI, and (c) with the reciprocating grid corrected by the grid-in FFCI. The direction of grid motion is from the top towards the bottom of the image.

Figure 11.

Close-up view of power spectra of the images shown in Figure 10 for the (a) grid-out and (b) with the reciprocating grid corrected by the grid-out FFCI, and (c) with the reciprocating grid corrected by the grid-in FFCI.

Figure 12.

Zero degree projection image of breast phantom. (a) Grid-out projection corrected by grid-out FFCI. (b) Grid-in projection image corrected by grid-out FFCI (c) Grid-in projection corrected by grid-in FFCI.

Figure 13.

Close-up of ROIs shown in Figure 8. (a) Grid-out projection corrected by grid-out FFCI. (b) Grid-in projection image corrected by grid-out FFCI (c) Grid-in projection corrected by grid-in FFCI.

Figure 14.

2D PS images of the 0° (a) grid-out projection corrected by grid-out FFCI, (b) grid-in projection image corrected by grid-out FFCI (c) grid-in projection corrected by grid-in FFCI.

Figure 15.

(a) shows the PS of the grid-in 0° projection image shown in Figure 13(b) over which a line is drawn along the grid peaks to show where a profile was drawn for (b) the grid-out PS image in Figure 13(a), (c) the grid-in image in Figure 13(b), and (d) the grid-in image in Figure 13(c).

Figure 16.

Close-up of slice 40 (slice where the worst artifacts are seen in the grid-in images).

Reconstruction slices from (a) grid-out projection data corrected by grid-out FFCI (b) grid-in projection image data corrected by grid-out FFCI, and (c) grid-in projection data corrected by grid-in FFCIs.

Figure 17.

2D PS images of the image slices shown in Figure 16. (a) Reconstruction slice from grid-out projection data corrected by grid-out FFCI. (b) Reconstruction slice from grid-in projection data corrected by grid-out FFCI. (c) Reconstruction slice from grid-in projections corrected by grid-in FFCIs.

Figure 18.

(a) PS_0 of the of the grid-in image in Figure 16(b). (b) Modified $PS_{mod,40}$ where grid peaks have been set to unity. (c) Custom filter that was produced by dividing (b) by (a). The dark spots in (c) correspond to the peak locations and have fractional pixel values while the surrounding pixels are equal to 1.

Figure 19.

(a) Close-up view of grid-in image seen in Figure 16(b), which has been filtered using Figure 18(c), the result of which is shown in (b).

Figure 20.

Close-up of slice from grid-in DBT scan where projection data were corrected by grid-out FFCI (a) before and (b) after spectral filtering. (c) Profiles along the lines shown in (a) and (b) are plotted in (c), and their difference is plotted in (d).

Figure 21.

DBT slice containing objects embedded in BR3D phantom (a) without the grid and (b) with the grid, correcting the projections using the grid-out FFCI, and filtering the slice. Speck groups, fibers, and masses that are visible in the grid-out image are circled by solid, dashed, and dotted lines, respectively, with the corresponding findings in the filtered grid-in image.

Table 1

Primary Transmission through Select ROIs of Grid Prototype (Average of five trials)

	Center of FOV (Minimum)	Beneath Focal Spot Projection (Maximum)	Average of all 9 ROIs
Stationary Grid	72.0%	75.1%	74.2%
Moving Grid	70.1%	73.9%	72.2%

Table 2

Comparison of prototype and full-size DBT antiscatter grids for the DMT scanner

	Height [mm]	Hole dimension [mm]	Septum thickness [mm]	Pitch [mm]	Rotation angle [deg]	Repeat distance [mm]	Focal distance [cm]	Approximate mass [g]
Prototype grid	3.85	1.0	0.1	1.1	47.29	19.46	80.0	570
Full-size grid	3.0	0.68	0.058	0.738	48.37	8.89	77.5	415