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Adaptive SPECT imaging with crossed-slit apertures

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Abstract

Preclinical single-photon emission computed tomography (SPECT) is an essential tool for studying the progression, response to treatment, and physiological changes in small animal models of human disease. The wide range of imaging applications is often limited by the static design of many preclinical SPECT systems. We have developed a prototype imaging system that replaces the standard static pinhole aperture with two sets of movable, keel-edged copper-tungsten blades configured as crossed (skewed) slits. These apertures can be positioned independently between the object and detector, producing a continuum of imaging configurations in which the axial and transaxial magnifications are not constrained to be equal. We incorporated a megapixel silicon double-sided strip detector to permit ultrahigh-resolution imaging. We describe the configuration of the adjustable slit aperture imaging system and discuss its application toward adaptive imaging, and reconstruction techniques using an accurate imaging forward model, a novel geometric calibration technique, and a GPU-based ultra-high-resolution reconstruction code.

Keywords

adaptive; adjustable; SPECT; crossed slit; silicon; double-sided strip detector; ¹²⁵I

1. INTRODUCTION

The design of SPECT systems requires careful consideration of performance metrics including resolution, sensitivity, field of view, and acquisition time. Unfortunately, these metrics are often inversely related, so that optimizing one aspect of the design will, in turn, limit others. As the number of preclinical SPECT applications increases, it becomes less likely that a single imaging configuration will be sufficient to adequately perform the desired set of imaging tasks. Current imaging system manufacturers must weigh the demand for each set of tasks with a range of possible design configurations to determine the most marketable imagers, but the limited audience for such devices invariably dictates that some imaging demands are not met. In recent years, interest has grown in the development of adaptive tomographic imaging systems to extend the imaging capabilities of SPECT systems.

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The intent of adaptive imaging is to select the imaging configuration that maximizes the performance of a detection or estimation task. For instance, a dynamic uptake study might benefit from larger pinholes and the resulting increase in sensitivity, while stem-cell studies where high resolution is essential may be better suited to imaging with a small pinhole and high magnification. Barrett *et al.* described a comprehensive approach to adaptive imaging that combines information from a preliminary scout scan with object and system statistics to select the imaging configuration that will provide the best information for the specific task.¹

This research explores the technical challenges of highly adaptive SPECT systems and how these challenges can be overcome through the use of an accurate geometric forward model and on-the-fly calculations with graphical processing units (GPUs). We particularly focus on the development of a system in which the full detector area can be used to best exploit its space-bandwidth product to collect information about the area of interest across a range of imaging configurations. The high-resolution detector and variable set of slit widths, magnifications, and fields of view present a significant calibration challenge. A precise representation of the system matrix is essential to perform high quality statistical image reconstruction, but direct measurement and storage of such a large matrix across a continuum of system settings is not a feasible approach. We discuss a simple calibration method from which we extract parameters describing the system's geometry, and a forward model that links these parameters to the system matrix. This method allows us to generate elements of the system matrix on the fly during reconstruction on a GPU, even as the overall system configuration changes.

2. IMAGING CONCEPT AND SYSTEM DESIGN

While clinical SPECT systems utilize parallel-hole collimators to form projection images, current preclinical SPECT imagers rely on imaging through pinhole apertures. While these apertures can be made with high precision, a drawback is often the radial symmetry of the magnification; the subject being imaged is rarely square, and much of the active area of the detector is lost to imaging the empty space around the object. The pinhole also limits the ability to adequately sample the object at the top and bottom of a cylindrical field of view.² Ideally, we would like to be able to magnify the object such that the majority of the detector is filled by the object of interest and increase the sampling capabilities across the field of view.

This type of anamorphic magnification can be accomplished by replacing a pinhole in a single plane with a pair of crossed slits, which decouples the magnification into separate and independent axial and transaxial elements.³ Separating the vertical and horizontal slits results in an anamorphic projection of the object onto the detector face, and the magnifications in the x and y directions are controlled by adjusting the relative distances between the object, slits, and detector. The position of the vertical slit with respect to the object and detector determines the transaxial (x) magnification; similarly, the position of the horizontal slit specifies the axial (y) magnification. When the vertical slit is closer to the object than the horizontal slit, the configuration produces images with increased transaxial resolution, and resulting reconstructions have been shown to possess fewer axial artifacts than reconstructions from pinhole-based systems.^{4, 5}

2.1 Crossed-slit design

The high-precision slits developed for this system consist of two pairs of five-millimeter-thick machinable 20% copper/ 80% tungsten blades. The standard double-keel edge design was discarded in favor of a single 30-degree keel edge and 280- μm bevel as shown in the inset in Figure 1a. This configuration allows us to position the horizontal and vertical slits very close to each other to obtain near-isotropic magnifications if desired. These blades are mounted to aluminum plates; one blade is fixed and the other is permitted to slide on brass ball bearings. Spring pressure between the blade plates maintains blade parallelism, and the slit width is determined by linear actuators mounted on the side of the slit frame. A close-up view of the inner slit construction is provided in Figure 1.

2.2 Silicon double-sided strip detector

While the selection of system characteristics such as pinhole diameter, magnification, and field of view define the performance range of SPECT systems, the detector resolution is the ultimate determining factor in the types of systems that can be produced with these components. Improved detector resolution not only contributes to increased spatial resolution in reconstructed images, but also *reduces* the number of detected events necessary to produce visually similar images compared with lower-resolution detectors. This was first performed in simulation by Muehllehner,⁶ and was later confirmed by Rolland and Barrett.^{7, 8} We used a high-resolution silicon double-sided strip detector in our system to incorporate these benefits into the system performance.

The bulk silicon crystal used in the DSSD was fabricated at SINTEF in Norway,⁹ and the full detector assembly was manufactured at Gamma-Medica IDEAS, also in Norway. The detector is composed of a 60 mm $\times$ 60 mm, 1 mm slab of crystalline silicon with 1024 20- μm -wide conducting strips on each side at a 59- μm pitch. The strips on one side are oriented orthogonally with respect to the strips on the opposite side of the detector, resulting in 1,048,576 “virtual” pixels; this detector is therefore the first true megapixel gamma-ray detector to be used in small-animal SPECT imaging. The silicon DSSDs were designed to detect photons between 10–60 keV. A photo of the detector module is shown in Figure 2a.

2.3 Geometric model

A small circular pinhole images an object at a point (x, y, z) in object space to a point (u, v) on the detector in image space. The relationship between these points can be expressed as

$$(u, v) = \left(-\frac{q}{p-z}x, -\frac{q}{p-z}y \right), \quad (1)$$

where p is the distance between the center of the field of view and the aperture, and q is the distance between the aperture and detector. In this imaging configuration, the magnification factor $m = -\frac{q}{p-z}$ is constant in the x and y directions.

In an imager using a crossed-slit collimator, mapping (x, y, z) to (u, v) occurs with separate axial and transaxial magnifications:

$$(u, v) = \left(-\frac{q_x}{p_x - z}x, -\frac{q_y}{p_y - z}y \right) \quad (2)$$

In the more general case where the detector is not located in a plane parallel to the slits, we calculate the intersection of the central ray with a tilted detector plane (x' , y' , z') and rotate the resulting point back to the normal position to determine the absolute pixel position (u , v):

$$\begin{bmatrix} u \\ v \\ s_d \end{bmatrix} = R_{-\psi} R_{-\phi} R_{-\theta} \begin{bmatrix} x' \\ y' \\ z' \end{bmatrix}, \quad (3)$$

where s_d is the distance in the z -direction between the center of rotation of the object and the center of the detector, and $R_{\psi} R_{\phi} R_{\theta}$ are the three-dimensional rotation matrices given by:

$$R_{\psi} R_{\phi} R_{\theta} = \begin{bmatrix} \cos\psi & 0 & -\sin\psi \\ 0 & 1 & 0 \\ \sin\psi & 0 & \cos\psi \end{bmatrix} \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\phi & \sin\phi \\ 0 & -\sin\phi & \cos\phi \end{bmatrix} \begin{bmatrix} \cos\theta & \sin\theta & 0 \\ -\sin\theta & \cos\theta & 0 \\ 0 & 0 & 1 \end{bmatrix}. \quad (4)$$

This model assumes that the slits are oriented orthogonally with respect to each other, and that the object's axis of rotation is parallel to the vertical slit. It should also be noted that the point (u , v) as defined above has not been converted from spatial coordinates and requires a further mapping to obtain the actual pixel indices.

2.4 Design specifications

In order to create a flexible imaging environment, we designed the system to allow adjustment of both slit and object locations. Each slit is mounted on an aluminum extrusion frame, and the detector is mounted in its opaque custom housing on a separate aluminum frame. Two linear stages, one rotation stage, and a motorized lab jack control the object position. The slits are mounted to a dual-lead-screw stage that determines the separation between the slits; this stage is mounted to an additional linear stage that positions the slit assembly between the source and detector. The linear stages can be positioned in steps as low as 2.5 μm ; two additional linear actuators control the slit widths with 50-nm stepping resolution.

The positions and length of the linear stages have been designed to accommodate imaging situations ranging from full-body imaging of a small mouse (30 mm wide $\times$ 70 mm long) to high-resolution mouse brain imaging (12 mm $\times$ 15 mm). A photograph of the assembled system in a high-magnification configuration is provided in Figure 2b. Tables 1a and 1b provide a list of the stage ranges, the parameters they influence, and the available range of values for the adjustable geometric parameters. The system limits are determined by the mechanical design of the slits: the vertical slit cannot be closer than 40 mm to the object, and the horizontal slit cannot be closer than 50 mm to the detector.

2.5 Geometric resolution

The geometric resolution of a circular pinhole is the result of convolution between the aperture resolution and the intrinsic resolution of the detector, scaled by the system magnification. This is generally approximated as a root sum of squares:¹⁰

$$R_o \approx [R_g^2 + (p/q)^2 R_i^2]^{1/2}, \quad (5)$$

where R_i is the intrinsic detector resolution. The geometric resolution of the aperture R_g is given by

$$R_g = a_e(p+q)/q, \quad (6)$$

and

$$a_e = [a(a + 2\mu^{-1}\tan(\alpha/2))]^{1/2} \quad (7)$$

where p and q are as defined in the previous section. The parameter a_e is the effective diameter of the pinhole, which accounts for penetration through a double-keel edge pinhole with cone angle α . In contrast, the geometric resolution given by the crossed-slit configuration used in this system must be calculated separately in the x and y directions:

$$R_{ox} \approx [R_{gx}^2 + (p_x/q_x)^2 R_i^2]^{1/2} \quad (8a)$$

$$R_{oy} \approx [R_{gy}^2 + (p_y/q_y)^2 R_i^2]^{1/2}, \quad (8b)$$

where

$$R_{gx} = \frac{a_{ex}(p_x + q_x)}{q_x} \quad (9a)$$

$$R_{gy} = \frac{a_{ey}(p_y + q_y)}{q_y}, \quad (9b)$$

and

$$a_{ex} = [a_x(a_x + 2\mu^{-1}\tan(\alpha/2))]^{1/2} \quad (10a)$$

$$a_{ey} = [a_y(a_y + 2\mu^{-1}\tan(\alpha/2))]^{1/2}. \quad (10b)$$

It should be noted that the expressions for a_{ex} and a_{ey} are approximations to the effective slit widths in the x and y directions; the half-keel-edge blades increase the effective width of the aperture. Figure 3 demonstrates the projection images acquired for different slit widths, highlighting the high intrinsic resolution of the detector.

2.6 Geometric sensitivity

The sensitivity is a useful figure of merit for SPECT systems, as it is a measure of the amount of radiation collected by the imaging aperture that, in turn, specifies the amount of time an object must be imaged to acquire a desired number of detected gamma-ray events. The geometric sensitivity for a circular aperture can be expressed as a function of aperture width and the separation between the object and aperture: ¹¹

$$S \propto \frac{a^2}{16p^2} \cos^3 \theta. \quad (11)$$

Here, a is the diameter of the pinhole, and θ represents the angle formed between a point in the object, the center of the aperture, and the central imaging axis.

In a similar fashion, Ayan et. al. derived the sensitivity for ideal crossed-slit collimators: ¹²

$$S_{\text{crossed slit}} = \frac{a_{ex} a_{ey}}{4\pi p_x p_y} \frac{\cos^3 \theta}{\left[1 + \left[\left(\frac{p_y}{p_x} \right)^2 - 1 \right] \sin^2 \theta \sin^2 \phi \right]^{3/2}} \quad (12)$$

Here, the angle θ continues to refer to the off-axis angle formed by the object, aperture, and central imaging axis, while ϕ is the angle a projection of the point makes with respect to the xy plane. Sensitivity plots for several imaging configurations are provided in Figure 4.

3. GEOMETRIC CALIBRATION

Reconstructions based on projection data determine the estimated maximum-likelihood object that had the highest probability of producing measured data, according to the imaging equation

$$\mathbf{g} = \mathbf{H}\mathbf{f} + \mathbf{n}, \quad (13)$$

where $\mathbf{f}$ is an $N \times 1$ vector of voxels in the object, $\mathbf{g}$ is an $M \times 1$ vector consisting of a concatenation of all pixels acquired of $\mathbf{f}$, and $\mathbf{n}$ represents noise in the measurement process. The system matrix $\mathbf{H}$ is an $M \times N$ matrix that maps a region of interest in object space into the projected data of image space, where each element $h_{m,n}$ represents the probability of a point source in voxel n being detected in pixel m . It is important to note that the dimensions of $\mathbf{H}$ increase linearly with increased resolution in both object space *and* image space. Our system uses 60 one-megapixel projections to produce a high-resolution reconstruction, which, for an object space of $101 \times 101 \times 101$ voxels, requires a system matrix $\mathbf{H}$ consisting of over 64 trillion elements.

A single-precision floating-point matrix representing 101^3 voxels viewed over 60 projections would require 235 TB of storage space. We can reduce the needed storage space by storing the sparse data, where only the relevant portion of the detector area on which the point source is projected (221 GB), or simplify even further by storing each nominally rectangular projection image as a set of six coefficients that describe the center location, height, width, amplitude, and rotation angle (1.38 GB). However, if we characterize the system as a whole by estimating the 11 geometric parameters that define the relative positions of the object, slits, and detector, and calculate the elements of $\mathbf{H}$ as needed, we need only 0.05 kB of storage space. These storage required are tabulated in Table 2. The geometric forward model described in Section 2.3 can be leveraged to compute the geometric parameters from a series of calibration measurements.

3.1 Calibration data acquisition and estimation

We have developed a geometric calibration method that uses one of several brands of therapeutic brachytherapy rods, which are nominally 5 mm long with an 800- μm outer diameter. These rods are miniature sealed sources that minimize radioactive contamination risks during extended calibration procedures. Five rods are inserted into bores in a rapid-prototyped geometric calibration phantom, which allows us to reproducibly position the rods in height and radial spacing to cover the desired FOV. The calibration phantom allows us to define a cylindrical FOV up to 35 mm in height with a 12-mm maximum radius.

We acquire five sets of 60 projection images corresponding to a six-degree rotation step size, moving the slit translation stage five millimeters towards the detector after each set of 60 projections. Measuring at multiple magnifications with the same source-to-detector distance s_d provides supplemental information necessary to estimate the set of desired geometric parameters when analyzing the acquired data. Acquisition times are determined at the beginning of the calibration procedure such that each brachytherapy-rod projection contains at least 4500 events. Since the brachytherapy rods can by no means be considered point sources, we process each projection image by estimating the region corresponding to each rod and calculating its center position in detector pixel units.

If we assume normal (Gaussian) statistics, the maximum-likelihood contracting-grid method minimizes the differences between a measured set of centroids $(u_{m,i}, v_{m,i})$ and predicted forward projections $(u_{p,i}, v_{p,i})$ of the brachytherapy rod centers. The smallest squared-distance sum acquired from a grid of sampled parameter values indicates the starting parameter values for the next iteration of the contracting grid.¹³ Each iteration samples new parameters from a smaller region of the subspace surrounding the best result from the previous iteration. Geometric calibration data and the resulting estimates are shown in Figure 5. The rms distance between the sets of measured and predicted centroids after 15 iterations is 149.1 μm , or approximately 2.5 pixels. In order to accurately estimate the system parameters, we must also estimate a number of “nuisance” parameters to account for offsets in the brachytherapy phantom. The full list of estimated parameters are listed in Table 3. Iterating over the subsets $\{(p_x, p_y, q_x, q_y), (m_x, m_y, e_x, e_y), (\theta, \phi, \psi), (\delta_x, \delta_z), (o_1, o_2, o_4, o_5), (p, \alpha, \beta)\}$ provided the fastest convergence to an estimated set of geometric parameters.¹⁴

3.1.1 PSF model—The rectangular aperture defined by the crossed slits, when illuminated by a point source, will produce a rectangular irradiance pattern on the detector. This geometric point spread function is easily calculated for any point within the field of view of the system by forward projecting four rays through the effective corners of the slit aperture to define the corners of the projected rectangle. This is accomplished by replacing the slit offset parameters m_x and m_y by $m_x \pm \frac{\varepsilon_x}{2}$ and $m_y \pm \frac{\varepsilon_y}{2}$ and forward-projecting the ray using the expressions presented in Equations 2 and 3. The lines connecting the corner points define the geometric boundary of the PSF function. The image of the PSF can be generated by evaluating each pixel i in the region of the PSF to determine the fractional area A_i that is contained within the geometric boundary. The PSF value for each pixel is proportional to the fractional area scaled by factors that account for angle-dependent detector thickness and radiometric falloff across the field of view:

$$\text{PSF}_i \propto \frac{(1 - e^{-d_{Si} \mu_{Si} \sec \theta}) \cos \theta}{4\pi R^2} A_i \quad (14)$$

where d_{Si} is the thickness of the silicon detector, μ_{Si} is the attenuation coefficient of silicon, and where θ and R are shown schematically in Figure 6. Since the dimensions of the pixels are very small compared to R , we assume that the irradiance is constant across a given pixel.

In a procedure we developed, once the relevant intersection points $\mathbf{p}_i = [x_i \ y_i]^\top$ have been calculated, they are sorted in counterclockwise order. The list defines a closed ring of points where $\mathbf{p}_N = \mathbf{p}_0$. The fractional area enclosed by the N vertices is then calculated with Gauss' area formula:

$$A_i = \frac{1}{2} \sum_{n=0}^{N-1} \det \begin{bmatrix} \mathbf{p}_n & \mathbf{p}_{n+1} \end{bmatrix}, \quad (15)$$

where $\det \begin{bmatrix} \mathbf{p}_n & \mathbf{p}_{n+1} \end{bmatrix}$ refers to the determinant of a matrix containing the elements of vectors $\mathbf{p}_n$ and $\mathbf{p}_{n+1}$. These calculations compute the exact fractional pixel area, but the result comes at a high computational cost. We have determined that it is possible to substitute several approximations into the exact forward model that can significantly improve the calculation speed.

3.2 System matrix generation

We previously discussed the massive memory demands of storing even a sparse form of the system matrix $\mathbf{H}$ for this high-resolution system.¹⁴ However, the parallel processing power available in commercial GPU hardware allows us to bypass these constraints: we can use the GPU to calculate (and discard) elements of the system matrix as needed during the reconstruction process.

4. MAXIMUM-LIKELIHOOD ESTIMATION METHODS

We chose to use maximum-likelihood expectation-maximization methods (MLEM) to reconstruct the measured data. MLEM reconstruction techniques were first described for emission tomography by Shepp and Vardi in 1982,¹⁵ although Richardson¹⁶ and Lucy¹⁷ had

previously developed similar techniques for astronomical imaging. The MLEM algorithm can be modified to reduce computation time.

4.1 MLEM

The MLEM update equation, as described by Shepp and Vardi, is:

$$\hat{f}_n^{(k+1)} = \frac{\hat{f}_n^{(k)}}{\sum_{m'=1}^M h_{m'n}} \sum_{m=1}^M h_{mn} \frac{g_m}{\sum_{n'=1}^N h_{mn'} \hat{f}_{n'}^{(k)}} \quad (16)$$

The equation, when examined piecewise, provides a straightforward map for implementing the reconstruction routine. An initial estimate of the voxel data ($\mathbf{f}^{(k)}$) is forward-projected through the system described by $\mathbf{H}$ to generate $\hat{\mathbf{g}}$, where

$$\hat{g}_m = \sum_{n'=1}^N h_{mn'} \hat{f}_{n'}^{(k)}. \quad (17)$$

We can consider $\hat{\mathbf{g}}$ as the set of noiseless projection images that would have been acquired if an object described by $\hat{\mathbf{f}}$ was imaged in the SPECT system. The measured data $\mathbf{g}$ is then scaled by $\hat{\mathbf{g}}$, and the resulting set of values is back-projected through the imaging system. The backprojection is scaled by the sensitivity of the system, and the resulting value multiplies the estimate of $\mathbf{f}^{(k)}$ to generate the next estimate $\mathbf{f}^{(k+1)}$.

4.2 OSEM

Hudson and Larkin demonstrated¹⁸ that MLEM reconstructions will converge faster if the routine is modified to accommodate ordered subsets (OS) of the measured data:

$$\hat{f}_n^{(k+1)} = \frac{\hat{f}_n^{(k)}}{\sum_{m' \in S_k} h_{m'n}} \sum_{m \in S_k} h_{mn} \frac{g_m}{\sum_{n'=1}^N h_{mn'} \hat{f}_{n'}^{(k)}}, \quad (18)$$

where S_k denotes the subset of pixels corresponding to the projections that are included in iteration k . The OSEM algorithm converges most quickly when the measured data in each subset samples equally from the entire projection space. For a reconstruction performed with L OSEM subsets, the output often converges to the MLEM reconstruction in $1/L$ times as many passes through the full data.

It is worthwhile to note that ML estimation is employed to first establish the system matrix $\mathbf{H}$ during the calibration process, where we use knowledge of the calibration object $\mathbf{f}$ to estimate

$$\hat{\mathbf{H}} = \underset{\mathbf{H}}{\operatorname{argmax}} \operatorname{pr}(\mathbf{g}|\mathbf{H}, \mathbf{f}). \quad (19)$$

We then leverage the knowledge of $\mathbf{H}$ during the reconstruction process, where we calculate

$$\hat{\mathbf{f}} = \underset{\mathbf{f}}{\operatorname{argmax}} \operatorname{pr}(\mathbf{g}|\mathbf{H}, \mathbf{f}) \quad (20)$$

for the unknown object $\mathbf{f}$.

5. GPU-BASED RECONSTRUCTION

We are able to perform the large number of computations required for these reconstructions by implementing the reconstruction algorithm on GPUs. We performed all GPU computations using a single NVIDIA GeForce GTX 580 desktop graphics card, which features 512 computing cores and 3GB of onboard memory.

5.1 Multithreaded forward model

The EM algorithms of Section 4 were implemented with a series of three major processing kernels: one that performs the forward projection operation, one that back-projects the comparison between the computed and measured forward projection, and a third that precomputes the sensitivity before the iterative reconstruction begins. A simple fourth kernel performs an element-by-element division, scaling the backprojection by the sensitivity at the end of each iteration. The sensitivity is precomputed before the iterative reconstruction using a variation of the backprojection kernel, assuming a uniform voxel volume with unit activity in each voxel.

At the core, all three main kernels rely on applying the forward model to each voxel to calculate the elements of $\mathbf{H}$. The steps for calculating a single PSF using the parallel processing capabilities of the GPU are as follows:

System matrix generation: For each voxel j (block level):

1. Forward-project the voxel center through the center of the slits to determine the center of the PSF.
2. Forward-project the voxel center along slit edges to determine the corners that define the PSF boundaries.
3. Identify the geometric center of each pixel and calculate the radiometry and absorption corresponding to this point using Equation 14.

For each pixel i in the PSF (thread level):

1. Evaluate the fraction of the pixel area that is illuminated by the PSF. To reduce computation time, we approximate the fractional area by summing the number of subpixels that have centers contained within the PSF boundary.

2. Multiply that value by the radiometry and detection efficiency scaling factor calculated at the beginning of the block execution to generate one $\mathbf{H}$ matrix element h_{ij} .

A graphical representation of this process is presented in Figure 7.

5.2 Brachytherapy rod phantom reconstruction

Figure 8 shows cross-sectional slices through a reconstruction of a radial brachytherapy rod phantom imaged through 250- μm slits. The phantom contains 11 IsoAid ^{125}I brachytherapy seeds. The $255 \times 255 \times 255$ -voxel reconstruction was performed with 60 15-minute projections acquired with 250- μm slits using five iterations of OSEM with five subsets. The reconstructed field of view is a cube 15 mm on a side. The non-radioactive center of each ^{125}I -coated rod is clearly visible. We have previously estimated that the resolution of this reconstruction is $\sim 225 \mu\text{m}$ in the axial direction and $\sim 175 \mu\text{m}$ in the transaxial direction.²⁶

5.3 Reconstructing data from adaptive acquisitions

The ability to generate the system matrix on the fly also provides the freedom to incorporate object-dependent adaptive techniques in the object reconstruction.

5.3.1 Variable fields of view—We are not limited to reconstructing the object from a set of acquisitions with identical fields of view. As long as the object and slit positions corresponding to each image acquisition are recorded, we have all the information required to generate each projection's contribution to the system matrix. In this way we can perform reconstructions where the field of view is not constant over the set of projections. Information from the initial scout scan can also be incorporated into the reconstructions to minimize artifacts occurring due to activity outside the field of view of the primary area of interest.

5.3.2 Variable voxel size—The total reconstruction time is dependent on the number of voxels in the reconstructed object. Currently, we reconstruct each object using a fixed voxel volume across the field of view, but this does not have to be the case. It stands to reason that if we can define an area of interest about which we perform an adaptive scan, we can also assign regions of the volume that are reconstructed at specified resolutions. The ability to generate the system matrix on the fly enables us to create reconstructions with varying voxel sizes, for example, using the technique described by Meng and Li.²⁷ The current reconstruction method must be modified to include an additional array that specifies the size of each voxel in $\mathbf{f}$ to calculate the proper activity per voxel volume.

5.4 Kernel execution times

The EM execution time varies as a function of the number of voxels, projections, subpixels, and PSF grid size specified for a given reconstruction. Table 4 lists the execution time for the three computation-intensive CUDA kernels in the reconstruction code for a variety of voxels and subpixel numbers. The execution time per projection as measured for

reconstructions with 101^3 voxels and one subpixel are approximately the same as those reported by Miller.²³

While increasing the number of subpixels causes an increase in execution time, it is clear that the execution time of each kernel call is most dependent on the overall number of voxels in the reconstruction volume.

When multiple data sets require processing, it is not always advantageous to produce reconstructions with the highest-possible resolution. The system variability must also be extended to the reconstruction software: the reconstruction parameters should be selected to best match the anticipated performance of the imaging configuration.

6. CONCLUSIONS

We have developed a highly adjustable SPECT system that provides a unique opportunity to explore a continuum of imaging configurations. The ability to vary the position and widths of the crossed-slit apertures permits the selection of custom resolutions, sensitivities, and fields of view for a desired imaging acquisition. This flexibility is possible through the use of a calibration technique that estimates the geometric parameters of the system so that the large system matrix can be generated on the fly during GPU reconstruction.

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Figure 1.

Close-up of the internal construction of the adjustable slit design (a). Five-millimeter-thick copper-tungsten blades are mounted in an aluminum frame. One blade is fixed and the other is spring-tensioned and able to slide along brass ball bearings to adjust the blade separation. External views of the front and back sides of the adjustable slits (b). The slits were designed to permit close contact between the horizontal and vertical apertures for uniform magnification applications.

Figure 2.

The high-resolution silicon double-sided strip detector used in the anamorphic SPECT system described in this paper (a). A photograph of the anamorphic SPECT system with movable lead shielding in place. Two orthogonal slits are independently adjusted to permit maximum magnification of the object onto the high-resolution double-sided strip detector.

Figure 3.

Internal structure of the brachytherapy seeds used in these projections (a). Rapid-prototyping techniques permit the design and fabrication of test phantoms in which the seeds can be arranged in known and repeatable positions (b). The phantom shown in (b) was imaged through our system to demonstrate the intrinsic resolution of the DSSD and the effect of the adjustable slit widths (c)–(d).

Figure 4.

Sensitivity across the field of view as a function of angular position for two magnifications. Each magnification is shown for 250 μm , 500 μm , and 750 μm slit widths. The low magnification ((a)–(c)) corresponds to a field of view suitable for imaging a 30 mm $\varnothing$ $\times$ 70 mm mouse onto the detector. The high magnification ((d)–(f)) corresponds to a field of view suitable for imaging a 12 mm $\varnothing$ $\times$ 15 mm rodent brain.

Figure 5.

The geometric phantom used during calibration (a). In this example, we acquired five sets of calibration data obtained with five rods located at a radial distance of 6 mm from the center of rotation and vertically separated in 7.5 mm increments. A total of 1500 centroid measurements (superimposed over a single projection image) extracted from brachytherapy rod projections form the data set from which the geometric parameters are extracted (b). A contracting-grid algorithm employs maximum-likelihood methods to minimize the difference between measured (red) and estimated centroid positions (green) (c).

Figure 6.

Schematic diagram of the crossed-slit configuration. Radiometric calculations are performed using the distance R between the source point and the center of the PSF on the detector face. The angle θ is calculated as the angle between $\mathbf{R}$ and the normal distance z between the source and detector planes.

Figure 7.

During reconstruction, the required portions of the $\mathbf{H}$ matrix are computed on the fly in the GPU. One computational block considers a subset of pixels containing a single PSF (blue); each thread within the block computes the approximate irradiance detected at one pixel (red) within the PSF block by evaluating the number of subpixel centers that lie within the PSF boundary.

Figure 8.

Cross-sectional slices through a reconstruction of 11 ^{125}I brachytherapy rods in the custom phantom shown in Figure 3b. Transaxial slices (blue) through the volume demonstrate slightly higher resolution than axial slices (green), though the internal structure of the rods is evident at all rod orientations.

Table 1

Range of motion of the linear stages in the system and the geometric parameters influenced by each stage's position (a). Range of achievable geometric parameter values (b).

(a) Motion stage ranges		
Stage name	Travel range	Geometric parameter
Object x position	0–50 mm	center of FOV
Object y position	0–50 mm	$p_x, p_y, s_d, \text{mag}_x, \text{mag}_y$
Object z position	0–120 mm	center of FOV
Slit separation	5–100 mm	$p_x, p_y, q_x, q_y, \text{mag}_x, \text{mag}_y$
Slit translation	0–200 mm	$p_x, p_y, q_x, q_y, \text{mag}_x, \text{mag}_y$
Horizontal slit width	0–4 mm	resolution, sensitivity
Vertical slit width	0–4 mm	resolution, sensitivity

(b) Geometric parameter ranges	
Geometric parameter	Value range
p_x	$40 \rightarrow (s_d - 55)$ mm
p_y	$40 \rightarrow (s_d - 55)$ mm
q_x	$50 \rightarrow (s_d - 45)$ mm
q_y	$50 \rightarrow (s_d - 45)$ mm
s_d	$110 \rightarrow 230$ mm
mag_x	$0.85 \rightarrow 4$
mag_y	$0.85 \rightarrow 4$

Table 2

Storage requirements for system matrix using single precision floating point numbers.

Storage type	Description	Storage size
single full PSF	1024×1024 pixels	32 MB
full measured system matrix	60 projections, 101×101×101 voxels	235 TB
sparse matrix storage	assumes 31×31 pixels/PSF	221.31 GB
coefficient storage	6 parameters/PSF	1.38 GB
geometric system parameters	11 parameters total	0.05 kB

Table 3

Parameters estimated using contracting grid methods. Eleven geometric parameters are necessary to compute the system matrix, but ten additional nuisance parameters must also be determined to provide the best geometric estimates.

(a) Geometric parameters	
Geometric parameters	Description
p_x	Distance between center of rotation and vertical slit
p_y	Distance between center of rotation and horizontal slit
q_x	Distance between vertical slit and detector plane
q_y	Distance between horizontal slit and detector plane
s_d	Center of FOV to detector distance (redundant)
m_x	Mechanical offset of vertical slit in the x direction
m_y	Mechanical offset of horizontal slit in the y direction
e_x	Mechanical offset of detector in the x direction
e_y	Mechanical offset of detector in the y direction
θ	Rotation about detector center in the $x-y$ plane
ψ	Rotation parallel to y axis through the detector center
φ	Rotation parallel to x axis through the detector center

(b) Nuisance parameters	
Nuisance parameters	Description
	Phantom rotation offset
(δ_x, δ_z)	Phantom center displacement from axis of rotation
p	Phantom axis pivot point $(0, 0, p)$
α, β	Phantom pivot angles about $(0, 0, p)$, axis of rotation
$o_{1,2,4,5}$	Seeds 1,2,4,5 vertical position offset

Table 4

Timing measurements for GPU kernels for varying numbers of voxels and subpixels. Reconstructions were performed using 60 projections of 1024×1024 pixels and 31×31 pixels per forward-projected PSF.

(a) Sensitivity kernel run time				
	55 ³ voxels	101 ³ voxels	155 ³ voxels	255 ³ voxels
1 subpixel	27.329 s	156.678 s	565.197 s	2464.097 s
2 ² subpixels	32.766 s	189.909 s	684.590 s	3003.287 s
3 ² subpixels	48.907 s	303.596 s	1100.996 s	4916.496 s

(b) Forward project kernel run time				
	55 ³ voxels	101 ³ voxels	155 ³ voxels	255 ³ voxels
1 subpixel	10.205 s	61.595 s	223.502 s	999.863 s
2 ² subpixels	25.631 s	158.659 s	574.332 s	2584.195 s
3 ² subpixels	48.485 s	300.130 s	1085.542 s	4853.381 s

(c) Back project kernel run time				
	55 ³ voxels	101 ³ voxels	155 ³ voxels	255 ³ voxels
1 subpixel	10.785 s	67.529 s	244.442 s	1095.552 s
2 ² subpixels	28.002 s	172.592 s	624.568 s	2805.933 s
3 ² subpixels	51.026 s	315.067 s	1139.787 s	5121.851 s