Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2012 August 12; 8508: . doi:10.1117/12.966810.

Progress in BazookaSPECT: High-Resolution, Dynamic Scintigraphy with Large-Area Imagers

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Abstract

We present recent progress in BazookaSPECT, a high-resolution, photon-counting gamma-ray detector. It is a new class of scintillation detector that combines columnar scintillators, image intensifiers, and CCD (charge-coupled device) or CMOS (complementary metal-oxide semiconductors) sensors for high-resolution imaging. A key feature of the BazookaSPECT paradigm is the capability to easily design custom detectors in terms of the desired intrinsic detector resolution and event detection rate. This capability is possible because scintillation light is optically amplified by the image intensifier prior to being imaged onto the CCD/CMOS sensor, thereby allowing practically any consumer-grade CCD/CMOS sensor to be used for gamma-ray imaging. Recent efforts have been made to increase the detector area by incorporating fiber-optic tapers between the scintillator and image intensifier, resulting in a 16× increase in detector area. These large-area BazookaSPECT detectors can be used for full-body imaging and we present preliminary results of their use as dynamic scintigraphy imagers for mice and rats. Also, we discuss ongoing and future developments in BazookaSPECT and the improved event-detection rate capability that is achieved using Graphics Processing Units (GPUs), multi-core processors, and new high-speed, USB 3.0 CMOS cameras.

Keywords

BazookaSPECT; dynamic scintigraphy; gamma-ray imaging; small-animal imaging; columnar scintillators

1. INTRODUCTION

During the last several years, there has been a growing emphasis on developing high-resolution gamma-ray detectors for use in molecular imaging and pre-clinical applications such as small-animal scintigraphy and single-photon emission computed tomography (SPECT). Accordingly, a new class of scintillation detector has emerged that leverages the

advances made in CCD/CMOS sensor technology, yielding gamma cameras with hundreds of thousands to millions of individual sensor channels.^{1–8} In photomultiplier tube (PMT)-based detectors, scintillation light is spread out over a large area and over relatively few PMTs. PMT signals are then used to estimate the location and energy of the gamma-ray interaction. In contrast, CCD/CMOS-based detectors image scintillation light onto the sensor pixels. This approach provides fine sampling of the light distribution leading to less uncertainty when estimating the interaction location. The 2D/3D interaction location is estimated (to sub-pixel accuracy), using either a centroid calculation on pixel data or optimally using maximum-likelihood (ML) estimation techniques, to achieve intrinsic detector resolution $\sim 100\ \mu\text{m}$.^{1,9}

An integral component of CCD/CMOS-based gamma cameras are structured scintillators such as micro-columnar CsI(Tl).^{10, 11} Columnar scintillators are grown using an evaporative deposition process that produces arrays of needle-like structures ranging in size from 10–30 μm in diameter.¹¹ The columnar structure aids in restricting the lateral spread of scintillation light while channelling it via total internal reflection towards an exit face, somewhat analogous to optical fibers. Even though the lateral spread of light is significantly reduced, it still leaks into neighboring columns resulting in signal spread over a continuous region of multiple pixels, which we call a cluster. The size of the cluster varies with the energy of the gamma-ray photon and the interaction depth, which can be used to estimate the 3D interaction location.¹ Columnar scintillators can be grown in large areas (e.g., $47\times 47\ \text{cm}^2$) and thicknesses ranging from tens of microns to several millimeters. In addition to columnar CsI(Tl), a fabrication technique has also been developed by Radiation Monitoring Devices, Inc. (RMD) for growing columnar LaBr₃:Ce.^{12–14} An image frame of data with columnar LaBr₃:Ce illuminated by 140 keV gamma rays from ^{99m}Tc is shown in Figure 1b. Seven event clusters are visible in the image.

A variety of pathways exist for imaging the scintillation light onto the CCD/CMOS sensor, each with various trade-offs. As shown in Figure 2, the first pathway choice entails whether or not to optically amplify the light before imaging onto the CCD/CMOS sensor. If the light is not amplified, then detection of the relatively low optical signal requires the use of a low noise, high quantum efficiency (QE) CCD or the use of CCDs that provide electronic amplification, such as electronic-multiplying CCDs (EMCCDs). Low noise, high QE CCDs typically have a slow readout rate, which severely limits the event detection rate.³ EMCCDs provide high QE and modest frame rates (e.g., 30–60 fps at full resolution) resulting in an improved event detection rate,^{1, 4, 6} but have a relatively small active area (e.g. $16\times 16\ \text{mm}^2$). The active area can be increased with the incorporation of an optical taper between the scintillator and the sensor,¹⁵ but light loss in the taper beyond a few magnification factors quickly makes event detection difficult as signal from the scintillation light approaches the sensor read noise level.

Following the path of applying optical gain before imaging onto the CCD/CMOS sensor leads to a number of benefits. As demonstrated by Meng et al.,⁷ the area of an EMCCD-based gamma can be increased (e.g., 80 mm diameter) using a demagnifying (DM) tube. The DM tube, essentially a first-generation image intensifier, uses a photocathode to convert scintillation photons into photoelectrons. Then using electrostatic focusing, the

photoelectrons are converted back into optical photons using a phosphor screen resulting in a gain of ~ 100 photons/photoelectron, depending on the operating voltage.

Another optical amplification pathway is to use microchannel plate (MCP) image intensifiers. Each channel of the MCP is $\sim 10 \mu\text{m}$ in diameter and while preserving spatial information from the scintillation light due to its small diameter, it also acts as an individual photomultiplier. MCP intensifiers provide optical gains ranging from $10^5 - 10^6$ (or even higher) depending on the number of MCP plates. Similar to the DM tube, the amplified charge is converted into optical photons using a phosphor screen and the output screen is then imaged onto the CCD/CMOS sensor. The key benefit of this approach is that because scintillation light is significantly amplified prior to imaging onto the CCD/CMOS sensor, practically any consumer-grade CCD/CMOS sensor can be used for gamma-ray imaging. Consequently, this approach offers detector design flexibility in terms of cost, resolution, and speed, and it fully utilizes the ongoing advances in CCD/CMOS technology, e.g., multi-megapixel sensors and cameras that operate at tens of thousands of frames per second (fps).⁹ The original prototype detector that demonstrated this pathway, shown in Figure 3a, resembles a Bazooka and the imaging paradigm was accordingly dubbed BazookaSPECT, even though the device no longer resembles a Bazooka and is not necessarily used for SPECT imaging or to detect only gamma rays.^{2, 9}

This paper discusses recent progress made with BazookaSPECT to increase the detector area by insertion of a fiber-optic (FO) taper between the scintillator and the image intensifier. The development of this large-area BazookaSPECT has enabled the capability for full-body dynamic planar imaging of small animals, such as mice and rodents, using parallel-hole collimators. We present results from the proof-of-concept experiments and discuss on-going plans to develop even larger-area imagers and higher-resolution collimators. We also discuss recent updates to the acquisition software that enables the use of new, low-cost SuperSpeed USB (USB 3.0) CMOS sensors for increased event-detection rate.

2. MATERIALS AND METHODS

2.1 Event Detection

As previously discussed, an interaction event from a single gamma-ray photon results as signal contained in a continuous region of pixels called a cluster. Using an algorithm, which we refer to as frame parsing,⁹ multiple events within a image frame are identified, associated pixel data are extracted, and the interaction location and energy of each event is estimated. One algorithm for frame parsing uses the following step:

1. A frame from the CCD/CMOS camera is acquired.
2. A median filter is applied to remove hot/noisy pixels.
3. The filtered image is thresholded above the noise, and individual clusters are identified via a fast, connected-components-labeling algorithm.¹⁶
4. Pixels corresponding to identified clusters are extracted (e.g., 5×5 region of pixels).
5. From pixel data, the 2D or 3D interaction location and energy are estimated, optimally using ML estimation.^{1, 17}

A visualization of this algorithm is shown in Figure 4 taken using a CCD frame with multiple events. The magnified region of the detector has five gamma-ray photon interactions. It is undesirable to store complete image frames as high-resolution sensors operating at even modest frame rates quickly leads to gigabytes of disk storage for a short acquisition. With the use of graphics processing units (GPUs), frame parsing is implemented in real time. For each event, a super listmode¹⁷ entry is created that contains a time stamp, pixels associated with the event cluster, and an estimate of the 2D interaction location (centroid) and energy (summed pixels). Since only information relevant to an event are stored to disk, storage challenges become trivial with frame parsing. Also, if desired, further processing is possible with super listmode data, such as energy filtering or 3D position estimation using ML estimation.

2.2 Large-Area BazookaSPECT

MCP intensifier tubes are available in a range of diameters from standard sizes such as 18 and 25 mm (active area) typically used in military, night-vision devices to 40, 75, and 150-mm diameter custom tubes. Many night-vision devices use electrostatic focusing for image inversion as well as to increase the active area by focusing a large-area photocathode onto a smaller-area MCP (e.g. 50 mm photocathode/40 mm MCP see Figure 3b), but the electrostatic focusing introduces distortion at the edges and dark counts at the center of the field of view. Other devices, such as scientific-grade tubes, place the photocathode, MCP(s), and phosphor screen in close proximity resulting in a more compact design (proximity-focused intensifier) and have reduced distortion.

As can be seen in Figure 3, the overall length of BazookaSPECT has been significantly reduced by replacing the two-lens imaging optics (Figure 3a), a macro imaging configuration, with a single lens that has a short focal length (Figure 3b). Further reduction in length is achieved using a proximity-focused image intensifier (Figure 3c). The proximity-focused image intensifier has two MCPs (chevron stacked) for high gain up to 10^6 .

If the scintillator is placed in direct contact with the intensifier, then the area of the detector is limited by the active area of the intensifier. At first glance, it may appear that the most logical choice for increasing detector area is to use large-diameter intensifiers (>50 mm), however, these devices are expensive custom products and may be prohibitively expensive for most applications such as those motivating this work.

Another, and more cost effective, approach for increasing detector area is to insert a fiber-optic taper between the scintillator and intensifier. The taper serves to relay the image plane at the scintillator surface onto the entrance face (a fiber-optic faceplate) of the image intensifier. The fabrication process for both FO faceplates and tapers involves taking a block of parallel optical fibers and welding them together through a fusing or pressing operation.¹⁸ When one end of the fused block of fibers is in contact with a surface, an image of the surface is relayed to the opposite end ($1\times$ magnification for the faceplate). The FO taper imparts a magnification >1 or a minification <1 depending on which end is in contact with the imaging plane. The taper is made by heating the bundle in the middle, and then slowly pulling and stretching the block from both ends until the middle of bundle is the desired

diameter. Now having an hour-glass shape, the bundle is cut in half to make two fiber-optic tapers.¹⁸ Ends of the taper are cut and polished, and the magnification or minification is given as a ratio of the two end diameters. This imaging principle is demonstrated in Figure 5a.

The numerical aperture (NA) of a fiber is:

$$NA = n_0 \sin(\theta_{max}) = \sqrt{n_1^2 - n_2^2}, \quad (1)$$

where n_0 is the refractive index of the medium outside the fiber ($n_0 = 1$ for air), n_1 is the refractive index of the core (typically $n_1 = 1.81$), and n_2 is the refractive index of the cladding (typically $n_2 = 1.52$).¹⁸ NA is a measure of the maximum incident angle that light rays are transmitted down the fiber via total internal reflection (TIR). When using a fiber taper to increase detector area, such as can be done by directly bonding to a CCD sensor or to an image intensifier, light entering the taper from the larger diameter can be lost. As demonstrated in Figure 5b, a light ray that enters the taper at an angle greater than the critical angle, θ_{max} , will reflect down a fiber at a decreasing incident angle until the ray no longer undergoes TIR and leaks into the cladding material. To control this stray light, an extramural absorption material (EMA) is typically added to the fibers.¹⁸ The effective NA of a tapered fiber is given as,

$$NA_{eff} = NA_{max} \times \frac{D_{min}}{D}, \quad (2)$$

where D_{min} is the diameter of the smaller end of the taper, D is the diameter of the large end, and NA_{max} is the same as in Equation 1. As a result, minification reduces the fiber's acceptance cone by a factor of $1/m^2$, where $m = D_{min}/D$. If incident light is from a diffuse source, the resultant loss in efficiency using a fiber taper for increased detector area will also be $1/m^2$. Typically, this hit in efficiency would steer one away from using tapers with a high magnification/minification factor for use in low-light experiments. However, if incident light is forward peaked such that it is well coupled to the acceptance cone defined by NA_{eff} , high-magnification tapers make an excellent choice for increasing detector area.

Research is underway to understand the radiometry of scintillation light produced in columnar scintillators. Initial studies indicate a Lambertian response,¹⁹ and consequently would not suggest using high-magnification factor tapers for imaging due to the inherent light loss of the taper. Surprisingly, however, to date we have successfully verified that we can increase detector area $16\times$ using 4:1 tapers and 2 MCP intensifiers. This was demonstrated with GOS and columnar CsI(Tl) scintillators using low-energy sources such as ^{125}I .

Components of the large-area BazookaSPECT are shown in Figure 6. These include a $\phi 25$ mm, 2 MCP intensifier (Figure 6a) from ProxiVision GmbH that has a bialkali photocathode on a fiber-optic input window and P43 phosphor on clear glass output window, and a 100mm:25mm (4:1) FO taper from Incom, Inc. (Figure 6b). The manufacturer's claimed gross distortion of the taper is low at 2% max. The CCD camera is a Point Grey Research,

Inc. Dragonfly Express that uses a 640×480, 200 fps sensor (Figure 6a). A large-area columnar CsI(Tl) scintillator (10×10 cm²), fabricated by RMD, Inc., is placed in direct contact with the FO taper. The RMD columnar CsI(Tl) scintillator, shown in Figure 6c, is 750-μm thick and provides a detection efficiency of ~24% at 140 keV.

2.3 Imaging Collimators and Dynamic Scintigraphy

Large-area BazookaSPECT detectors (>100 mm diameter active area) allow us to perform full-body, high-resolution dynamic scintigraphy. Two common methods for imaging the distribution of radiotracer, each with trade-offs, are using pinholes and parallel-hole collimators.^{20, 21} Pinhole imaging provides increased sensitivity and spatial resolution (ultimately limited by the pinhole diameter) over parallel-hole collimation when the object is in close proximity to the pinhole and using magnification (especially required when using low-resolution detectors).^{21, 22} Increasing the field of view (FOV) in pinhole imaging is accomplished by placing the object at a distance farther away from the collimator at the expense of sensitivity and resolution. Another method to increase the FOV is to enlarge the acceptance cone of the pinhole, however, non-uniform magnification across the FOV is amplified as gamma rays from the edge of the FOV enter the pinhole at highly incident angles introducing distortion and parallax effects in the image. Also, thinning material around the pinhole to enlarge the FOV increases noise in the image by increased pinhole edge-penetration effects.²³

With parallel-hole collimators, the area of the detector defines the FOV and the sensitivity is fairly independent of distance from the collimator surface. Also, geometric magnification is uniform across the FOV. The geometric spatial resolution falls off approximately as

$R_0 + R_0 \frac{D}{t_{col}}$, where D is the distance from the collimator, R_0 is the geometric resolution at the surface of the collimator, and t_{col} is collimator thickness.²⁴ Therefore, since the imaging subject is in close proximity and of the order of the thickness of the collimator for small-animal imaging of rodents, the geometric resolution is relatively constant.

These features of uniform sensitivity, magnification, and resolution across the FOV in parallel-hole collimators, motivated their use in demonstrating the capability for full-body, high-resolution dynamic scintigraphy with the large-area BazookaSPECT. For dynamic scintigraphy we utilize super listmode data, where every event entry contains a timestamp. From the timestamp we generate sequences of planar projection images for any given time interval, limited only by the CCD/CMOS sensor integration period, e.g., 5 ms per frame with the camera running at 200 fps.

3. RESULTS

To demonstrate the dynamic imaging capability of the detector, we performed imaging tests using ^{99m}Tc tracers methylene-diphosphonate (MDP), which is preferentially taken up by bone, and mercaptoacetyltriglycine (^{99m}Tc MAG3), typically used in renal scintigraphy. The studies used a low-energy, medium resolution collimator (1.44 mm at 1 cm) and a BazookaSPECT (100-mm diameter) with ~200 μm intrinsic spatial resolution. The imaging subjects were anesthetized and tracers were injected through a tail-vein catheter connected to a syringe pump. Image acquisitions were started a few seconds prior to tracer infusion.

Figure 7 shows the dynamic uptake of MDP in a mouse over the course of 96 minutes. The duration of each projection image is two minutes (decay corrected). Note the rapid uptake and washout in the kidneys followed by eventual uptake in the knees and skull. Figure 8 shows dynamic projection images using ^{99m}Tc MAG3 at one-second and ten-second intervals (with decay correction). Note in the one-second interval images, tracer can be seen traveling down the tail vein towards the heart. Within a few seconds uptake in the kidney begins, and around 70–80 seconds after injection we begin to see uptake in the bladder.

From ^{99m}Tc MAG3 data we generated time-activity curves with selected regions of interest, namely kidneys and bladder, that are shown in Figure 9. The plots are generated at one-second temporal resolution and show the fast uptake and eventual washout in the kidneys and slow uptake and eventual leveling off in the bladder. The same imaging experiment was performed with a rat and planar projection images at 30-second intervals are shown in Figure 10.

4. ONGOING AND FUTURE DEVELOPMENTS

Since the rat is larger than the diameter of the prototype large-area BazookaSPECT detector, the FOV was limited to the abdomen region, but plans are underway to assemble a detector capable of full-body rat scintigraphy. This will be done using a 40-mm diameter intensifier coupled to a 40mm:160mm FO taper. Images of the FO taper and intensifier are shown in Figure 11. We are also considering developing even larger area imagers using BazookaSPECTs with square tapers that can be tiled.²⁵

The relatively poor spatial resolution of typical PMT-based gamma cameras (~ 3 mm) is the limiting resolution factor when imaging with parallel-hole collimators (e.g., 1-mm pitch between bores). However, because of the more than an order-of-magnitude increase in spatial resolution with CCD/CMOS-based gamma cameras, detector resolution no longer becomes the limiting factor and this is prompting the development of large-area high-resolution collimators. To illustrate this point, a mouse kidney was imaged using a medium-resolution collimator with ~ 1 -mm square bores and 200- μm thick septa followed by imaging with a high-resolution collimator that has 260- μm square bores and 120- μm thick septa.²⁶ Figure 12 shows images taken with both collimators. Notice the visible collimator pattern when imaging using the medium-resolution collimator. The high-resolution collimator has an active area of 25×25 mm² and plans are underway to fabricate 100-mm and 160-mm diameter high-resolution collimators.

Even though we have successfully demonstrated that event detection is possible at high minification using FO tapers, even at 4:1, the limit of this approach remains an open question. We will continue to investigate this approach, e.g. using 6:1 or even 8:1 FO tapers, and seek to quantify how the light-loss impacts detection performance. We are actively seeking to fabricate thicker (several millimeters) microcolumnar scintillators for increased detector efficiency. Additionally, we will continue to investigate and better understand the radiometric properties of columnar scintillators and how they are influenced by deposition growth parameters.

Other areas of on-going and future developments with BazookaSPECT include increasing the event-detection rate. We are currently updating the acquisition software to accommodate new USB 3.0 CMOS cameras. USB 3.0, also called SuperSpeed USB, provides a data transfer speed of 5 Gbits/second. Low-cost cameras (<\$1k USD) have recently entered the market and can acquire images at 150 fps, 1280×1024 pixels. An image of this camera, Point Grey Research, Inc. Flea[®] 3 USB 3.0, is shown in Figure 13. With pixel binning (640×512), it operates ~500 fps. Real-time frame parsing at these frame rates is possible using GPUs and multi-core CPUs.

Using the method developed by Furenlid et al.²⁷ to estimate the event detection rate for pixellated gamma-ray detectors, we can estimate the counting-rate capability of a 100-mm diameter large-area BazookaSPECT that uses these new USB 3.0 CMOS cameras. Assuming uniform illumination, an event cluster size ~340- μ m in diameter, and a reasonable 10% probability of event cluster overlap, the detector can acquire $\sim 3.4 \times 10^5$ events/second running at 150 fps (1024×1024 pixels), and $\sim 1.1 \times 10^6$ events/second at 490 fps (512×512 pixels).

5. SUMMARY AND CONCLUSION

CCD/CMOS-based gamma-ray detectors represent a paradigm shift from traditional PMT-based gamma cameras. They are of great interest in pre-clinical (and potentially clinical) medical imaging applications as they provide more than an order-of-magnitude increase in detector resolution over traditional cameras. One of the challenges with these detectors has been to develop large-area imagers. We have successfully demonstrated that large-area BazookaSPECT detectors can be assembled by incorporation of a fiber-optic taper between the scintillator and image intensifier.

Using parallel-hole collimators, we assembled a proof-of-concept BazookaSPECT imager for performing full-body rodent imaging. The timestamp associated with each event allows for sequences of planar projection images to be generated at a desired temporal resolution. From these data we generated time activity curves (one-second temporal resolution) that show the dynamic uptake/washout of radiotracer from organs of interest.

On-going and future work with BazookaSPECT will be to assemble even larger area imagers for full-body rat scintigraphy as well as quantify the highest taper magnification ratio (>4:1) that can be used with these detectors. Plans are underway to increase detector efficiency by fabricating thicker columnar scintillators and to better understand their radiometric properties. The high intrinsic detector spatial resolution of BazookaSPECT is prompting the need to fabricate high-resolution parallel-hole collimators, and we are in the process of obtaining large-area (100–160 mm in diameter) collimators for high-resolution dynamic scintigraphy. Finally, we are currently updating the BazookaSPECT acquisition software to accommodate USB 3.0 CMOS cameras that will significantly increase the event rate capability to $>10^6$ events/second.

Acknowledgments

This work is supported by the Center for Gamma-Ray Imaging, NIH Grant P41-EB002035. Updates to BazookaSPECT acquisition software to accommodate new USB 3.0 CMOS cameras described in this paper were conducted under the Laboratory Directed Research and Development Program at Pacific Northwest National Laboratory, a multiprogram national laboratory operated by Battelle for the U.S. Department of Energy. BWM is grateful for the support of a Linus Pauling Distinguished Postdoctoral Fellowship at PNNL. We would also like to thank Christy Barber for assistance with small-animal imaging experiments.

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Figure 1.
(a) 60-mm diameter columnar LaBr₃:Ce Source fabricated by RMD, Inc. (b) 25-mm diameter area of scintillator illuminated by seven, 140-keV gamma rays from ^{99m}Tc.

Figure 2.
Imaging pathways of various CCD/CMOS-based scintillation gamma cameras.⁹

Figure 3.

Evolution of the BazookaSPECT detector.⁹ (a) Original BazookaSPECT detector configuration using two lenses coupled in a macro-photography configuration. (b) Compact BazookaSPECT detector that uses a single, 6-mm lens to image a 50-mm input/40-mm output MCP intensifier onto a 640×480 CCD. (c) Compact BazookaSPECT detector using a 6-mm lens to image a 25-mm diameter proximity-focused, 2-MCP intensifier onto a 640×480 CCD.

Figure 4.

Frame-parsing, event detection algorithm.⁹ (a) Acquired image, (b) median-filtered image where noisy pixels are removed, (c) filtered image where individual clusters are identified using a connected-components labeling algorithm¹⁶ (individual events shown here are identified by color), and (d) the estimated 2D interaction location. The 2D position estimate, summed pixel values, time stamp, and associated event cluster pixels are written to disk as a super listmode¹⁷ entry.

Figure 5.

(a) Imaging using a 2:1 (50mm:25mm) fiber-optic taper. The FO taper is used to relay an image plane from one spatial location to another and either magnifies (right) or minifies (center) depending on which end of the taper is in contact with the image plane. (b) Light loss in FO tapers occurs when light enters a fiber at an angle greater than the maximum acceptance angle, θ_{max} .

Figure 6.

Large-area BazookaSPECT detector components: (a) Bialkali, 2-MCP (chevron) image intensifier with a P43 phosphor screen that is imaged onto a 640×480 CCD camera (same configuration as Figure 3c), (b) Incom Inc. 100mm:25mm (4:1) FO taper, and (c) an RMD, Inc. $750\text{-}\mu\text{m}$ thick, $10 \times 10\text{ cm}^2$ columnar CsI(Tl) scintillator.

Figure 7.

Full-body mouse dynamic scintigraphy using ^{99m}Tc MDP (2.5 mCi). Detector Configuration: 0.75-mm thick columnar CsI(Tl), medium resolution collimator (1.44 mm at 1 cm), and 100-mm diameter BazookaSPECT detector with $\sim 200\text{-}\mu\text{m}$ spatial resolution. The temporal window for each projection image is two minutes (decay corrected).

Figure 8.

Full-body mouse dynamic scintigraphy using ^{99m}Tc MAG3 (~3.0 mCi). Detector Configuration: Same as Figure 7. Both high spatial and temporal resolution images are generated using super listmode data. Top: Twenty, one-second images beginning at start of tracer infusion. Bottom: Twenty, ten-second images. Note the dynamic uptake in the heart, kidneys, and bladder.

Figure 9.

Dynamic scintigraphy with ^{99m}Tc MAG3 (~ 3.0 mCi). Time-activity curves (one-second temporal resolution) are shown for mouse kidneys and bladder over a course of 30 minutes, beginning at time of tracer injection.

Figure 10.

Rat dynamic scintigraphy (10 minutes) using ^{99m}Tc MAG3 (~4.8 mCi). Same detector configuration as Figure 7. Twenty, thirty-second images post injection.

Figure 11.

Large-area BazookaSPECT detector components for full-body rat scintigraphy: (a) 40-mm diameter (active area) 2-MCP image intensifier and (b) 160-mm diameter 4:1 FO taper.

Figure 12.

Interchangeable collimators allow imaging experiments to be tailored for desired sensitivity and resolution. $^{99\text{m}}\text{Tc}$ MAG3 resolution comparison of kidneys imaged using high-resolution and medium-resolution parallel-hole collimators. The medium-resolution collimator has ~1-mm square bores with ~200- μm septa and the high-resolution collimator has 260- μm square bores with 120- μm septa.²⁶

Figure 13. Point Grey Research, Inc. Flea[®] 3 USB 3.0 CMOS camera. The camera has 1280×1024, 4.8×4.8- μm^2 pixels and operates at 150 fps (full resolution).