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Development and characterization of a round hand-held silicon photomultiplier based gamma camera for intraoperative imaging

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Abstract

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This paper describes the development of a hand-held gamma camera for intraoperative surgical guidance that is based on silicon photomultiplier (SiPM) technology. The camera incorporates a cerium doped lanthanum bromide (LaBr₃:Ce) plate scintillator, an array of 80 SiPM photodetectors and a two-layer parallel-hole collimator. The field of view is circular with a 60 mm diameter. The disk-shaped camera housing is 75 mm in diameter, approximately 40.5 mm thick and has a mass of only 1.4 kg, permitting either hand-held or arm-mounted use. All camera components are integrated on a mobile cart that allows easy transport. The camera was developed for use in surgical procedures including determination of the location and extent of primary carcinomas, detection of secondary lesions and sentinel lymph node biopsy (SLNB). Here we describe the camera design and its principal operating characteristics, including spatial resolution, energy resolution, sensitivity uniformity, and geometric linearity. The gamma camera has an intrinsic spatial resolution of 4.2 mm FWHM, an energy resolution of 21.1 % FWHM at 140 keV, and a sensitivity of 481 and 73 cps/MBq when using the single- and double-layer collimators, respectively.

Keywords

Biomedical imaging; Gamma-ray detectors; Instrumentation and measurement

I. Introduction

Small hand-held gamma cameras offer surgeons the ability to assess the distribution of injected radiotracers or radiopharmaceuticals within the surgical patient, a task currently left to inspection of preoperative scintigraphic or single-photon emission computed tomography (SPECT) images and the intraoperative use of non-imaging gamma probes. Others have reported using gamma imagers with small or medium fields of view (FOV) in sentinel lymph node biopsy (SLNB) [1-14] and recently in radioguided occult lesion localization (ROLL) [15]. These imagers have in most cases been used to assist the standard non-imaging gamma probes. Additional information potentially available from intraoperative gamma cameras includes better delineation of closely spaced surgical targets (e.g. lymph nodes or tumors), straightforward, user-independent quantification of activity in excised targets, and image-based documentation of pre- and post-excisional radiotracer distribution. Recently a 'freehand SPECT' system that generates 3-dimensional maps of radiotracer distribution using a motion-tracked hand-scanned non-imaging probe has become commercially available, thus permitting intraoperative imaging [16]. Intraoperative image information could be especially useful in complex situations involving multiple closely spaced SLNs (e.g. head and neck surgery) or for combined SLNB and ROLL (SNOLL).

Our group has evaluated the use of an arm-mounted, medium FOV (12.5 cm × 12.5 cm) gamma camera for SLNB among melanoma patients and found that intraoperative imaging provides additional information when standard techniques fail or are ambiguous [2]. Based on the experience gained in that study, our surgeon collaborators suggested the addition of a more compact camera with a round shape for easier positioning in tight locations. Ideally, the camera would be capable of both hand-held and mechanical arm-mounted operation. Facilitated by the recent increased availability of silicon photomultipliers (SiPMs), a round

format, lightweight gamma camera was designed and built. Here we present results of bench-top performance evaluations of the imaging characteristics of a SiPM-based gamma camera designed for hand-held intraoperative use.

II. MATERIALS AND METHODS

A. Gamma camera components summary

The scintillator used in this gamma camera is a round monolithic crystal of 5 % cerium doped LaBr_3 , commercially available as BrillanCe 380 from Saint-Gobain Crystals, Hiram, OH. The crystal is 6 cm in diameter and 6 mm thick and is in a hermetic package with a 1 mm thick glass window light-guide, a 0.5 mm aluminum window, and with an overall diameter of 7.5 cm and height of 9.5 mm. The SiPM array contains eighty multi-photon pixel counters (MPPCs) with an active area of $3 \times 3 \text{ mm}^2$ and a fill factor of 61.5 % each (model S10362-33-050P from Hamamatsu Photonics, Hamamatsu, Japan). The MPPCs were arranged in a circular pattern with a central 8×8 square array bounded on each side by a 1×4 array (Fig. 1). The MPPC pitch is 6 mm.

Although the inter-MPPC gaps are not minimized, this pitch is adequate to sample the broad light distribution of the LaBr_3 scintillator. LaBr_3 is ~ 1.4 times as dense as thallium doped sodium iodide (NaI:Tl), has ~ 1.6 times the light yield of NaI:Tl and has a 16 ns decay time (NaI:Tl has a 230 ns decay time). LaBr_3 is a good spectral match with the photon detection efficiency (PDE) of the MPPCs.

The scintillator is coupled to the MPPC array using polydimethylsiloxane (PDMS) coupling compound to improve the scintillation light transfer to the MPPCs and insure a sturdy coupling. The design of the amplifier board and assembly of the MPPC array were executed by AiT Instruments (Newport News, VA). In this setup, each MPPC is supplied with its own readout channel. The diameter of the printed circuit board (PCB) module is ~ 65 mm and the thickness with MPPCs is ~ 5 mm (Fig. 1). To supply their bias voltage, the MPPCs are split into five groups of 16 MPPCs (Fig. 2) with each group powered by a separate SIP90 power supply from EMCO High Power Corporation, Sutter Creek, CA. The position of the incident gamma photon is calculated from the individual digitized MPPC signals using a center of gravity (COG) algorithm. During testing, the rightmost MPPC in the second row failed and we were unable to replace it for these tests.

The sum of the signals from the 80 SiPMs is sent to a leading-edge discriminator (Fig. 3) whose input is buffered and amplified by a wide bandwidth op-amp and then compared to a potentiometer-set threshold in a high-speed comparator MAX901. The final output signals a trigger for the analog-to-digital converters (ADCs). The output of the comparator triggers a one-shot circuit that generates a potentiometer-controlled pulse. A second one-shot circuit is also triggered that inhibits re-triggering the first for a fixed time period, thereby avoiding undesirable ringing trigger response. The one-shot inhibit time is approximately 100 ns (47 pF, 2.00 K Ω). The estimated dead time from our count rate experiments is $\sim 85 \mu\text{s}$. Until the event rate reaches the range of 2×10^3 counts per second, this dead time has an inconsequential effect on sensitivity. Each of the measurements reported here was performed at counting rates lower than 2 kHz.

The camera analog data are acquired by two field-programmable gate array (FPGA)-based units designed and built by AiT Instruments, Newport News, VA [17]. They incorporate simultaneous sampling gated integrators and 12-bit ADCs. One unit (model DAQ6400U) contains 64 channels and digitizes the inner square group of 8×8 MPPCs. The other unit (model DAQ1600U) has 16 channels and digitizes the four outer groups of 4 MPPCs. The ADC modules' design permits event timing synchronization via CAT5e connection between the units [18]. The ADCs transmit conversion data over the universal serial bus (USB) to the acquisition computer where core acquisition functions are performed with custom written software. Basic data-acquisition functionality is implemented with a set of Java based routines, and Java extensions are used to interface with a graphical user interface (GUI). The GUI is written in the data-acquisition programming environment Kmax (Sparrow Corporation, Daytona Beach, FL). Java modules are used for energy and COG position calculation, while the GUI displays raw and uniformity-corrected images based on various user-defined energy or positional conditions.

The camera housing, designed at Jefferson Lab, is built from CMW-1000 machinable tungsten. There is an inside housing layer made of black Delrin® (polyoxymethylene) that has a dual purpose – it forms a light-tight enclosure for the scintillator and the MPPC board, and it mechanically stabilizes the camera electronics in the center of the tungsten housing during hand-held operation.

The camera body is capped with a custom designed tungsten-polymer composite two-part collimator fabricated by Mikro Systems (Charlottesville, VA) with an active diameter of 60 mm and overall diameter of 75 mm. The collimator has a unique magnet-based quick assembly/disassembly alignment mechanism [19] that allows aligned stacking of two layers of the collimator to accommodate either high-resolution (double-layer) or high-sensitivity (single-layer) imaging (Fig. 4).

The matrix size of the raw images is 270×270 pixels for the FOV of 60 mm, resulting in an effective pixel size of 0.24 mm. Additionally, the images can be displayed and saved in $5 \times$ and $10 \times$ binned modes with image matrix sizes of 54×54 and 27×27 pixels and image pixel sizes of 1.2 and 2.4 mm, respectively.

The collimator hole pitch was chosen to be 1.0 mm. Fig. 5 contains plots of the theoretical collimator efficiency and spatial resolution for a 1.0 mm hole pitch, but varying the hole size. For each hole size, the hole length was adjusted to achieve 5 % septal penetration [20], assuming a nominal linear attenuation coefficient of 21.9 cm^{-1} for the tungsten-polymer composite material. The figure shows that given these constraints, the maximum efficiency (4.8×10^{-4}) occurs for a hole size of 0.6 mm. The corresponding hole length is 5.5 mm, which was on this basis chosen as the single layer thickness. Fig. 6 plots collimator efficiency and spatial resolution for changing hole length, while keeping the pitch and hole size constant. Vertical lines show the thickness corresponding to one and two layers. The figure shows that for single-layer operation the predicted collimator resolution at a source-to-collimator separation of 5 cm is 7.2 mm full width at half maximum (FWHM) and the predicted geometric efficiency is 4.8×10^{-4} . For two 5.5 mm layers stacked, the predicted resolution is 3.7 mm FWHM and the predicted geometric efficiency is 1.0×10^{-4} .

The camera head, including the two-layer collimator is approximately 40.5 mm thick and its mass is 1.4 kg, which permits comfortable hand-held use by the surgeons in the operating room (Fig. 7) [21].

B. Measurement of camera imaging properties

1) Linear attenuation coefficient—The linear attenuation coefficient of the tungsten polymer material used to manufacture the collimator for the gamma camera was measured using thin tungsten polymer samples acquired from Mikro Systems. A small acrylic lesion (9.9 mm outer diameter (OD)) was filled with ^{99m}Tc -pertechnetate solution (activity of 37 MBq) and positioned 1 cm above a 40° acceptance angle 1 mm pinhole collimator from Mikro Systems. The rest of the pinhole collimator is 10.6 mm thick tungsten designed to completely cover the parallel-hole collimator of the gamma camera. The tungsten polymer samples ranged in thickness from 170 μm to 530 μm and were combined to produce materials of thickness up to 2 mm in between the source and the pinhole collimator, covering the pinhole completely. The ^{99m}Tc -pertechnetate source, pinhole collimator and the gamma camera were at fixed positions during this experiment. For each thickness of the material, an image was acquired for a fixed number of total counts in the image. For each acquired image, the sum of the counts in a fixed circular region of interest (ROI) was measured, normalized by the respective acquisition time and plotted against the thickness of the tungsten.

2) Spatial resolution—The intrinsic spatial resolution of the gamma camera was measured using a ^{99m}Tc -pertechnetate point source placed at a distance greater than five times the useful field of view (UFOV) from the detector, with the collimator removed and with a 0.3 mm wide tungsten slit placed at a distance of 3 mm from the scintillator. The UFOV is defined as a circular area with a diameter that is the largest inscribed circle within the collimated field of view [22]. The extrinsic resolution of the camera was also measured by translating a small capillary (Kimble 71900-50 μL , 1 mm inner diameter (ID)) filled with ^{99m}Tc -pertechnetate away from the collimator and obtaining images at discrete distances from the collimator. A program in Interactive Data Language (IDL, Exelis Visual Information Solutions, Inc, Boulder, CO) was used to fit a Gaussian to the line spread function (LSF) of the resulting images. Measurements were made with both one and two collimator layers.

3) Sensitivity—The system sensitivity, defined as the number of detected counts per second per MBq of source activity, was experimentally measured and calculated following the NU 1-2007 National Electrical Manufacturers Association (NEMA) standard [23]. A flat 26 mm ID plastic dish was filled with a shallow (2 – 3 mm) layer of ^{99m}Tc -pertechnetate solution and placed 100 mm above the collimator with the dish positioned at the center of the FOV of the gamma camera. An image was acquired with half million counts with the energy acceptance window set from 50 – 200 keV. This measurement was performed for both one and two collimator options. The sensitivity at 100 mm was then calculated as the ratio of the total number of counts in the image as measured in ImageJ (National Institutes of Health, Bethesda, MD) and the total activity in the plastic dish.

4) Energy resolution—The camera energy resolution was characterized using a solution of ^{99m}Tc -pertechnetate (activity of 3.7 MBq) in a 75 mm diameter Petri dish placed on top of the collimator to produce a uniform low-scatter flood source. The Petri dish extends ~ 15 mm in diameter beyond the face of the gamma camera. A high-statistics flood image was acquired with user defined ROIs in different parts of the image for purposes of examining the energy resolution over different parts of the FOV of the detector. The energy spectra were plotted in ImageJ and a Gaussian function was fit to the 140 keV photopeak to obtain the FWHM energy resolution for each region separately and for the full FOV.

To account for varying gain properties of SiPMs in different regions and permit a single energy discrimination to be applied, the energy spectra from different regions of the FOV are normalized. In this process, the energy spectrum for each pixel in the $5\times$ binned image is recorded during acquisition of the high-statistics flood image. A look-up table is then formed containing the location of the energy spectra peak for each pixel in the $5\times$ binned image. This table is used to scale the summed signals of each detected event to determine whether or not they fall within the global energy discrimination window, and to generate a normalized full-field energy spectrum.

5) Image geometric linearity—An aluminum phantom was constructed to hold two orthogonal layers of 10 mm spaced ^{99m}Tc -pertechnetate solution filled capillaries (Kimble 71900-50) with inner diameters 1 mm as shown in Fig. 8. This setup was used to characterize the geometric nonlinearities within the image. Algorithms to correct for image nonlinearities were then developed using the image data.

6) Image sensitivity uniformity—A Petri dish containing ^{99m}Tc -pertechnetate solution (activity of 3.7 MBq) was placed on top of the collimator to produce a uniform low-scatter source. A high-statistics flood image (Flood_{hs}) was acquired in this fashion and used as a flat-field correction map. This was then used to correct subsequent images on a pixel-to-pixel basis according to the formula: $\text{Corrected Image} = (\text{Uncorrected Image} * \text{Average}(\text{Flood}_{\text{hs}})) / \text{Flood}_{\text{hs}}$. The sensitivity uniformity of the corrected image was judged by measuring and calculating the integral uniformity in the useful and central fields of view (UFOV and CFOV). The CFOV is defined as a circular area with a diameter that is 75 percent of the diameter of the UFOV.

III. RESULTS AND DISCUSSION

A. Linear attenuation coefficient and its impact on septal penetration

The linear attenuation coefficient of the tungsten polymer was measured to be 18.7 cm^{-1} , a value differing from the value of 21.9 cm^{-1} used in the collimator parameter calculations, which was measured using earlier tungsten-polymer samples. This lower linear attenuation coefficient negatively affects the septal penetration of the collimator which was targeted to be less than 5 % when using one collimator layer. Using the more recently measured value of 18.7 cm^{-1} , the theoretical values for single- and double-layer septal penetration are 7.6 % and 0.6 %, respectively. A value of $\mu = 18.7 \text{ cm}^{-1}$ is used in all calculations below.

B. Spatial resolution

The intrinsic spatial resolution was measured to be 4.2 mm FWHM. Fig. 9 shows the measured FWHM extrinsic spatial resolution at varying distances for both single- and double-layer collimator configurations. The theoretical resolution is also plotted. The extrinsic spatial resolution at 5 cm working distance is 10.3 mm FWHM.

C. Sensitivity

The measured sensitivities using single- and double-layer collimator configurations are 481 and 73 cps/MBq, respectively. This efficiency ratio of 6.6 between the single- and double-layer options can be compared to the theoretical ratio of geometric collimator efficiencies of 5.0. The larger than anticipated ratio is the result of higher septal penetration for the single-layer configuration (see section III. A.).

D. Energy spectra

Fig. 10 shows a high-statistics flood image acquired with user defined ROIs in five regions of the image for the purpose of examining the energy-resolution variation over the FOV of the detector. The FWHM energy resolution for the entire FOV was found to be 24.3 %, with measured FWHM energy spectra in the custom defined ROIs calculated at 28.8 % (bottom ROI), 22 % (left ROI), 19.1 % (middle ROI), 21.3 % (top ROI) and 23.5 % (right ROI). Energy normalization improved the energy FWHM resolution for the full FOV to 21.1 %.

E. Image geometric linearity

To improve the image geometric linearity, the raw capillary phantom image was loaded in Matlab (Matrix laboratory, MathWorks, Natick, MA) and all grid intersections were selected. These intersections were compared to target grid intersection points for an ideal grid model using the cp2tform image processing toolbox. Two third-degree 2D polynomial equations:

$$\text{new_x_position} = a_0 + a_1y + a_2x + a_3y^2 + a_4x^2 + a_5xy + a_6x^2y + a_7xy^2 + a_8x^3 + a_9y^3,$$

and

$$\text{new_y_position} = b_0 + b_1y + b_2x + b_3y^2 + b_4x^2 + b_5xy + b_6x^2y + b_7xy^2 + b_8x^3 + b_9y^3,$$

were used to fit the original data to the ideal grid with least-square fitting. To improve linearity, counts from each pixel in each original uncorrected image are transformed according to the above equations to shift their positions accordingly. The dewarping equations are implemented in the Kmax GUI for gamma camera image acquisition and image correction in real-time. As an example a raw and corrected image of the 2-layer capillary phantom are shown in Fig. 11.

F. Image sensitivity uniformity

Fig. 12 shows examples of a low-statistics uncorrected image (left), a high-statistics flood image (center), and the result of uniformity correction of the low-statistics image following correction using the high-statistics image (right). To characterize the sensitivity uniformity of the gamma images, the integral uniformity is calculated as:

$$\text{Integral uniformity} = \frac{Pixel_{max} - Pixel_{min}}{Pixel_{max} + Pixel_{min}} \cdot 100 \quad (1)$$

where $Pixel_{max}$ and $Pixel_{min}$ are the maximum and minimum pixel values in the selected FOV.

The symmetry of the full 60 mm diameter FOV is compromised by a dead SiPM in the upper right corner, resulting in a full FOV integral uniformity of 100 %. However, if the UFOV is defined to be a circular area in the center of the image with a diameter (51 mm) that is the largest inscribed circle within the collimated field of view not containing the missing SiPM, then the integral uniformity is 26 % over the UFOV and 15 % over the 38.3 mm CFOV. The dead MPPC reduced only the useful sensitive area. The absence of signal from that region did not affect the determination of the centroids of events in the other regions of the detector.

G. Discussion

One of the challenges of the SiPM technology is that the SiPM gain and count rate are affected by slight changes in temperature [24]. Fig. 13 shows the shift in the photopeak channel position with changing temperature. A plot of the photopeak position versus the time after initial application of the bias voltage with the camera and the source at room temperature is shown in Fig. 14. Fig. 15 shows the fluctuation in photopeak position in seven surgical cases during a pilot clinical study using the SiPM camera for SLNB for melanoma patients [21].

A temperature compensation method to maintain the MPPC gain by changing the bias voltage based on camera's internal temperature is being implemented. Two LM73CIMK-0 (National Semiconductor Corporation, Santa Clara, CA) sensors have been mounted on a dedicated PCB on top of the SiPM array inside the camera housing (Fig. 16).

IV. SUMMARY AND CONCLUSIONS

In conclusion, a novel round-FOV compact hand-held gamma camera with imaging and logistical properties designed for intraoperative guidance has been built and tested. The camera is the result of collaboration among groups from West Virginia University, Thomas Jefferson National Accelerator Facility and the University of Virginia. The camera's compact size and low mass make it convenient for hand-held operation. It has an intrinsic spatial resolution of 4.2 mm FWHM, an energy resolution of 21.1 % FWHM at 140 keV, and a sensitivity of 481 and 73 cps/MBq when using the single- and double-layer parallel-hole collimators, respectively.

For reference, Table I [2,5,6,7,8,11,14,25,26,27] presents a non-exhaustive list of some previous intraoperative gamma cameras, listing a few of their physical and imaging properties.

Intraoperative imaging systems like those listed in Table I have the potential to guide surgical procedures in real time by helping surgeons visually assess the spatial distribution of injected radiotracers or radiopharmaceuticals within the patient. The most useful size, weight, and shape of such gamma cameras is a topic of discussion and almost certainly depends on the nature of the particular surgical task (SLNB, tumor excision, the particular organ or anatomical region of interest, etc.). The SiPM-based gamma camera described here has reasonably small FOV, is round, and is light enough for hand-held operation, in accordance with the desires of melanoma surgeons at our institution for use in SLNB. The camera was evaluated in the operating room during a human study of melanoma SLNB [21] and is scheduled to be used in two additional human SLNB studies in the near future. During the previous clinical study, a second larger gamma camera (20 cm $\times$ 20 cm FOV) was also used preoperatively to identify nodal basins containing SLNs and establish the overall surgical plan. In order to keep the gamma camera sterile intraoperatively, the hand-held SiPM camera was sheathed in a sterile disposable surgical drape such as those used with ultrasound probes, laparoscopic cameras and C-Trak gamma probes. The combination of large FOV preoperative survey imaging immediately prior to surgery, followed by small FOV intraoperative spot imaging for SLN localization and post-excision assessment proved to provide a useful balance among the needs of the various stages of melanoma SLNB.

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Fig. 1. Hamamatsu MPPC arrangement of 80 units (top); round PC board connected to the readout system (bottom left) and thickness measurement (~ 5 mm) of the printed circuit board with mounted MPPCs (bottom right).

Fig. 2.
Map of five regions of 16 MPPCs. The colors identify the five regions, each powered by a separate SIP90 power supply. The MPPCs of each region are numbered for clarity.

Fig. 3.
Schematic of data-acquisition hardware used to read out the SiPM gamma camera.

Fig. 4. Parallel-hole collimator with the magnetic locking mechanism on the edges (one layer shown).

Fig. 5.
Theoretical collimator efficiency and resolution versus the hole size at fixed septal penetration and hole pitch.

Fig. 6.
The calculated efficiency and resolution trade-off for changing overall hole length at fixed hole size of 0.6 mm and hole pitch of 1.0 mm, with marked single- and double-layer collimator options.

Fig. 7.
Demonstration of hand-held operation of the gamma camera.

Fig. 8. Custom phantom holding perpendicular layers of parallel capillaries filled with ^{99m}Tc -pertechnetate solution at fixed pitch of 10 mm.

Fig. 9. Experimental and theoretical values of the gamma camera's FWHM spatial resolution.

Fig. 10.

Display of the Kmax-based data-acquisition user interface. Shown is a flood image with several user-defined ROIs (left) and their respective energy spectra (right). The full-FOV not-normalized energy spectrum is shown in the top left histogram and Region 0 to 4 histograms correspond to bottom, left, middle, right and top ROI energy spectra respectively. The distortion from a regular circular active area seen in the top right of the flood image is caused by the failure of one of the MPPCs.

Fig. 11. Barrel distortion is apparent in the raw gamma camera image of the phantom shown in Fig. 8 (left). The same image following dewarping (right) shows improved linearity across the field of view.

Fig. 12.

Image of a raw low-statistics flood image (left), the high-statistics flood image that was used for the correction (middle), and the resulting corrected low statistics image (right).

Fig. 13.
Nominal Hamamatsu MPPC temperature effects - photopeak position versus temperature.

Fig. 14. Photopeak channel position drift arising from temperature change due to camera self-heating.

Fig. 15.
Photopeak channel position drift due to temperature changes in normal camera use in the OR during a pilot clinical study.

Fig. 16.
Gamma camera assembled in a Delrin® enclosure with the temperature sensor PCB on top.

TABLE I

Devices used for clinical trials of SLNB.

Device and investigator	Spatial resolution (FWHM) [mm]		Sensitivity [cps/MBq]	Energy resolution (FWHM) [%]	Field of view [cm × cm]	ESR working distance [cm]	Handling mode
	ISR	ESR					
6cm round SiPM gamma camera (described here)	4.2	7.5 10.3	481	21.1% at 140 keV	6cm diameter round	3 5	Handheld, 1.4kg mass
GammaCAM Aarsvod <i>et al</i> 2006	NS	NS	61	NS	12.5 × 12.5	NS	Arm-mounted
Sentinella Vermeeren <i>et al</i> 2009	1.8	7 21	319 18.5	15.9% (50 - 200 keV)	4 × 4 20 × 20	3 15	Arm-mounted
12.5 × 12.5 MGC prototype Dengel <i>et al</i> 2011	NS	NS	405	NS	13 × 13	NS	Arm-mounted
2020tc Digirad <i>et al</i> 2007	NS	18	133	<14% (60-300 keV)	21 × 21	15	Arm-mounted
SSGC Tsuchimochi <i>et al</i> 2003	NS	1.56 6.3	300	7.8% at 140 keV	4.48 × 4.48	0 5	Arm-mounted
IP-824 Li-Tech Scopinaro <i>et al</i> 2008	NS	3	NS	NS	2.54 × 2.54	1	Handheld, 1kg Mass
CarollReS Mathelin <i>et al</i> 2008	3	10	1.6	45% at 122 keV	5 × 5	5	Handheld, 2.5kg mass
NodeView Olcott <i>et al</i> 2007	1.8	4	270	12% at 140 keV	5 × 5	3	Handheld, 1.1kg mass
Minicam II Garbay <i>et al</i> 2009	2.5	NS	200	<10% at 140 keV	4 × 4	1	Handheld, 0.7kg mass
POCI Pitre <i>et al</i> 2003	3.2	3.9 7.6	290	32% at 140 keV	4cm diameter round	1 5	Handheld, 1.2kg mass

FWHM, full width at half maximum; ISR, intrinsic spatial resolution; ESR, extrinsic spatial resolution; NS, not specified.