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(54) **NEAR 2Pi COMPTON CAMERA FOR MEDICAL IMAGING**

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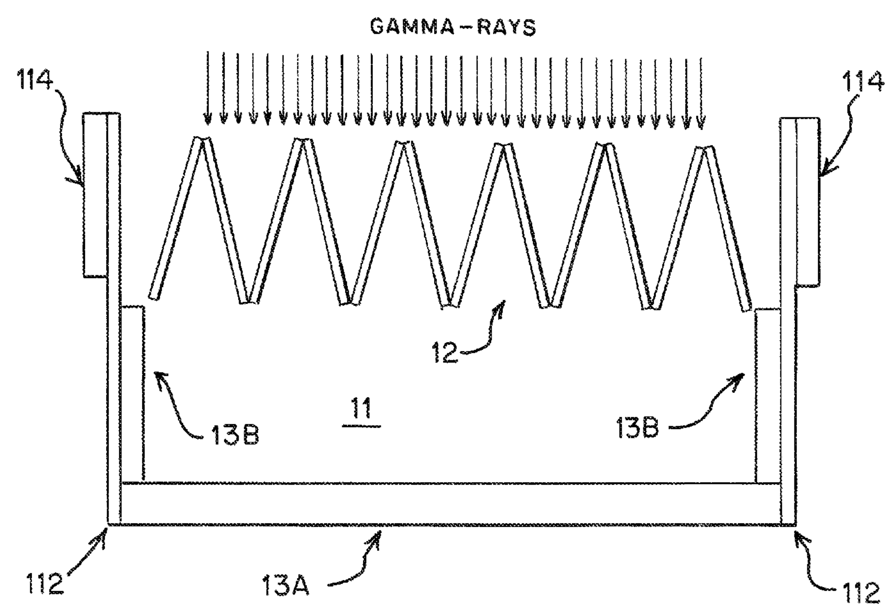
(57) **ABSTRACT**

To capture more emitted photons with a Compton camera, the scatter detector is tilted (non-orthogonal angle) relative to a radial from the isocenter of the imaging system. The tilt creates a greater volume for scatter interaction. To capture more scatter photons, the catcher detector is non-planar, such as a multi-faced detector at least partially surrounding a volume behind the scatter detector. The tilted scatter detector alone, the non-planar catcher detector alone, or the tilted scatter detector and the non-planar catcher detector are used in the Compton camera.

17 Claims, 8 Drawing Sheets



(51) **Int. Cl.**

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A61B 6/42	(2024.01)
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G01T 1/164	(2006.01)
G01T 1/24	(2006.01)
G01T 1/29	(2006.01)

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CPC ***A61B 6/4411*** (2013.01); ***A61B 6/4452*** (2013.01); ***A61B 6/461*** (2013.01); ***G01T 1/1642*** (2013.01); ***G01T 1/242*** (2013.01); ***G01T 1/2985*** (2013.01); *A61B 2562/18* (2013.01)

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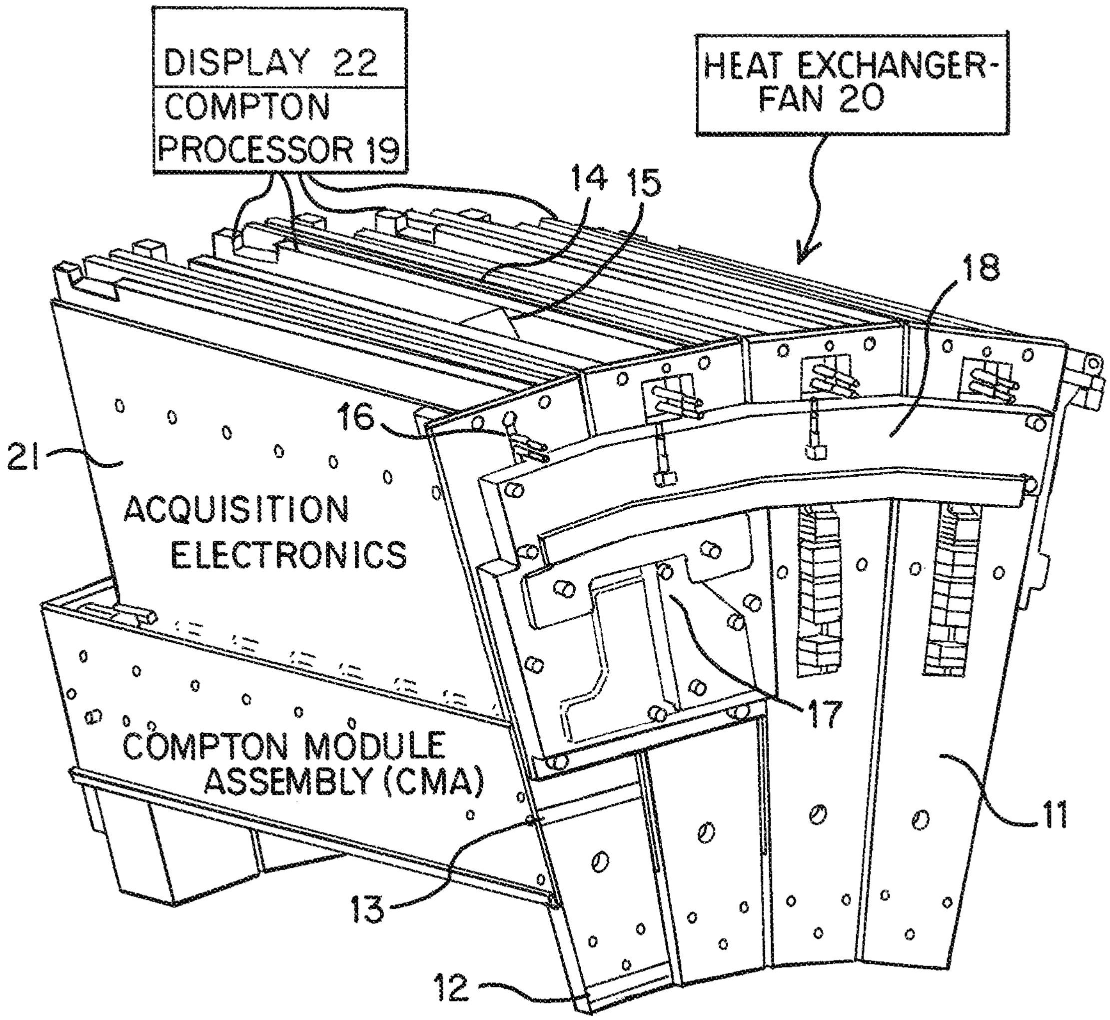
FIG. 1


DISPLAY 22
COMPTON
PROCESSOR 19
HEAT EXCHANGER-
FAN 20
14
15
18
16
21
ACQUISITION
ELECTRONICS
17
COMPTON MODULE
ASSEMBLY (CMA)
11
13
12

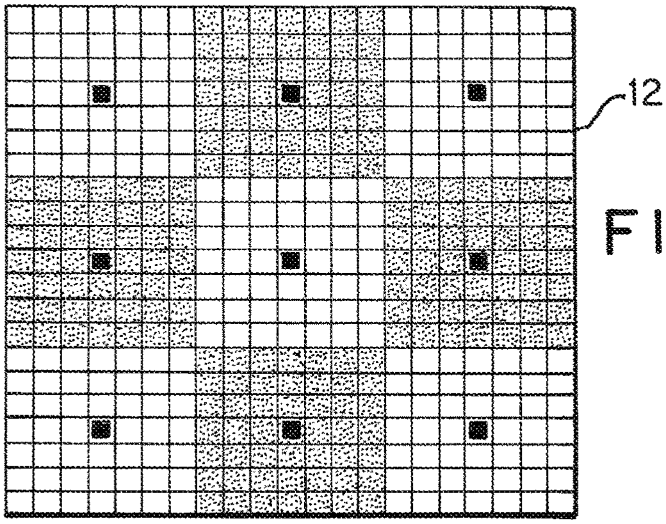


FIG. 2

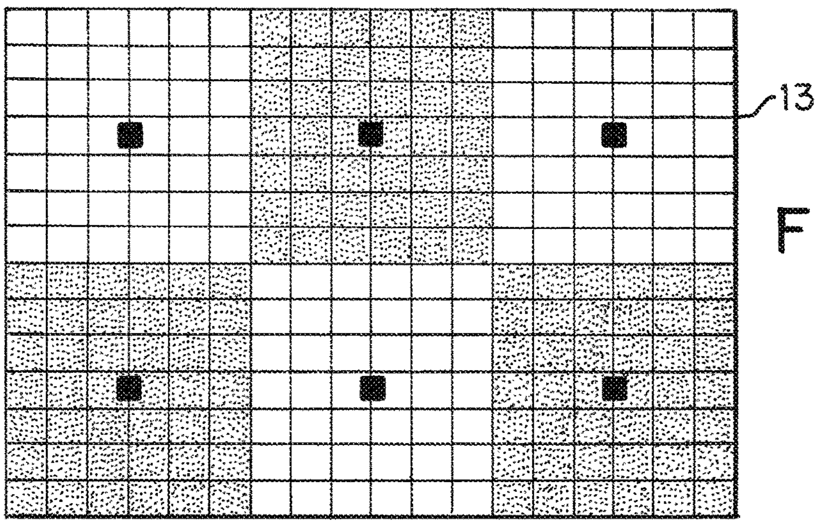


FIG. 3

FIG. 4A

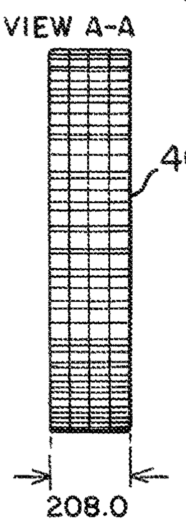


FIG. 4B

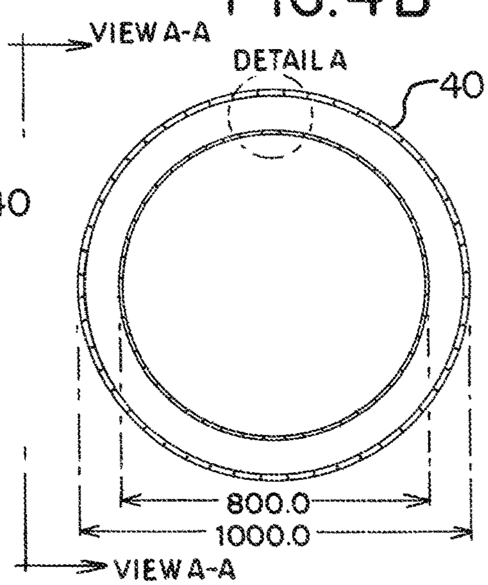


FIG. 4C

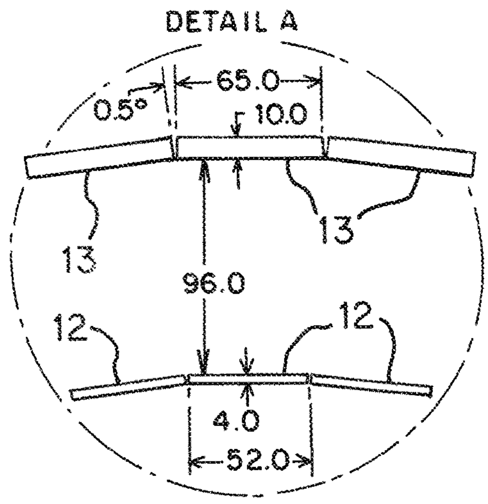


FIG. 5

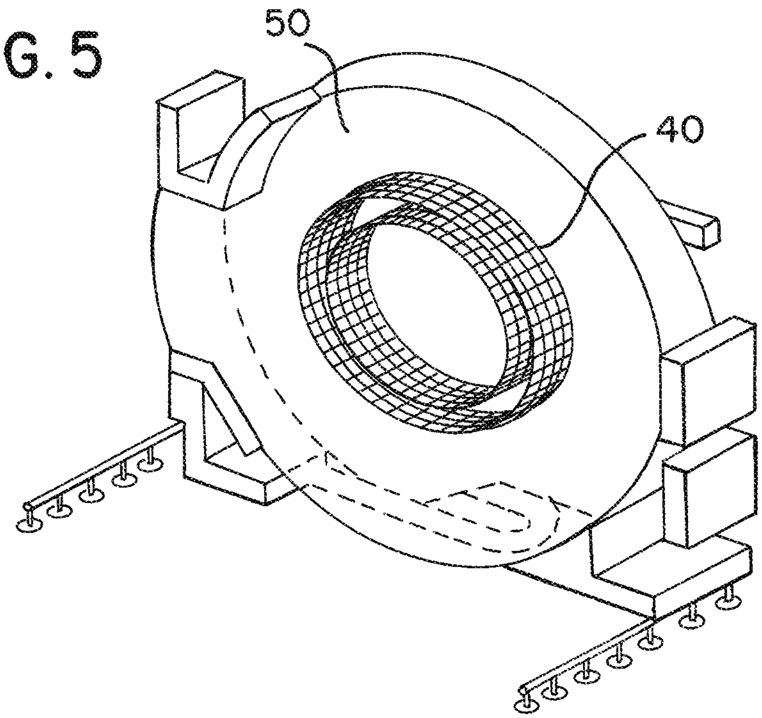


FIG. 6

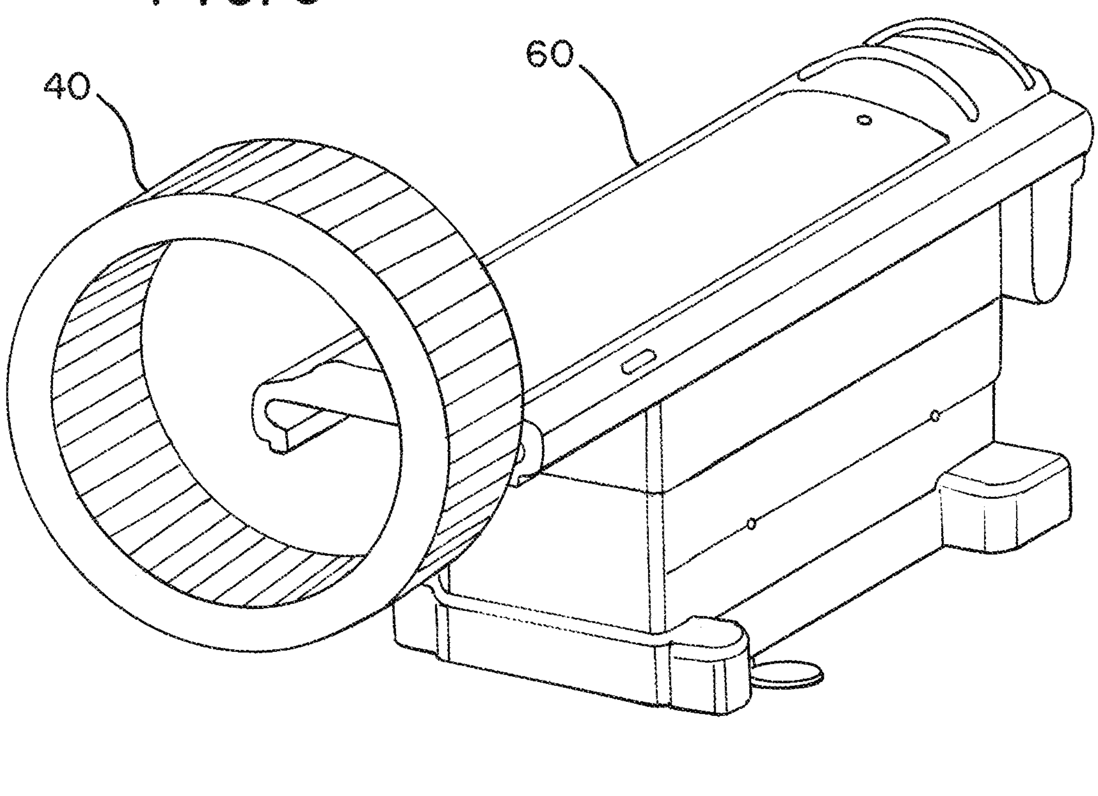


FIG. 7

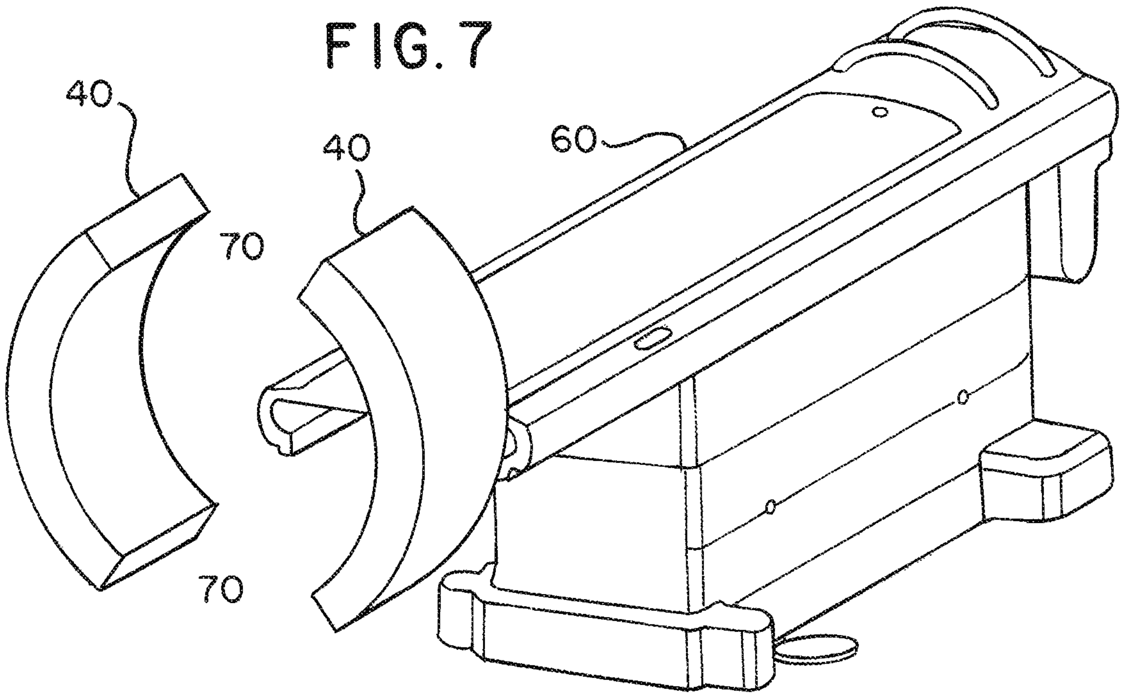


FIG. 8

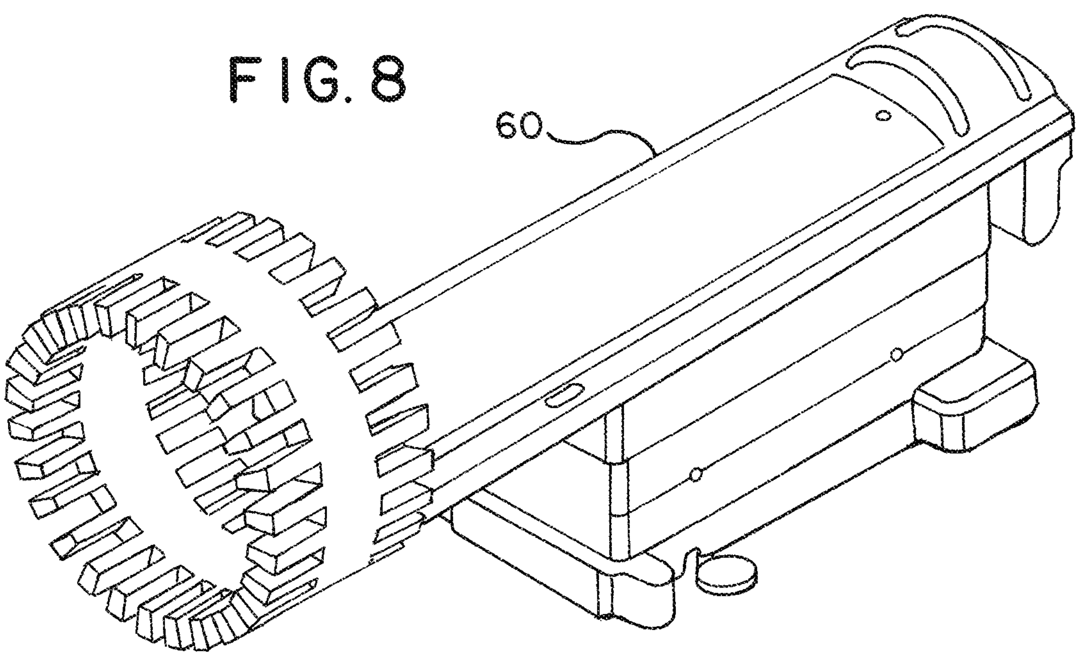


FIG. 9

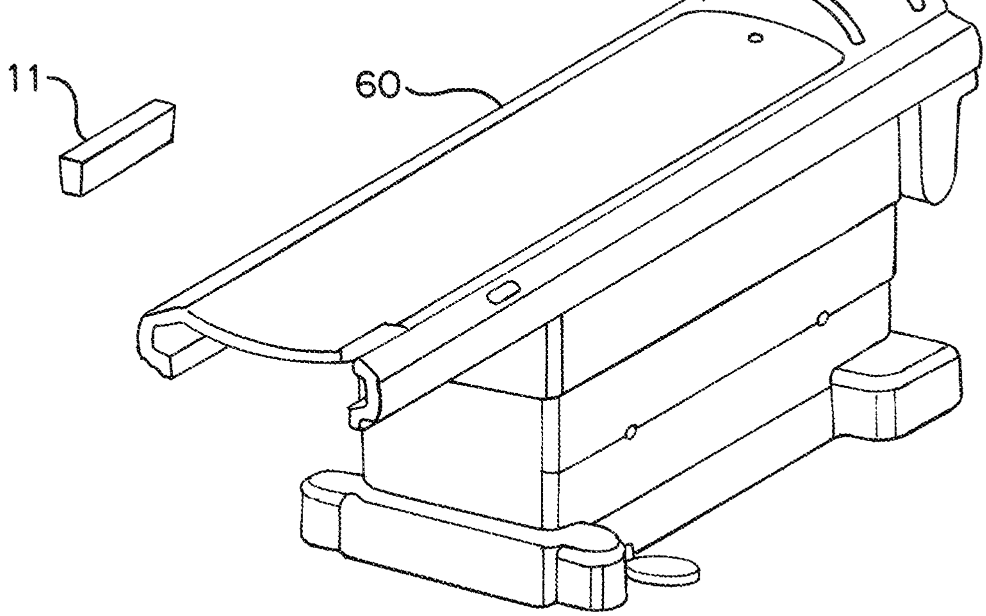

11
60

FIG. 10

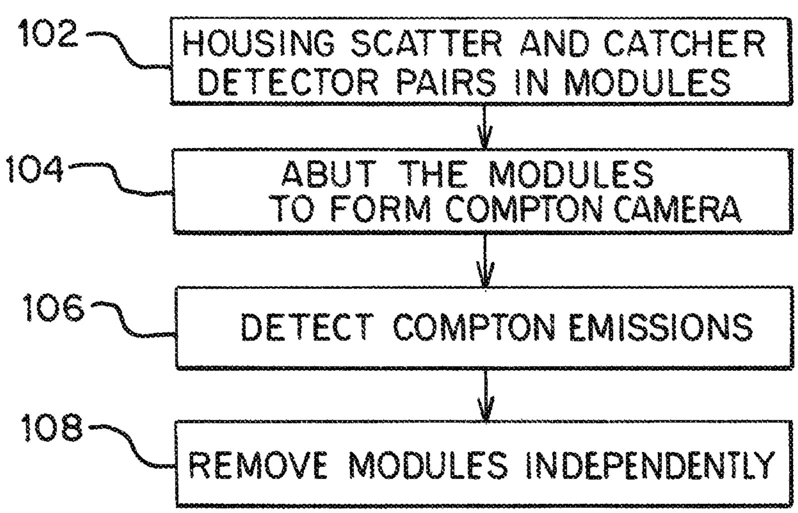

102
HOUSING SCATTER AND CATCHER DETECTOR PAIRS IN MODULES
104
ABUT THE MODULES TO FORM COMPTON CAMERA
106
DETECT COMPTON EMISSIONS
108
REMOVE MODULES INDEPENDENTLY

FIG. 11

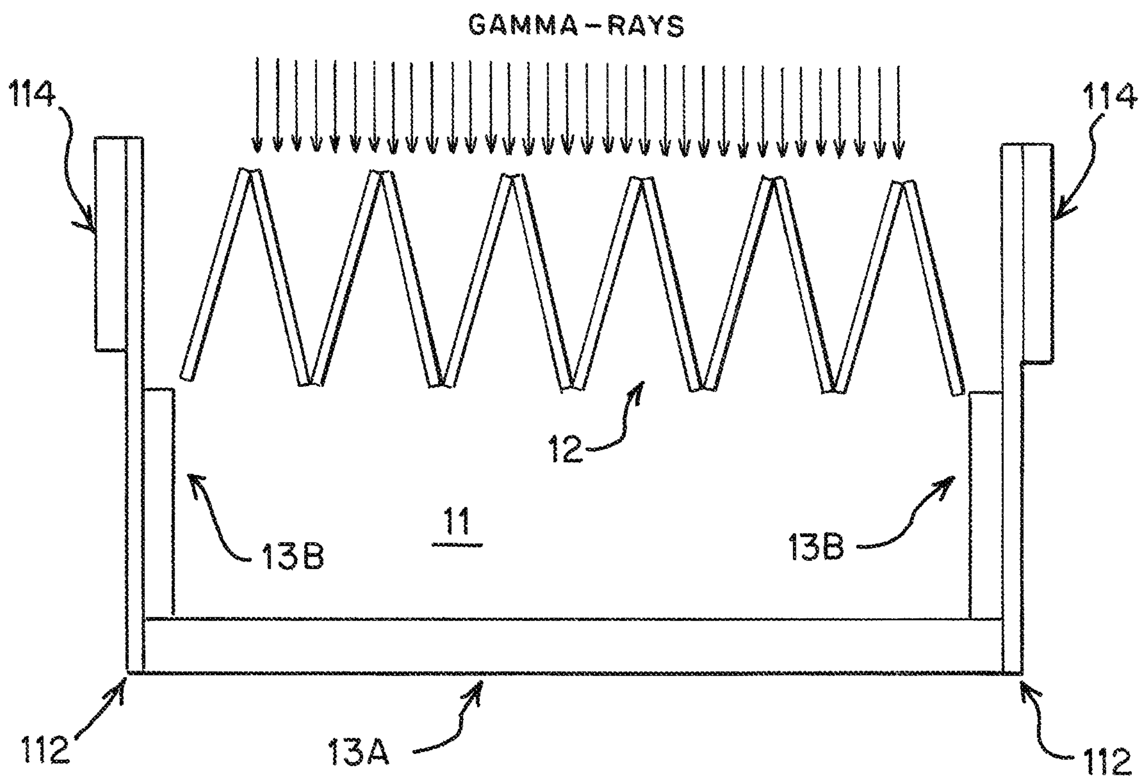


FIG. 12A
FRONT-VIEW
BUMP-BONDED ASIC

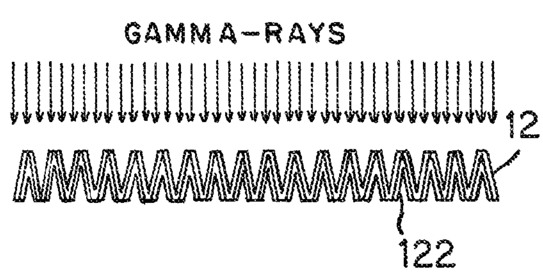


FIG. 12B
FRONT-VIEW
FLEX-CABLE ASIC

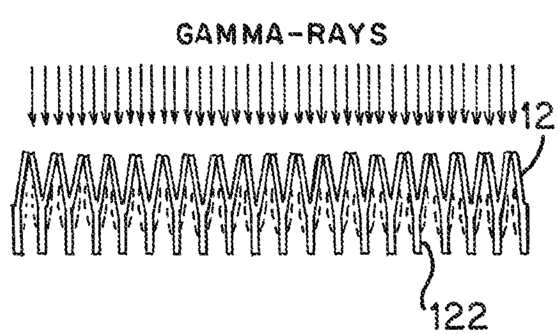


FIG. 13A

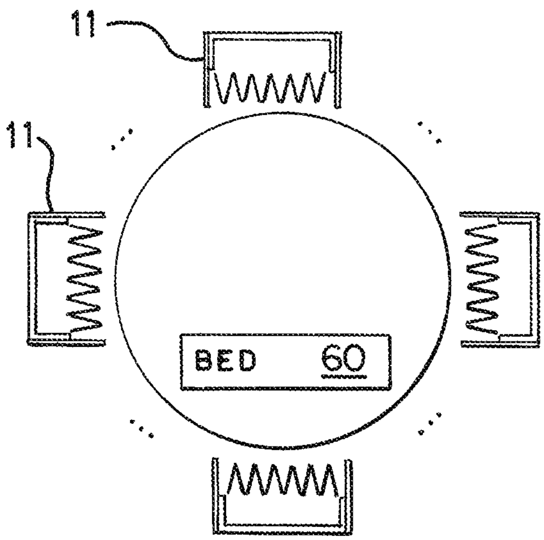


FIG. 13B

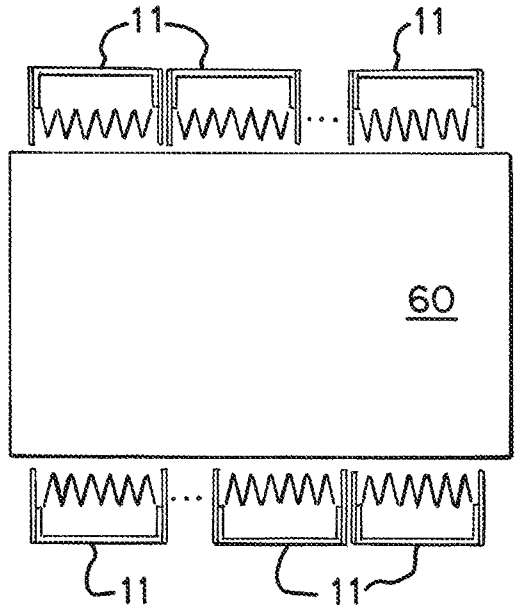


FIG. 14A

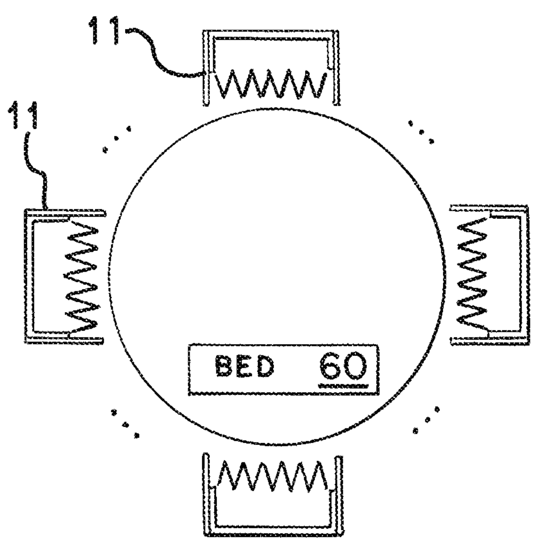


FIG. 14B

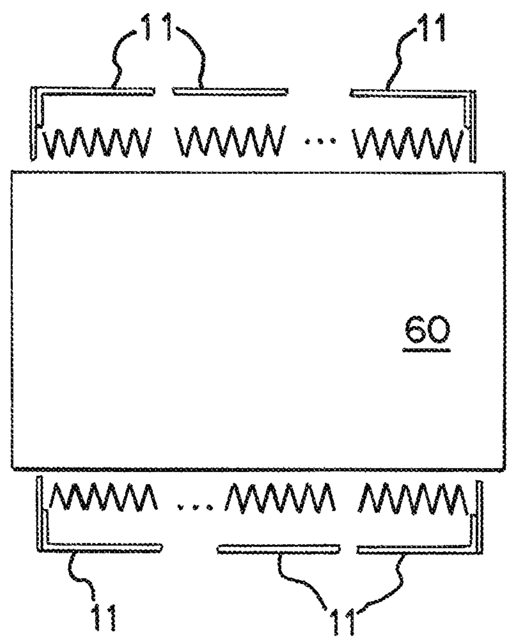


FIG. 15

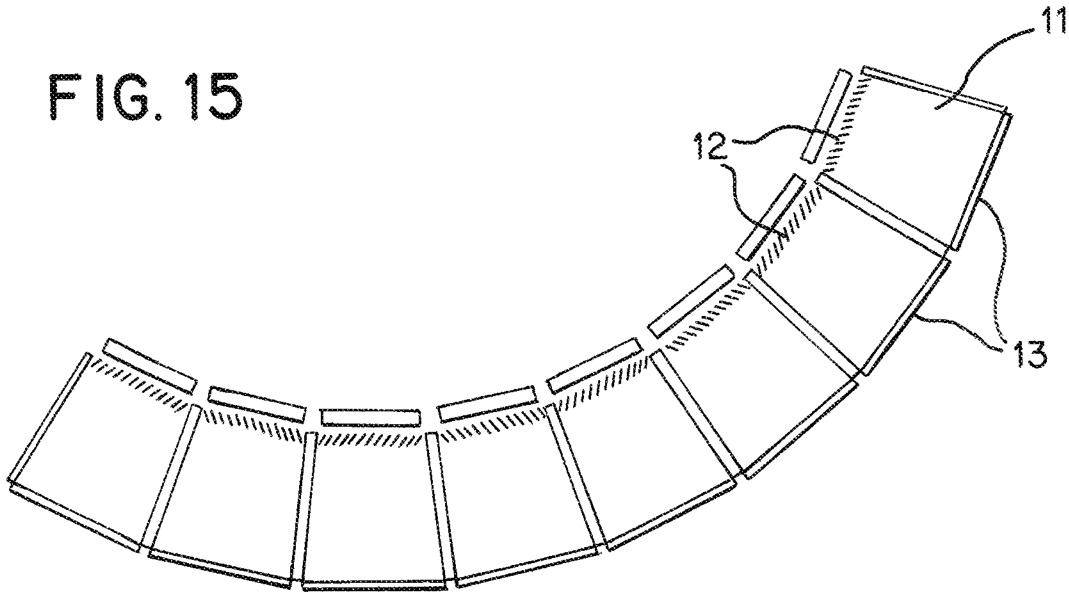


FIG.16A

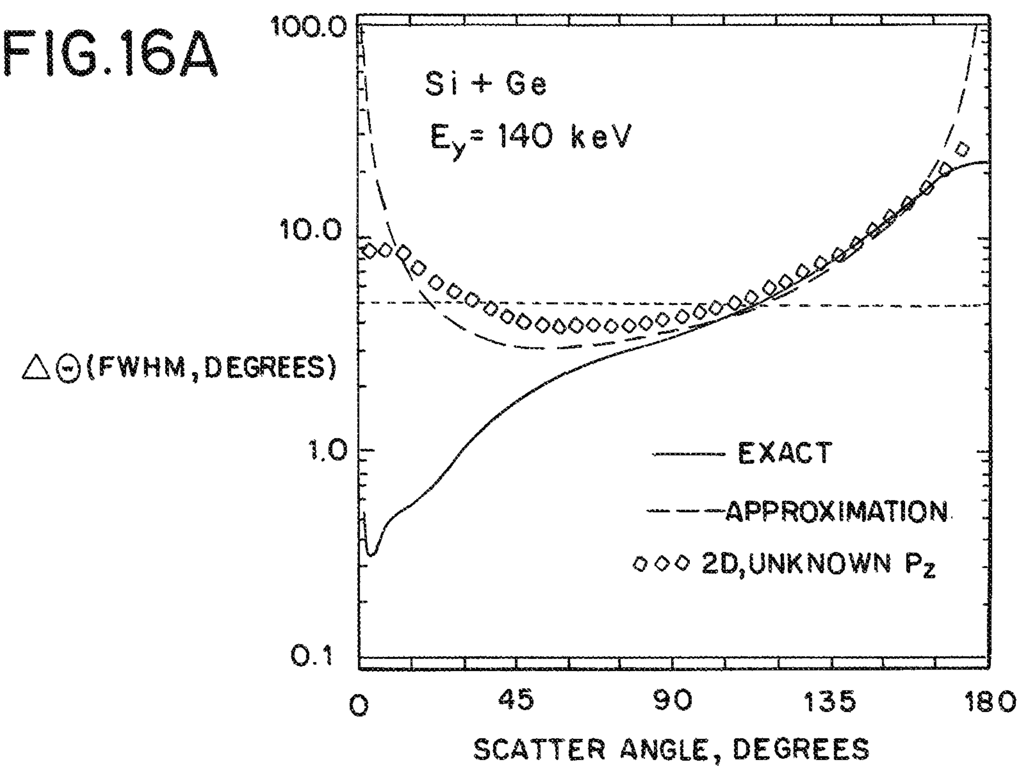
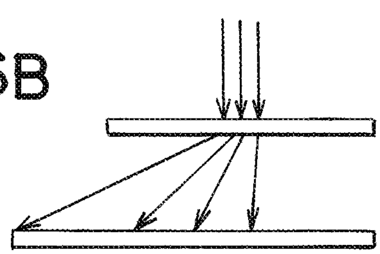


FIG. 16B



NEAR 2Pi COMPTON CAMERA FOR MEDICAL IMAGING

BACKGROUND

The present embodiments relate to medical imaging using the Compton effect. The Compton effect allows for imaging higher energies than used for single photon emission computed tomography (SPECT). Compton imaging systems are constructed as test platforms, such as assembling a scatter layer and then a catcher layer mounted to a large framework. Electronics are connected to detect Compton-based events from emissions of a phantom. Compton imaging systems have failed to address design and constraint requirements for practical use in any commercial clinical settings. Current proposals lack the ability to be integrated into imaging platforms in the clinic or lack the design and constraint requirements (i.e., flexibility and scalability) to address commercial and diagnostic needs.

Compton-cameras may have low sensitivity ($) and poor image quality (IQ). The absolute number of scattered photons in the scatter layer is low due to the geometry (e.g., source-scatter solid angle $\Omega<<4\pi$), material (e.g., low scatter fraction in the detection material which favors photoelectric effect), and detector fabrication limitations (e.g., practical detector thickness that can be manufactured for both scatter and catcher layers is bounded, such as a maximum of ~1 mm for Si detectors and 2 mm . . . 10 mm for CZT detectors). The number of caught scattered photons in the catcher layer is low due to geometry (e.g., scatter-catcher solid angle $\Omega<<4\pi$). Doppler broadening degrades image quality of Compton cameras. The contribution of Doppler broadening to the Compton angle uncertainty depends on incident photon energy E_0, scattered angle θ, and the energy of moving electrons bound to the target atom. Limited detector energy resolution causes additional Compton angle uncertainties. Limited detector position resolution in both scatter and catcher layers causes additional Compton cone annular offsets.

SUMMARY

By way of introduction, the preferred embodiments described below include methods and systems for medical imaging. To capture more emitted photons with a Compton camera, the scatter detector is tilted (non-orthogonal angle) relative to a radial from the isocenter of the imaging system. The tilt creates a greater volume for scatter interaction. To capture more scatter photons, the catcher detector is non-planar, such as a multi-faced detector at least partially surrounding a volume behind the scatter detector. The tilted scatter detector alone, the non-planar catcher detector alone, or the tilted scatter detector and the non-planar catcher detector are used in the Compton camera.

In a first aspect, a Compton camera is provided for medical imaging. A bed is for a patient space having an iso center axis. A first module has a first scatter detector and a first catcher detector spaced from the first scatter detector. The first scatter detector has an outer surface facing the iso center axis where the outer surface is away from orthogonal by an angle of at least 20 degrees to a radial line extending perpendicular from the iso center axis through a center of the first scatter detector. The first catcher detector forms a substantially semi-spherical surround behind the first scatter detector relative to the patient space. An image processor is configured to determine angles of incidence for Compton events from the first scatter detector and the first catcher detector.

In a second aspect, a medical imaging system includes a Compton camera with a scatter detector arranged to receive emissions from a patient. The scatter detector has an outer surface facing the patient where the outer surface is away from orthogonal by an angle of at least 20 degrees to a radial line extending perpendicular from a longitudinal axis of the patient through the scatter detector.

In a third aspect, a medical imaging system includes a Compton camera with a scatter detector and a catcher detector. The scatter detector is arranged to receive emissions from a patient. The catcher detector is arranged to receive scatter from the scatter detector due to the emissions from a patient. The catcher detector includes a multi-sided detection surface positioned behind the scatter detector relative to the patient.

The present invention is defined by the following claims, and nothing in this section should be taken as a limitation on those claims. Further aspects and advantages of the invention are discussed below in conjunction with the preferred embodiments and may be later claimed independently or in combination.

BRIEF DESCRIPTION OF THE DRAWINGS

The components and the figures are not necessarily to scale, emphasis instead being placed upon illustrating the principles of the invention. Moreover, in the figures, like reference numerals designate corresponding parts throughout the different views.

FIG. **1** is perspective view of multiple modules of a Compton camera according to one embodiment;

FIG. **2** illustrates an example scatter detector;

FIG. **3** illustrates an example catcher detector;

FIG. **4**A is a side view of one embodiment of a Compton camera,

FIG. **4**B is an end view of the Compton camera of FIG. **4**A, and FIG. **4**C is a detail view of a part of the Compton camera of FIG. **4**B;

FIG. **5** is a perspective view of one embodiment of a Compton camera in a medical imaging system;

FIG. **6** is a perspective view of one embodiment of a full-ring Compton camera in a medical imaging system;

FIG. **7** is a perspective view of one embodiment of a partial-ring Compton camera in a medical imaging system;

FIG. **8** is a perspective view of one embodiment of a full-ring Compton camera with partial-rings in axial extension in a medical imaging system;

FIG. **9** is a perspective view of one embodiment of a single module-based Compton camera in a medical imaging system;

FIG. **10** is a flow chart diagram of an example embodiment of a method for forming a Compton camera;

FIG. **11** illustrates one embodiment of a module with a tilted scatter detector and a near 2π catcher detector;

FIG. **12**A illustrates one embodiments of a titled scatter detector with parallel application specific integrated circuits, and FIG. **12**B illustrates the tilted scatter detector with the application specific integrated circuits in a non-parallel arrangement;

FIGS. **13**A and B show orthogonal cross-sections of a multi-ring configuration of modules in a Compton camera according to a first embodiment;

FIGS. **14**A and B show orthogonal cross-sections of a multi-ring configuration of modules in a Compton camera according to a second embodiment;

FIG. **15** illustrates different tilting of scatter detectors in different modules according to one embodiment; and

FIG. **16**A shows an example graph of full width, half maximum (FWHM) by scatter angle for Compton imaging, and FIG. **16**B shows example scatter angles.

DETAILED DESCRIPTION OF THE DRAWINGS AND PRESENTLY PREFERRED EMBODIMENTS

FIGS. **1-9** are directed to a multi-modality compatible Compton camera. A modular design is used to form the Compton camera for use with various other imaging modalities. FIGS. **11-15** are directed to a Compton camera with a tilted scatter detector and/or near 2π catcher detector. The titled scatter detector and/or near 2π catcher detector are used in modules of FIGS. **1-9**, other modules, or without modules. After a summary of the tilted scatter detector and/or near 2π catcher detector embodiments, the Compton camera of FIGS. **1-9** is described. Many of the features and components of the Compton camera of FIGS. **1-9** may be used in the tilted scatter detector and/or near 2π catcher detector embodiments later described for FIGS. **11-15**.

A more efficient Compton camera is provided by the tilted scatter detector and/or near 2π catcher detector. Sensitivity ($) and/or image quality (IQ) may be improved. Synchronization and triggering limitations between modules may be avoided by capturing photons at higher rates within a module. Tilting the scatter detector and/or using a near 2π catcher detector may improve the sensitivity ($) as compared to the parallel plate scatter and catcher detectors of FIG. **1** by ~15 times. The absolute number of scattered photons may be increased by ~3-5 times using the tilted scatter detector, and the number of caught photons may be increased by ~3-5 times using the near 2π catcher detector.

The tilted scatter detector and/or near 2π catcher detector may be applied to any Compton-Camera regardless of the detection materials used, readout electronics and/or size of imaging object. The design configuration of each module may be re-arranged and optimized for different imaging tasks, assuming a quantized number of different modules that may be swapped in the system for different tasks during design. Using modularized smaller Compton-cameras forming a larger imaging system, with reduced or near zero cross-talk between modules due to shielding, a lower requirement for electronics (e.g., ASIC/FPGA) cross-talk and inter-module triggers at high rates results.

Referring to FIGS. **1-9**, a medical imaging system includes a multi-modality compatible Compton camera with segmented detection modules. The Compton camera, such as a Compton camera ring, is segmented into modules that house the detection units. Each module is independent, and when assembled into a ring or partial ring, the modules may communicate with each other. The modules are independent yet can be assembled into a multi-module unit that produces Compton scattering-based images. Cylindrically symmetric modules or spherical shell segmented modules may be used.

The scatter-catcher pair, modular arrangement allows efficient manufacturing, is serviceable in the field, and is cost and energy efficient. The modules allow for the design freedom to change the radius for each radial detection unit, angular span of one module, and/or axial span. The scatter-catcher pair modules are multi-modality compatible and/or form a modular ring Compton camera for clinical emission imaging. This design allows flexibility, so the Compton camera may be added to existing computed tomography (CT), magnetic resonance (MR), positron emission tomography (PET) or other medical imaging platforms, either as axially separated systems or as fully integrated systems. Each module may address heat dissipation, data collection, calibration, and/or allow for efficient assembly as well as servicing.

Each scatter-catcher paired module is formed from commercially suitable solid-state detector modules (e.g., Si, CZT, CdTe, HPGe or similar), allowing for an energy range of 100-3000 keV. Compton imaging may be provided with a wider range of isotope energies (>2 MeV), enabling new tracers/markers through selection of the scatter-catcher detectors. The modularity allows for individual module removal or replacement, allowing for time and cost-efficient service. The modules may be operated independently and isolated or may be linked for cross-talk, allowing for improved image quality and higher efficiency in detecting Compton events using a scatter detector of one module and a catcher detector of another module.

The modularity allows for flexible design geometry optimized to individual requirements, such as using a partial ring for integration with a CT system (e.g., connected between the x-ray source and detector), a few modules (e.g., tiling) used for integration with a single photon emission computed tomography gamma camera or other space limited imaging system, or a full ring. Functional imaging based on Compton-detected events may be added to other imaging systems (e.g., CT, MR, or PET). Multiple full or partial rings may be placed adjacent to each other for greater axial coverage of the Compton camera. A dedicated or stand-alone Compton-based imaging system may be formed. In one embodiment, the modules include a collimator for lower energies (e.g., <300 keV), providing for multichannel and multiplexed imaging (e.g., high energies using the scatter-catcher detectors for Compton events and low energies using one of the detectors for SPECT or PET imaging). The modules may be stationary or fast rotating (0.1 rpm<<ω<<240 rpm). The dimensional, installation, service, and/or cost constraints are addressed by the scatter-catcher paired modules.

FIG. **1** shows one embodiment of modules **11** for a Compton camera. Four modules **11** are shown, but additional or fewer modules may be used. The Compton camera is formed from one or more modules, depending on the desired design of the Compton camera.

The Compton camera is for medical imaging. A space for a patient relative to the modules is provided so that the modules are positioned to detect photons emitted from the patient. A radiopharmaceutical in the patient includes a radio-isotope. A photon is emitted from the patient due to decay from the radio-isotope. The energy from the radio-isotope may be 100-3000 keV, depending on the material and structure of the detectors. Any of various radio-isotopes may be used for imaging a patient. Modules **11** optimized for different isotopes may be interleaved to cover any range (e.g., the entire) of energy spectrum. For example, a 1^{st} module for 100-400 keV, a second module for 300-600 keV, a third module for 500-above, a 4th module for 100-400 keV, . . . covering the entire full ring and/or partially populating the ring.

Each of the modules **11** includes the same or many of the same components. A scatter detector **12**, a catcher detector **13**, circuit boards **14**, and baffle **15** are provided in a same housing **21**. Additional, different, or fewer components may be provided. For example, the scatter detector **12** and catcher detector **13** are provided in the housing **21** without other components. As another example, a fiber optic data line **16** is provided in all or a sub-set of the modules **11**.

The modules **11** are shaped for being stacked together. The modules **11** mate with each other, such as having matching indentation and extensions, latches, tongue-and-grooves, or clips. In other embodiments, flat or other surfaces are provided for resting against each other or a divider. Latches, clips, bolts, tongue-and-groove or other attachment mechanisms for attaching a module **11** to any adjacent modules **11** are provided. In other embodiments, the module **11** attaches to a gantry or other framework with or without direct connection to any adjacent modules **11**.

The connection or connections to the other modules **11** or gantry may be releasable. The module **11** is connected and may be disconnected. The connection may be releasable, allowing removal of one module **11** or a group of modules **11** without removing all modules **11**.

For forming a Compton camera from more than one module **11**, the housing **21** and/or outer shape of the modules **11** is wedge shaped. The modules **11** may be stacked around an axis to form a ring or partial ring due to the wedge shape. The part closer to the axis has a width size that is narrower along a dimension perpendicular to the axis than a width size of a part further from the axis. In the modules **11** of FIG. **1**, the housings **21** have the widest part furthest from the axis. In other embodiments, the widest part is closer to the axis but spaced away from the narrowest part closest to the axis. In the wedge shape, the scatter detector **12** is nearer to the narrower part of the wedge shape than the catcher detector **13**. This wedge shape in cross-section along a plane normal to the axis allows stacking of the modules **11** in abutting positions, adjacently, and/or connected to form at least part of a ring about the axis.

The taper of the wedge provides for a number N of modules **11** to form a complete ring around the axis. Any number N may be used, such as N=10-30 modules. The number N may be configurable, such as using different housings **21** for different numbers N. The number of modules **11** used for a given Compton camera may vary, depending on the design of the Compton camera (e.g., partial ring). The wedge shape may be provided along other dimensions, such as having a wedge shape in a cross-section parallel to the axis.

The modules **11** as stacked are cylindrically symmetric as connected with a gantry of a medical imaging system. A narrowest end of the wedged cross-section is closest to a patient space of the medical imaging system and a widest end of the wedged cross-section may be furthest from the patient space. In alternative embodiments, other shapes than wedge allowing for stacking together to provide a ring or generally curved shape of the stack may be provided.

The housing **21** is metal, plastic, fiberglass, carbon (e.g., carbon fiber), and/or other material. In one embodiment, different parts of the housing **21** are of different materials. For example, tin is used for the housing around the circuit boards **14**. Aluminum is used to hold the scatter detector **12** and/or catcher detector **13**. In another example, the housing **12** is of the same material, such as aluminum.

The housing **21** may be formed from different structures, such as end plates having the wedge shape, sheets of ground plane housing the circuit boards **14**, and separate structure for walls holding the scatter detector **12** and catcher detector **13** where the separate structure is formed of material through which photons of a desired energy from a Compton event may pass (e.g., aluminum or carbon fiber). In alternative embodiments, walls are not provided for the modules **11** between the end plates for a region where the scatter detector **12** and/or catcher detector **13** are positioned, avoiding interference of photons passing from the scatter detector **12** of one module **11** to a catcher detector **13** of another module **11**. The housing **21** by and/or for holding the detectors **12**, **13** is made of low attenuating material, such as aluminum or carbon fiber.

The housing **21** may seal the module or includes openings. For example, openings for air flow are provided, such as at a top of widest portion of the wedge shape at the circuit boards **14**. The housing **21** may include holes, grooves, tongues, latches, clips, stand-offs, bumpers, or other structures for mounting, mating, and/or stacking.

Each of the solid-state detector modules **11** includes both scatter and catcher detectors **12**, **13** of a Compton sensor. By stacking each module, the size of the Compton sensor is increased. A given module **11** itself may be a Compton sensor since both the scatter detector **12** and catcher detector **13** are included in the module.

The modules **11** may be separately removed and/or added to the Compton sensor. For a given module **11**, the scatter detector **12** and/or catcher detector **13** may be removable from the module **11**. For example, a module **11** is removed for service. A faulty one or both detectors **12**, **13** are removed from the module **11** for replacement. Once replaced, the refurbished module **11** is placed back in the medical imaging system. Bolts, clips, latches, tongue-and-groove, or other releasable connectors may connect the detectors **12**, **13** or part of the housing **21** for the detectors **12**, **13** to the rest of the module **11**.

The scatter detector **12** is a solid-state detector. Any material may be used, such as Si, CZT, CdTe, HPGe, and/or other material. The scatter detector **12** is created with wafer fabrication at any thickness, such as about 4 mm for CZT. Any size may be used, such as about 5×5 cm. FIG. **2** shows a square shape for the scatter detector **12**. Other shapes than square may be used, such as rectangular. For the modules **11** of FIG. **1**, the scatter detector **12** may be rectangular extending between two wedge-shaped end-plates.

In the module **11**, the scatter detector **12** has any extent. For example, the scatter detector **12** extends from one wedge-shaped end wall to the other wedge-shaped end wall. Lesser or greater extent may be provided, such as extending between mountings within the module **11** or extending axially beyond one or both end-walls. In one embodiment, the scatter detector **12** is at, on, or by one end wall without extended to another end wall.

The scatter detector **12** forms an array of sensors. For example, the 5×5 cm scatter detector **12** of FIG. **2** is a 21×21 pixel array with a pixel pitch of about 2.2 mm. Other numbers of pixels, pixel pitch, and/or size of arrays may be used.

The scatter detector **12** includes semiconductor formatted for processing. For example, the scatter detector **12** includes an application specific integrated circuit (ASIC) for sensing photon interaction with an electron in the scatter detector **12**. The ASIC is collocated with the pixels of the scatter detector **12**. The ASIC is of any thickness. A plurality of ASICs may be provided, such as 9 ASICS in a 3×3 grid of the scatter detector **12**.

The scatter detector **12** may operate at any count rate, such as >100 kcps/mm. Electricity is generated by a pixel due to the interaction. This electricity is sensed by the application specific integrated circuit. The location, time, and/or energy is sensed. The sensed signal may be conditioned, such as amplified, and sent to one or more of the circuit boards **14**. A flexible circuit, wires, or other communications path carries the signals from the ASIC to the circuit board **14**.

Compton sensing operates without collimation. Instead, a fixed relationship between energy, position, and angle of a photon interaction at the scatter detector **12** relative to a photon interaction at the catcher detector **13** is used to determine the angle of the photon entering the scatter detector **12**. A Compton process is applied using the scatter detector **12** and the catcher detector **13**.

The catcher detector **13** is a solid-state detector. Any material may be used, such as Si, CZT, CdTe, HPGe, and/or other material. The catcher detector **13** is created with wafer fabrication at any thickness, such as about 10 mm for CZT. Any size may be used, such as about 5×5 cm. The size may be larger along at least one dimension than the scatter detector **12** due to the wedge shape and spaced apart positions of the scatter detector **12** and the catcher detector **13**. FIG. **3** shows a rectangular shape for the catcher detector **13** but other shapes may be used. For the modules **11** of FIG. **1**, the catcher detector **13** may be rectangular extending between two end-plates where the length is the same as and the width is greater than the scatter detector **12**.

The catcher detector **12** forms an array of sensors. For example, the 5×6 cm catcher detector **13** of FIG. **3** is a 14×18 pixel array with a pixel pitch of about 3.4 mm. The pixel size is larger than the pixel size of the scatter detector **12**. The number of pixels is less than the number of pixels of the scatter detector **12**. Other numbers of pixels, pixel pitch, and/or size of arrays may be used. Other relative pixels sizes and/or numbers of pixels may be used.

In the module **11**, the catcher detector **13** has any extent. For example, the catcher detector **13** extends from one wedge-shaped end wall to the other wedge-shaped end wall. Lesser or greater extent may be provided, such as extending between mountings within the module **11** or extending axially beyond one or both end-walls. In one embodiment, the catcher detector **13** is at, on, or by one end wall without extending to another end wall.

The catcher detector **13** includes semiconductor formatted for processing. For example, the catcher detector **13** includes an ASIC for sensing photon interaction with an electron in the catcher detector **13**. The ASIC is collocated with the pixels of the catcher detector **13**. The ASIC is of any thickness. A plurality of ASICS may be provided, such as 6 ASICS in a 2×3 grid of the catcher detector **13**.

The catcher detector **13** may operate at any count rate, such as >100 kcps/mm. Electricity is generated by a pixel due to the interaction. This electricity is sensed by the ASIC. The location, time, and/or energy is sensed. The sensed signal may be conditioned, such as amplified, and sent to one or more of the circuit boards **14**. A flexible circuit, wires, or other communications path carries the signals from the ASIC to the circuit board **14**.

The catcher detector **13** is spaced from the scatter detector **12** by any distance along a radial line from the axis or normal to the parallel scatter and catcher detectors **12**, **13**. In one embodiment, the separation is about 20 cm, but greater or lesser separation may be provided. The space between the catcher detector **13** and the scatter detector **12** is filled with air, other gas, and/or other material with low attenuation for photons at the desired energies.

The circuit boards **14** are printed circuit boards, but flexible circuits or other materials may be used. Any number of circuit boards **14** for each module may be used. For example, one circuit board **14** is provided for the scatter detector **12** and another circuit board **14** is provided for the catcher detector **13**.

The circuit boards **14** are within the housing **21** but may extend beyond the housing **21**. The housing **21** may be grounded, acting as a ground plane for the circuit boards **14**. The circuit boards **14** are mounted in parallel with each other or are non-parallel, such as spreading apart in accordance with the wedge shape. The circuit boards are positioned generally orthogonal to the catcher detector **13**. Generally is used to account for any spread due to the wedge shape. Brackets, bolts, screws, and/or stand-offs from each other and/or the housing **21** are used to hold the circuit boards **14** in place.

The circuit boards **14** connect to the ASICS of the scatter and catcher detectors **12**, **13** through flexible circuits or wires. The ASICs output detected signals. The circuit boards **14** are acquisition electronics, which process the detected signals to provide parameters to the Compton processor **19**. Any parameterization of the detected signals may be used. In one embodiment, the energy, arrival time, and position in three-dimensions is output. Other acquisition processing may be provided.

The circuit boards **14** output to each other, such as through a galvanic connection within a module **11**, to the data bridge **17**, and/or to a fiber optic data link **16**. The fiber data link **16** is a fiber optic interface for converting electrical signals to optical signals. A fiber optic cable or cables provide the acquisition parameters for events detected by the scatter and catcher detectors **12**, **13** to the Compton processor **19**.

The data bridge **17** is a circuit board, wires, flexible circuit, and/or other material for galvanic connection to allow communications between modules **11**. A housing or protective plate may cover the data bridge **17**. The data bridge **17** releasably connects to one or more modules **11**. For example, plugs or mated connectors of the data bridge **17** mate with corresponding plugs or mated connectors on the housing **21** and/or circuit boards **14**. A latch, clip, tongue-and-groove, screw, and/or bolt connection may be used to releasably hold the data bridge **17** in place with the modules **11**.

The data bridge **17** allows communications between the modules. For example, the fiber data link **16** is provided in one module **11** and not another module **11**. The cost of a fiber data link **16** in every module **11** is avoided. Instead, the parameters output by the other module **11** are provided via the data bridge **17** to the module **11** with the fiber data link **16**. The circuit board or boards **14** of the module **11** with the fiber data link **16** route the parameter output to the fiber data link **16**, using the fiber data link **16** to report detected events from more than one module **11**. In alternative embodiments, each module **11** includes a fiber data link **16**, so the data bridge **17** is not provided or communicates other information.

The data bridge **17** may connect other signals between the modules **11**. For example, the data bridge **17** includes a conductor for power. Alternatively, a different bridge provides power to the modules **11** or the modules **11** are individually powered. As another example, clock and/or synchronization signals are communicated between modules **11** using the data bridge **17**.

In the embodiment of FIG. **1**, a separate clock and/or synchronization bridge **18** is provided. The clock and/or synchronization bridge **18** is a circuit board, wires, flexible circuit, and/or other material for galvanic connection to allow communication of clock and/or synchronization signals between modules **11**. A housing or protective plate may cover the clock and/or synchronization bridge **18**. The clock and/or synchronization bridge **18** releasably connects to one or more modules **11**. For example, plugs or mated connectors of the clock and/or synchronization bridge **18** mate with corresponding plugs or mated connectors on the housing **21** and/or circuit boards **14**. A latch, clip, tongue-and-groove, screw, and/or bolt connection may be used to releasably hold the clock and/or synchronization bridge **18** in place with the modules **11**.

The clock and/or synchronization bridge **18** may connect with the same or different grouping of modules **11** as the data bridge **17**. In the embodiment shown in FIG. **1**, the data bridge **17** connects between pairs of modules **11** and the clock and/or synchronization bridge **18** connects over groups of four modules **11**.

The clock and/or synchronization bridge **18** provides a common clock signal and/or synchronization signals for synchronizing clocks of the modules **11**. One of the parameters formed by the circuit boards **14** of each module **11** is the time of detection of the event. Compton detection relies on pairs of events—a scatter event and a catcher event. Timing is used to pair events from the different detectors **12**, **13**. The common clocking and/or synchronization allows for accurate pairing where the pair of events are detected in different modules **11**. In alternative embodiments, only scatter and catcher events detected in a same module **11** are used, so the clock and/or synchronization bridge **18** may not be provided.

Other links or bridges between different modules **11** may be provided. Since the bridges **17**, **18** are removable, individual modules **11** may be removed for service while leaving remaining modules **11** in the gantry.

Each module **11** is air cooled. Holes may be provided for forcing air through the module **11** (i.e., entry and exit holes). One or more baffles **15** may be provided to guide the air within the module **11**. Water, conductive transfer, and/or other cooling may be alternatively or additionally provided.

In one embodiment, the top portion of the wedge-shape module **11** or housing **21** is open (i.e., no cover on the side furthest from the patient area). One or more baffles **15** are provided along the centers of one or more circuit boards **14** and/or the housing **21**. A fan and heat exchanger **20** force cooled or ambient temperature air into each module **11**, such as along one half of the module **11** at a location spaced away from the catcher detector **13** (e.g., top of the module **11**). The baffles **15** and/or circuit boards **14** guide at least some of the air to the airspace between the scatter detector **12** and the catcher detector **13**. The air then passes by the baffles **15** and/or circuit boards **14** on another part (e.g., another half) of the module **11** for exiting to the heat exchanger **20**. Other routing of the air may be provided.

The heat exchanger and fan **20** is provided for each individual module **11**, so may be entirely or partly within the module **11**. In other embodiments, ducting, baffles, or other structure route air to multiple modules **11**. For example, groups of four modules **11** share a common heat exchanger and fan **20**, which is mounted to the gantry or other framework for cooling the group of modules **11**.

For forming a Compton sensor, one or more modules **11** are used. For example, two or more modules **11** are positioned relative to a patient bed or imaging space to detect photon emissions from the patient. An arrangement of a greater number of modules **11** may allow for detection of a greater number of emissions. By using the wedge shape, modules **11** may be positioned against, adjacent, and/or connected with each other to form an arc about the patient space. The arc may have any extent. The modules **11** directly contact each other or contact through spacers or the gantry with small separation (e.g., 10 cm or less) between the modules **11**.

In one example, four modules **11** are positioned together, sharing a clock and/or synchronization bridge **18**, one or more data bridges **17**, and a heat exchanger and fan **20**. One, two, or four fiber data links **16** are provided for the group of modules **11**. Multiple such groups of modules **11** may be positioned apart or adjacent to each other for a same patient space.

Due to the modular approach, any number of modules **11** may be used. Manufacturing is more efficient and costly by building multiple of the same component despite use of any given module **11** in a different arrangement than used for others of the modules **11**.

The fiber data links **16** of the modules **11** or groups of modules **11** connect to the Compton processor **19**. The Compton processor **19** receives the values for the parameters for the different events. Using the energy and timing parameters, scatter and catcher events are paired. For each pair, the spatial locations and energies of the pair of events are used to find the angle of incidence of the photon on the scatter detector **12**. The event pairs are limited to events in the same module **11** in one embodiment. In another embodiment, catcher events from the same or different modules **11** may be paired with scatter events from a given module **11**. More than one Compton processor **19** may be used, such as for pairing events from different parts of a partial ring **40**.

Once paired events are linked, the Compton processor **19** or another processor may perform computed tomography to reconstruct a distribution in two or three dimensions of the detected emissions. The angle or line of incidence for each event is used in the reconstruction. The reconstructed distribution of emissions is used to generate a Compton image.

The display **22** is a CRT, LCD, projector, printer, or other display. The display **22** is configured to display the Compton image. The image or images are stored in a display plane buffer and read out to the display **22**. The images may be displayed separately or are combined, such as displaying the Compton image overlaid with or adjacent to a SPECT image.

FIGS. **4A**-**6** shows one example arrangement of modules **11**. The modules **11** form a ring **40** surrounding a patient space. FIG. **4A** shows four such rings **40** stacked axially. FIG. **4B** shows the scatter detectors **12** and corresponding catcher detectors **13** of the modules **11** in the ring **40**. FIG. **4C** shows a detail of a part of the ring **40**. Three modules **11** provide corresponding pairs of scatter and catcher detectors **12**, **13**. Other dimensions than shown may be used. Any number of modules **11** may be used to form the ring **40**. The ring **40** completely surrounds the patient space, but gaps with less than ½ module width may be provided. Within a housing of a medical imaging system, the ring **40** connects with a gantry **50** or another framework as shown in FIG. **5**. The ring **40** may be positioned to allow a patient bed **60** to move a patient into and/or through the ring **40**. FIG. **6** shows an example of this configuration.

The ring may be used for Compton-based imaging of emissions from a patient. FIG. **7** shows an example of using the same type of modules **11** but in a different configuration. A partial ring **40** is formed. One or more gaps **70** are provided in the ring **40**. This may allow for other components to be used in the gaps and/or to make a less costly system by using fewer modules **11**.

FIG. **8** shows another configuration of modules **11**. The ring **40** is a full ring. Additional partial rings **80** are stacked axially relative to the bed **60** or patient space, extending the axial extent of detected emissions. The partial rings **80** are in an every other or every group of N modules **11** (e.g., N=4) distribution rather than the two gaps **70** partial ring **40** of FIG. **7**. The additional rings may be full rings. The full ring **40** may be a partial ring **80**. The different rings **40** and/or partial rings **80** are stacked axially with no or little (e.g., less than ½ a module's **11** axial extent) apart. Wider spacing may be provided, such as having a gap of more than one module's **11** axial extent.

FIG. **9** shows yet another configuration of modules **11**. One module **11** or a single group of modules **11** is positioned by the patient space or bed **60**. Multiple spaced apart single modules **11** or groups (e.g., group of four) may be provided at different locations relative to the bed **60** and/or patient space.

In any of the configurations, the modules **11** are held in position by attachment to a gantry, gantries, and/or other framework. The hold is releasable, such as using bolts or screws. The desired number of modules **11** are used to assemble the desired configuration for a given medical imaging system. The gathered modules **11** are mounted in the medical imaging system, defining or relative to the patient space. The result is a Compton sensor for imaging the patient.

The bed **60** may move the patient to scan different parts of the patient at different times. Alternatively or additionally, the gantry **50** moves the modules **11** forming the Compton sensor. The gantry **50** translates axially along the patient space and/or rotates the Compton sensor around the patient space (i.e., rotating about the long axis of the bed **60** and/or patient). Other rotations and/or translations may be provided, such as rotating the modules **11** about an axis non-parallel to the long axis of the bed **60** or patient. Combinations of different translations and/or rotations may be provided.

The medical imaging system with the Compton sensor is used as a stand alone imaging system. Compton sensing is used to measure distribution of radiopharmaceutical in the patient. For example, the full ring **40**, partial ring **40**, and/or axially stacked rings **40**, **80** are used as a Compton-based imaging system.

In other embodiments, the medical imaging system is a multi-modality imaging system. The Compton sensor formed by the modules **11** is one modality, and another modality is also provided. For example, the other modality is a single photon emission computed tomography (SPECT), a PET, a CT, or a MR imaging system. The full ring **40**, partial ring **40**, axially stacked rings **40**,**80**, and/or singular module **11** or group of modules **11** are combined with the sensors for the other type of medical imaging. The Compton sensor may share a bed **60** with the other modality, such as being positioned along a long axis of the bed **60** where the bed positions the patient in the Compton sensor in one direction and in the other modality in the other direction.

The Compton sensor may share an outer housing with the other modality. For example, the full ring **40**, partial ring **40**, axially stacked rings **40**,**80**, and/or singular module **11** or group of modules **11** are arranged within a same imaging system housing for the sensor or sensors of the other modality. The bed **60** positions the patient within the imaging system housing relative to the desired sensor. The Compton sensor may be positioned adjacent to the other sensors axially and/or in a gap at a same axial location. In one embodiment, the partial ring **40** is used in a computed tomography system. The gantry holding the x-ray source and the x-ray detector also holds the modules **11** of the partial ring **40**. The x-ray source is in one gap **70**, and the detector is in another gap **70**. In another embodiment, the single module **11** or a sparse distribution of modules **11** connects with a gantry of a SPECT system. The module **11** is placed adjacent to the gamma camera, so the gantry of the gamma camera moves the module **11**. Alternatively, a collimator may be positioned between the modules **11** and the patient or between the scatter and catcher detectors **12**, **13**, allowing the scatter and/or catcher detectors **12**, **13** of the modules **11** to be used for photoelectric event detection for SPECT imaging instead of or in addition to detection of Compton events.

The module-based segmentation of the Compton sensor allows the same design of modules **11** to be used in any different configurations. Thus, a different number of modules **11**, module position, and/or configuration of modules **11** may be used for different medical imaging systems. For example, one arrangement is provided for use with one type of CT system and a different arrangement (e.g., number and/or position of modules **11**) is used for a different type of CT system.

The module-based segmentation of the Compton sensor allows for more efficient and costly servicing. Rather than replacing an entire Compton sensor, any module **11** may be disconnected and fixed or replaced. The modules **11** are individually connectable and disconnectable from each other and/or the gantry **50**. Any bridges are removed, and then the module **11** is removed from the medical imaging system while the other modules **11** remain. It is cheaper to replace an individual module **11**. The amount of time to service may be reduced. Individual components of a defective module **11** may be easily replaced, such as replacing a scatter or catcher detector **12**, **13** while leaving the other. The modules **11** may be configured for operation with different radioisotopes (i.e., different energies) by using corresponding detectors **12**, **13**.

FIG. **10** shows one embodiment of a flow chart of a method for forming, using, and repairing a Compton camera. The Compton camera is formed in a segmented approach. Rather than hand assembling the entire camera in place, scatter detector and catcher detector pairs are positioned relative to each other to form a desired configuration of the Compton camera. This segmented approach may allow different configurations using the same parts, ease of assembly, ease of repair, and/or integration with other imaging modalities.

Other embodiments form a combination of a Compton camera and a SPECT camera. The segmented modules **11** of FIG. **11** are used. The modules of FIGS. **13-16** may be used for forming a SPECT camera. The detector arrangement of FIG. **11** may be used.

The method may be implemented by the system of FIG. **1** to assemble a Compton sensor as shown in any of FIGS. **4-9**. The method may be implemented by the system of FIG. **11** to assemble a Compton sensor as shown in any of FIGS. **13-16**. Other systems, modules, and/or configured Compton sensors may be used.

The acts are performed in the order shown (i.e., top to bottom or numerically) or other orders. For example, act **108** may be performed as part of act **104**.

Additional, different, or fewer acts may be provided. For example, acts **102** and **104** are provided for assembling the Compton camera without performing acts **106** and **108**. As another example, act **106** is performed without other acts.

In act **102**, scatter and catcher detector pairs are housed in separate housings. Modules are assembled where each module includes both a scatter detector and a catcher detector. A machine and/or person manufactures the housings.

The modules are shaped to abut where the scatter and catcher detector pairs of different ones of the housings are non-planar. For example, a wedge shape and/or positioning is provided so that the detector pairs from an arc, such as shown in FIG. **4**C. The shape allows and/or forces the arc shape when the modules are positioned against one another.

In act **104**, the housings are abutted. A person or machine assembles the Compton sensor from the housings. By stacking the housings adjacent to each other with direct contact or contact through spacers, gantry, or framework, the abutted housings form the arc. A full ring or partial ring is formed around and at least in part defines a patient space. Based on the design of the Compton camera or Compton-SPECT camera, any number of housings with the corresponding scatter and catcher detector pairs are positioned together to form a camera.

The housings may be abutted as part of a multi-modality system or to create a single imaging system. For a multi-modality system, the housings are positioned in a same outer housing and/or relative to a same bed as the sensors for the other modality, such as SPECT, PET, CT, or MR imaging system. The same or different gantry or support framework may be used for the housings of the Compton camera and the sensors for the other modality. For other embodiments, the modules provide the multi-modality by providing for both a Compton camera and the SPECT imaging system.

The configuration or design of the Compton camera defines the number and/or position of the housings. Once abutted, the housings may be connected for communications, such as through one or more bridges. The housings may be connected with other components, such as an air cooling system and/or a Compton processor.

In act **106**, the assembled Compton camera detects emissions. A given emitted photon interacts with the scatter detector. The result is scattering of another photon at a particular angle from the line of incidence of the emitted photon. This secondary photon has a lesser energy. The secondary photon is detected by the catcher detector. Based on the energy and timing of both the detected scatter event and catcher event, the events are paired. The positions and energies for the paired events provides a line between the detectors and an angle of scattering. As a result, the line of incidence (e.g., Compton cone of incidence) of the emitted photon is determined.

To increase the likelihood of detecting the secondary photon, the catcher events from one housing may be paired with the scatter events of another housing. Due to the angles, scatter from one scatter detector may be incident on the paired catcher detector in the same housing or a catcher detector in another housing. By the housings being open in the detector region and/or using low photon attenuating materials, a greater number of Compton events may be detected.

The detected events are counted or collected. The lines of response or lines along which the different Compton events occur are used in reconstruction. The distribution in three dimensions of the emissions from the patient may be reconstructed based on the Compton sensing. The reconstruction does not need a collimator as the Compton sensing accounts for or provides the angle in incidence of the emitted photon.

The detected events are used to reconstruct the locations of the radioisotope. Compton and/or photoelectric images are generated from the detected events and corresponding line information from the events.

In act **108**, a person or machine (e.g., robot) removes one of the housings. When one of the detectors or associated electronics of a housing fails or is to be replaced for detecting at different energies, the housing may be removed. The other housings are left in the medical imaging system. This allows for easier repair and/or replacement of the housing and/or detectors without the cost of a greater disassembly and/or replacement of the entire Compton camera.

FIGS. **11-15** are directed to a Compton camera with a tilted scatter detector and/or a near 2π catcher detector. Using the modules of FIGS. **1-9** or another Compton camera, the scatter and/or catcher layers are arranged to capture a greater percentage of emissions from a patient and/or scatter. The scatter layer is configured in a tilted configuration. The catcher layer is formed as a near 2π catcher layer. Collimation may be used between modules to exclude large Compton angle events, which degrade image quality, thus improving signal-to-noise in the image and reducing demands on the ASIC/FPGA.

FIG. **11** shows one embodiment of a module **11** of a Compton camera for a medical imaging system. A tilted scatter layer and a near 2π catcher layer are provided. The tilted scatter layer results in a greater volume for scattering. The near 2π catcher detector results in greater opportunity to catch scattered photons by catching scatter over a wider range of angles.

The module **11** may be the entire Compton camera or multiple such modules **11** form the Compton camera. The module **11** of the medical imaging system includes a tilted scatter layer of scatter detectors **12**, a bottom catcher detector **13**A, and side catcher detectors **13**B, inter-module shielding **112** to reduce cross-talk, and inter-module slits and/or slats (i.e., collimator) **114** to block large scatter Compton angle events and reduce the load on the ASICs or FPGAs of the detectors **12**, **13**. Additional, different, or fewer components may be provided. For example, the tilted scatter detector **12** is provided without the near 2π catcher detectors **13**A, **13**B or vise versa. As another example, the shielding **112** and/or slits or slats **114** are not provided. In another example, the ASICs or FPGAS, circuit boards, housing, or other components are provided.

The dimensions in the drawing are arbitrary and sized for explanation. Other relative sizes may be used. Since FIG. **11** is a cross-section, other components may be provided in front or beyond FIG. **11**. For example, side wall catcher detectors **13**B, shielding **112**, and/or slits or slats **114** are provided on side walls in parallel with the plane of the drawing sheet. In other embodiments, one or more of the side walls does not include side wall catcher detectors **13**B, shielding **112**, and/or slits or slats **114**.

The module **11** is positioned relative to a patient space, bore of the medical imaging system, and/or bed **60** as discussed above for FIGS. **5-9** or another configuration. The patient bed **60** supports the patient in the patient space. The bed **60** may be moveable, such as a robot or roller system for moving the patient into and out of the medical imaging system. The outer housing of the medical imaging system and/or scatter layer create a bore into which the patient bed **60** is positioned. The bore defines the patient space for imaging the patient. The bore may be of any dimension in a cross-sectional plane orthogonal to a longitudinal or iso center axis, such as 70 cm. The center of the bore along a longitudinal axis of the cylindrical shaped bore or center of the patient space is the iso center axis. The bed **60** moves along the iso center axis.

The scatter detector **12** is arranged to receive emissions from a patient. In FIG. **11**, gamma rays are shown in parallel directed to a front face of the scatter detector **12**. The gamma rays emitted from the patient are not all parallel, so may arrive at the front face at any of various angles. The module **11** is positioned so that gamma rays directed to the module **11** are likely to intersect the scatter detector **12** before the catcher detector **13**. The scatter detector **12** has an outer surface facing the iso center axis.

For tilting, the outer surface is away from orthogonal by an angle of at least 10-80 degrees (e.g., at least 20, 30 or 45 degrees) to a radial line extending perpendicular from the iso center axis through a center or other part of the scatter detector **12**. Any angle may be used, such as being 35, 45, 65, or 75 degrees. Where the scatter detector **12** is a plate orthogonal to the radial line, a given area may fit in the module **11**. By tilting, the outer surface facing the patient may have a greater area. This results in a greater likelihood of scatter as the volume for interaction is greater. Greater angles result in greater area of the face and volume of the detector **12**, resulting in greater likelihood of scatter for any given photon.

The tilt is with respect to the radial from perpendicular to the iso center axis. No tilt corresponds to the scatter detector **12** being a plate orthogonal to a radial, which is perpendicular to the iso center axis. The tilt may be relative to the front or back surface of the module **11** or the back or rear wall catcher detector **13**A.

The absolute number of scattered photons is increased by: a) using scatter layer detector materials that favor Compton scatter vs. photoelectric effect (low-Z materials where low Z is 30 or less); b) adding more scatter material in the scatter layer; and c) maximizing the number of scattered events that contribute to have a better overall image quality by eliminating, using physical or digital collimation, Compton events that have large Compton angle uncertainties. By increasing the number of scattered photons escaping from the scatter layers and reaching the catcher layer (tilting), the probability that Compton events escape from the scatter layer is increased by reducing the mean-free path of those events at the scatter layer. The increase in scatter increases the number of scattered photons reaching the catcher layer. Fewer scatter detector modules (per absolute number of scatter events reaching the catcher layer) may be used due to a larger escape probability using tilted geometry. The tilting results in a larger number of pixels in projection orthogonal to the radial in the scatter detector **12**, increasing resolution.

In the embodiment shown in FIG. **11**, the scatter detector **12** is formed from a plurality of plates positioned in an accordion arrangement. To increase the number of scatter events, the plates of the scatter detector **12** are configured in a tilted configuration with no gaps between detector plates. The plates are abutted to create the accordion arrangement. Gaps are provided in other embodiments. The plates tilt in a repeating sequence of two angles. Other arrangements with sequences of three or more plates and three or more corresponding angles may be used. In alternative embodiments, the scatter detector is a single plate that is tilted. In yet other embodiments, one or more plates of the detectors are orthogonal to the radial (e.g., parallel with the rear wall catcher detector **13**A) while other plates are tilted. Alternatively, the plates do not abut, but instead are all tilted at the same angle and in parallel to each other (see FIG. **15**).

Using plates and the accordion arrangement, the same arrangement extends between side walls into and out of the plane of the drawing. In other embodiments, the scatter detector **12** varies in angle or tilting into and out of the plane of the drawing as well as through the cross section of the drawing. Any 3D surface that is non-planar may be used.

The tilting maximizes the absolute number of scattered photons for the scatter layer, thus increasing the absolute number of scattered photons reaching the catcher layer. Since scattering is more likely to occur due to the tilting, a larger variety of detector materials may be used, such as Si, HPGe, CdTe, CZT, GaAs, TIBr and others, due to inherent easiness of fabrication of more uniform detectors with thinner thickness. The tilt counteracts some of the loss of scatter volume in a thinner detector. Fewer scatter detector modules are required per absolute number of scatter events reaching the catcher due to a larger escape probability using tilted geometry. Due to the tilting, there is a greater pixel density of the scatter detector relative to viewing from the iso center axis. Position resolution improves using the tilted geometry due to a larger number of pixels (ASIC channels) per unit of projected area along the radials.

The catcher detector **13** is positioned behind the scatter detector **12** relative to the patient space. The catcher detector **13** is spaced from the scatter detector **12**, forming a volume between the detectors **12**, **13**. One or more parts of the catcher detector **13** may contact one or more parts of the scatter detector **12**, such as at the sides of the module **11**. Gaps may be provided between the scatter detector **12** and the catcher detector **13** with no contact. The catcher detector **13** may extend to be at a same z-depth as part of the scatter detector **12**, such as due to the tilt of the scatter detector **12** and/or due to the surround shape of the catcher detector **13**.

The catcher detector **13** forms a substantially semi-spherical surround behind the scatter detector **12** relative to the patient space. Substantially is used to account for gaps at joints, one or two of four sides not including a side wall catcher detector **13**B, the side wall catcher detectors **13**B starting at a z-depth (i.e., along the radial perpendicular to the iso center axis) that is 30 degrees or less from a plane orthogonal to the radial at a depth of the deepest part of the scatter detector **12**, and/or the side wall catcher detectors **13**B starting at a z-depth (i.e., along the radial perpendicular to the iso center axis) that is 10 cm or less deeper than the plane orthogonal to the radial at a depth of the deepest part of the scatter detector **12**. A multi-sided detector front surface is provided. The catcher detector **13** is non-planar. A front face or surface through which the scattered photons enter the detector **13** is non-planar. The catcher detector **13** is formed in a cup, box, or semi-sphere of three or more sides to at least partially (i.e. substantially) surround a volume behind the scatter detector **12**. This provides a near 2π structure behind the scatter detector **12** relative to the patient.

In one embodiment, the substantially semi-spherical surround is formed from a plurality of planar plates, such as the rear wall catcher detector **13**A and two or more side wall catcher detectors **13**B (e.g., the rear wall catcher detector **13**A and four side wall catcher detectors **13**B forming a cuboid of five sides with one open side directed to the scatter detector **12**). The plates are at right angles to each other, but greater or lesser angles may be provided. The plates are substrates positioned in non-parallel planes within the module **11**. Semiconductor or other processes for forming the detectors as slabs may be used to form the plates, which are then positioned in the module **11** to provide the substantially semi-spherical surround. Four-sided square or rectangular plates are used in FIG. **11**, but three, five, six, or other numbers of sides may be used.

The catcher layer geometry at least partly surrounds the space behind the scatter layer, such a providing a near 2π solid angle geometry. This surround may result in catching more scatter photons. For example, near 100% scattered/caught ratio may be achieved by adding catcher layers in-between modules (i.e., the side wall catcher detectors

13B). A larger field of view of the Compton-camera in the Z-direction (bed direction) is provided. The absolute number of scattered/caught fraction of photons is increased by: a) increasing the solid angle between the scatter layer and the catcher layer; b) increasing the area of the catcher layer; c) reducing the distance between scatter and catcher layers; d) increasing the effective thickness of the catcher layer; e) and/or selecting materials favoring photoelectric effect vs. Compton scatter in the catcher layer. By shaping the catcher detector **13** as multi-sided in a surround of the volume behind the scatter detector **12**, the solid angle is increased, the area of the front face of the catcher detector is increased, the distance between the scatter detector **12** and the catcher detector **13** may be decreased, and the effective thickness of the catcher layer is increased.

The shields **112** are shielding material, such as lead or tungsten. The shields **112** are gamma ray shields. Materials and thickness that are opaque to a given percentage (e.g., 75-100%) of the emissions at the isotope energies are positioned at the side walls (e.g., covering part or the entire wall) or are the side walls of the module **11**. The shielding material is provided over the entire side wall, such as adjacent ends and over the z-axis extent (i.e., depth along the radial) of the scatter detector **12** and the catcher detector **13**. In other embodiments, the shielding material has a lesser extent, such as beginning on the side walls at a deepest extent of the scatter detector **12** to the catcher detector **13** or to a deepest part of the catcher detector **13**.

The shields **112** may be on all walls but a side of the module **11** facing the patient space. Alternatively, one or more side walls and/or the rear wall do not include the shields **112**. Where multiple modules **11** abut each other, one shield **112** between them may be provided rather than having abutting shields **112** from the two modules **11**. The shielding material separates the modules **11**. Inter-module cross-talk is reduced by adding inter-module shielding.

The slits and/or slats **114** are collimators formed from plates. The plates are positioned in parallel to each other, forming slits through which photons may pass. In other embodiments, the slits and/or slats **114** are collimators with holes at a desired angle. Any size holes may be used.

The slits and/or slats **114** are angled to allow photons at some angles to pass and absorbing or blocking photons at other angles. For example, the slits and/or slats **114** are angled to absorb large angle scattered photons (e.g., 80-110 degrees from the radial). The large angle scattered photons contain large angular uncertainty and may be noise for adjacent modules. The slits and/or slats **114** may improve image quality and reduce ASIC or FPGA **122** requirements by reducing noise-related photons.

The slits and/or slats **114** are provided adjacent to the scatter detector **12**, such as with a same depth extent on two or more side walls of the module **11**. Other extents or positioning may be provided. The slits and/or slats **114** may be part of the module **11** or positioned between modules **11**.

The scatter detector **12** includes application specific integrated circuits (ASIC) or field programmable gate arrays (FPGA) for reading the scatter detector **12**. A separate ASIC or FPGA is provided for each group of pixels in the scatter detector **12**. The ASIC or FPGA is formed as part of the substrate with the scatter detector **12** or may be formed separately. The ASIC or FPGA **122** is positioned in parallel with the scatter detector **12** as shown in FIG. **12**A. In this position, some scatter photons are caught in the ASIC or FPGA **122**, resulting in loss of the Compton event.

FIG. **12**B shows another embodiment. The ASICs or FPGAs **122** electrically connect to the scatter detector **12** by traces on flexible circuit material or by wires. The ASICs or FPGAs **122** are positioned in parallel with the radial lines from the iso center axis, minimizing the area of interaction with scatter. The ASICs or FPGAs **122** are plates or substrate positioned to be non-parallel with the outer or front facing surface of the scatter detector **12**. In other embodiments, the ASICs or FPGAs **122** of the scatter detector **12** are removed from the field of view, such as being behind the catcher detector **13** relative to the patient. The ASICs or FPGAs **122** are positioned to reduce the effect on Compton kinematics.

FIG. **11** shows one module **11**. The Compton camera is formed from the single module **11** or from multiple modules **11**. Each module **11** includes the tilted scatter detector **12** and/or near 2π catcher detector **13**. The modules **11** may have housing shapes to stack or abut for forming a ring or partial ring, such as having the wedge shape of FIG. **1**. Other shapes may be used.

For a multi-module Compton camera, the scatter layer is formed from a plurality of scatter detectors **12**, such as using the modular system of FIGS. **1-9**. Similarly, the catcher layer **13** is formed from a plurality of catcher detectors **13**. For example, eighteen modules **11** provide for eighteen pairs of scatter and catcher detectors **12**, **13**. More or fewer modules **11** may be used. The modules **11** have any arrangement, such as one or more axially spaced rings and/or partial rings or one or more sparsely distributed modules **11** or groups of modules. The modules **11** may be part of a multi-modality imaging system or for a Compton-camera only system. The scatter and catcher detectors **12**, **13** (e.g., modules **11**) are positioned to receive emissions from a patient on the patient bed **60** or otherwise in the patient space.

The module **11** may be positioned adjacent to none, one, two, three, or four other modules **11**. Where the scatter detector **12** is tilted with a repeating pattern along one dimension, the modules **11** may be abutted or positioned adjacent to each other along two sides. Where the scatter detector **12** is tilted with a repeating pattern along two dimensions, the modules **11** may be abutted or positioned adjacent to each other along four sides. In other embodiments, the module **11** may abut along 1-4 sides regardless of the tilt arrangement.

In one embodiment, the modules **11** are positioned to form one or more axially spaced partial or full rings. FIGS. **13**A and **13**B show at least three full or partial rings around the patient space and the patient bed **60**. Additional or fewer rings and/or partial rings may be provided. FIG. **13**A is a view of a cross section orthogonal to the iso center axis at one of the rings or partial rings. FIG. **13**B is cross-section view of a plane parallel with the bed **60** and along the iso center axis.

The near 2π catcher detectors **13** of the modules include side wall catcher detectors **13**B positioned on sides abutting modules **11** in the same ring and modules **11** of other rings. Each module **11** operates independently so that scatter from one module **11** is not paired with capture in the catcher detector **13** of another module **11**. Timing, power, or other information may be or may not be shared between modules. Since the modules **11** are fully isolated modules, the modules **11** may stack or abut on any of four sides.

In alternative embodiments, Compton events may be formed from scatter from one module **11** and capture of the scatter photon in another module **11**. FIGS. **14**A and **14**B show cross sections where the side wall catcher detectors **13**B for the sides adjacent to other rings or partial rings are removed or not provided. The modules **11** within a given ring or partial ring are isolated from each other. The adjacent modules **11** across rings share a common synchronization and/or clock for event detection by the ASICs or FPGAs of the detectors **12**, **13**, allowing a Compton pair of events using scatter in one module **11** and capture of the scatter photon in another module **11**. The surround of the catcher detectors **13** in the modules of the axially outer rings or partial rings have side wall catcher detectors **13**B on three sides. The surround of the catcher detectors **13** in the axially inner rings or partial rings have side wall catcher detectors **13**B on two sides (sides adjacent modules **11** in the same ring or partial ring). In alternative embodiments, modules are not isolated within a ring or partial ring but are isolated between rings or partial rings. In yet other embodiments, one or more modules **11** may not be isolated within a ring and between rings. The near 2π catcher layer is formed from the catcher detectors **13** of multiple modules **11**.

FIG. **15** shows a partial ring of modules **11**. The modules **11** have wedge shapes for closer stacking. FIGS. **13** and **14** show cube shapes where gaps are provided at least further from the iso center axis when the modules **11** are stacked adjacent each other in the ring or partial ring.

In FIG. **15**, the scatter detectors **12** are tilted. The plates of detectors forming the tilted scatter detector **12** are arranged in parallel planes. All the plates of a scatter detector **12** in a module **11** tilt in the same direction or towards a given side. Within the ring or partial ring, the scatter detectors **12** of the different modules **11** tilt the same direction or in different directions. Different patterns or tilt structures may be provided between different modules **11** or by groups of modules **11**, such as the every-other pattern of opposite tilting between adjacent modules **11** of FIG. **15**. Any grouping of tilt pattern across modules **11** and/or within modules **11** may be used.

The pattern or grouping may correspond to isolation or cross-talk between modules **11**. For example, pairs of modules **11** with opposite tilt of the scatter detector **12** share a synchronization signal and/or clock but are isolated from other pairs. The side wall catcher detectors **13**B between modules **11** sharing the synchronization signal and/or clock (i.e., with cross talk) are not provided. The side wall catcher detectors **13**B between modules **11** not grouped are provided. In other embodiments, the paired or grouped modules **11** for cross talk have a same tilt as each other. The tilt for other groups is the same and/or different.

The Compton processor **19** (e.g., image processor) is configured to generate a Compton image from Compton events detected from the tilted scatter and near 2π catcher detectors **12**, **13**. The electronics of the modules **11** or other electronics output events detected from the detectors **12**, **13**. The location, energy, and time of the events are received by the Compton processor **19**. These events are paired using the location, energy, and/or time. Based on the pairing, location, and energy, an angle of incidence of the emission from the patient onto the scatter detector **12** is determined. The angle may be expressed probabilistically, such as a cone of incidence. Using reconstruction from many detected Compton events and the angles of incidence, a spatial distribution in patient or object space of the emissions is determined. A Compton image is rendered from the spatial distribution.

The display **22** displays the Compton image. Other images may be displayed with the Compton image. The tilted scatter detector **12** and/or near 2π catcher detector **13** result in a greater number of Compton events being captured, so the resulting Compton image has more information. This better image quality results in a diagnostically improved image.

The Compton processor **19** is configured to perform digital collimation. Once events are paired, the angle of the scatter from the scatter detector **12** for a given event is determined. The relationship of energy and angle and the positions of the paired events indicates the angle of the scatter photon. Compton events may be rejected based on the angle, such as applying one or more scatter angle thresholds. The Compton image is generated from the Compton events that are not rejected. In other embodiments, digital collimation is not used.

FIG. **16**A shows angular uncertainties in the Compton angle as a function of Compton angle. Compton events with some scatter angles may result in worse image quality. For example, the FWHM of a back projected cone is to be at a desired level, such as represented by the horizontal dashed line. The FWHM for a given Compton event is above or below the desired FWHM based on the scatter angle. For example, angles between 40 degrees and 120 degrees provide information with sufficient FWHM. FIG. **16**B shows different scatter angles given emissions orthogonal to the scatter detector. Compton events for lesser (e.g., less than 40 degrees) and/or greater (e.g., greater than 120 degrees) scatter angles are not used (i.e., rejected by digital collimation). The remaining Compton events are used to generate the Compton image.

In one example, a CZT scatter detector **12** and CZT catcher detector **13** have a 30 cm distance between scatter and catcher layers with a 70 cm bore diameter. A PSF with FWHM <40.0 mm is produced by rejecting events with Compton angle greater than ~40°. Other thresholds may be used.

While the invention has been described above by reference to various embodiments, it should be understood that many changes and modifications can be made without departing from the scope of the invention. It is therefore intended that the foregoing detailed description be regarded as illustrative rather than limiting, and that it be understood that it is the following claims, including all equivalents, that are intended to define the spirit and scope of this invention.

We claim:

1. A Compton camera for medical imaging, the Compton camera comprising:

a bed for a patient space having an iso center axis;

a first module having a first scatter detector and a first catcher detector spaced from the first scatter detector, the first scatter detector formed from a plurality of plates positioned back and forth in an accordion arrangement, wherein each of the plates of the plurality of plates include an outer surface facing the iso center axis where the outer surface is tilted away from orthogonal by an angle of at least 20 degrees to a radial line extending perpendicular from the iso center axis through a center of the first scatter detector, the first catcher detector forming a substantially semi-spherical surround behind the first scatter detector relative to the patient space;

an image processor configured to determine angles of incidence for Compton events from the first scatter detector and the first catcher detector.

2. The Compton camera of claim **1** wherein the angle is at least 45 degrees to the radial line extending perpendicular from the iso center axis through the center of the first scatter detector.

3. The Compton camera of claim **1** wherein the first scatter detector comprises a tilted arrangement relative to the radial line.

4. The Compton camera of claim **1** wherein the substantially semi-spherical surround comprises a five-sided cuboid with an open side adjacent to the first scatter detector.

5. The Compton camera of claim 1 wherein the substantially semi-spherical surround comprises a plurality of planar catcher substrates positioned on non-parallel planes within the first module.

6. The Compton camera of claim 1 further comprising application specific integrated circuits or field programmable gate arrays for reading the first scatter detector, the application specific integrated circuits or field programmable gate arrays forming plates positioned to be non-parallel with the outer surface.

7. The Compton camera of claim 1 further comprising shielding material on sidewalls of the first module.

8. The Compton camera of claim 7 further comprising a second module having a second scatter detector and a second catcher detector spaced from the second scatter detector, the shielding material separating the first module from the second module.

9. The Compton camera of claim 1 further comprising a second module having a second scatter detector and a second catcher detector spaced from the second scatter detector, the surround including the second catcher detector.

10. The Compton camera of claim 1 further comprising additional modules having additional scatter detectors and additional catcher detectors spaced from the additional scatter detectors, the first module and additional modules forming a ring or partial ring around the patient space.

11. The Compton camera of claim 1 wherein the image processor is configured to generate a Compton image from the Compton events and the angles of incidence, and further comprising a display configured to display the Compton image.

12. A medical imaging system comprising:

a Compton camera comprising a scatter detector arranged to receive emissions from a patient, the scatter detector formed from a plurality of plates positioned back and forth in an accordion arrangement, wherein each plate of the plurality of plates includes an outer surface facing the patient where the outer surface is away from orthogonal by an angle of at least 20 degrees to a radial line extending perpendicular from a longitudinal axis of the patient through the scatter detector.

13. The medical imaging system of claim 12 wherein the Compton camera further comprises a near a structure behind the scatter detector relative to the patient.

14. The medical imaging system of claim 12 wherein the scatter detector is in a module, and further comprising gamma ray shielding material on a side of the module.

15. A medical imaging system comprising:

a Compton camera comprising a scatter detector and a catcher detector, the scatter detector arranged to receive emissions from a patient, the catcher detector arranged to receive scatter from the scatter detector due to the emissions from a patient, the catcher detector comprising a multi-sided detection surface positioned behind the scatter detector relative to the patient, wherein the scatter detector is formed from a plurality of plates positioned back and forth in an accordion arrangement, wherein each plate of the plurality of plates includes an outer surface facing the patient where the outer surface is away from orthogonal by an angle of at least 20 degrees to a radial line extending perpendicular from a longitudinal axis of the patient through the scatter detector.

16. The medical imaging system of claim 15 wherein the multi-sided detection surface comprises a near 2π a structure.

17. The medical imaging system of claim 15 wherein the scatter detector and catcher detector are in a module, and further comprising gamma ray shielding material on a side of the module.

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