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(54) **Title**: IMPROVED SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY SYSTEM AND METHOD

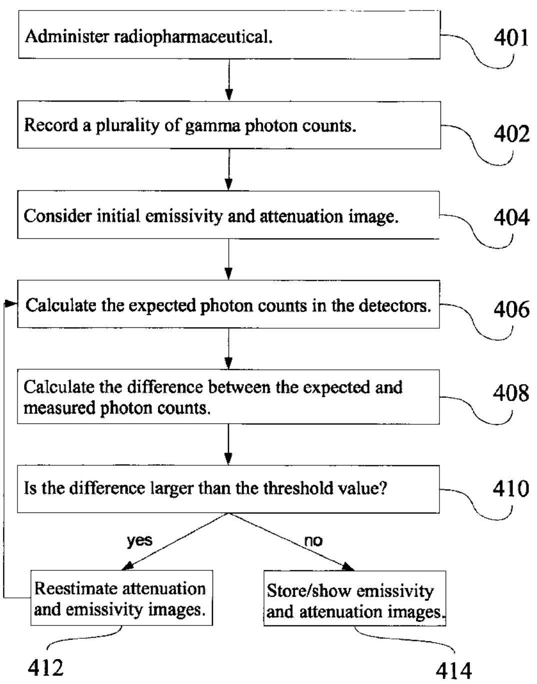

FIG. 4

(57) **Abstract**: The present invention relates to an improved system and method to obtain single photon emission computed tomography (SPECT) images without the use of collimator. The gamma radiation emitted by the radiopharmaceutical may be detected at a plurality of directions around the object of interest without collimation. The disclosed method uses only the photons which may be detected having substantially the same energy as the gamma emission. Limiting the data analysis only to these photons may minimize contribution of elastically scattered gamma radiation to the image and it may also simplify image reconstruction. The absorption of the gamma radiation and the inverse-square relationship between distance and intensity may provide a dependence of the detector illumination on the distance from the source. Based on the above distance dependence, the gamma photon emissivity image of the object of interest may be reconstructed using the gamma illumination of the detector elements at a plurality of positions outside the object of interest.

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IMPROVED SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY SYSTEM AND METHOD

FIELD OF THE INVENTION

This invention pertains to imaging methods and systems, and more particularly to an improved system and method of recording single photon emission computed tomography (SPECT) images without using collimator.

BACKGROUND OF THE INVENTION

Single photon emission computed tomography (SPECT) is a non-invasive nuclear imaging technique. SPECT is capable of reconstructing images representing the distribution of gamma emitting radioactive tracers in a living body.

SPECT is extensively used both in medical practice and biomedical research. It is a powerful technique for visualizing physiological processes and anatomical details of living creatures.

SPECT can be combined with x-ray computed tomography (CT). After recording a CT image, the SPECT image reconstruction can use corrections for the attenuation of the emitted gamma rays inside the body.

The SPECT technique requires the administration of a gamma emitting radioisotope into the subject or sample to be studied. In some applications of the SPECT technique the radioisotope can be used directly, but for the most part, the marker radioisotope is attached to a molecule of interest. The molecule labeled with the radioactive isotope (a radiopharmaceutical) will follow physiological transport pathways, or bind to certain target within the subject to be studied. Due to these processes, the radiopharmaceutical will accumulate in certain places of interest in the body. The distribution of the radioisotope inside the body can then be followed by detecting the gamma-emission of the isotope.

The radiation emitted by the radioactive tracer isotope is projected through one or more collimator slits on a gamma camera, which records two dimensional projection images of the gamma emission. A commonly used gamma camera type is the Anger camera. In this camera, a scintillation plate produces a flash of visible or ultraviolet light upon absorbing gamma photon energy. The gamma photon may transfer all of its energy to the crystal, or only part of it. In this later case a gamma photon with lower energy leaves the crystal. The strength of the light flash is proportional to the energy released by the gamma photon in the crystal. This flash is detected by many individual light detectors for simultaneously detecting every individual gamma photon impact. The data acquisition system then accurately determines the location and the energy transmitted to the crystal by the incident gamma photon.

Gamma emitting isotopes emit photons at discrete energies only. For example all photons emitted by 99mTc have 140 keV energy. The appearance of photons with lower energy in the gamma radiation emanating from the object of interest is due to the energy loss of photons in inelastic scattering on charged particles. Taking into account only scintillation flashes that correspond to the emitted gamma energy eliminates the effect of inelastic scattering from the image.

A projection image of the radiating organ source may be obtained by the camera at a specific measurement position.

SPECT combines several projections recorded from different directions (angles of view) to produce cross section images of the object of interest. The cross section images can be used to reconstruct a three dimensional image of the object of interest. The created three dimensional or cross section images are typically visualized on a computer screen, printed, or reproduced on a film.

The collimator simplifies image reconstruction by a projection mapping of the gamma emitting isotope on the detector. The collimator is built of an absorber material with one or more slits bored through it. It allows radiation to pass through the holes, and absorbs the radiation that does not travel along the desired path. The method has several inevitable disadvantages, including: (i) the collimator strongly reduces the sensitivity of the measurement by absorbing most of the emitted photons, and (ii) some part of the radiation does pass through the absorber material thereby reducing image contrast.

U.S. Pat. No. 5,818,050 to Dilmanian and Barbour (US '050) describes a collimator-free SPECT method. The gamma photon emissivity of the radiopharmaceutical distributed inside the patient is measured externally with an uncollimated gamma camera at a plurality of positions. It is important for the method of US '050 that the energy spectra of the incident gamma photons is recorded in each detector position.

The SPECT image is calculated by “effectively utilizing the probabilistic information in the entire energy spectrums” (column 6, line 23-24 of US ‘050) Photon detection spectra in the detector are predicted using an initial guess of an image of the source. The predicted and measured photon emissivities are then compared to obtain differences between them. Prediction and comparison is iterated by updating the image prediction until the differences are below a pre-set threshold for obtaining a final prediction of the source image. This invention suffers from several shortcomings. It has to measure, store, predict, and approximate by iteration the energy spectrum measured in every detection position. Unscattered photons are detected as photons with undegraded full energy. Photons undergoing single or multiple scattering either in the patient or in the detector will be detected at degraded lower energy levels. The observed spectra are determined by the probability of the different scattering effects, such as elastic scattering, photoelectric effect or Compton scattering. The probability of these effects depends on the composition of the different tissues and it is difficult to predict. The method described in US ‘050 uses a water filled phantom to estimate how emissions from different volume elements contribute to the spectra measured at every detector element. From a mathematical point of view this means the experimental determination of the Green function of the arrangement, using a phantom. Apart from the obvious problem that water may be a rather poor model of several tissues (e.g. bones), the procedure also makes the measurement very difficult. The phantom should represent accurately the shape of tissues and the interactions of the tissues with the emitted gamma radiation for the actual patient examined. Practically a new phantom has to be made for every patient. To determine the Green function, a point-like radiation source is moved inside the phantom from volume element to volume element, and the complete spectral detection has to be recorded in all the detector cells at all radiation source positions. The resulting data amount is enormous.

U.S. Pat. No. 7,253,416 discloses the use of a special detector which is sensitive to the direction of the incident radiation to eliminate the need for the collimator. This method has several drawbacks, including, without limitation, that the special detector is expensive, and that it is not energy selective.

SUMMARY OF THE INVENTION

The objective of the invention is to provide a new, improved system and method of recording single photon emission computed tomography (SPECT) images without the use of collimator.

In one embodiment, the present invention provides for a method for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, characterized in that said method comprises: (a) administering the at least one radioisotope to the at least one object of interest, said at least one radioisotope being capable of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object; (b) measuring an array of gamma photon counts from the at least one radioisotope inside the plurality of voxels at a plurality of positions around the at least one object of interest using one or more collimator free or partially collimated gamma detectors, said measuring counting gamma photons having substantially the same photon energy as a gamma emission of said at least one radioisotope; (c) calculating gamma photon emissions from the plurality of voxels based on the array of measured photon counts; and (d) generating the at least one image of the distribution of the at least one radioisotope inside the at least one object of interest based on the calculated gamma photon emissions.

In another embodiment, the present invention provides for a system for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, said radioisotope being capable of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object, characterized in that said system comprises: (a) at least one collimator free or partially collimated gamma detection means for measuring at least one array of gamma photon counts from the plurality of voxels at a plurality of positions around the at least one object of interest, said measured gamma photons having substantially the same energy as a gamma emission of said at least one radioisotope; (b) means for calculating gamma photon emissions from the plurality of voxels based on the measured at least one gamma photon counts; and (c) means for generating the at least one image of the distribution of the at least one radioisotope inside the at least one object of interest based on the calculated gamma photon emissions.

In another embodiment, the present invention provides for a system for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, said radioisotope being capable of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object, characterized in that said system comprises: (a) at least one collimator free or partially collimated gamma detection means for measuring at least one array of gamma photon counts from the plurality of voxels at a plurality of positions around the at least one object of interest, said measured gamma photons having substantially the same energy as a gamma emission of said at least one radioisotope; (b) means for calculating at least one estimated gamma emission image of the at least one distribution of the at least one emitting radioisotopes from the voxels; (c) means for calculating at least one expected array of gamma photon counts at the plurality of detector positions using the at least one estimated gamma emission image; (d) means for comparing the measured and the expected arrays of gamma photon counts at the plurality of detector positions to obtain at least one number representing differences between the measured and the expected arrays of gamma photon counts; (e) means for updating the at least one estimated gamma emission image; and (f) means for reconstructing at least one final image of the at least one distribution of the at least one radioisotope inside the at least one object using at least one updated gamma emission image.

BRIEF DESCRIPTION OF THE DRAWINGS

The invention will be described with reference to the embodiments shown in the drawings. It should be understood that the intention is not to limit the invention only to the particular embodiments shown, but rather to cover all alterations, modifications and equivalent arrangements possible within the scope.

The object of interest imaged by the embodiments described in this provisional patent application may be any living organism or part of a living organism.

The invention will now be described in more detail, by way of example only, with reference to the accompanying drawings, in which:

FIG. 1 is a schematic representation of a combined CT and SPECT imaging system in accordance to one embodiment of the present invention.

FIG. 2 is a flowchart of data acquisition, processing and image reconstruction in the system of FIG. 1.

FIG. 3 is a schematic representation of a SPECT system in accordance to one embodiment of the present invention.

FIG. 4 is a flowchart of data acquisition, processing and image reconstruction in the system of FIG. 3.

DRAWINGS – Reference numerals

102 x-ray source

104 x-ray emitted by the x-ray source

106 collimator and filter

108 x-ray beam before the object of interest

110 rotational subsystem

112 translational subsystem

114 object of interest

116 part of the x-ray beam that passed through and around the object of interest

118 x-ray detector

120 gamma emitting isotope

122 gamma photons emitted by the gamma emitting isotope

124 fraction of the gamma photons that reach the gamma detector

126 gamma detector

150 digital computer

152 x-ray source controller

156 collimator controller

160 rotational subsystem controller

162 translational subsystem controller

168 x-ray detector controller

176 gamma detector controller

180 system controller unit

Step 201 Administering radiopharmaceutical to the object of interest.

Step 202 Generating x-rays for imaging.

Step 204 Shaping the x-ray beam to desired intensity distribution in the object of interest.

Step 206 Passing the x-ray beam through the object of interest at a plurality of directions.

Step 208 Detecting the x-ray passed through the object of interest.

Step 210 Reconstructing the distribution of the x-ray absorbance inside the object of interest.

Step 212 Calculating gamma absorbance image of the object of interest.

Step 214 Recording a plurality of gamma photon count measurements.

Step 216 Providing an initial gamma photon emissivity image of the object of interest.

Step 218 Calculating the expected photon counts in the detector elements.

Step 220 Calculating the difference between the expected and measured photon counts.

Step 222 Is the difference large?

Step 224 YES: Reestimating the gamma photon emissivity image of the object of interest, and going back to step 218.

Step 226 NO: Storing or visualizing the calculated image.

302 translational subsystem

304 object of interest

306 gamma emitting isotope

308 gamma photons emitted by the gamma emitting isotope

310 fraction of the gamma photons that reach the gamma detector

312 gamma detector

350 digital computer

352 translational subsystem controller

362 gamma detector controller

380 system controller unit

Step 401 Administering radiopharmaceutical to the object of interest.

Step 402 Recording a plurality of gamma photon count measurements.

Step 404 Providing initial gamma photon emissivity and gamma absorption images of the object of interest.

Step 406 Calculating the expected photon counts in the detectors.

Step 408 Calculating the difference between the expected and measured photon counts.

Step 410 Is the difference larger than the threshold value?

Step 412 YES: Re-estimating both the gamma photon absorption and emissivity image of the object of interest, and going back to Step 406.

Step 414 NO: Storing or visualizing the calculated image.

DETAILED DESCRIPTION OF THE INVENTION

Overview

The present invention relates to new methods and new systems for generating SPECT images. The SPECT method of the present invention may achieve higher sensitivity by omitting the use of collimator and by using novel data collection and analysis method. The method may evaluate directly the gamma illumination falling on a detector that is comprised of a plurality of detector elements. The method of the present invention in several aspects does exactly the opposite of what is proposed in US ‘050. US ‘050 uses the information in the spectra to reconstruct the SPECT image. The method and system of the present invention only uses the photons that may be detected having substantially the same energy as the gamma emission. US ‘050 determines the green function of the patient-instrument system by direct measurements on manufactured phantoms. Described herein are methods to calculate the Green function. This is possible because limiting analysis to the photons having substantially the same energy as the gamma emission may minimize unwanted contribution of scattered gamma radiation to the image and it may also simplify image reconstruction. The absorption of the gamma radiation and the inverse-square relationship between distance and intensity may provide a dependence of the detector illumination on the distance from the source. Based on the above distance dependence, the gamma photon emissivity image of the object of interest may be reconstructed from the gamma illumination of the detector elements at a plurality of positions outside the object of interest.

As such, the present invention relates to a method for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest. In one embodiment of the present invention, the method may start by administering the at least one radioisotope to the object of interest. In aspects of the invention, the at least one radioisotope may be included in a pharmaceutical compound. The at least one radioisotope may be capable of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object. An array of gamma photon counts from the at least one radioisotope inside the plurality of voxels may then be measured at a plurality of positions around the at least one object of interest. The gamma photon counts may be measured or recorded using one or more collimator free or partially collimated gamma detectors. Only those gamma photons having substantially the same photon energy as an unscattered gamma emission of the at least one radioisotope may be measured. The measured photon counts may then be used to calculate gamma photon emissions from the plurality of voxels. The at least one image of the distribution of the at least one radioisotope inside the at least one object of interest may then be generated based on the calculated gamma photon emissions.

The present invention relates also to a system for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest. In one embodiment of the present invention, the system for generating at least one image of a distribution of at least one isotope inside at least one object of interest may include at least the following: (a) at least one collimator free or partially collimated gamma detection means for measuring at least one array of gamma photon counts from the plurality of voxels at a plurality of positions around the at least one object of interest, said measured gamma photons having substantially the same energy as an unscattered gamma emission of said at least one radioisotope; (b) means for calculating gamma photon emissions from the plurality of voxels based on the measured at least one gamma photon counts; and (c) means for generating the at least one image of the distribution of the at least one radioisotope inside the at least one object of interest based on the calculated gamma photon emissions.

Various changes may be made in the embodiments and operating methods presented below without departing from the spirit or scope of the invention. All matter contained in the descriptions or shown in the accompanying drawings should be interpreted as illustrative and not in a limiting sense.

<u>Combined x-ray Computed Tomography (CT) and Single Photon Emission Computed Tomography (SPECT) System</u>

FIG. 1 shows a diagram of an exemplary embodiment of a system 10 of the present invention: a combined x-ray CT and SPECT system. This combination system 10 may be capable of collecting measurements and reconstruct at least one image of the distribution of at least one radiopharmaceutical inside at least one object of interest. The object of interest may be any living creature or part of a living creature, including, without limitation, a human.

At least one gamma emitting radiopharmaceutical may be administered to an object of interest. A commonly used gamma emitting isotope may be 99mTc, but a person of ordinary skill in the art understands that other radiotracers may also be used. Radiopharmaceuticals may follow physiological and biochemical pathways leading to a distribution of the emissivity inside the body of the object.

Gamma emitting isotopes may emit photons at discrete energies only. The emitted photons may lose energy in inelastic scattering on charged particles. The actual SPECT part of the present embodiment may record only gamma photons having substantially full energy to minimize the contribution of the scattered gamma radiation to the image. Elastic scattering is not removed by this procedure, but the effect of elastic scattering on image quality is less significant than inelastic scattering.

The presented embodiment may determine the x-ray attenuation distribution inside the object of interest by a CT scan. Attenuation coefficients for the gamma emission of the specific administered radionuclide may be estimated from the measured x-ray attenuation coefficients. It is contemplated that a straightforward method to estimate the distribution of the gamma attenuation may be the CT scan. Any other method, however, that gives a useful estimation of the gamma absorption in the voxels rendered to the object of interest may also be used. It is contemplated that other morphological imaging methods, including, without limitation, MRI, Optical tomography or UltraSound, may be used to create a 3D image of the object of interest. The anatomic details may then be combined with a table or database that assigns gamma attenuation values to the tissues in the 3D image.

In one aspect, this embodiment may use the recorded gamma illumination of the detector elements and the attenuation coefficients determined as described above to reconstruct the distribution of the gamma photon emissivity inside the body.

Referring to FIG. 1, the illustrated system 10 includes an x-ray source 102 capable of substantially emitting x-radiation 104. The x-ray source 102 may be connected to an x-ray source controller 152. The x-ray source controller 152 may control the timing and intensity of the emission of the x-ray source. It is contemplated that any source of high energy photons may be used as the x-ray source. These x-ray sources may include one or more gamma emitting isotopes, or less traditional x-ray emitters, including, without limitation, x-ray sources using nanotubes as cathode. In this exemplary embodiment typically the x-ray source 102 is an x-ray tube.

Adjacent to the x-ray source 102 a collimator 106 may be positioned, through which an x-ray beam 108 may reach the object of interest 114. The object of interest 114 may be a body or part of the body of any living creature, including, without limitation, a human. The collimator 106 may be controlled by a collimator controller 156. Other embodiments may include x-ray sources and detectors with different geometry, which may or may not use a collimator.

The portion of the x-ray beam 108 which passes through and around the object of interest is the x-ray beam 116, which hits an x-ray detector 118. The detector 118 may include at least one detector element, which is sensitive to x-radiation. The detector 118 in this exemplary embodiment may be an Anger camera. It is contemplated that instead of the Anger camera any other suitable detector may be used which is capable of detecting and measuring the spatial distribution of x-radiation intensity. The x-ray detector 118 may be coupled to an x-ray detector controller 168. In this exemplary embodiment, detector 118 produces electrical signals that represent the intensity of the incident x-ray beams. In this exemplary embodiment, the x-ray detector controller electronics 168 typically receives analogue electrical signals from the detector 118 and converts data to digital signals for subsequent processing by a computer 150. It is contemplated that in another embodiment an analogue to digital converter may also be incorporated in the detector, to obtain directly digital data from the detector.

The intensity measurements which may be produced by the detector 118 may undergo pre-processing and calibration to condition the data to represent the line integrals of the attenuation coefficients of the scanned object of interest. The obtained projection data may then be filtered and used to reconstruct at least one image of the scanned area or part of the scanned area.

The exemplary embodiment presented here may also involve a rotational subsystem 110 connected to a rotational subsystem controller 160 and a translational subsystem 112 connected to a translational subsystem controller 162. The rotational subsystem 110 and the translational subsystem 112 may allow collection of projections of the object of interest along different directions. Other embodiments may omit either or both of the rotational and translational subsystems, and may use immobile arrangements to collect different projections of the object of interest.

To allow SPECT imaging at least one radioisotope 120 may be administered to the object of interest 114. The at least one radioisotope 120 may be capable of substantially emitting gamma radiation 122. A portion 124 of the emitted gamma radiation 122 may reach gamma detector 126, which may be a collimator-free gamma detector or a gamma detector partially covered by a collimator.

The gamma detector 126, which may be collimator-free or partially covered by a collimator, may contain at least one detector element, which may be sensitive to the emitted gamma radiation from the object 114. The detector 126 may be used to detect only gamma photons that did not lose energy in inelastic scattering. The detector 126 may include a scintillation element, or a direct conversion material. The detector 126 in this exemplary embodiment may be an Anger type detector, which may be coupled to a gamma detector controller 176. In this exemplary embodiment, detector elements of the array produce electrical signals that represent the intensity of the incident gamma radiation. In this exemplary embodiment, the controller electronics 176 typically receives analogue electrical signals from the detector 126 and converts data to digital signals for subsequent processing by a computer 150. In another embodiment an analogue to digital converter may also be incorporated in the detector 126 so as to obtain directly digital data from the detector. In another embodiment the detectors 118 and 126 may be replaced by a single detector. The gamma photon counting measurements done by the detector 126 may undergo pre-processing and calibration.

The x-ray source controller 152, the rotational subsystem controller 160, the translational subsystem controller 162, the x-ray detector controller 168, and the gamma detector controller 176 may all be integrated in a system controller 180, which may be coupled to computer 150. In this exemplary embodiment, the system controller 180 may command operation of the imaging, data acquisition and preliminary data processing. In one embodiment the x-ray source controller 152, the rotational subsystem controller 160, the translational subsystem controller 162, the x-ray detector controller 168, and the gamma detector controller 176 units may be built of several sub-controllers performing parts of the tasks, or are integrated or grouped differently, may also be suitable.

The system controller 180 may perform several tasks connected to the data collection and processing. The system controller 180 may control the x-ray power emitted by the x-ray source 102. It may also command the data acquisition done with the x-ray detector 118 or gamma detector 126. System controller 180 may synchronize the movement generated by the rotational subsystem 110 and the translational subsystem 112 with data collection. It may also carry out various data processing and filtering tasks, it may adjust the dynamic ranges, or perform interleaving the digital image data. In this embodiment, the system controller 180 may include a general purpose or an application specific digital computer, with memory units for storing executable routines, settings, configuration parameters, collected data, and so forth.

FIG. 2 is a flowchart illustrating one embodiment of a method of conducting a combination x-ray and photon computed tomography of an object using the system 10 illustrated in FIG. 1 for generating at least one image of the distribution of at least one radiotracer inside the at least one object of interest. The method may also generate at least one image representing the distribution of the x-ray absorption inside the at least one object of interest. The produced images of the object of interest may be considered as attenuation and emission values rendered to a plurality of volumetric elements (voxels). The voxels collectively define the object of interest.

The method illustrated in FIG. 2, with continued reference to FIG. 1, may start in step 201 by administering to the at least one object of interest the at least one suitable gamma emitting radioisotope. If the object is a living creature, the suitable gamma emitting radioisotope may be included in a pharmaceutical compound which may be administered to the living creature by any conventional form, such as by injection, inhalation, or oral intake.

In step 202 x-rays 104 may be generated by the x-ray source 102, and shaped by a collimator and filter 106 in step 204. In step 206 the x-ray 108 may be passed through the object of interest 114 at a plurality of different directions. In step 208 the x-ray 116 that passed through the object of interest 114 may be detected by the detector 118.

In step 210 at least one image representing the x-ray attenuation distribution inside the object of interest may be determined. The exemplary embodiment described here may use weighted filtered back projection to determine the x-ray CT image. Several other methods exist that may be used to reconstruct three dimensional images from their x-ray projections. Other embodiments may use other reconstruction methods.

The contemplated algorithm may include a series of weighting, filtering and back projection steps for each projection measurement over the reconstruction volume. Weighting of the projection data may be performed by an element by element multiplication with an array containing the weighting factors. The filtering step may use a series of convolutions to decorrelate image data points. In the back projection step the projection measurements may be added to all voxels in the line of the projection line. Different x-ray beam geometries may be taken in account through the use of weighting factors in the back projection.

In step 212 the absorbance or attenuation distribution of the object of interest for the gamma photons emitted by the used isotope may be calculated. The presented embodiment may perform this calculation using a bilinear dependence of the attenuation of gamma radiation upon the attenuation of x-rays. Although the bilinear relationship is widely used, other embodiments may use other models to convert measured x-ray densities into gamma absorbance values. Any method may be suitable which allows the estimation of the attenuation of the emitted gamma photons inside the object of interest.

In step 214 gamma photon count measurements may be recorded at a plurality of positions surrounding the object of interest producing an array of gamma photon count data. The gamma photon count measurements may be recorded using a collimator free or partially collimated gamma detector. The detector may be used to detect only unscattered gamma photons which did not lose energy in inelastic scattering. Photon counts may reflect the intensity of the incident gamma radiation. The photon count reading of a detector, however, may vary even if the intensity falling on the detector element is constant. This variation may come partially from the inherent Poisson noise of counting events, and partially from other noise sources of the instrument. Longer data acquisition times may allow the collection of more photons, yielding a more accurate estimation of the intensity, but may also increase the examination time. Optimal data acquisition times may be set based on theoretical assumptions and calibration measurements.

Gamma photon emissions from the plurality of voxels may be calculated based on the measured photon counts. The gamma photon emissions may then be used to generate the at least one image of the at least one radioisotopes inside the at least one object of interest. There are several ways to calculate the gamma photon emissions form the plurality of voxels. The embodiment illustrated in FIG. 2 may use an image reconstruction process known as the inverse source problem. The objective of this reconstruction process is to reconstruct the distribution of sources based on measurements performed outside of the object of interest.

In step 216 at least one initial estimated distribution of the gamma photon emissivity inside the object of interest may be considered. This is an initial tomographic image, which may be used as a starting point for the iterative reconstruction of the image. In a simplest case the initial image may be a constant value for each volumetric element of the object of interest. The initial image may also take into account theoretical assumptions and earlier experience about the distribution of the gamma emissivity inside the object of interest.

Based on the at least one initial distribution of the gamma emissivity, the expected photon counts at the plurality of detected positions may be calculated in step 218. In one aspect of the present invention, the predicted photon counts m may be computed using the equation $m=G\times E$, where G is a matrix representing the Green's function of the mapping of emissivities on the detector; and E is the array of the estimated emissivities. The symbol $\times$ denotes matrix product. The result is a predicted or expected photon count array m, which is identical in size to the measured photon count data.

The Green function G may be calculated from the distance of the voxel and detector element and the gamma absorbance of the object of interest between the actual voxel and detector element.

In step 220 the difference between the expected 218 and measured photon counts 214 may be calculated, and compared to a preset threshold value indicative of little or no difference in step 222. The obtained difference may also be an array having the same dimension as the predicted and measured photon counts. The threshold value may be optimized for best image quality.

If the difference is substantially larger than the threshold value, the distribution of the gamma photon emissivity inside the object of interest may be re-estimated (i.e. updated) in step 224 and the process reiterated from step 218. The steps 218, 220, 222 and 224 may be cycled again and again, until at decision point 222 the difference between the measured and expected photon count becomes approximately similar to the threshold. If the difference becomes approximately similar to the threshold value, including about, equal to or below the threshold value, then the re-estimated distribution of the gamma photon emissivity may be saved or visualized as a reconstructed image in step 226. By repeating the predicting and comparing steps described above and by the sequential updating of the estimated image, a final approximation or prediction of the image of the source may be reconstructed.

Re-estimation of the image step 224 may be obtained by solving the equation d=GxD. The inverse problem that needs to be solved may be described by a system of linear equations. The image may be computed by solving the equation d=G×D, where d is the array of differences between the measured and predicted photon counts; G is a matrix representing the Green's function of the mapping of emissivities on the detector; and D is the array of differences between the true emissions from the voxels and the estimated emission image. The symbol × denotes matrix product.

The Green function G may be calculated from the distance of the voxel and detector element and the gamma attenuation of the object of interest between the actual voxel and detector element.

A number of conventionally known numerical computational methods exist for the solution of the Green's function equation d=G×D. The Green's function method may be used in each iterating step to produce the image difference array D. The prediction of the image may then be updated by incorporating (i.e. adding, using a multiplication factor, or any other suitable mathematics) to the present estimation the calculated difference D to provide the next updated estimated image.

A final image may be obtained after a sufficient number of iteration steps, when the compared difference d of the measured and expected photon counts becomes relatively small (i.e. approximately similar to, equal to or below the threshold value). It should be understood, however, that a final image may also be obtained when d is larger than the threshold value. The final image may be reconstructed from any of the re-estimated gamma emission or from an average of several re-estimated gamma emission images or any other method using one or more of the updated gamma emission images of the distribution of the at least one emitting radioisotopes from the voxels.

The present embodiment uses an iterative method to solve the inverse problem associated with the image reconstruction process. Several other iterative algorithms or methods based on the Fourier Slice Theorem are routinely used to solve such mathematical problems. We contemplate that any of these algorithms, or in very simple cases even elementary matrix calculus may also be successfully used.

Single Photon Emission Computed Tomograph (SPECT) System

FIG. 3 shows a diagram of an exemplary embodiment of a SPECT system 30. This embodiment may be capable of collecting measurements and reconstruct at least one image of the distribution of at least one radioisotope inside at least one object of interest. System 30 may also be capable to reconstruct an image representing the distribution of the gamma attenuation inside the object of interest. The object of interest may be any living creature or part of a living creature.

To allow SPECT imaging at least one radioisotope 306 may be administered to the object of interest. The radioisotope 306 may be capable of emitting gamma radiation 308. A portion 310 of the emitted gamma radiation 306 may reach a gamma detector 312.

Gamma detector 312 may be characterized by the absence of a collimator or it may be partially covered by a collimator. Gamma detector 312 may include at least one detector element, which may be sensitive to the emitted gamma radiation. The detector 312 may be used to detect only gamma photons which did not lose energy in inelastic scattering. The detector 312 may include a scintillation element, or a direct conversion material. The detector 312 in this exemplary embodiment may be an Anger type detector, which may be coupled to a gamma detector controller 362. In this exemplary embodiment, detector elements of the array may produce electrical signals that represent the intensity of the incident gamma radiation. In this exemplary embodiment, the controller electronics 362 may typically receive analogue electrical signals from the detector 312 and may convert data to digital signals for subsequent processing by a computer 350. In one embodiment an analogue to digital converter may also be incorporated in the detector 312, to obtain directly digital data from the detector 312. The measurements done by the detector 312 may undergo pre-processing and calibration.

The exemplary embodiment presented herein may also involve a translational subsystem 302 connected to a translational subsystem controller 352. The translational subsystem 302 may allow the movement of the object of interest and the detector compared to each other.

The translational subsystem controller 352 and the gamma detector controller 362 may all be integrated in a system controller 380, which is coupled to the computer 350. In this exemplary embodiment, the system controller 380 may command operation of the imaging, data acquisition and preliminary data processing. In one embodiment of the present invention the translational subsystem controller 352 and the gamma detector controller 362 units may be built of several sub-controllers performing parts of the tasks. Any other form of integration between the subsystem controller 352 and the gamma detector controller 362, may also be suitable.

The system controller 380 may perform several tasks connected to the data collection and processing. It may command the data acquisition done with the detector 312. The system controller 380 may synchronize the movement generated by the translational subsystem 302 with data collection. It may also carry out various data processing and filtering tasks, it may adjust the dynamic ranges, or perform interleaving the digital image data. In this embodiment, the system controller 380 may include a general purpose or an application specific digital computer, with memory units for storing executable routines, settings, configuration parameters, collected data, and so forth.

FIG. 4 is a flowchart including exemplary steps for generating at least one image of the distribution of at least one radiotracer inside the body of at least one object of interest. This embodiment may also determine one or more images representing the distribution of the gamma attenuation inside the object of interest. The produced images of the object of interest may be considered as attenuation and emission values rendered to a plurality of volumetric elements (voxels). The voxels collectively define the object of interest.

In step 401 at least one gamma emitting radioisotope may be administered to the object of interest.

In step 402 a plurality of gamma photon count measurements may be recorded at a plurality of positions surrounding the object of interest producing an array of photon count data. The gamma photon count measurements may be recorded using a collimator free or partially collimated gamma detector. The detector may be used to detect only gamma photons which did not lose energy in inelastic scattering. Photon counts may reflect the intensity of the incident gamma radiation. The photon count reading of a detector, however, may vary even if the intensity falling on the detector element is constant. This variation may come partially from the inherent Poisson noise of counting events, and partially from other noise sources of the instrument. Longer data acquisition times may allow the collection of more photons, yielding a more accurate estimation of the intensity, but may also increase the examination time. Optimal data acquisition times may be set based on theoretical assumptions and calibration measurements.

Gamma photon emissions from the plurality of voxels may be calculated based on the measured photon counts. The gamma photon emissions may then be used to generate the at least one image of the radioisotopes inside the at least one object of interest. There are several ways to calculate the gamma photon emissions form the plurality of voxels. The embodiment illustrated in FIG. 4 may use an image reconstruction process known as the inverse source problem. The goal of this process is to reconstruct the distribution of the gamma sources and absorbers based on measurements performed outside of the object of interest. An accurate measurement of the activity of the administered at least one radioisotope may provide a constrain which may also facilitate image reconstruction.

In step 404 at least one initial distribution of the gamma photon emissivity and gamma photon attenuation inside the object of interest may be considered. This may be initial tomograph images that may be used as a starting point for the iterative reconstruction of the image. In a simplest case the at least one initial image may be a constant value for each volumetric element of the object of interest. The initial image may also take into account theoretical assumptions and earlier experience about the distribution of the gamma emissivity inside the object of interest.

The expected gamma photon counts at the plurality of detected positions may be calculated in step 406 based on the at least one initial emissivity and attenuation distributions. The result is a predicted photon count array identical in size to the measured photon count data.

In step 408 the difference between the expected and measured photon counts may be calculated, and compared to a preset threshold value in step 410. The obtained difference is also an array having the same dimension as the predicted and measured photon counts. The threshold value may be optimized for best image quality based on earlier experience.

If the difference is substantially larger than the threshold value, the distribution of the gamma photon emissivity and the gamma attenuation inside the object of interest may be re-estimated in step 412, and the process reiterated from step 406. If the difference becomes approximately similar to the threshold, including about, equal to or below the threshold, the estimated distribution of the gamma photon emissivity and the gamma attenuation may be saved or visualized as reconstructed images of the object of interest in step 414. By repeating the predicting and comparing steps described above and by the sequential updating of the emission and attenuation image prediction, a final approximation or prediction of the image of the object of interest may be obtained.

Re-estimation of the image (step 412) may be an important part of the image reconstruction algorithm. The inverse problem that needs to be solved may be described by a system of linear equations. In one aspect of the present invention, the image may be computed by solving the equation $d=G(A)\times DE+d_AG^T\times E_{EST}\times DA$. Here d is the array of differences between the measured and predicted emissivities; A is the array describing the absorbance of the object of interest; E_{EST} is the estimated image array describing the emission of the object of interest; G(A) is the matrix representing the Green's function. (The Green's function depends on the array of gamma absorbances A inside the object of interest.) DE is the array of differences between the true and estimated emission image; DA is the array of differences between the true and estimated absorption image; d_AG^T is the transposed matrix of the derivative of G with respect to the absorbance array. The matrix d_AG^T may be calculated from the positions of the voxels and detector elements. Filling in instead of the true arrays A our estimation for it, we may receive a simple linear equation system to determine the arrays DA and DE. The prediction of the gamma emission and gamma attenuation image may be updated by adding to the present prediction the calculated difference DA and DE arrays to provide the next predicted image.

A final image may be obtained after a sufficient number of iteration steps, when the compared difference d of the measured and predicted emissivities becomes relatively small (i.e. approximately similar to the threshold value). It should be understood, however, that a final image may also be obtained when d is larger than the threshold value. The final image may be reconstructed from any of the re-estimated gamma emission images or from an average of several re-estimated gamma emission images or any other method using one or more of the updated gamma emission images of the distribution of the at least one emitting radioisotopes from the voxels.

As will be appreciated by those skilled in the art, the exemplary embodiments of the measuring devices and methods described above may involve extensive computer calculations. These calculations may include a listing of computer code containing executable instructions. This listing (program) may be embodied in any computer-readable information storage device, for use by or in connection with a system which can execute the instructions. The processing may be done local to the acquisition or local to the storage of the acquired data. Alternatively, some or all the calculations may be performed remotely. The computer-readable information storage device may be any means that can contain, store, communicate, propagate, transmit or transport information. The usable devices may apply electronic, magnetic, optical, electromagnetic, mechanic, nanotechnology-based media, but are not limited to these.

The presented embodiments are described here as exemplary systems only. It should be noted that the presented systems and methods are in no way limited to the actual arrangements described.

<u>Advantages</u>

The present invention may have several economical, technical and health care benefits. Described below are a few uses and benefits of the present invention for illustrative and exemplary purposes only. These descriptions should not be taken as limiting the scope of the invention.

An important advantage of the disclosed improved collimator-free method may be that no collimator is necessary thus a significantly higher fraction of the available radiation is evaluated and used for the imaging. The invention may achieve an increased efficiency, i.e. the available radiation may be used more efficiently for image generation. This may be beneficial in several different ways, including, without limitation: i) The dose to be administered to the patient may be smaller, thus the exposure of the patient and the cost of the examination may be reduced. ii) The acquisition time may be shortened, and faster physiological processes followed. This may open new diagnostic fields which were not available due to the slow data acquisition of the prior art.

Collimators are typically extremely delicate, position-sensitive and therewith very elaborate to manufacture. Omission of the SPECT collimator is a simplification in the technical complexity of the apparatus and may allow cost savings.

The setup and methods of the present invention may have several advantages compared to the state-of-the-art collimator-free methods. The methods of the present invention may be simpler, more robust, while some embodiments allow the extraction of attenuation images as well.

SPECT imaging already has a substantial market with several fields of use. Customers interested in SPECT imaging may find the new collimator-free SPECT method useful.

The new method of the present invention may require only minor modification of existing data acquisition modules and data processing procedures. This may have several advantages. i) The costs of the development necessary to integrate the new method into presently manufactured equipment may be small. ii) Manufacturing instruments in which the new method has been integrated may not be more expensive, or even cheaper than the present instruments lacking the new method. iii) Switching to manufacture instruments that use the new method may not need large changes in the production procedures. iv) Modifications required for the new method may inexpensively be added to imaging devices installed earlier. The new collimator-free method thus may be installed for established customers as well.

One embodiment presented herein may extract the gamma attenuation image simultaneously with the SPECT image. This image may contain diagnostic information similar to the x-ray CT images. We obtain the additional information compared to a conventional SPECT image without extra exposure to ionizing radiation.

Other embodiments

Variations or modifications to the design and construction of this invention, may occur to those skilled in the art upon reviewing this disclosure. Such variations or modifications, if within the spirit of this invention, are intended to be encompassed within this provisional patent application, as well as within the patent application intended to be filed based on it, and the resulting patent protection issuing upon this invention.

Conclusions, Ramifications, and Scope

The description above contains many specifications. These should not be construed as limiting the scope of the embodiments, but as merely providing illustrations of some of the presently preferred embodiments.

A goal of the disclosed method is to provide a more efficient way of imaging gamma emitting radioisotopes administered to objects of interest, particularly to living creatures by omitting the projecting collimator from the SPECT detection and imaging only photons with full energy.

We contemplate that work-arounds to the disclosed method may cover some part of the detector with a collimator or absorber, without any functional reason, while leaving large part of the detector surface uncovered for a collimator-free detection. We consider these solutions covered by the present invention disclosure.

The present disclosure presents two embodiments which reconstruct the distribution of at least one gamma emitting isotope. The methods may allow simultaneous imaging of several gamma emitting isotopes. The attenuation for the gamma photon emitted may be determined from the CT scan for every isotope. The gamma emission of the different isotopes may be measured separately based on their different photon energy. The images of the different isotopes may be reconstructed separately.

Both presented exemplary embodiments use a gamma detector that surrounds the object of interest on four sides. Other embodiments may use a simpler gamma camera which may not surround the object of interest. In some systems the gamma camera may be moved to different positions around the object of interest to record gamma intensities at more positions.

Gamma detectors of the presented embodiments measure the intensity of the gamma radiation in different detector positions by performing a gamma photon counting. Instead of photon counts, other embodiments could use any other physical quantity indicating the strength of the radiation (for example: intensity, energy absorbed in the detector, electric field strength, magnetic field strength, etc.), or any mathematical function of these quantities.

One of the presented embodiments uses CT scan to determine the x-ray attenuation image of the studied body. We contemplate that any CT scanner that can be integrated with the SPECT instrument may also be used.

Glossary of Technical Terms

Collimator – A collimator may be defined as a device which narrows a beam of particles or waves. In the case of high energy photons this means the absorption of the photons that do not travel in the desired narrow beam.

Green's function – The Green's function may be defined as a solution of a partial differential equation for the case of a point source of unit strength.

CLAIMS

We claim:

1. A method for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, characterized in that said method comprises:

(a) administering the at least one radioisotope to the at least one object of interest, said at least one radioisotope being capable of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object;

(b) measuring an array of gamma photon counts from the at least one radioisotope inside the plurality of voxels at a plurality of positions around the at least one object of interest using one or more collimator free or partially collimated gamma detectors, said measuring counting gamma photons having substantially the same photon energy a gamma emission of said at least one radioisotope;

(c) calculating gamma photon emissions from the plurality of voxels based on the array of measured photon counts; and

(d) generating the at least one image of the distribution of the at least one radioisotope inside the at least one object of interest based on the calculated gamma photon emissions.

2. The method of claim 1 characterized in that said calculating gamma photon emissions step (c) comprises (i) providing at least one estimated gamma emission image of the distribution of the at least one emitting radioisotopes from the plurality of voxels; (ii) based on the at least one estimated image calculating at least one expected array of gamma photon counts at the plurality of detector positions using the at least one estimated gamma emission image; (iii) comparing the measured and the expected arrays of gamma photon counts at the plurality of detector positions to obtain at least one number representing differences between the measured array of gamma photon counts and the at least one expected array of gamma photon counts; and (iv) reiterating the predicting and comparing steps by sequential updating said at least one estimated gamma emission image until the at least one number representing the array differences is approximately similar to a threshold value; and wherein said generating step (d) comprises reconstructing the at least one image of

the distribution of the at least one radioisotope inside the object using at least one of the updated estimated gamma emission images.

3. The method of claim 2 characterized in that said method further comprises providing at least one estimated gamma attenuation image, and wherein said at least one expected array of gamma photon count is obtained by solving an equation comprising the at least one estimated emission image, the at least one estimated gamma attenuation image and a distance between the plurality of voxels and the gamma detector.

4. The method of claim 1 characterized in that said method further comprises exposing the object of interest to penetrating radiation and obtaining at least one attenuation image of the penetrating radiation through the object.

5. The method of claim 2 characterized in that said method further comprises exposing the object of interest to penetrating radiation and obtaining at least one attenuation image of the penetrating radiation through the object, and wherein said at least one estimated gamma attenuation image is obtained using the at least one attenuation image of said penetrating radiation.

6. The method of claim 5 characterized in that said sequential updating of the estimated gamma emission image comprises incorporating at least one emission image difference array to at least one of the estimated gamma emission image to be updated, said at least one emission image difference array being obtained using a distance between the plurality of voxels and the detector and a gamma attenuation of the at least one object of interest between the plurality of voxels and the detector.

7. The method of claim 6 characterized in that said image difference array is obtained by solving an equation $d = G \times D$ wherein d is the array of differences between the measured and the expected photon counts, G is a Green's function mapping emissions from inside the voxels on the detector, D is an array of differences between a true gamma emissions from the plurality of voxels and the estimated emission image, and x is a matrix product.

8. The method of claim 3 characterized in that said updating of the estimated gamma emission image and gamma attenuation image comprises incorporating to the at least one estimated gamma emission and to the at least one gamma

attenuation images at least one emission image difference array and at least one attenuation image difference array respectively, said emission image difference array and attenuation image difference array being obtained using the distance between the plurality of voxels and detector and the gamma absorbance of the object of interest between the plurality of voxels and the detector.

9. The method of claim 8 characterized in that said emission image difference array and attenuation image difference array are obtained by solving the equation $d=G(A)\times DE+d_AG^T\times E_{EST}\times DA$, where d is the array of differences between the measured and predicted emissions, A is an array of the gamma photon absorbance of the object of interest, E_{EST} is an estimated array of the gamma photon emission of the object of interest; G(A) is a matrix representing the Green's function, DE is the array of differences between an actual gamma emission from the plurality of voxels and the estimated emission image; DA is the array of differences between an actual gamma absorbance from the plurality of voxels and the estimated attenuation image; d_AG^T is a transposed matrix of the derivative of G with respect to the absorbance array.

10. A system for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, said radioisotope being capable of of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object, characterized in that said system comprises:

(a) at least one collimator free or partially collimated gamma detection means for measuring at least one array of gamma photon counts from the plurality of voxels at a plurality of positions around the at least one object of interest, said measured gamma photons having substantially the same energy as a gamma emission of said at least one radioisotope;

(b) means for calculating gamma photon emissions from the plurality of voxels based on the measured at least one gamma photon counts; and

(c) means for generating the at least one image of the distribution of the at least one radioisotope inside the at least one object of interest based on the calculated gamma photon emissions.

11. The system of claim 10 characterized in that said system further comprises means for calculating at least one estimated attenuation image of gamma photons emitted by the at least one radioisotope.

12. The system of claim 11 characterized in that said system further comprises:

(d) at least one source capable of substantially emitting penetrating radiation;

(e) at least one detector sensitive to said penetrating radiation, said detector capable of producing a plurality of measurements related to the penetrating radiation passing through the object of interest; and

(f) at least one image reconstruction processor means for reconstructing at least one image of the at least one object of interest based on the plurality of measurements related to the penetrating radiation.

13. A system for generating at least one image of a distribution of at least one radioisotope inside at least one object of interest, said radioisotope being capable of of substantially emitting gamma ray photons from a plurality of voxels inside the at least one object, characterized in that said system comprises:

(a) at least one collimator free or partially collimated gamma detection means for measuring at least one array of gamma photon counts from the plurality of voxels at a plurality of positions around the at least one object of interest, said measured gamma photons having substantially the same energy as a gamma emission of said at least one radioisotope;

(b) means for calculating at least one estimated gamma emission image of the at least one distribution of the at least one emitting radioisotopes from the voxels;

(c) means for calculating at least one expected array of gamma photon counts at the plurality of detector positions using the at least one estimated gamma emission image;

(d) means for comparing the measured and the expected arrays of gamma photon counts at the plurality of detector positions to obtain at least one number representing differences between the measured and the expected arrays of gamma photon counts;

(e) means for updating the at least one estimated gamma emission image; and

(f) means for reconstructing at least one final image of the at least one distribution of the at least one radioisotope inside the at least one object using at least one updated gamma emission image.

14. The system of claim 13 characterized in that said means for calculating the at least one expected array of gamma photon counts comprises a Green's function solving means of the mapping of gamma emissions on the detector.

15. The system of claim 13 characterized in that said means for updating the at least one estimated gamma emission image comprises a Green's function solving means of the mapping of gamma emissions on the detector.

16. The system of claim 13 characterized in that said system further comprises means for calculating at least one estimated attenuation image of gamma photons emitted by the at least one radioisotope.

17. The system of claim 16 characterized in that said system further comprises:

(g) at least one source capable of substantially emitting penetrating radiation;

(h) at least one detector sensitive to said penetrating radiation, said detector capable of producing a plurality of measurements related to the penetrating radiation passing through the object of interest; and

(i) at least one image reconstruction processor means for reconstructing at least one image of the at least one object of interest based on the plurality of measurements related to the penetrating radiation.

18. The system of claim 17 characterized in that said means for calculating the expected attenuation uses at least one attenuation of the penetrating radiation through the at least one object of interest.

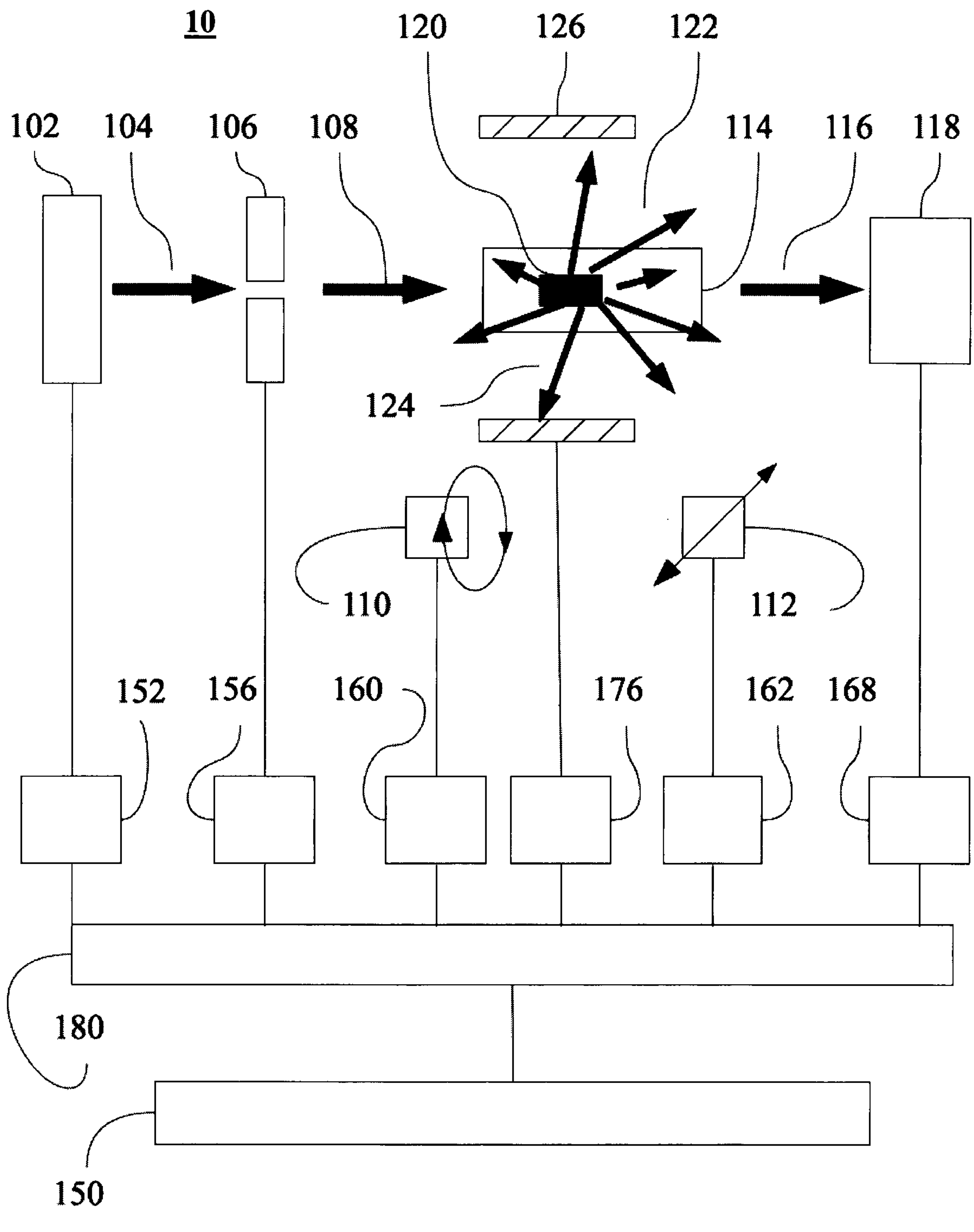
10
120
126
122
102
104
106
108
114
116
118
124
110
112
152
156
160
176
162
168
180
150

FIG. 1

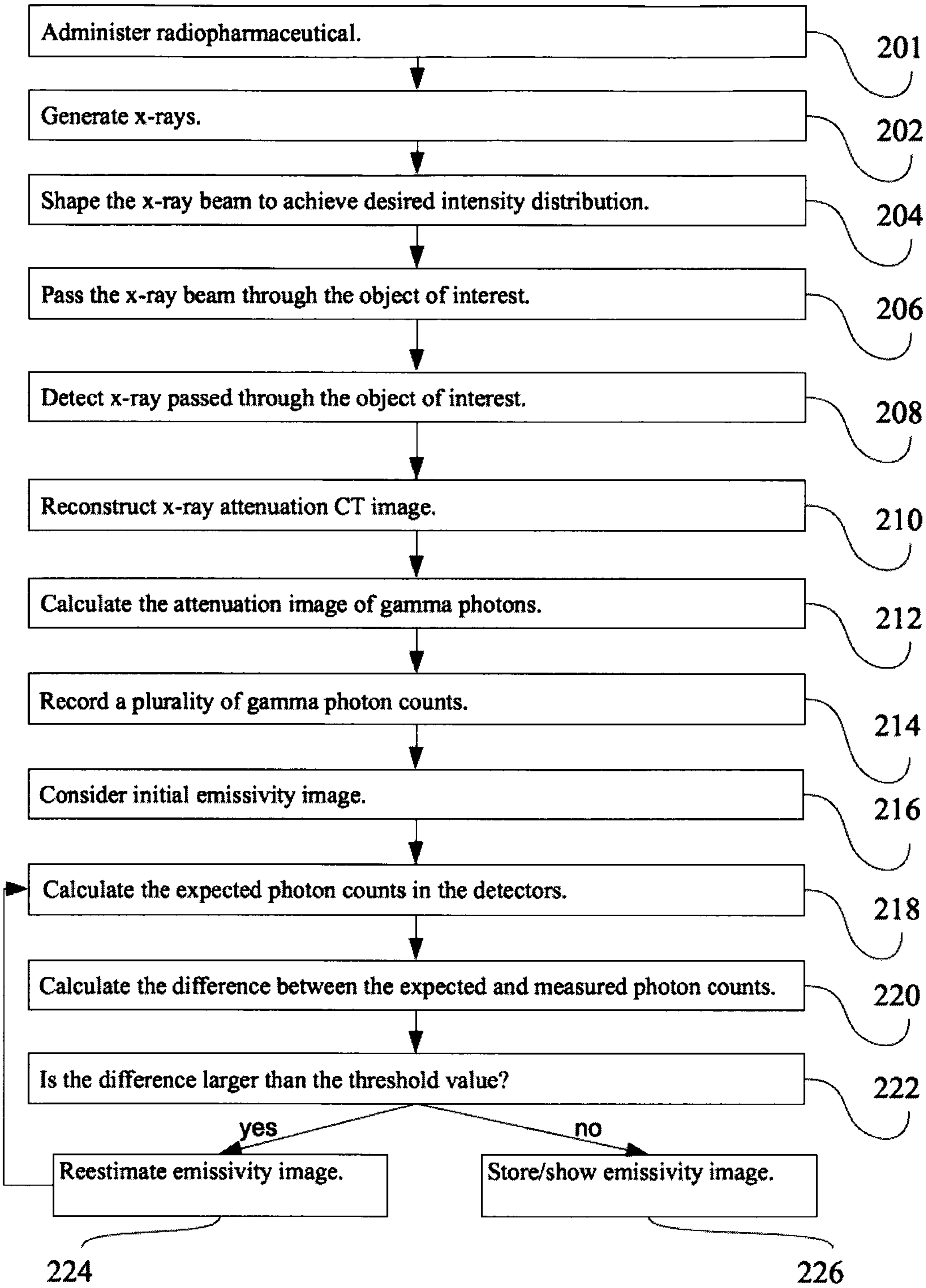
Administer radiopharmaceutical.
201
Generate x-rays.
202
Shape the x-ray beam to achieve desired intensity distribution.
204
Pass the x-ray beam through the object of interest.
206
Detect x-ray passed through the object of interest.
208
Reconstruct x-ray attenuation CT image.
210
Calculate the attenuation image of gamma photons.
212
Record a plurality of gamma photon counts.
214
Consider initial emissivity image.
216
Calculate the expected photon counts in the detectors.
218
Calculate the difference between the expected and measured photon counts.
220
Is the difference larger than the threshold value?
222
yes
no
Reestimate emissivity image.
Store/show emissivity image.
224
226

FIG. 2

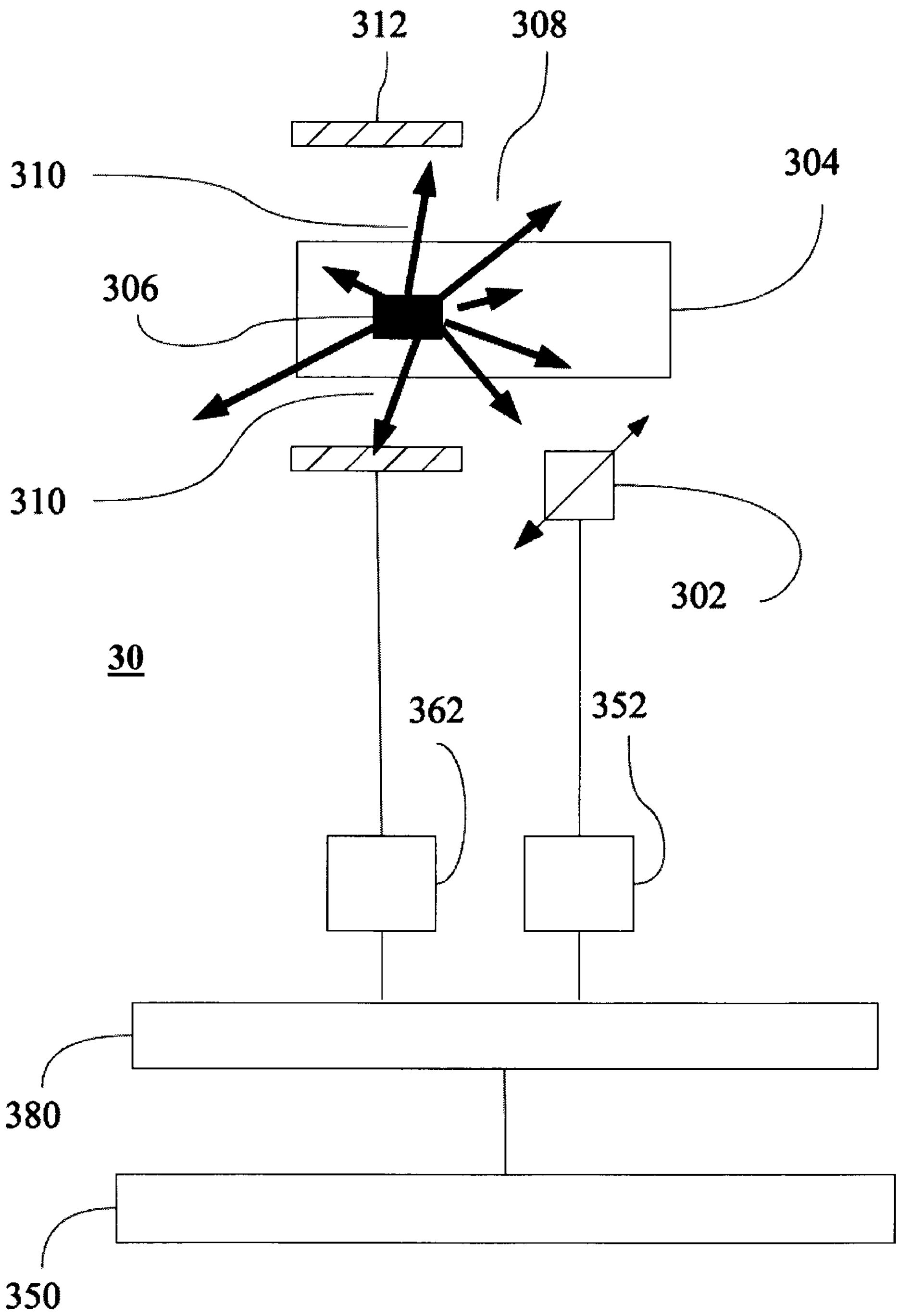

FIG. 3

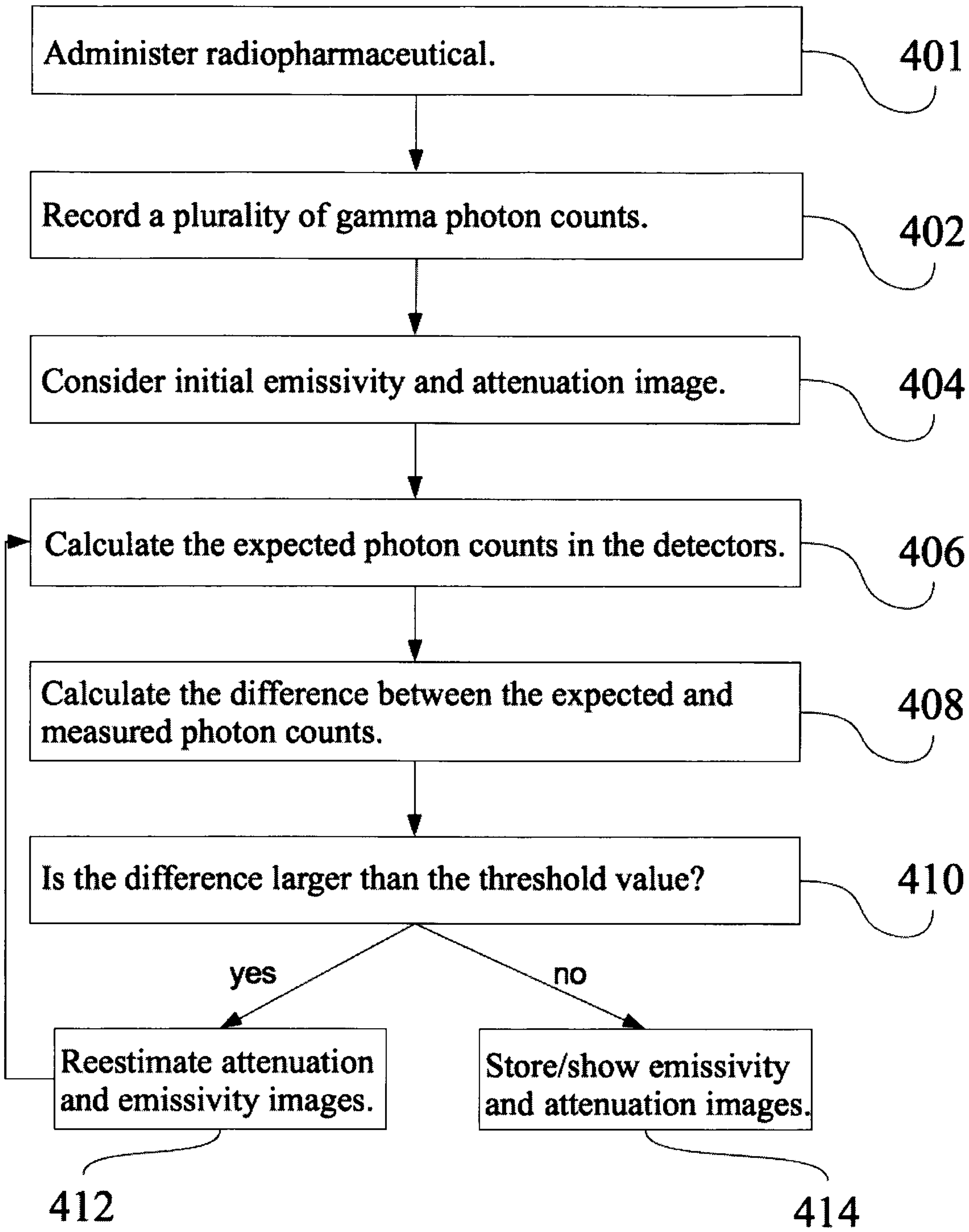
Administer radiopharmaceutical.
401
Record a plurality of gamma photon counts.
402
Consider initial emissivity and attenuation image.
404
Calculate the expected photon counts in the detectors.
406
Calculate the difference between the expected and measured photon counts.
408
Is the difference larger than the threshold value?
410
yes
no
Reestimate attenuation and emissivity images.
412
Store/show emissivity and attenuation images.
414

FIG. 4