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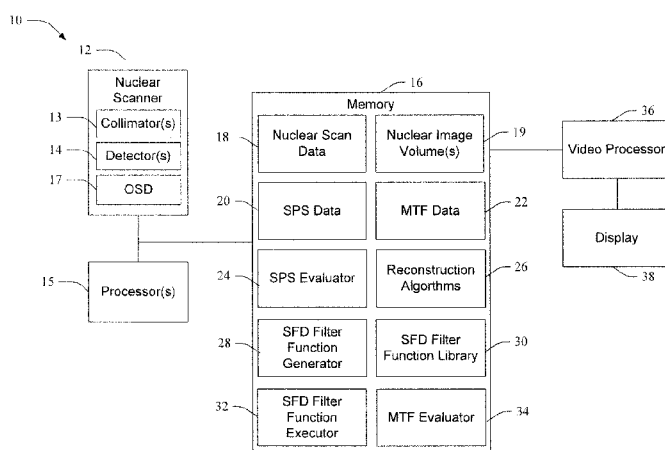
(54) Title: HIGH CONTRAST IMAGING AND FAST IMAGING RECONSTRUCTION

FIG. 1

(57) **Abstract:** When reconstructing low-collimation nuclear scan data (18) (e.g., SPECT) into a nuclear image volume (19), a spatial frequency-dependent (SFD) filter function is applied in Fourier space to the reconstructed image (19) to improve image resolution given a predefined number of reconstruction iterations and/or to reduce the number of reconstruction iterations required to achieve a predetermined level of image resolution. Size of an object to be imaged is determined, and the SFD filter function is determined or generated based on signal power spectrum (and/or modulated transfer function) data, object size, and desired image quality (or number of reconstruction iterations). The SFD filter function amplifies higher-energy components (e.g., corresponding to a lesion or tumor, or the like) of the spatial frequency spectrum to improve viability in a low collimated nuclear image (19) using fewer reconstruction iterations.



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HIGH CONTRAST IMAGING AND FAST IMAGING RECONSTRUCTION

DESCRIPTION

The present innovation finds particular application in anatomic and functional imaging systems, particularly nuclear imaging such as single photon emission computed tomography (SPECT), positron emission tomography (PET), and the like.

5 However, it will be appreciated that the described technique may also find application in other imaging systems, other image processing scenarios, other image reconstructing techniques, and the like.

SPECT cameras use collimators for imaging of gamma quanta. Every collimation elicits a trade-off between system efficiency and spatial resolution. The use
10 of low collimation, and thus higher signal-to-noise ratio (SNR), in combination with resolution recovery is valuable. Low collimation (short collimator septa) has an advantage in terms of SNR, specifically in the low frequency part of the detective quantum efficiency (DQE), a standard measure of image quality well known in x-ray imaging. The problem with low collimation is a low DQE in the medium and high
15 spatial frequency range. This impedes the detection of small structures, e.g. small lesions in oncology.

SPECT images are usually reconstructed in 2D transaxial slices or 2D projections, i.e. in a 2D reconstruction. The typical size of one reconstructed slice is 64 x 64 voxels in cardiac SPECT. For high quality images, especially in oncology, a larger
20 matrix size such as 128 x 128 is desirable. In other imaging modalities, larger matrix sizes are common. When quantitative information is important, an iterative reconstruction algorithm with a high number of iterations, typically 8 to 32, is useful to provide this information (See, e.g., H. Wiczorek, "SPECT Image Quality and Quantification," 2006 IEEE Nuclear Science Conference Record, pp. 2854-2858 (2006)).
25 For these reasons, classical iterative SPECT reconstruction approaches are highly time-consuming.

Large reconstructed volumes, such as a 128 x 128 x 128 matrix used for imaging a volume of interest, are desirable to detect small tumors or other small volumes, but they require about 16 times longer for reconstruction than the typical 64 x 64 x 64

matrix used in conventional Cardiac SPECT. The 8 to 32 iterations typically required for quantitative SPECT, a high number of iterations, typically 8 to 32, represents a 4-fold to 8-fold increase in reconstruction time compared to typical values of 2 to 4 iterations in cardiology.

5 Conventional reconstruction software is optimized for speed and with today's computers takes about 5 seconds per iteration on a 64 x 64 matrix. With 128 x 128 matrix size and a doubled number of projections, this results in more than one minute being required per iteration. For quantitative SPECT, half-hour reconstruction times are not uncommon, and for dual isotope imaging reconstruction times are on the order of
10 hours.

 The present application provides new and improved systems and methods for improving SPECT image quality and reducing reconstruction time, which overcome the above-referenced problems and others.

15 In accordance with one aspect, a nuclear image optimization system includes a nuclear scanner that acquires nuclear scan data of a volume of interest (VOI), and having one or more collimators that provide a low level of collimation during nuclear scan data acquisition, an object size detector that determines a size of the VOI, and a
20 filter function library having a lookup table that identifies one or more object size-dependent filter functions for each of a plurality of object sizes. The system further includes a processor that performs an iterative reconstruction algorithm, executes the object size-dependent filter function after executing a predetermined number of reconstruction algorithm iterations, and outputs an enhanced nuclear image volume
25 representation.

 In accordance with another aspect, a method of increasing nuclear image reconstruction speed for a predetermined number of reconstruction iterations includes acquiring nuclear scan data of an object, determining a size of the object, and identifying
30 an object size-dependent filter function for application to a reconstructed image of the object based on the object size and the predetermined number of reconstruction iterations. The method further includes performing the predetermined number of reconstruction iterations to reconstruct a nuclear image of the object, applying the object size-dependent

filter function to the reconstructed image after a selected number of reconstruction iterations to enhance resolution, and storing the enhanced reconstructed image to memory.

5 In accordance with another aspect, a method of decreasing the number of reconstruction iterations executed to achieve a desired level of image resolution includes acquiring low-collimation nuclear scan data of an object, determining a size of the object, and identifying an object size-dependent filter function for application to a reconstructed image of the object based on the object size and the desired image resolution. The method further includes performing a number of reconstruction iterations to reconstruct a
10 nuclear image of the object, applying the object size-dependent filter function during reconstruction to enhance resolution, and storing the enhanced reconstructed image to memory.

One advantage is that low collimation improves system efficiency and therefore improves signal-to-noise ratio.

15 Another advantage resides in lower spatial resolution for low collimation that is partly recovered by resolution recovery, which inherently gives an extra noise reduction.

Another advantage resides in enhancing contrast for small objects using an additional filter in Fourier space.

20 Another advantage resides in improved image quality, specifically contrast to noise ratio, even with a dual matched filter applied.

Still further advantages of the subject innovation will be appreciated by those of ordinary skill in the art upon reading and understand the following detailed description.

25

The innovation may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for purposes of illustrating various aspects and are not to be construed as limiting the
30 invention.

FIGURE 1 illustrates a system that facilitates optimizing nuclear images, such as SPECT images, for contrast and noise.

FIGURE 2 illustrates a graph showing signal power spectra for standard collimation with resolution recovery (e.g., iterative reconstruction with object position dependent collimator spatial resolution information applied in the forward projection of a dual matrix reconstructor) for an object with a 150 mm diameter.

5 FIGURE 3 illustrates a graph showing signal power spectra for low collimation with resolution recovery (e.g., iterative reconstruction with object position dependent collimator spatial resolution information applied in the forward projection of a dual matrix reconstructor) for an object with a 150 mm diameter.

10 FIGURES 4A-D illustrate examples of reconstructed SPECT images of a volume of interest with a simulated lesion or tumor under various combinations of collimation level and resolution recovery at 16 iterations of a reconstruction algorithm.

FIGURE 5 illustrates a graph showing signal power spectra determined from a simulated object (e.g., having a diameter of approximately 150 mm) on a logarithmic scale.

15 FIGURE 6 illustrates a graph showing signal power spectra determined from a simulated object (e.g., having a diameter of approximately 150 mm) on a linear scale.

20 FIGURES 7A-D illustrate a plurality of reconstructed images of a cross-section of a phantom with a simulated lesion, and associated central profile plots, generated with varying numbers of reconstruction iterations and optional SFD filtering.

FIGURE 8 illustrates a curve representing a noise power spectrum for reconstructed image after 2 iterations with SFD filtering.

25 FIGURE 9 illustrates a method for reducing nuclear image reconstruction time using SFD filter functions applied to the image in Fourier space after iterative reconstruction.

30 FIGURE 10 illustrates a method for improving nuclear image quality for a given number of reconstruction algorithm iterations using SFD filter functions applied to the reconstructed image in Fourier space after the given number of reconstruction iterations.

FIGURE 1 illustrates a system **10** that facilitates optimizing nuclear images, such as SPECT images, for contrast and noise. The system uses low collimation in combination with iterative reconstruction, resolution recovery and an object size dependent filter applied in Fourier space. The extra filtering step in Fourier space gives
5 increased contrast for small structures and facilitates optimizing for lesion detection and noise performance. Additionally, the system **10** facilitates iterative nuclear image reconstruction that reduces a number of iterations performed for a given level of image quality. In one embodiment, the system **10** replaces a large number of iterations by a much smaller number of iterations with subsequent filtering in Fourier space. Exact filter
10 parameters are specified according to object size and other a priori information, and are pre-determined and stored in an object library.

The system **10** includes a nuclear scanner **12** having one or more collimators **13** for collimating rays incident to a respective nuclear detector **14** that acquires patient data during a scan of a patient or subject. In SPECT imaging, incoming
15 radiation is collimated to define spatial trajectories along which the radiation events originated. The higher the collimator septa, the smaller the viewing cone through each aperture and the finer the resolution, but the less the data that successfully passes the collimator. Conversely, reducing collimation dilates the viewing cone and increases the amount of data detected, and thus increases the signal-to-noise ratio (SNR), but reduces
20 resolution. In order to recover lost resolution due to reduced collimation, the system **10** employs an iterative reconstruction technique that applies position dependent collimator resolution information (e.g., collimator-object voxel distance information) in the forward projection step of a dual matrix reconstructor.

The system further includes a processor **15** that executes, and a memory
25 **16** that stores, computer-executable instructions for carrying out the various steps and/or providing the various functions described herein. The scanner **12** additionally includes an object size detector (OSD) **17** (e.g., a laser gauge, a priori knowledge, one or more proximity sensors, a camera sensor, a CT scanning device, the nuclear detector(s), etc.) that determines the size of a volume of interest or the like. In one embodiment, the object
30 size detector employs a priori knowledge based on acquired nuclear scan data, the angular dependence of the object size in respective transaxial slices of the acquired nuclear scan data, etc. The scanner **12** scans an object or subject and acquires nuclear

scan data **18** that is reconstructed into one or more nuclear image volumes **19**. In one embodiment, object size is determined from the acquired nuclear scan data and/or the image volume data.

The memory **16** stores signal power spectrum (SPS) data **20** that is derived
5 from the scan data **18** and/or the OSD **17** and is a function of object (e.g., a tumor or other small volume of interest) size. Additionally or alternatively, the memory stores modulation transfer function (MTF) data **22** (e.g., normalized SPS data), which is a function of object size. The processor **15** executes an SPS evaluation algorithm **24** (e.g., a set of computer-executable instructions) to assess SPS data as a function of imaged
10 object size and a number of iterations of a nuclear image reconstruction algorithm **26** to facilitate low-collimation (e.g., 10-12 mm collimation, 18-20 mm collimation, etc.) imaging with resolution recovery during iterative reconstruction.

The processor **15** executes a spatial frequency-dependent (SFD) filter function generation algorithm **28** that identifies or generates SFD filter functions as a
15 function of the evaluated power spectra and/or object size. SFD filter functions are stored to a filter function library **30** in the memory **16** for recall and application during iterative execution of the reconstruction algorithm(s) **26**. In one embodiment, the filter function library includes a lookup table that the processor accesses to identify a filter function for a given object size. The processor **16** invokes an SFD filter function
20 executor **32** that applies one or more SFD filter functions in Fourier space during reconstruction of the nuclear image volume **19**, which enhances image quality and improves small structure (e.g., tumor) visibility. For instance, the filter function, which is selected as a function of object size, amplifies or boosts one or more high frequency components of the signal power spectrum of the object.

25 In another embodiment, the system **10** enhances iterative reconstruction by Fourier space filtering to reduce the number of reconstruction iterations needed to achieve a given image resolution. For instance, the processor **15** executes one or both of the SPS evaluation algorithm **24** and an MTF evaluation algorithm **34** to evaluate reconstruction parameters, SPS data **20**, and/or MTF data **22**, as a function of object size,
30 number of reconstruction iterations, and optionally object shape and/or contrast. The processor **15** executes the SFD filter function generation algorithm **28** that determines, identifies, and/or generates SFD filter functions as a function of the evaluated power

spectra and/or the evaluated modulated transform functions, and object size. SFD filter functions are stored to the filter function library **30** in the memory **16** for recall and application during iterative execution of the reconstruction algorithm(s) **26**. The processor then executes a small number of iterations of the reconstruction algorithm(s) **26** and invokes the SFD filter function executor **32** to apply one or more identified SFD filter functions in Fourier space during reconstruction of the nuclear image volume **19**. In this manner, the signal power spectra for the acquired scan data is enhanced, thereby improving image quality. Once generated, the nuclear image volume **19** is rendered by a video processor **36** for viewing on a display **38** (e.g., a graphical user interface, a video monitor at a user workstation, etc.).

Thus, object size (e.g., pixel count, signal distribution, or some other suitable object size indicator) is used to select an appropriate filter function that reduces a number of reconstruction iterations required to achieve a desired or predetermined image quality, and/or improves image quality for an image reconstructed using a predetermined number of reconstruction iterations. By reducing the number of required reconstruction iterations, reconstruction speed is improved, which facilitates rapid reconstruction of images generated using dual isotopes, quantitative image reconstruction, and any other image that require multiple iterations of a reconstruction algorithm.

In another embodiment, the inverse function of the selected or identified SFD filter function is applied to the reconstructed image in object space. It will be appreciated that object size-dependent filter functions may thus be applied to the reconstructed image in object space and/or as spatial frequency-dependent filter functions applied in Fourier space, as desired by a user for a particular application. Additionally, since such filter functions are the Fourier inverse of each other, object size-dependent filter functions for application in object space may be generated and stored to the filter function library and subsequently inverted to generate corresponding SFD filters, or vice versa.

In another embodiment, a user is permitted to select the degree of filtering applied to the reconstructed image.

FIGURE 2 illustrates a graph **50** showing signal power spectra for standard collimation with resolution recovery (e.g., iterative reconstruction with object position-dependent collimator spatial resolution information applied in the forward

projection of a dual matrix reconstructor) for an object with a 150 mm diameter. From left to right, the solid lines (labeled **1i**, **2i**, **4i**, **8i**, **16i**, and **32i**) represent signal power spectra for 1, 2, 4, 8, 16, and 32 iterations of the reconstruction algorithm, respectively, with resolution recovery. The dotted lines represent corresponding signal power spectra for 1, 2, 4, 8, 16, and 32 iterations of the reconstruction algorithm, respectively, calculated for SPECT projections without collimator resolution.

FIGURE 3 illustrates a graph **60** showing signal power spectra for low collimation with resolution recovery (e.g., iterative reconstruction with object position dependent collimator spatial resolution information applied in the forward projection of a dual matrix reconstructor) for an object with a 150 mm diameter. From left to right, the solid lines (labeled **1i**, **2i**, **4i**, **8i**, **16i**, and **32i**) represent signal power spectra for 1, 2, 4, 8, 16, and 32 iterations of the reconstruction algorithm, respectively, with resolution recovery. The dotted lines represent corresponding signal power spectra for 1, 2, 4, 8, 16, and 32 iterations of the reconstruction algorithm, respectively, calculated for SPECT projections without collimator resolution.

Figures 2 and 3 thus represent examples of power spectra that are evaluated when the processor **15** (Fig. 1) executes the SPS evaluator **24** to assess basic reconstruction properties. It can be seen that, even after 32 iterations, frequencies above 0.2 cycles per pixel are not recovered using low collimation, whereas standard collimation shows a better signal transfer in the medium frequency range. In order to recover resolution, the processor executes the SDF filter function generator **28** to generate one or more filter functions that are applied after iterative reconstruction. For example, the low collimation SPS curve **16i** (Fig. 3) for **16** iterations can be selected by the SPS evaluator **24**, and the SPS in the frequency range of 0.11-0.16 can be increased by the SDF filter function generator **28**, which generates an appropriate normalizing filter, up to a factor of two. Higher frequencies are increased by a constant factor of two to keep image noise low. When the filter generated for a 16-iteration reconstruction process is executed, the information content of the reconstructed image volume **19** is enhanced so that small volumes are more visible.

FIGURES 4A-D illustrate examples of reconstructed SPECT images of a volume of interest **70** with a simulated lesion or tumor **72** under various combinations of collimation level and resolution recovery at 16 iterations of a reconstruction algorithm.

FIGURE 4A illustrates an image **74** of the volume of interest **70** and simulated lesion **72**, generated using standard collimation with resolution recovery. The image exhibits undesirable noise and a low contrast-to noise ratio for the lesion **72**.

FIGURE 4B illustrates an image **76** of the volume of interest **70** and
5 simulated lesion **72**, generated using standard collimation with resolution recovery and a dual-match filter. Application of the dual matched filter results in considerably reduced noise but tends to produce blotchy images and artifacts due to the high content of low frequency noise.

FIGURE 4C illustrates an image **78** of the volume of interest **70** and
10 simulated lesion **72**, generated using low collimation with resolution recovery. Low collimation with resolution recovery reduces lower noise but may limit visibility of small structures.

FIGURE 4D illustrates an image **78** of the volume of interest **70** and
15 simulated lesion **72**, generated using low collimation with resolution recovery and an SFD filter. The appropriate SFD filter applied in Fourier space enhances the contrast for small objects. Noise may be slightly increased but gives a much better image than the dual matched filter (Fig. 4B) due to the different noise power spectrum.

FIGURE 5 illustrates a graph **90** showing signal power spectra determined from a simulated object (e.g., having a diameter of approximately 150 mm) on a
20 logarithmic scale. From left to right, the solid lines (labeled **2i**, **4i**, and **8i**, respectively) represent signal power spectra for 2, 4, and 8 iterations of the reconstruction algorithm, respectively. A plotted curve **92** represents a signal power spectrum after 2 iterations of the reconstruction algorithm with SFD filtering. As illustrated, the SFD-filtered power spectrum **92** is amplified at the higher-frequency end of the spatial frequency spectrum
25 relative to the unfiltered 2-iteration power spectrum **2i**.

FIGURE 6 illustrates a graph **100** showing signal power spectra determined from a simulated object (e.g., having a diameter of approximately 150 mm) on a linear scale. From left to right, the solid lines (labeled **2i**, **4i**, and **8i**, respectively) represent signal power spectra for 2, 4, and 8 iterations of the reconstruction algorithm,
30 respectively. A plotted curve **102** represents a signal power spectrum after 2 iterations of the reconstruction algorithm with SFD filtering.

An object size-dependent filter function **104** is selected according to the object size, as determined from the SPS for the object at a desired number of reconstruction iterations (two iterations in this example). For instance, at least one corner frequency for the SFD filter can be determined from the analyzed SPS (or MTF) data.

- 5 The object size-dependent filter function may include an additional high-frequency component **106** to mitigate frequency application in a desired spectral region. Once the object size-dependent filter function is determined, a corresponding SFD filter function is selected for application in Fourier space during image reconstruction.

10 Figures 5 and 6 thus represent examples of signal power spectra that are evaluated when the processor **15** (Fig. 1) executes the SPS evaluator **24** and/or the MTF evaluator **34** to assess basic reconstruction properties. Assessment of signal power spectra (or modulation transfer function, when normalized) shows the development of the transfer function based on the number of iterations. In the example of Fig. 5, only the lowest spatial frequencies are transferred after two iterations (curve **2i**), but the frequency
15 range of 0 - 0.15 cycles per pixel is well transferred after eight iterations (curve **8i**). This is a general behavior of iterative reconstruction. In this example, SPS are recovered within the full frequency range after 16 to 32 iterations (not shown).

The exact shape of the SPS depends on the size of the object given as a number of voxels. For small objects, the recovery of frequency dependent information is
20 faster than for large objects. Typically, for each halving of object size, the number of reconstruction iterations can be reduced by a factor of 2 to 3.

For the specific object size used in the example of Figures 5 and 6, the lower frequency part of the SPS is approximated after two iterations by the object size-dependent filter function curve **104** shown in Figure 6. In normalized form, the curve
25 **104** has a negative slope from 1×10^3 at $f = 0$ to 0.2 at $f = 0.2$, and is constant for $f = 0.2$ to 0.5. In one embodiment, the SFD filter function used for image correction is specified as the inverse of the object size-dependent filter function curve **104**, which may be generated by the SFD filter function generator **28** (Fig. 1) upon execution of the SPS evaluator **24** and/or the MTF evaluator **34**, and stored to the SFD filter function library **30**
30 for use when reconstructing the nuclear image **19**.

The processor **15** (Fig. 1) applies the SFD filter function after a selected number of reconstruction iterations. For example, when only two iterations are used and

then the filter function is applied, the information content of the reconstructed image is enhanced by a factor of 1 to 5 in the frequency range $f = 0$ to 0.2, and by a constant factor of 5 in the frequency range $f = 0.2$ to 0.5. The resulting SPS curve is shown as curve **102** in Figs. 5 and 6, with data points shown together with a B-spline fit.

5 FIGURES 7A-D illustrate a plurality of reconstructed images of a cross-section of a phantom with a simulated lesion, and associated central profile plots, generated with varying numbers of reconstruction iterations and optional SFD filtering. FIGURE 7A illustrates an image **110** and corresponding central profile plot **112** for the cross-section of the phantom **114** and simulated lesion **115** after two reconstruction
10 iterations without SFD filtering. The central profile plot exhibits a lesion peak **116** that corresponds to the simulated lesion **115** and smaller phantom peaks **117** that correspond to the phantom **114**. The image **110** appears “soft” due to the absence of higher spatial frequencies.

FIGURE 7B illustrates an image **120** and corresponding central profile
15 plot **122** for the cross-section of the phantom **114** and simulated lesion **115** after four reconstruction iterations without SFD filtering. The central profile plot exhibits the lesion peak **116** that corresponds to the simulated lesion **115** and smaller phantom peaks **117** that correspond to the phantom **114**. The peaks **116** and **117** are sharper than those shown in Figure 7A, due to overshoot at sharp edges in the object.

20 FIGURE 7C illustrates an image **130** and corresponding central profile plot **132** for the cross-section of the phantom **114** and simulated lesion **115** after eight reconstruction iterations without SFD filtering. The central profile plot exhibits the lesion peak **116** that corresponds to the simulated lesion **115** and smaller phantom peaks **117** that correspond to the phantom **114**. The peaks **116** and **117** are sharper still than
25 those shown in Figure 7A or 7B, due to increased overshoot at sharp edges in the object after eight iterations.

FIGURE 7D illustrates an image **140** and corresponding central profile
plot **142** for the cross-section of the phantom **114** and simulated lesion **115** after two
reconstruction iterations with SFD filtering. The central profile plot exhibits the lesion
30 peak **116** that corresponds to the simulated lesion **115** and smaller phantom peaks **117** that correspond to the phantom **114**. The peaks **116** and **117** are sharper than those shown in Figure 7C (8 iterations without SFD filtering), due to the SFD filter function

applied in Fourier space after two reconstruction iterations. Thus, a sharper image is achieved using the described filtering techniques with fewer reconstruction iterations than can be achieved using classical approaches, which both improves image quality and reduces reconstruction time. The slight ring artifact around the perimeter of the phantom
5 **114** in Figure 7D is caused by the angular structure of the filter function. If desired, a more refined filter function, as well as assessment of function stability with respect to object shape, size, and contrast can be generated to remove such artifacts.

FIGURE 8 illustrates a graph **150** showing noise power spectra for reconstructed images after 2, 4, and 8 iterations without SFD filtering, labeled **2i**, **4i**, and
10 **8i**, respectively. A curve **152** represents a noise power spectrum for reconstructed image after 2 iterations with SFD filtering. Noise properties of the filtered image are comparable to noise in an image reconstructed with 4 to 8 iterations.

FIGURE 9 illustrates a method for reducing nuclear image reconstruction time using SFD filter functions applied to the image in Fourier space after iterative
15 reconstruction. At **160**, signal power spectra information for a nuclear image is assessed as a function of the imaged object size and a number of reconstruction iterations to be performed. At **162**, an SFD filter function is generated as a function of the object size and iteration number, and is stored to memory. At **164**, the SFD filter function is recalled and applied in Fourier space after the desired number of reconstruction iterations to
20 enhance the visibility of the object using fewer iterations of a reconstruction algorithm than would be necessary without the SFD filter. By reducing the number of iterations required to achieve a desired level of reconstructed image quality, the application of the SFD filter reduces image reconstruction time.

FIGURE 10 illustrates a method for improving nuclear image quality for a
25 given number of reconstruction algorithm iterations using SFD filter functions applied to the reconstructed image in Fourier space after the given number of reconstruction iterations. At **170**, reconstruction properties (e.g., a number of iterations, etc.), SPS properties, and/or MTF properties are evaluated along with object size and optionally object shape and/or contrast. At **172**, one or more SFD filter functions are generated
30 based on the evaluated properties, and are stored to a filter function library. At **174**, one or more stored filters are applied in Fourier space to an image that has undergone the predetermined number of reconstruction iterations to refine the image.

In this manner, low collimation techniques are combined with resolution recovery techniques and object size-dependent filtering in Fourier space for image enhancement in nuclear imaging devices. Using object size-dependent filtering in Fourier space facilitates recovering higher-frequency information and adapting lesion
5 detectability and noise power spectra to application parameters.

Additionally, the described systems and methods can be applied in SPECT, PET and other imaging modalities using iterative reconstruction, such as x-ray CT or volume imaging. There is no restriction with regard to collimation geometry since re-binning of pinhole or fan beam data to parallel beam geometry facilitates using
10 standard parallel beam reconstruction techniques. In addition to ordered subset expectation maximization (OSEM), the described systems and methods can be used in combination with any other type of iterative reconstruction, such as maximum likelihood expectation maximization (MLEM), algebraic reconstruction techniques (ART), maximum a posteriori (MAP) techniques, etc.

15 The innovation has been described with reference to several embodiments. Modifications and alterations may occur to others upon reading and understanding the preceding detailed description. It is intended that the innovation be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

CLAIMS

Having thus described the preferred embodiments, the invention is now claimed to be:

1. A nuclear image optimization system (10), including:
 - a nuclear scanner (12) that acquires nuclear scan data (18) of a volume of interest (VOI), and having one or more collimators that provide a low level of collimation during nuclear scan data acquisition;
 - an object size detector (17) that determines a size of the VOI;
 - a filter function library (30) having a lookup table that identifies one or more object size-dependent filter functions for each of a plurality of object sizes; and
 - a processor (15) that performs an iterative reconstruction algorithm (26), executes the object size-dependent filter function after executing a predetermined number of reconstruction algorithm (26) iterations, and outputs an enhanced nuclear image volume (19) representation.
2. The system according to claim 1, wherein the object size-dependent filter function is a spatial frequency-dependent (SFD) filter function and is applied to the image volume after the predetermined number of reconstruction algorithm iterations.
3. The system according to claim 2, wherein the processor (15) generates the SFD filter function based on at least one of signal power spectrum data (20) and modulation transfer function (MTF) data (22) acquired from nuclear scan data (18) of the VOI or from scan data acquired on appropriate test phantoms .
4. The system according to claim 1, wherein the object size detector (17) determines object size using at least one of a proximity sensor, a camera sensor, a laser gauge, a computed tomography (CT) imaging device, the acquired nuclear scan data, or angular dependence of the object size in respective transaxial slices of the acquired nuclear scan data.

5. A method of increasing image reconstruction speed for a predetermined number of reconstruction iterations, including:

acquiring scan data of an object;

determining a size of the object;

identifying an object size-dependent filter function for application to a reconstructed image of the object based on the object size and the predetermined number of reconstruction iterations;

performing the predetermined number of reconstruction iterations to reconstruct an image of the object;

applying the object size-dependent filter function to the reconstructed image after a selected number of reconstruction iterations to enhance resolution; and

storing the enhanced reconstructed image to memory (16).

6. The method according to claim 5, wherein the object size-dependent filter function is a spatial frequency-dependent (SFD) filter function and is applied in Fourier space.

7. The method according to claim 6, wherein the scan data is nuclear scan data acquired using a low collimation technique.

8. The method according to claim 7, further including:

generating one or more SFD filter functions based on object size and on the predetermined number of reconstruction iterations.

9. The method according to claim 8, further including:

analyzing at least one of signal power spectrum (SPS) data (20) and modulation transfer function (MTF) data (22) when generating the one or more SFD filter functions; and

determining at least one corner frequency for each SFD filter function from the analyzed spectrum data.

10. The method according to claim 8, further including:

storing the one or more SFD filter functions to a filter function library (30) for subsequent lookup when identifying an SFD filter function for a determined object size and number of reconstruction iterations.

11. A computer-readable medium (16) having stored thereon software for controlling one or more computers to perform the method according to claim 5.

12. A processor programmed to control one or more computers configured to perform the method of claim 5.

13. A single photon emission computed tomography (SPECT) scanner including a processor (15) programmed to perform the method of claim 5.

14. A method of decreasing a number of reconstruction iterations executed to achieve a desired level of image resolution, including:

acquiring low-collimation nuclear scan data (18) of an object;

determining a size of the object;

identifying an object size-dependent filter function for application to a reconstructed image of the object based on the object size and the desired image resolution;

performing a number of reconstruction iterations to reconstruct a nuclear image (19) of the object;

applying the object size-dependent filter function during reconstruction to enhance resolution; and

storing the enhanced reconstructed image (19) to memory (16).

15. The method according to claim 14, wherein the object size-dependent filter function is a spatial frequency-dependent (SFD) filter function and is applied in Fourier space.

16. The method according to claim 15, further including:

generating one or more SFD filter functions based on object size and on the predetermined number of reconstruction iterations and evaluation of at least one of signal

power spectrum (SPS) data (20) and modulation transfer function (MTF) data (22) for the object; and

determining at least one corner frequency for each SFD filter function from the analyzed spectrum data.

17. The method according to claim 16, further including:

storing the one or more SFD filter functions to a filter function library for subsequent lookup when identifying an SFD filter function for a determined object size and number of reconstruction iterations.

18. The method according to claim 14, wherein object size is determined as a function of pixel count for the object.

19. A computer-readable medium (16) having stored thereon software for controlling one or more computers to perform the method according to claim 14.

20. A processor programmed to control one or more computers configured to perform the method of claim 14.

21. A single photon emission computed tomography (SPECT) scanner including a processor (15) programmed to perform the method of claim 14.

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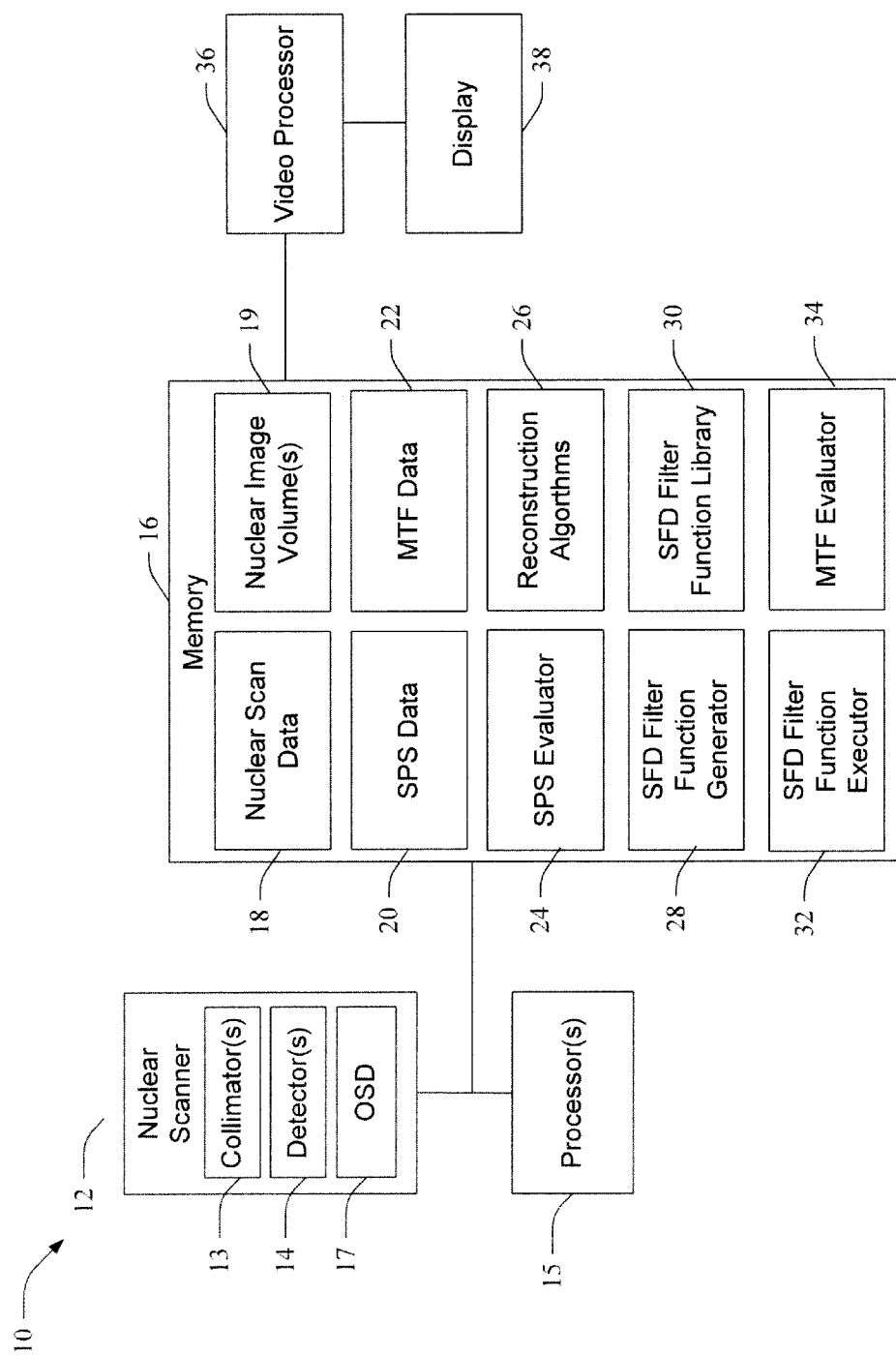


FIG. 1

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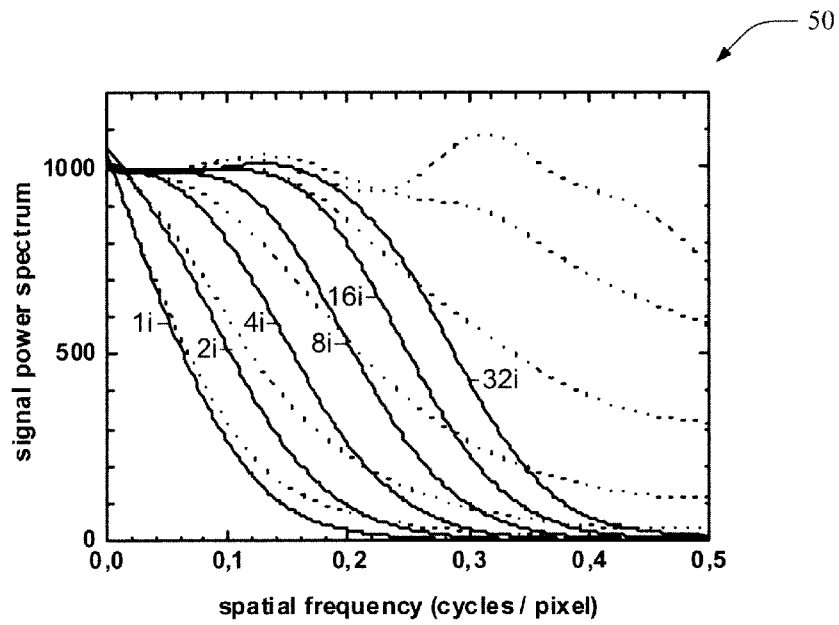


FIG. 2

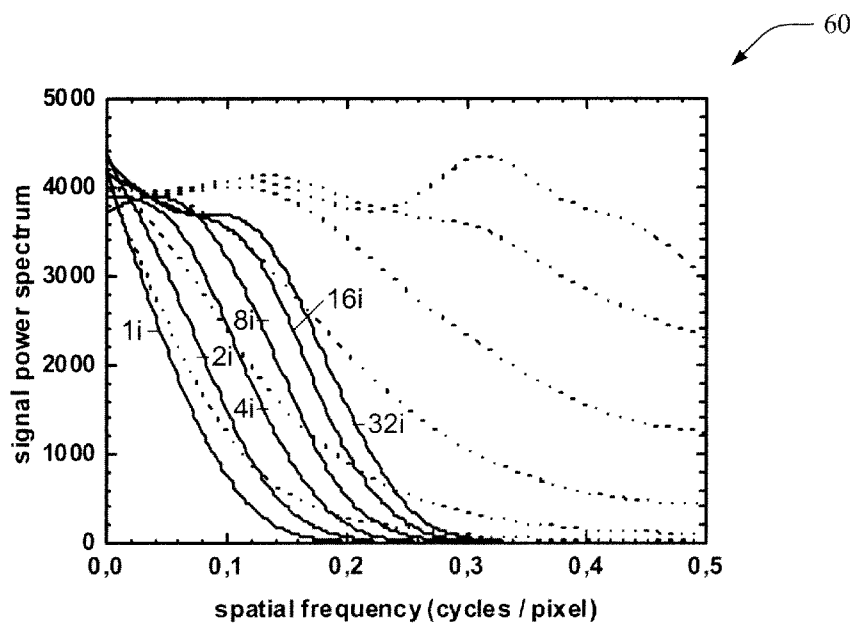


FIG. 3

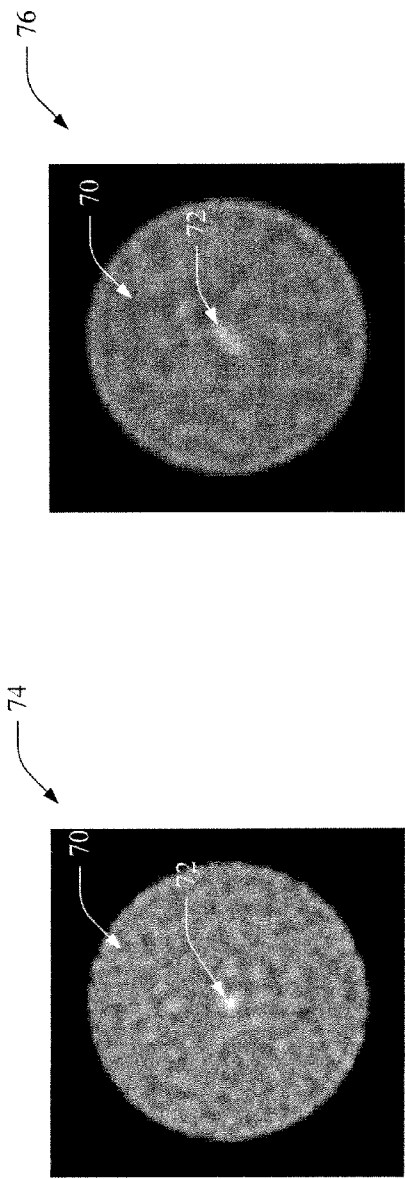


FIG. 4B

FIG. 4A

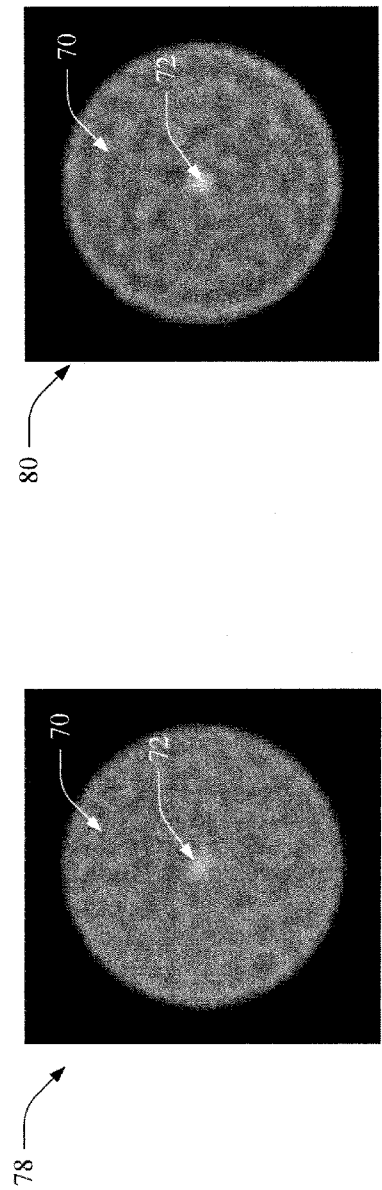


FIG. 4D

FIG. 4C

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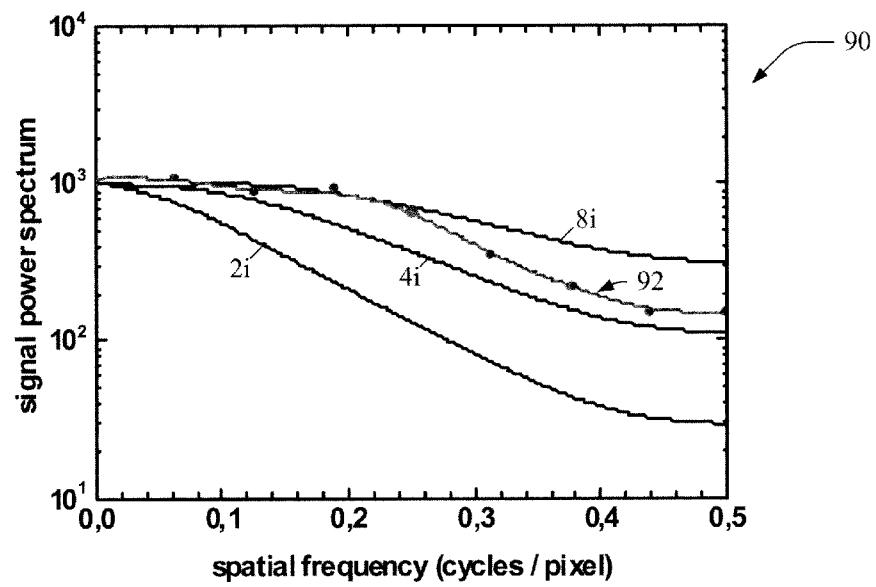


FIG. 5

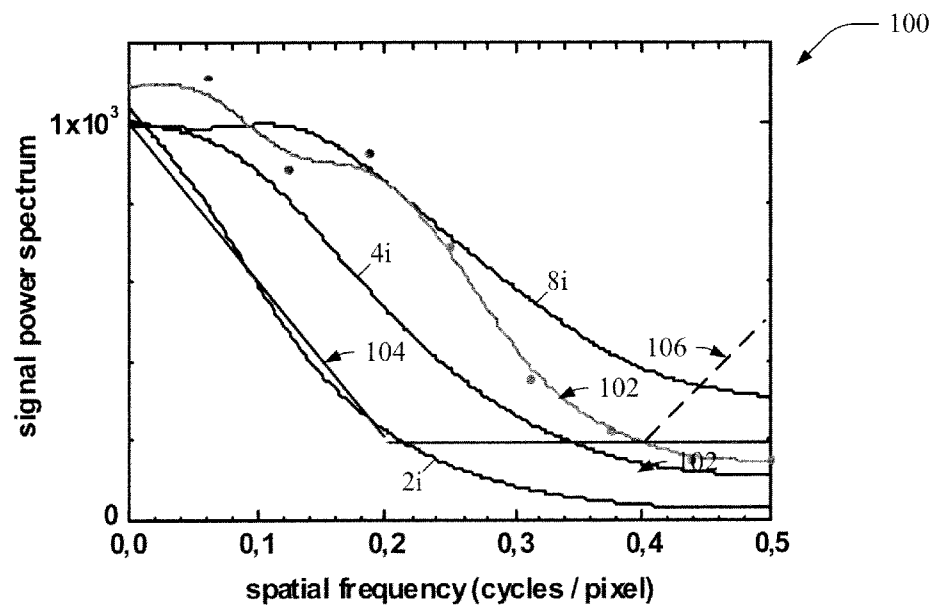


FIG. 6

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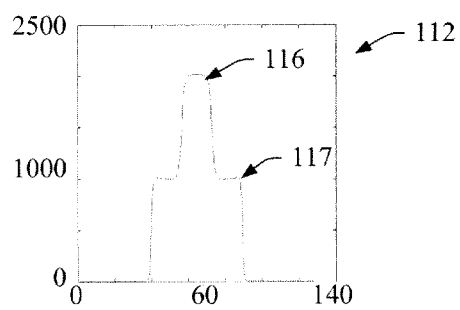
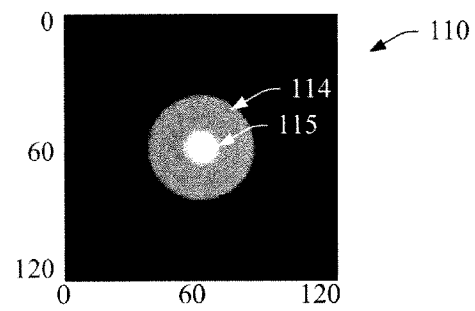


FIG. 7A

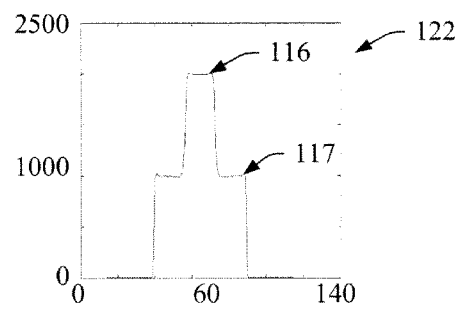
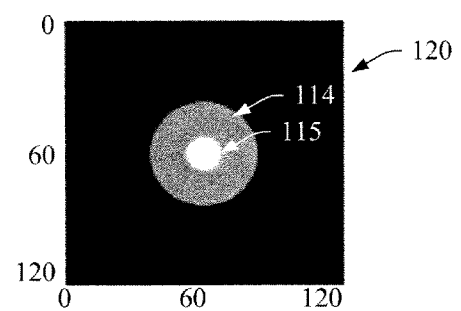


FIG. 7B

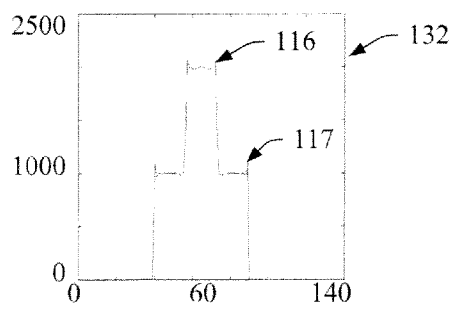
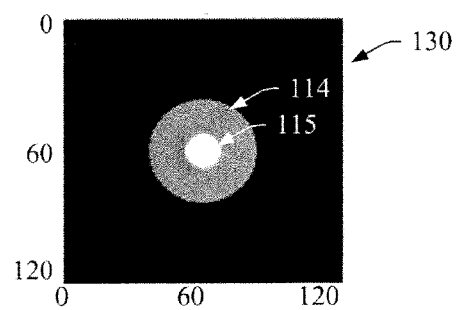


FIG. 7C

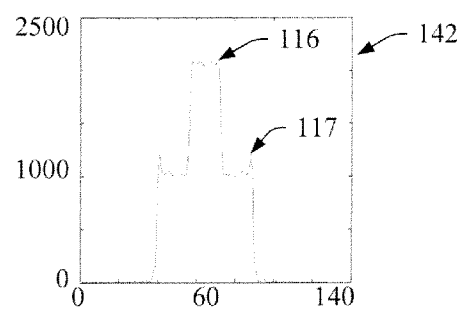
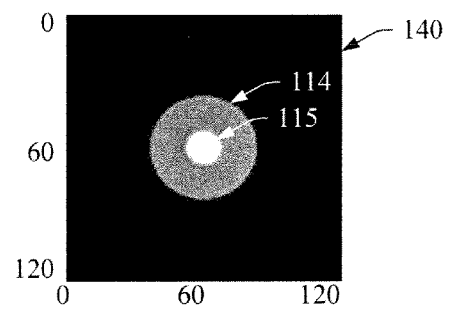


FIG. 7D

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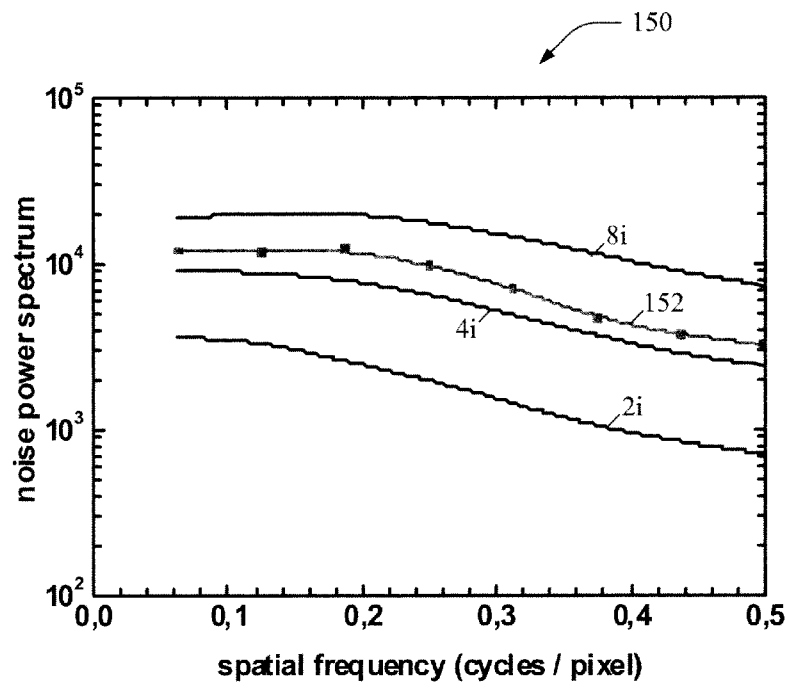


FIG. 8

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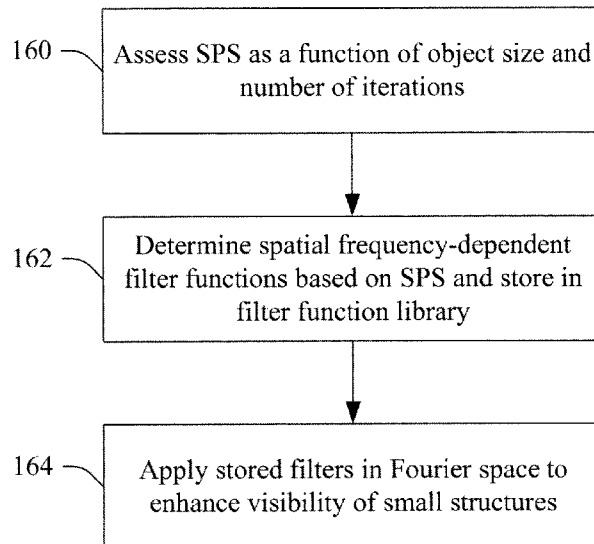


FIG. 9

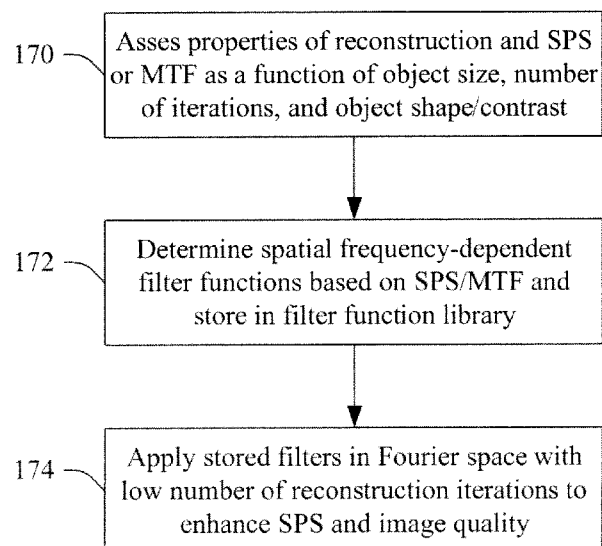


FIG. 10

INTERNATIONAL SEARCH REPORT

International application No

PCT/IB2009/054366

A. CLASSIFICATION OF SUBJECT MATTER
 INV. G06T5/00 G06T11/00

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
 G06T

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, COMPENDEX, INSPEC, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	O. G. Rousset and H. Zaidi: "Correction for Partial Volume Effects in Emission Tomography" In: Habib Zaidi: "Quantitative Analysis in Nuclear Medicine Imaging" 11 July 2006 (2006-07-11), Springer , US , XP007911332 ISBN: 9780387238548 , pages 236-271 the whole document ----- -/---	1-21

☒ Further documents are listed in the continuation of Box C.

☐ See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search

29 January 2010

Date of mailing of the international search report

09/02/2010

Name and mailing address of the ISA/

European Patent Office, P.B. 5818 Patentlaan 2
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 Fax: (+31-70) 340-3016

Authorized officer

Werling, Alexander

INTERNATIONAL SEARCH REPORT

International application No

PCT/IB2009/054366

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	PRETORIUS P H ET AL: "Reducing the influence of the partial volume effect on SPECT activity quantitation with 3D modelling of spatial resolution in iterative reconstruction" PHYSICS IN MEDICINE AND BIOLOGY IOP PUBLISHING UK, vol. 43, no. 2, February 1998 (1998-02), pages 407-420, XP007911366 ISSN: 0031-9155 the whole document	1-21
A	HOFFMAN E J ET AL: "QUANTITATION IN POSITRON EMISSION COMPUTED TOMOGRAPHY PART 1 EFFECT OF OBJECT SIZE" JOURNAL OF COMPUTER ASSISTED TOMOGRAPHY, vol. 3, no. 3, 1979, pages 299-308, XP009128688 ISSN: 0363-8715 abstract; figure 2	1-21