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Arenson et al.

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(54) **X-RAY FLUX MANAGEMENT DEVICE**

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(51) **Int. Cl.**
G21K 3/00 (2006.01)

(52) **U.S. Cl.** **378/159**

(58) **Field of Classification Search** 378/156, 378/159
See application file for complete search history.

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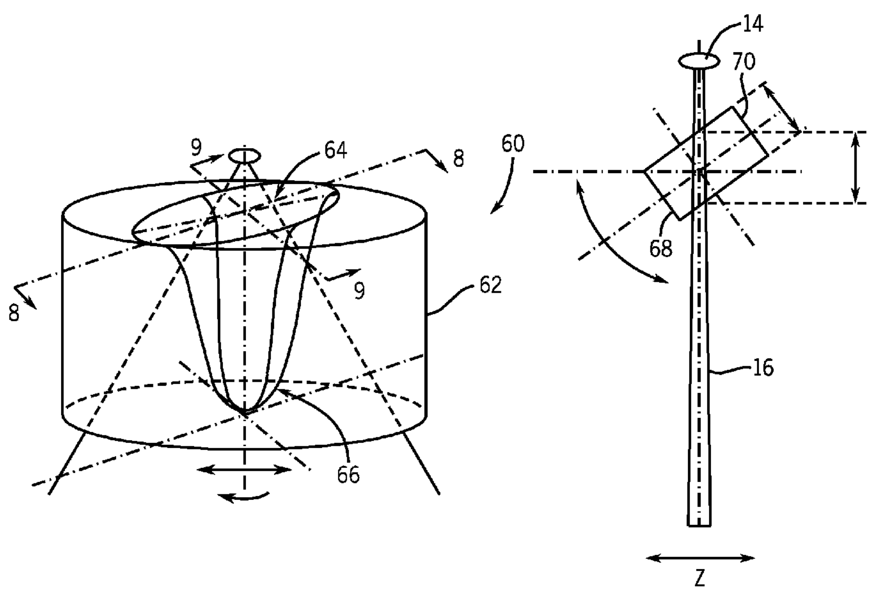
(57) **ABSTRACT**

The invention is directed to an x-ray flux management device that adaptively attenuates an x-ray beam to limit the incident flux reaching a subject and radiographic detectors in potentially high-flux areas while not affecting the incident flux and detector measurements in low-flux regions. While the invention is particularly well-suited for CT, the invention is also applicable with other x-ray imaging systems. In addition to reducing the required detector system dynamic range, the present invention provides an added advantage of reducing radiation dose.

20 Claims, 6 Drawing Sheets



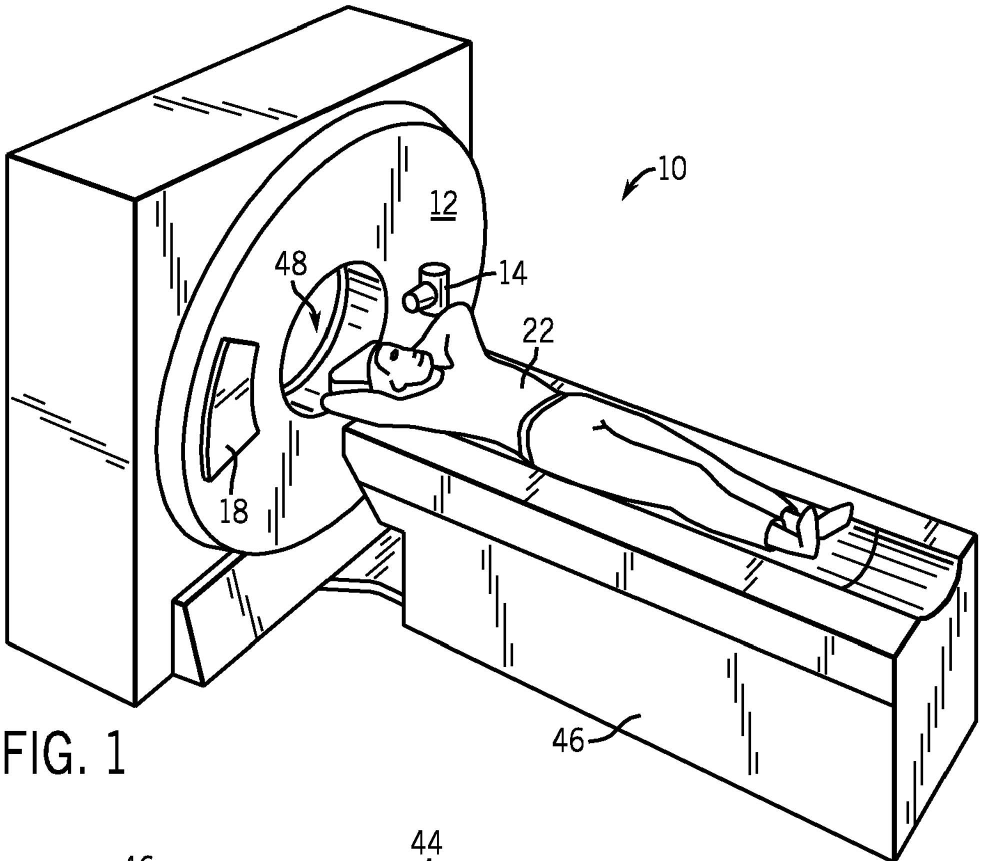

10
12
48
14
22
18
46

FIG. 1

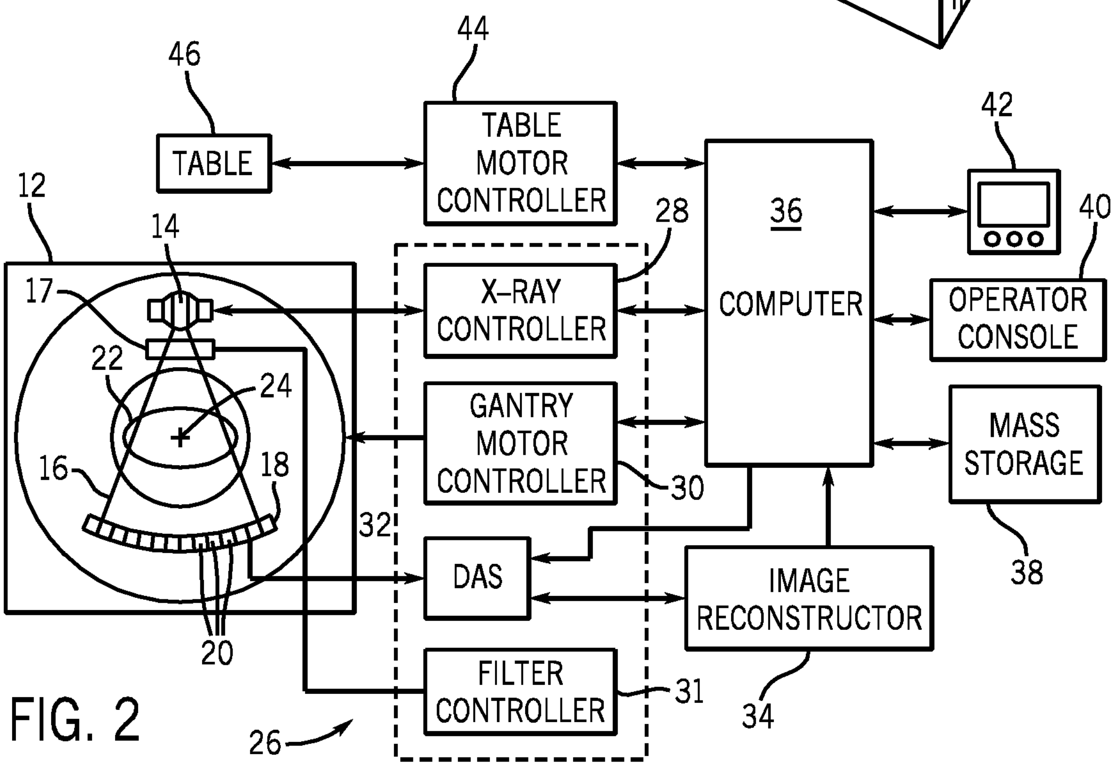

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TABLE
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TABLE MOTOR CONTROLLER
42
36
COMPUTER
40
OPERATOR CONSOLE
12
14
17
22
24
16
18
20
32
28
X-RAY CONTROLLER
GANTRY MOTOR CONTROLLER
30
DAS
FILTER CONTROLLER
31
IMAGE RECONSTRUCTOR
34
MASS STORAGE
38
26

FIG. 2

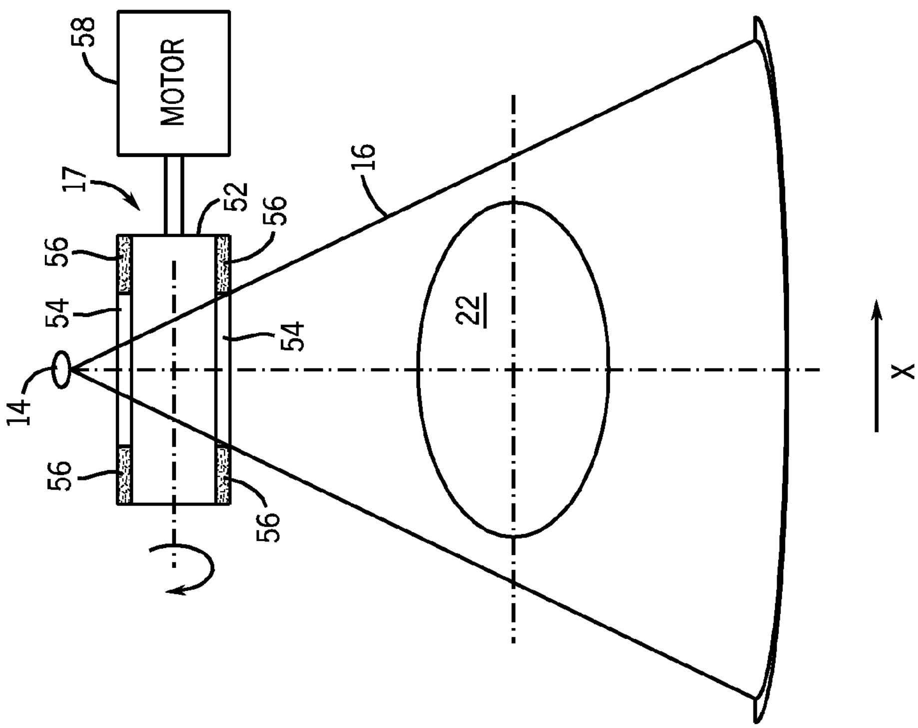


FIG. 4

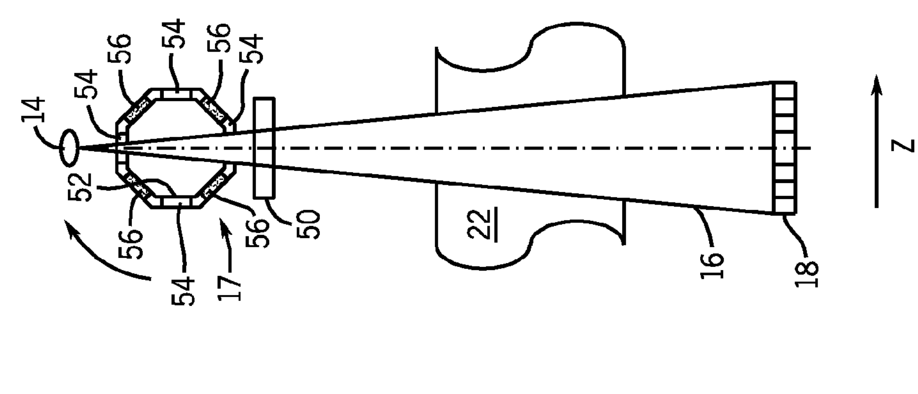


FIG. 3

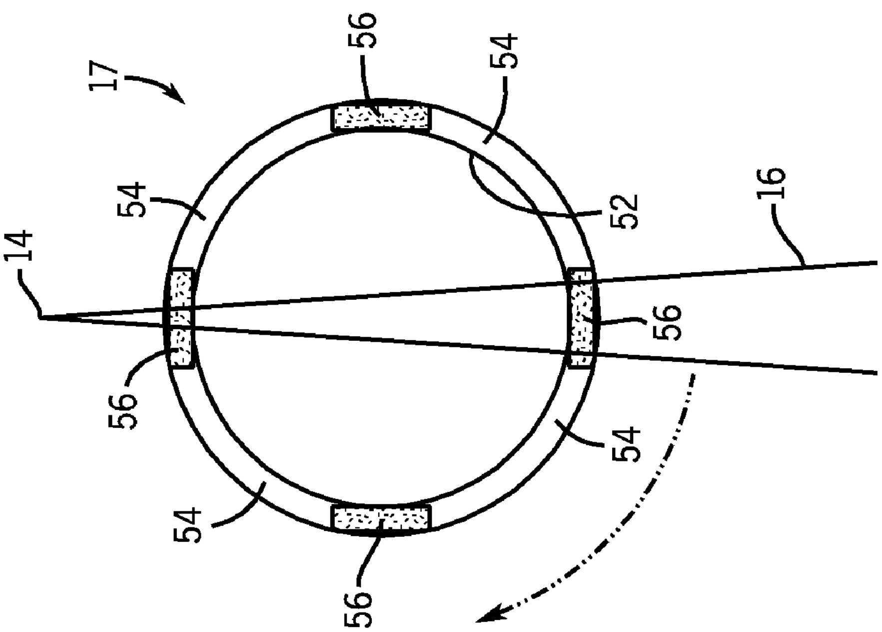
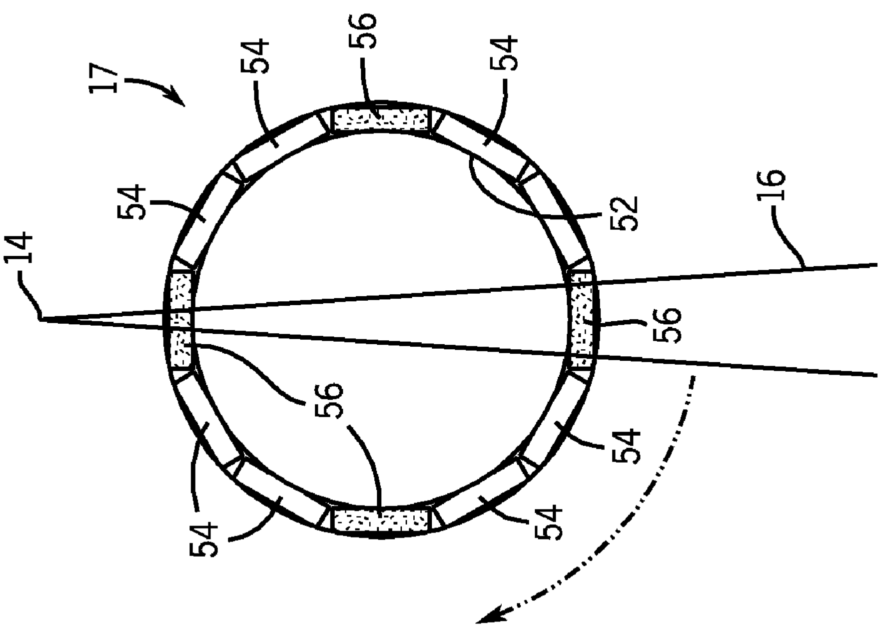

17
56
54
54
52
16
14
56
56
54
54
56

FIG. 6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54
56
54
54
52
16
14
56
56
56
54
54
54

FIG. 5

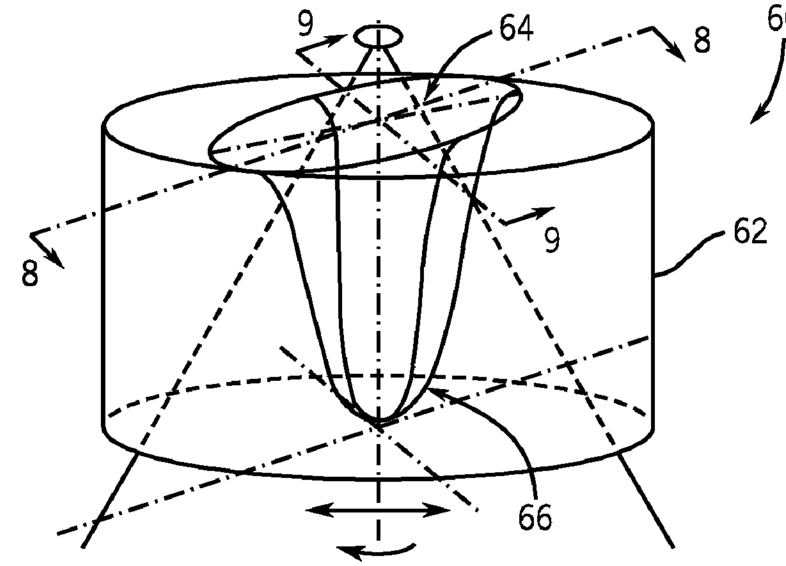


FIG. 7

FIG. 8

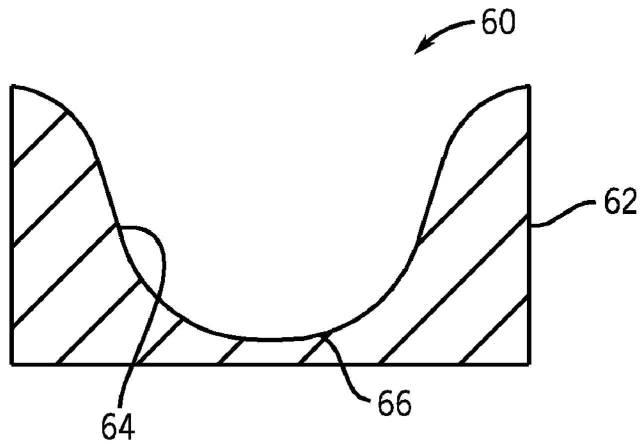
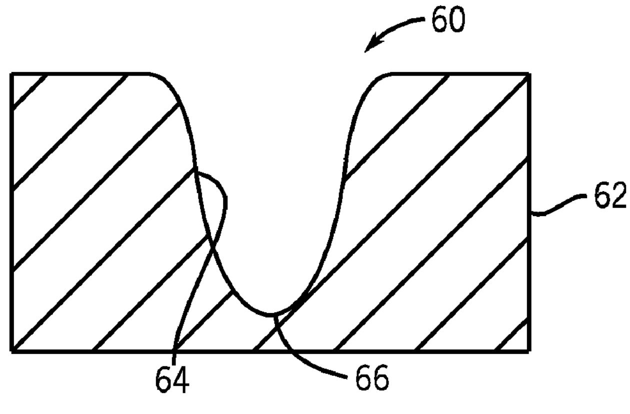


FIG. 9



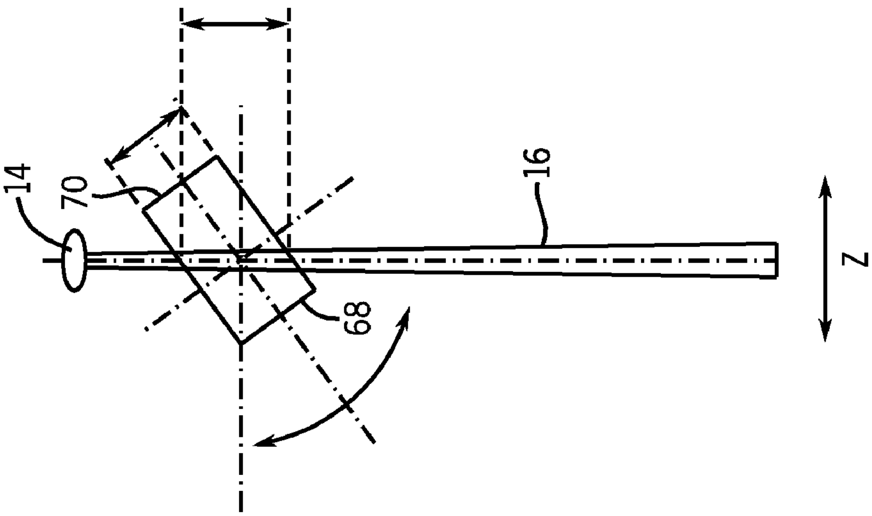


FIG. 11

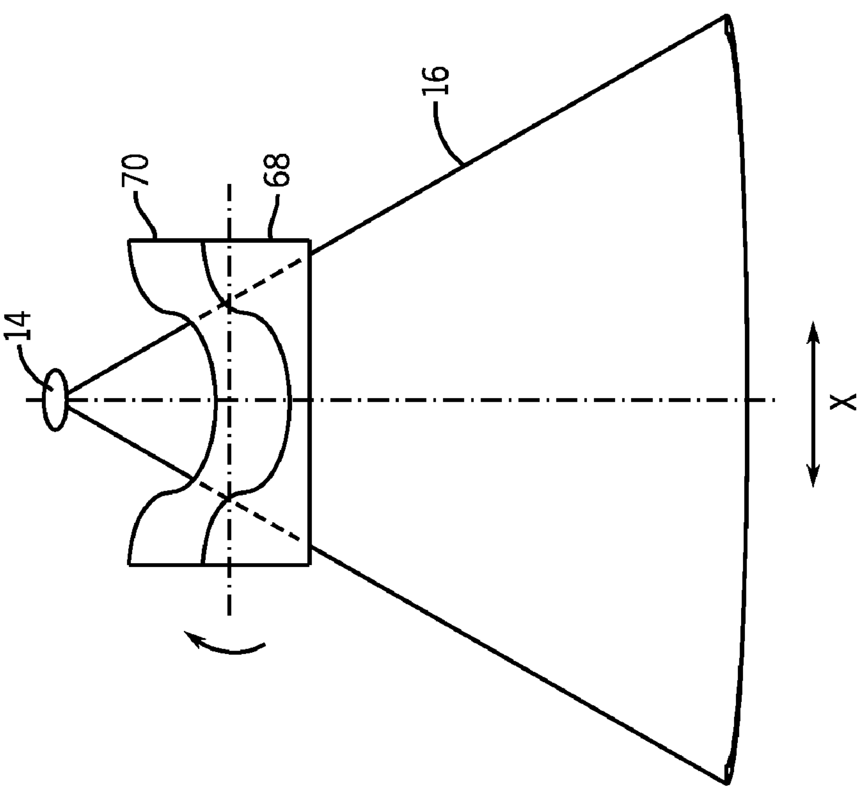


FIG. 10

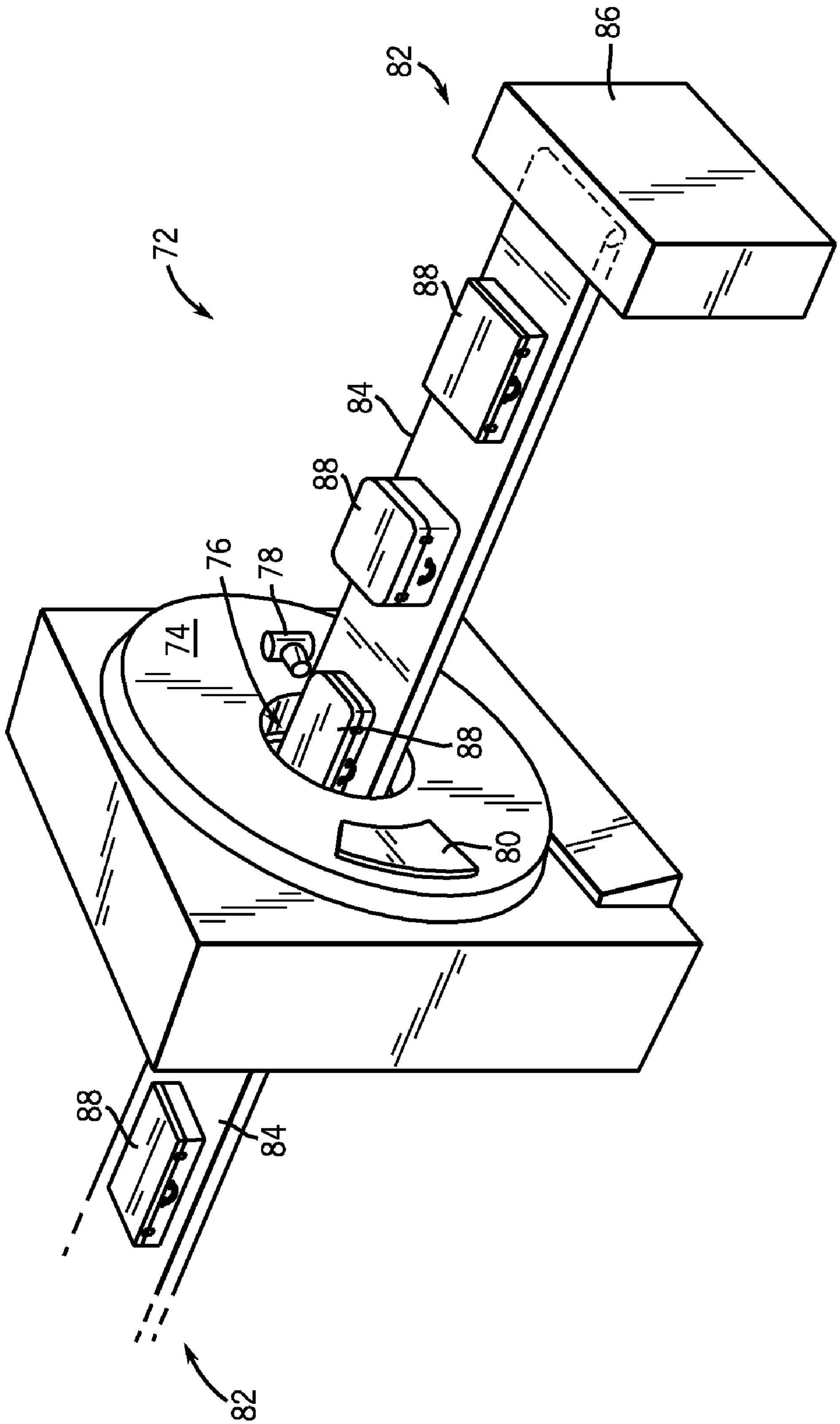


FIG. 12

X-RAY FLUX MANAGEMENT DEVICE

CROSS-REFERENCE TO RELATED APPLICATION

The present application is a divisional of and claims priority of U.S. Ser. No. 11/164,121 filed Nov. 10, 2005, the disclosure of which is incorporated herein by reference.

BACKGROUND OF THE INVENTION

The present invention relates generally to radiographic imaging and, more particularly, to a beam chopper for a radiographic imaging system. The invention is also directed to an x-ray filter. The present invention is particularly related to photon counting and/or energy discriminating radiation detectors.

Typically, in radiographic systems, an x-ray source emits x-rays toward a subject or object, such as a patient or a piece of luggage. Hereinafter, the terms "subject" and "object" may be interchangeably used to describe anything capable of being imaged. The x-ray beam, after being attenuated by the subject, impinges upon an array of radiation detectors. The intensity of the radiation beam received at the detector array is typically dependent upon the attenuation of the x-rays through the scanned object. Each detector element of the detector array produces a separate signal indicative of the attenuated beam received by each detector element. The signals are transmitted to a data processing system for analysis and further processing which ultimately produces an image. Generally, the x-ray source and the detector array are rotated about the gantry within an imaging plane and around the subject. X-ray sources typically include x-ray tubes, which emit the x-ray beam at a focal point. X-ray detectors typically include a collimator for collimating x-ray beams received at the detector, a scintillator for converting x-rays to light energy adjacent the collimator, and photodiodes for receiving the light energy from the adjacent scintillator and producing electrical signals therefrom.

In a similar fashion, radiation detectors are employed in emission imaging systems such as used in nuclear medicine (NM) gamma cameras and Positron Emission Tomography (PET) systems. In these systems, the source of radiation is no longer an x-ray source, rather it is a radiopharmaceutical introduced into the body being examined. In these systems each detector of the array produces a signal in relation to the localized intensity of the radiopharmaceutical concentration in the object. Similar to conventional x-ray imaging, the strength of the emission signal is also attenuated by the inter-lying body parts. Each detector element of the detector array produces a separate signal indicative of the emitted beam received by each detector element. The signals are transmitted to a data processing system for analysis and further processing which ultimately produces an image.

In most computed tomography (CT) imaging systems, the x-ray source and the detector array are rotated about a gantry encompassing an imaging volume around the subject. X-ray sources typically include x-ray tubes, which emit the x-rays as a fan or cone beam from the anode focal point. X-ray detector assemblies typically include a collimator for reducing scattered x-ray photons from reaching the detector, a scintillator adjacent to the collimator for converting x-rays to light energy, and a photodiode adjacent to the scintillator for receiving the light energy and producing electrical signals therefrom. Typically, each scintillator of a scintillator array converts x-rays to light energy. Each photodiode detects the light energy and generates a corresponding electrical signal. The outputs of the photodiodes are then transmitted to the data acquisition system and then to the processing system for image reconstruction.

Conventional CT imaging systems utilize detectors that convert x-ray photon energy into current signals that are integrated over a time period, then measured and ultimately digitized. A drawback of such detectors is their inability to provide independent data or feedback as to the energy and incident flux rate of photons detected. That is, conventional CT detectors have a scintillator component and photodiode component wherein the scintillator component illuminates upon reception of x-ray photons and the photodiode detects illumination of the scintillator component, and provides an integrated electrical current signal as a function of the intensity and energy of incident x-ray photons. While it is generally recognized that CT imaging would not be a viable diagnostic imaging tool without the advancements achieved with conventional CT detector design, a drawback of these integrating detectors is their inability to provide energy discriminatory data or otherwise count the number and/or measure the energy of photons actually received by a given detector element or pixel. Accordingly, recent detector developments have included the design of an energy discriminating detector that can provide photon counting and/or energy discriminating feedback. In this regard, the detector can be caused to operate in an x-ray counting mode, an energy measurement mode of each x-ray event, or both.

These energy discriminating detectors are capable of not only x-ray counting, but also providing a measurement of the energy level of each x-ray detected. While a number of materials may be used in the construction of an energy discriminating detector, including scintillators and photodiodes, direct conversion detectors having an x-ray photoconductor, such as amorphous selenium or cadmium zinc telluride, that directly convert x-ray photons into an electric charge have been shown to be among the preferred materials. A drawback of photon counting detectors, however, is that these types of detectors have limited count rates and have difficulty covering the broad dynamic ranges encompassing very high x-ray photon flux rates typically encountered with conventional CT systems. Generally, a CT detector dynamic range of 1,000,000 to one is required to adequately handle the possible variations in photon flux rates. In the very fast scanners now available, it is not uncommon to encounter x-ray flux rates of over 10^8 photons/mm^2/sec when no object is in the scan field, with the same detection system needing to count only 10's of photons that manage to traverse the center of large objects.

The very high x-ray photon flux rates ultimately lead to detector saturation. That is, these detectors typically saturate at relatively low x-ray flux levels. This saturation can occur at detector locations wherein small subject thickness is interposed between the detector and the radiographic energy source or x-ray tube. It has been shown that these saturated regions correspond to paths of low subject thickness near or outside the width of the subject projected onto the detector array. In many instances, the subject is more or less cylindrical in the effect on attenuation of the x-ray flux and subsequent incident intensity to the detector array. In this case, the saturated regions represent two disjointed regions at extremes of the detector array. In other less typical, but not rare instances, saturation occurs at other locations and in more than two disjointed regions of the detector. In the case of a cylindrical subject, the saturation at the edges of the array can be reduced by the imposition of a bowtie filter between the subject and the x-ray source. Typically, the filter is constructed to match the shape of the subject in such a way as to equalize total attenuation, filter and subject, across the detector array. The flux incident to the detector is then relatively uniform across the array and does not result in saturation. What can be problematic, however, is that the bowtie filter may not be optimum given that a subject population is significantly less than uniform and not exactly cylindrical in shape nor centrally located in the x-ray beam. In such cases, it is possible for one or more disjointed regions of saturation to occur or conversely to over-filter the x-ray flux and unnecessarily create regions of very low flux. Low x-ray flux in the projection results in a reduction in information content which will ultimately contribute to unwanted noise in the reconstructed image of the subject.

Moreover, a system calibration method common to most CT systems involves measuring detector response with no subject whatsoever in the beam. This "air cal" reading from each detector element is used to normalize and correct the preprocessed data that is then used for CT image reconstruction. Even with ideal bowtie filters, high x-ray flux now in the central region of the detector array could lead to detector saturation during the system calibration phase.

A number of imaging techniques have been proposed to address saturation of any part of the detector. These techniques include maintenance of low x-ray flux across the width of a detector array, for example, by modulating tube current or x-ray voltage during scanning. However, this solution leads to increased scanned time. That is, there is a penalty that the acquisition time for the image is increased in proportion to the nominal flux needed to acquire a certain number of x-rays that meet image quality requirements. Other solutions include the implementation of over-range algorithms that may be used to generate replacement data for the saturated data. However, these algorithms may imperfectly replace the saturated data as well as contribute to the complexity of the CT system.

It would therefore be desirable to design an x-ray flux management device that is effective in reducing detector saturation under high x-ray flux conditions while not compromising data acquisition under low x-ray flux conditions.

BRIEF DESCRIPTION OF THE INVENTION

The present invention is a directed an x-ray flux management device that overcomes the aforementioned drawbacks.

The impact of radiographic detector design on radiographic image quality is foremost an issue of properly handling low-flux conditions (to effectively measure the limited x-ray transmission through thicker imaging regions). At the same time, the higher flux areas in these scans (such as detector readings through air and partially within the subject contours) also need to be correctly evaluated. If insufficient detector dynamic range is available, these high-flux detector channels tend to over-range and enter a saturated state. Since these over-range areas are typically in air or in highly irradiated portions of the subject, the exact measurement of each photon in these high-flux regions is not as critical as for the low-flux areas where each photon contributes an integral part to the total collected photon statistics and improved imaging quality. Subsequently, the invention addresses the specific needs of low- and high-flux regions and thereby provides the opportunity to use low dynamic range detectors for radiographic imaging.

In this regard, the invention includes an x-ray flux management device that adaptively attenuates an x-ray beam to limit the incident flux reaching the subject and the radiographic detectors in the potentially high-flux areas while not affecting the incident flux and detector measurements in low-flux regions. While the invention is particularly well-suited for CT, the invention is also applicable with other x-ray imaging systems. In addition to reducing the required detector system dynamic range, the present invention provides an added advantage of reducing radiation dose.

Therefore, in accordance with one aspect, the invention includes an x-ray beam chopper for a radiographic imaging apparatus. The beam chopper has a rotatable frame and at least one x-ray transmission window disposed in the rotatable frame that allows a generally free transmission of x-rays. The chopper also has at least one x-ray filtering window disposed in the rotatable frame that filters x-rays.

In accordance with another aspect, the invention is directed to a radiographic imaging apparatus that includes an x-ray source and an x-ray detector. The apparatus further has a segmented filtering assembly having a generally annular frame with at least one low x-ray flux segment and at least one high x-ray flux segment, and a filtering assembly controller that causes the low x-ray flux segment to be in an x-ray beam path during a low x-ray flux data acquisition view and causes the high x-ray flux segment to be in the x-ray beam path during a high x-ray flux data acquisition view.

According to another aspect, the invention includes an x-ray filter having a 3D semi-cylindrical rotatable filter body formed of x-ray attenuating matter. The filter also has a semi-conical bore formed in the 3D semi-cylindrical rotatable filter. The semi-conical bore has an elliptically shaped base.

According to yet another aspect, the invention includes an x-ray filter assembly having a bowtie filter having an effective beam profile. The assembly further has a filter controller that tilts the bowtie filter during data acquisition to change the effective beam profile during data acquisition.

Various other features and advantages of the present invention will be made apparent from the following detailed description and the drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

The drawings illustrate one preferred embodiment presently contemplated for carrying out the invention.

In the drawings:

FIG. **1** is a pictorial view of a CT imaging system.

FIG. **2** is a schematic diagram of the system illustrated in FIG. **1**.

FIG. **3** is a schematic diagram of an x-ray beam chopper positioned relative to the z-axis according to the present invention.

FIG. **4** is a schematic diagram of an x-ray beam chopper positioned relative to the x-axis according to the present invention.

FIG. **5** is a schematic of an x-ray beam chopper according to an alternate embodiment of the present invention.

FIG. **6** a schematic of an x-ray beam chopper according to yet another alternate embodiment of the present invention.

FIG. **7** is a perspective view of a 3D bowtie filter according to the present invention.

FIG. **8** is a cross-sectional view of the bowtie filter of FIG. **7** taken along line **8-8** thereof.

FIG. **9** is a cross-sectional view of the bowtie filter of FIG. **7** taken along line **9-9** thereof.

FIG. **10** is a schematic view of a tiltable bowtie filter assembly positioned relative to the x-axis according to the present invention.

FIG. **11** is a schematic view of the tiltable bowtie filter of FIG. **10** shown relative to the z-axis according to the present invention.

FIG. **12** is a pictorial view of a CT system for use with a non-invasive package inspection system.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

The operating environment of the present invention is described with respect to a four-slice computed tomography (CT) system. However, it will be appreciated by those skilled in the art that the present invention is equally applicable for use with single-slice or other multi-slice configurations. Moreover, the present invention will be described with respect to the detection and conversion of x-rays. However, one skilled in the art will further appreciate that the present invention is equally applicable for the detection and conversion of other high frequency electromagnetic energy.

Referring to FIGS. **1** and **2**, an exemplary computed tomography (CT) imaging system **10** is shown as including a gantry **12** representative of a "third generation" CT scanner. Gantry **12** has an x-ray source **14** that projects a beam of x-rays **16** through an x-ray flux management assembly **17** toward a detector array **18** on the opposite side of the gantry **12**. The x-ray flux management assembly will be described in greater detail with respect to FIGS. **3-12**. Detector array **18** is formed by a plurality of detectors **20** which together sense the projected x-rays that pass through a medical patient **22**. Each detector **20** produces an electrical signal that represents the intensity of an impinging x-ray beam and hence the attenuated beam as it passes through the patient **22**. During a scan to acquire x-ray projection data, gantry **12** and the components mounted thereon rotate about a center of rotation **24**.

Rotation of gantry **12** and the operation of x-ray source **14** are governed by a control mechanism **26** of CT system **10**. Control mechanism **26** includes an x-ray controller **28** that provides power and timing signals to an x-ray source **14** and a gantry motor controller **30** that controls the rotational speed and position of gantry **12**. A data acquisition system (DAS) **32** in control mechanism **26** samples analog data from detectors **20** and converts the data to digital signals for subsequent processing. An image reconstructor **34** receives sampled and digitized x-ray data from DAS **32** and performs high speed reconstruction. The reconstructed image is applied as an input to a computer **36** which stores the image in a mass storage device **38**.

Computer **36** also receives commands and scanning parameters from an operator via console **40** that has a keyboard. An associated cathode ray tube display **42** allows the operator to observe the reconstructed image and other data from computer **36**. The operator supplied commands and parameters are used by computer **36** to provide control signals and information to DAS **32**, x-ray controller **28**, gantry motor controller **30**, and filter controller **31**. In addition, computer **36** operates a table motor controller **44** which controls a motorized table **46** to position patient **22** and gantry **12**. Particularly, table **46** moves portions of patient **22** through a gantry opening **48**.

The present invention is directed to an x-ray beam chopper that may be incorporated with the CT system described above or other radiographic systems, such as x-ray systems and the like.

Generally, high-sensitivity photon counting radiation detectors are constructed to have a relatively low dynamic range. This is generally considered acceptable for proton counting detector applications since high flux conditions typically do not occur. In CT detector designs, low flux detector readings through the subject are typically accompanied by areas of high irradiation in air, and/or within the contours of the scan subject requiring CT detectors to have very large dynamic range responses. Moreover, the exact measurement of photons in these high-flux regions is less critical than that for low-flux areas where each photon contributes an integral part to the total collected photon statistics. Notwithstanding that the higher flux areas may be of less clinical or diagnostic value, images reconstructed with over-ranging or saturated detector channel data can be prone to artifacts. As such, the handling of high-flux conditions is also important.

The present invention includes an x-ray flux management device designed to prevent saturation of photon counting x-ray systems having detector channels characterized by low dynamic range. Dynamic range of a detector channel defines the range of x-ray flux levels that the detector channel can handle to provide meaningful data at the low-flux end and not experience over-ranging or saturating at the high flux end. Notwithstanding the need to prevent over-ranging, to provide diagnostically valuable data, the handling of low-flux conditions, which commonly occur during imaging through thicker cross-sections and other areas of limited x-ray transmission, is also critical in detector design. As such, the x-ray flux management device described herein is designed to satisfy both high flux and low flux conditions.

Referring now to FIG. **3**, an x-ray flux management device according to one embodiment of the invention is shown. As illustrated, the device **17**, which is shown relative to the z-axis or long axis of subject **22**, is operative as an x-ray beam chopper that is positioned between x-ray tube **14** and z-plane collimator **50**. In a preferred embodiment, the beam chopper **17** has a generally annular frame or tube **52** with two types of windows alternatively arranged along an outer rim thereof. In the illustrated exemplary embodiment, the generally annular frame is polygonal. One type of window is a transmission window **54** that provides unobstructed transmission of x-rays **16** and, as such, is designed to be placed in the x-ray beam path during low x-ray flux conditions, e.g. when a thicker subject cross-section is being imaged. The other window type is an x-ray filtering window **56** that filters or attenuates x-rays **16** when placed in the x-ray beam path and, as such, is designed to be placed in the x-ray beam path during high x-ray flux conditions, e.g. when a thinner subject cross-section is being imaged. The x-ray transmission windows **54** are preferably constructed to not effect the energy of the x-ray beam. In one embodiment, each x-ray filtering window **56** is composed of a block of x-ray filtering or attenuating material with holes (not shown) formed therein.

In the exemplary embodiment of FIG. **3**, the beam chopper has an octagonal frame. In this regard, the chopper is constructed to have four x-ray transmission windows **54** and four x-ray filtering windows **56**. With this construction, the x-ray transmission windows **54** and x-ray filtering windows are alternately formed about the frame. As such, each x-ray transmission window is adjacent a pair of x-ray filtering windows.

As further illustrated in FIG. **3**, the transmission x-ray and x-ray filtering windows **54**, **56** are arranged relative to or integrally formed within frame **52** such that the x-ray beam **16** passes through a pair of transmission windows **54** or a pair of filtering windows **56**. With this orientation, transition times between adjacent windows are advantageously reduced. For example, for an octagonal beam chopper having four x-ray transmission windows and four x-ray filtering windows of substantially equal size, only a one-quarter rotation per data acquisition view is required. As such, a rotational speed of 30,000 rpm for one-half second scanners having 1,000 views per 360 degrees of acquisition is possible.

As described above, the x-ray transmission windows **54** are placed in the x-ray beam path when the current data acquisition view is from a thicker subject cross-section. Conversely, the x-ray filtering windows **56** are placed in the x-ray beam path when the current data acquisition view is from a thinner subject cross-section. Accordingly, rotation of the chopper is dynamically controlled by controller **31**, FIG. **2**, to provide synchronization between chopper rotation and data acquisition. In this regard, it is contemplated that the chopper may be caused to rotate continuously at a fixed rotational speed or at a variable rotational speed. Additionally, it contemplated that the chopper may be initially held stationary with x-ray transmission windows placed in the x-ray beam. In this regard, saturation of the x-ray detector can be monitored and if the detector is at or near saturation, the chopper can be incrementally rotated such that x-ray filtering windows are placed in the x-ray path. For the next acquisition, the chopper is again rotated such that x-ray transmission windows are placed in the x-ray beam path. Saturation is again monitored and, if need be, a subsequent incremental rotation of the chopper. Accordingly, x-ray filtering windows are not placed in the x-ray beam path unless saturation is imminent or has occurred.

Referring now to FIG. **4**, position of the beam chopper **17** relative to the x-axis of subject **22** is illustrated. For purposes of simplicity, collimator **50**, FIG. **3**, is not shown. As illustrated, for the current data acquisition view, a pair of low x-ray flux or x-ray transmission windows **54** is positioned in the x-ray beam **16**. At high x-ray flux conditions, the beam chopper **17** will be rotated by motor **58** to rotate x-ray filtering windows **56** into the x-ray beam path **16**. In addition to rotating the beam chopper, it is contemplated that motor **58** may translate the beam chopper in the x-direction to accommodate asymmetrical subjects and variations in subject contours. In one preferred embodiment, motor **58** is a stepper motor.

Referring now to FIG. **5**, an alternate embodiment of beam chopper **17** is illustrated. In the illustrated embodiment, there are more x-ray transmission windows **54** than x-ray filtering windows **56**. As shown, there is a 2:1 relationship between the number of x-ray transmission windows and the number of x-ray filtering windows. In this regard, only every third view would be attenuated if the beam chopper is continuously rotated. Accordingly, there is not an alternating between high x-ray flux views and low x-ray flux views as in the embodiment illustrated in FIG. **3**. One skilled in the art will appreciate that such a 2:1 relationship between transmission and filtering views may be equivalently achieved with a chopper having equal number of transmission and filtering windows, but through variable rotational speed of the chopper such that the transmission windows are in the x-ray beam twice as long as the filtering windows.

Also, it is contemplated that the beam chopper **17** may be constructed such that every Nth view is attenuated. In this regard, it is contemplated that the beam chopper can be designed to have NX transmission windows, where N is a number greater than one and X is the number of filtering windows.

Referring now to FIG. **6**, another embodiment of the beam chopper is illustrated. Similar to that illustrated in FIGS. **3** and **5**, the beam chopper of FIG. **6** also has a generally annular frame **52** about which x-ray transmission windows **54** and x-ray filtering windows **56** are formed. Unlike the polygonal constructions previously described, the beam chopper **17** of FIG. **6** has a fixed radius. Notwithstanding this distinction, operation of the filter is similar to that previously described. The beam chopper **17** is rotated such that x-ray transmission windows **52** are in the x-ray beam path **16** during low x-ray flux conditions and x-ray filtering windows **54** are in the x-ray beam path **16** during high x-ray flux conditions. In the exemplary beam chopper illustrated in FIG. **6**, there is a 2:1 relationship between transmission windows and filtering windows; however, it is contemplated that the beam chopper may have less than or more than a 2:1 ratio.

As described above, it is contemplated that detector saturation readings may be acquired for a given view and if the detector has saturated (or will saturate), the beam chopper can be caused to rotate to place x-ray filtering windows in the x-ray beam. Thus, it is contemplated that for a saturated or near-saturated view, data may be acquired with the x-ray filtering windows in the x-ray beam path and that data can be used not only for image reconstruction but to correct the otherwise saturated data.

Additionally, while the beam chopper has been described such that either two x-ray transmission windows or two x-ray filtering windows are in the x-ray beam at any given moment, it is contemplated that the beam chopper may be constructed such that only one transmission or only one filtering window is in the beam path. That is, it is contemplated that the windows may be formed on a hemispherical frame such that through pendulum-like translation, different attenuation profiles may be presented. In this regard, it is further contemplated that more than two types of windows may be supported by the frame. The invention contemplates that various windows of different attenuation power may be supported by the frame whereby the continuum of attenuation windows ranges from a free transmission window of zero attenuation to a maximum attenuation window. Moreover, it is contemplated that such a hemispherical frame could be caused to rotate clockwise as well as counterclockwise and at a fixed or variable speed. Additionally, it is contemplated that a mechanical shutter of x-ray filtering material may be dynamically presented in the x-ray beam during high x-ray flux conditions.

The present invention also includes an inventive bowtie filter. Standard bowtie filters have a symmetrical, one-dimensional shape. To overcome limitations associated with these standard bowtie filters, the present invention is also directed to a 3D semi-cylindrical rotatatable bowtie filter. This multi-dimensional filter **60**, shown in FIG. **7**, has a cylindrical frame **62** with a semi-conic bore **64** formed therein. The bore **64** has an elliptical base **66**. This is in stark contrast to conventional bowtie filters which have a circular base. Additionally, also in contrast to conventional bowtie filters, filter **60** is not symmetrical. This is illustrated by the cross-sectional views of FIGS. **8** and **9**.

Referring now to FIG. **8**, cross-sectional views of filter **60** taken along lines **8-8** and lines **9-9**, respectively, are shown. As illustrated, filter **60** is constructed to have a bore **64** formed within frame **62**. The width of the bore **64** cut along line **8-8**, however, is greater than that of bore cut along line **9-9**. This results in a different absorption profile for any rotational angle of the filter **60**. Also, it is contemplated that the filter may be dynamically repositioned during data acquisition so that the resulting profile can be matched to the subject's body and, in particular, centered for non-centered subjects. In this regard, it is contemplated that precise positioning of the subject can be measured and used to control translation of the filter. Precise positioning can be determined from positioning sensors, scout scan data, and the like. By doing so, the present invention supports rotation and translation of the filter during data acquisition to track subject profile. It is also contemplated that multiple filters in a stacked arrangement could be used and moved in tandem or independently to achieve a desired attenuation profile. This can be particularly advantageous when imaging two legs and other anatomical structures that require a relatively complex attenuation profile.

Referring now to FIGS. **10-11**, a filter assembly in accordance with another embodiment of the present invention is shown. In this embodiment, a pair of bowtie filters **68**, **70** are shown relative to the x-axis and in x-ray beam **16**. Each filter **68**, **70** is thicker in the z-direction than conventional bowtie filters. In contrast to conventional bowtie filters, however, filter **68**, **70** are designed to be tilted by a tilt mechanism (not shown) to effectively change the attenuation profile of the filters. In addition to being tilted, the filters may also be moved laterally in the x-direction to better match a given subject's contours or accommodate a non-centered subject.

Additionally, while two filters stacked on top of another are shown, it is contemplated that less than two or more than two filters may be used.

As illustrated in FIG. **11**, filters **68**, **70** are tiltable relative to the z-axis. In this regard, the attenuation profile generated by the filters **68**, **70** can be dynamically controlled to match a desired attenuation profile. The tilt angle (and translation) position of the bowtie filters can be changed during data acquisition to track a given subject profile. In a preferred embodiment, the filters can be titled a maximum ninety degrees. This ninety degree tilt range defines a minimum absorption profile at zero degrees to a maximum absorption profile at ninety degrees.

Referring now to FIG. **12**, package/baggage inspection system **72** includes a rotatable gantry **74** having an opening **76** therein through which packages or pieces of baggage may pass. The rotatable gantry **74** houses a high frequency electromagnetic energy source **78** as well as a detector assembly **80**. A conveyor system **82** is also provided and includes a conveyor belt **84** supported by structure **86** to automatically and continuously pass packages or baggage pieces **88** through opening **76** to be scanned. Objects **88** are fed through opening **76** by conveyor belt **84**, imaging data is then acquired, and the conveyor belt **84** removes the packages **88** from opening **76** in a controlled and continuous manner. As a result, postal inspectors, baggage handlers, and other security personnel may non-invasively inspect the contents of packages **88** for explosives, knives, guns, contraband, etc.

Therefore, in accordance with one embodiment, the invention includes an x-ray beam chopper for a radiographic imaging apparatus. The beam chopper has a rotatable frame and at least one x-ray transmission window disposed in the rotatable frame that allows a generally free transmission of x-rays. The chopper also has at least one x-ray filtering window disposed in the rotatable frame that filters x-rays.

In accordance with another embodiment, the invention is directed to a radiographic imaging apparatus that includes an x-ray source and an x-ray detector. The apparatus further has a segmented filtering assembly having a generally annular frame with at least one low x-ray flux segment and at least one high x-ray flux segment, and a filtering assembly controller that causes the low x-ray flux segment to be in an x-ray beam path during a low x-ray flux data acquisition view and causes the high x-ray flux segment to be in the x-ray beam path during a high x-ray flux data acquisition view.

According to another embodiment, the invention includes an x-ray filter having a 3D semi-cylindrical rotatable filter body formed of x-ray attenuating matter. The filter also has a semi-conical bore formed in the 3D semi-cylindrical rotatable filter. The semi-conical bore has an elliptically shaped base.

According to yet another embodiment, the invention includes an x-ray filter assembly having a bowtie filter having an effective beam profile. The assembly further has a filter controller that tilts the bowtie filter during data acquisition to change the effective beam profile during data acquisition.

While the present invention is applicable with a number of radiographic imaging systems, it is particularly well-suited for CT systems and, especially, those systems having detectors with relative small dynamic range, such as photon counting and energy discriminating detectors. In this regard, the present invention is believed to be a key enabler for the use of direct conversion and energy discriminating/photon counting detectors with conventional CT systems. Additionally, as the beam chopper and filters selectively limit radiation exposure, the invention advantageously reduces subject dose without sacrificing image quality.

The present invention has been described in terms of the preferred embodiment, and it is recognized that equivalents, alternatives, and modifications, aside from those expressly stated, are possible and within the scope of the appending claims.

What is claimed is:

1. An x-ray filter comprising:

a 3D cylindrical rotatable filter body formed of x-ray attenuating matter; and

a semi-conical bore formed in the 3D cylindrical rotatable filter body, the semi-conical bore having an elliptically shaped base.

2. The x-ray filter of claim **1** wherein the semi-conical bore of the filter body has a first cross-section of the filter body having a first parabolic profile and a second cross-section, perpendicular to the first cross-section, having a second parabolic profile that is different from the first parabolic profile.

3. The x-ray filter of claim **2** wherein the second parabolic profile has a midpoint width less than that of the first parabolic profile.

4. The x-ray filter of claim **1** wherein the 3D cylindrical rotatable filter body comprises a first absorption profile when placed at a first position with respect to an x-ray source and comprises a second absorption profile, different from the first absorption profile, when translated to a second position with respect to the x-ray source along an x-axis transverse to a beam of x-rays emitting from the x-ray apparatus.

5. The x-ray filter of claim **1** wherein the 3D cylindrical rotatable filter body comprises a first absorption profile when placed at a first orientation with respect to an x-ray source and comprises a second absorption profile, different from the first absorption profile, when rotated to a second orientation with respect to the x-ray source.

6. The x-ray filter of claim **1** wherein the filter is moveable based on data received from one of a positioning sensor and a scout scan.

7. The x-ray filter of claim **6** further comprising a filter profile formed in the 3D cylindrical rotatable filter matched to a subject body after dynamically positioning the filter.

8. The x-ray filter of claim **7** wherein the filter profile is centered about an imaging subject.

9. A method of fabricating a CT imaging system filter comprising:

providing a cylindrical body of x-ray attenuating material;

forming a semi-conical bore having an elliptically shaped base in the cylindrical body; and

positioning the cylindrical body between an x-ray detector and an x-ray source.

10. The method of claim **9** further comprising forming a first cross-section of the attenuating material having a first parabolic profile and forming a second cross-section of the attenuating material having a second parabolic profile that is different from the first parabolic profile.

11. The method of claim **9** further comprising the step of rotating the cylindrical body to track a subject profile.

12. The method of claim **9** further comprising the step of translating the cylindrical body to track a subject profile.

13. The method of claim **12** further comprising dynamically positioning the cylindrical body during data acquisition.

14. The method of claim **12** further comprising positioning the cylindrical body based on one of positioning sensors and scout scan data.

15. An x-ray filter assembly comprising:

a first bowtie filter having an effective beam profile; and

a second bowtie filter stacked on top of the first bowtie filter in a first direction parallel to a direction of travel of an x-ray through the first and second bowtie filters;

wherein the first and second bowtie filters are laterally translatable in a second direction orthogonal to the first direction to change an effective beam profile during image data acquisition; and

wherein the first and second bowtie filters are tiltable to change the attenuation profile such that an x-ray beam passes through both the first bowtie filter and the second bowtie filters.

16. The x-ray filter assembly of claim **15** wherein the first and second bowtie filters are moveable in tandem.

17. The x-ray filter assembly of claim **15** wherein the first and second bowtie filters are moveable independently from one another.

18. The x-ray filter assembly of claim **15** wherein the first and second bowtie filters are positioned to be centered on an imaging subject positioned in an x-ray imaging system.

19. The x-ray filter assembly of claim **15** wherein the x-ray filter assembly is dynamically positionable during data acquisition.

20. The x-ray filter assembly of claim **15** wherein the x-ray filter assembly is selected to match a given subject profile.

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