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(71) **Applicants** *(for all designated States except US)*: **THE UNIVERSITY OF NORTH CAROLINA AT CHAPEL HILL** [US/US]; 308 Bynum Hall, Campus Box 4105, Chapel Hill, NC 27599-4105 (US). **NORTH CAROLINA STATE UNIVERSITY** [US/US]; 920 Main Campus Drive, Suite 400, Campus Box 8210, Raleigh, NC 27695-8210 (US).

(72) **Inventors; and**
(75) **Inventors/Applicants** *(for US only)*: **ZHOU, Otto, Z.** [US/US]; 204 Glenhill Lane, Chapel Hill, NC 27514 (US). **YANG, Guang** [US/US]; 118 Friar Lane, Carrboro, NC 27510 (US). **LU, Jianping** [US/US]; 109 Glen Haven Drive, Chapel Hill, NC 27516 (US). **LALUSH, David** [US/US]; 103 Bell Vista Drive, Cary, NC 27513 (US).

(74) **Agent: WILSON, Jeffrey, L.**; Jenkins, Wilson, Taylor & Hunt, P.A., Suite 1200, University Tower, 3100 Tower Boulevard, Durham, NC 27707 (US).

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(54) **Title:** STATIONARY X-RAY DIGITAL BREAST TOMOSYNTHESIS SYSTEMS AND RELATED METHODS

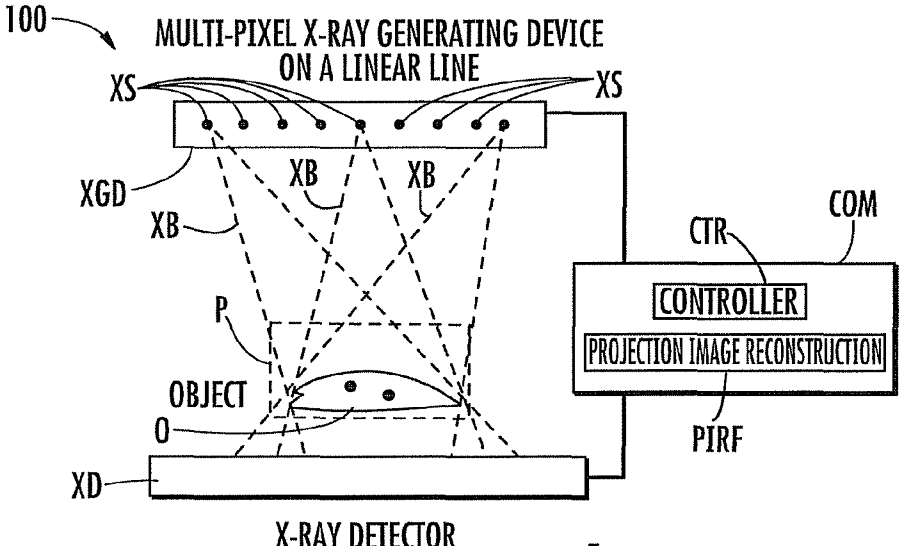

FIG. 1

(57) **Abstract:** Stationary x-ray digital breast tomosynthesis systems and related methods are disclosed. According to one aspect, the subject matter described herein can include an x-ray tomosynthesis system having a plurality of stationary field emission x-ray sources configured to irradiate a location for positioning an object to be imaged with x-ray beams to generate projection images of the object. An x-ray detector can be configured to detect the projection images of the object. A projection image reconstruction function can be configured to reconstruct tomography images of the object based on the projection images of the object.

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DESCRIPTION

STATIONARY X-RAY DIGITAL BREAST TOMOSYNTHESIS SYSTEMS AND RELATED METHODS

RELATED APPLICATION

The presently disclosed subject matter claims the benefit of U.S. Provisional Patent Application Serial No. 60/961,175, filed July 19, 2007, the disclosure of which is incorporated herein by reference in its entirety.

GOVERNMENT INTEREST

This presently disclosed subject matter was made with U.S. Government support under Grant No. US4CA119343 awarded by the National Cancer Institute. Thus, the U.S. Government has certain rights in the presently disclosed subject matter.

TECHNICAL FIELD

The subject matter described herein relates to x-ray radiography. More specifically, the subject matter describes stationary x-ray digital breast tomosynthesis systems and related methods.

BACKGROUND

Mammography is currently the most effective screening and diagnostic tool for early detection of breast cancer, and has been attributed to the recent reduction of breast cancer mortality rate. However, the nature of the two-dimensional mammogram makes it difficult to distinguish a cancer from overlying breast tissues, and the interpretation can be variable among radiologists. A higher rate of false positive and false-negative test results exist because the dense tissues interfere with the identification of abnormalities associated with tumors. Digital breast tomosynthesis (DBT) is a three-dimensional imaging technique that is designed to overcome this problem. It is a limited angle tomography technique that provides reconstruction planes in the breast using projection images from a limited angular range.

Several prototype DBT scanners have been manufactured by commercial vendors. The system designs are based on a full-field digital mammography (FFDM) unit. A mammography x-ray tube is used to collect the projection images by moving 10 - 50 degrees around the object. The reported total scanning time is 7 - 40 seconds depending on the number of views and the thickness of the breast, which is much longer than that of the regular mammography. The long imaging time can cause patient motion blur which degrades image quality and can make patients uncomfortable. Further, the power of the x-ray source, gantry rotating speed and detector frame rate limit the scanning speed of the current DBT systems.

DBT systems utilize the standard mammography x-ray tube with about a 300 μm x-ray focal spot size. Due to the gantry rotation and mechanical instability, the effective focal spot size during image acquisition is larger than the static value which degrades the image resolution. Two gantry rotation modes have been developed. One commercially-available system uses a stop-and-shoot technique. The gantry makes a full stop before taking each projection image. Acceleration/deceleration can cause mechanical instability of the system. A continuous rotation mode is used in other commercially available systems. The gantry keeps a constant rotation speed during the whole imaging process. In this case, the x-ray focal spot size is enlarged along the motion direction. The value of the enlargement depends on the rotation speed and the exposure time. It has been reported that the x-ray focal spot moves about 1 mm in a typical scan. This does not leave room for further reduction of the total scanning time, which will require a faster gantry rotation and a larger focal spot blurring.

It would be beneficial to provide x-ray imaging systems and methods having reduced data collection times and improvements for patient comfort. One or more such improvements can enable new applications for x-ray imaging of breast tissue as well as other objects. Accordingly, it is desirable to provide x-ray imaging systems and methods having one or more of these improvements.

In addition, current clinical mammography scanners use polychromatic x-ray radiation with slight energy filtering. It is known that monochromatic and quasi-monochromatic radiation provides better imaging quality and can potentially reduce the imaging dose. Currently, there is no effective way, however, to generate monochromatic or quasi-monochromatic radiation in a clinical environment that can provide sufficient x-ray photo flux. Accordingly, it is desirable to provide x-ray imaging systems and methods that can perform monochromatic or quasi-monochromatic imaging in a clinically acceptable scanning speed.

SUMMARY

It is an object of the presently disclosed subject matter to provide novel stationary x-ray digital breast tomosynthesis systems and related methods.

An object of the presently disclosed subject matter having been stated hereinabove, and which is achieved in whole or in part by the presently disclosed subject matter, other objects will become evident as the description proceeds when taken in connection with the accompanying drawings described hereinbelow.

BRIEF DESCRIPTION OF THE DRAWINGS

The subject matter described herein will now be explained with reference to the accompanying drawings, of which:

Figure 1 is a schematic diagram of a MBFEX system according to an embodiment of the subject matter described herein;

Figure 2 is a schematic diagram of a MBFEX system having x-ray sources positioned along a linear line according to an embodiment of the subject matter described herein;

Figure 3 is a schematic diagram of a MBFEX system having x-ray sources positioned along a two-dimensional plane according to an embodiment of the subject matter described herein;

Figure 4A is a schematic diagram of a MBFEX system having x-ray sources positioned along a line, evenly spaced, and angled for directing x-ray beams towards an object according to an embodiment of the subject matter described herein;

Figure 4B is a schematic diagram of two x-ray source pixels of the system shown in Figure 4A according to an embodiment of the subject matter described herein;

Figure 5 is a flow chart of an exemplary process of acquiring object images according to an embodiment of the subject matter disclosed herein;

Figure 6 is a flow chart of an exemplary process of sequentially acquiring object images utilizing a MBFEX system according to an embodiment of the subject matter disclosed herein;

Figure 7 is a flow chart of an exemplary process of multiplexing object images utilizing a MBFEX system according to an embodiment of the subject matter disclosed herein;

Figure 8 is a timing diagram of a detector trigger and x-ray source pixel triggers according to an embodiment of the subject matter described herein;

Figure 9 is an image of a MBFEX system in accordance with the subject matter disclosed herein;

Figure 10 is a schematic diagram showing the spatial relationship between an x-ray detector, a phantom, and x-ray sources of an x-ray generating device in accordance with the subject matter disclosed herein;

Figure 11A is a perspective view of an x-ray source according to an embodiment of the subject matter disclosed herein;

Figures 11B is a schematic diagram of an x-ray source according to an embodiment of the subject matter disclosed herein;

Figure 12 is a circuit diagram of a controller configured to control the emission of x-ray beams from a plurality of x-ray sources in accordance with the subject matter described herein;

Figure 13 is an image of a MBFEX x-ray source array according to an embodiment of the subject matter described herein;

Figure 14 is a graph of an experimentally measured energy spectrum of the system shown in Figure 9;

Figure 15 is a graph of anode current as a function of gate voltage for the system shown in Figure 9;

Figure 16 is a projection image of a cross phantom obtained in accordance with the subject matter described herein; and

Figure 17 is a graph showing the line profiles of the two wires obtained in accordance with the subject matter described herein.

DETAILED DESCRIPTION

The subject matter disclosed herein is directed to multi-beam field emission x-ray (MBFEX, also referred to as multi-pixel field emission x-ray) systems and techniques that can utilize a plurality of field emission x-ray sources, an x-ray detector, and projection image reconstruction techniques. Particularly, the systems and techniques disclosed herein according to one aspect can be applied to x-ray digital tomosynthesis. In accordance with one embodiment, a plurality of field emission x-ray sources can irradiate a location for positioning an object to be imaged with x-ray beams for generating projection images of the object. An x-ray detector can detect the projection images of the object. A projection image reconstruction function can reconstruct tomography images of the object based on the projection images of the object. The subject matter disclosed herein enables increased scanning speed, a simplification of system design, and image quality enhancements.

In one application, the subject matter disclosed herein can be a stationary digital breast tomosynthesis (DBT) system that utilizes a carbon nanotube based MBFEX system. The MBFEX system can include an array of individually programmable x-ray pixels that can be substantially evenly placed to cover a wide field of view. Projection images can be acquired by electronically switching on and off the individual x-ray pixels without mechanical motion of any of the x-ray source, the detector, or the object.

In one embodiment of the subject matter described herein, projection images of an object can be collected sequentially, one at a time, from different viewing angles by electronically switching on and off individual x-ray source pixels. The x-ray source pixels can be spatially distributed. Each pixel can be switched on for a predetermined time and a predetermined current to deliver a predetermined amount of dosage to the object. The transmitted x-ray intensity from a particular x-ray source pixel can be detected and recorded by an x-ray detector. The spacing between the x-ray beam pixels and the number of pixels can be varied to provide angular coverage and the number of projection images desired. The projection images collected from different viewing angles can be processed to reconstruct tomography images of the object to reveal the internal structure of the object. In one example, the x-ray source can include a total of between about ten and one-hundred x-ray focal spots (e.g., twenty-five (25) x-ray source pixels) positioned along an arc that can cover a viewing range of between about 10 and 100 degrees (e.g., 30-50 degree viewing range). The focal spots define a plane that is substantially perpendicular to an imaging plane of the x-ray detector.

In one embodiment of the subject matter described herein, one or a plurality of monochromators can be used to generate monochromatic x-ray radiation for imaging an object. Such monochromatic x-ray radiation can be produced using Bragg diffraction. Quasi-monochromatic x-ray beams can be generated by placing filters in front of an x-ray window that receives polychromatic x-ray radiation. By selecting the filtering material and thickness of the material, quasi-monochromatic radiation with a narrow energy window can be produced. This, however, typically includes the use of 200^{th} to 500^{th} value layer filtering material. This means that 99.5 to 99.8% of the x-ray intensity is attenuated by the filter. The low x-ray flux has prevented the use of monochromatic x-ray radiation for clinical imaging.

In one example of monochromatic x-ray radiation, the generated monochromatic x-ray radiation can be utilized for imaging a breast. An advantage of monochromatic and quasi-monochromatic x-ray radiation includes improved imaging quality at reduced x-ray dose, which is important for breast imaging. The subject matter described herein can enable physicians to image human breasts using quasi-monochromatic x-ray radiation at an imaging speed that is comparable to commercially-available DBT scanners with polychromatic x-ray radiation.

One technique to overcome the obstacles of low flux and therefore long imaging time is to combine multi-beam field emission x-ray source with multiplexing x-ray imaging. Cone beam quasi-monochromatic radiation can be produced by heavy filtering. The pixilated and spatially distributed MBFEX source can generate x-ray beams from multiple projection angles without mechanical motion. A stationary DBT scanner operating in the sequential scanning mode can provide a full scan of 25 views using 85 mAs total dose with a speed that is a factor of 10 faster than the C-arm based DBT scanners at a comparable dose. Experiments have also shown that the parallel multiplexing imaging process provides a factor of N/2 (N = number of x-ray pixels) increase of the imaging speed comparing to the conventional serial imaging technique used for tomography. The combination of the gains from stationary design and multiplexing described herein (about x 100) can compensate for the loss of x-ray flux due to the use of heavy filtering (100^{th} value layer) which enables the qM-DBT scanner to operate at a comparable scanning time as commercially-available C-arm based system, but with a better imaging quality and a reduced imaging dose.

As referred to herein, the term "nano-structured" or "nanostructure" material can designate materials including nanoparticles with particle sizes less than 100 nm, such as nanotubes (e.g., carbon nanotubes). These types of materials have been shown to exhibit certain properties that have raised interest in a variety of applications.

As referred to herein, the term "multi-beam x-ray source" can designate devices that can simultaneously or sequentially generate multiple x-ray beams, For example, the "multi-beam x-ray source" can include a field emission based multi-beam x-ray source having electron field emitters. The electron field emitters can include nano-structured materials based materials.

Figure 1 is a schematic diagram of a MBFEX system generally designated **100** according to an embodiment of the subject matter described herein. Referring to Figure 1, system **100** can include a computer **COM** having a controller **CTR** configured to control an x-ray generating device **XGD** and an x-ray detector **XD** for imaging an object **O** to be imaged. X-ray generating device **XGD** can include a plurality of individually-controllable, field emission x-ray sources **XS** configured to irradiate object **O** with x-ray beams **XB** for generating projection images of object **O**.

X-ray sources **XS** can be positioned for directing x-ray beams **XB** towards a location or position **P** (designated by broken lines) at which object **O** can be placed. The x-ray beams can be directed towards position **P** from several different angles. Further, x-ray sources **XS**, x-ray detector **XD**, and position **P** are positioned such that the generated projection images are detected by x-ray detector **XD**. X-ray sources **XS** are positioned along a substantially straight line formed by x-ray generating device **XGD** such that the generated x-ray beams are directed substantially towards position **P** and can pass through the area within position **P**. The line can be parallel to an imaging plane of the x-ray detector. As described in further detail below, x-ray sources **XS** can be arranged in any suitable position such that the x-ray beams are directed substantially towards position **P** and the projection images are detected by x-ray detector **XD**. The x-ray sources and x-ray detector can be stationary with respect to one another during irradiation of an object by the x-ray sources and detection of the projection images by the x-ray detector. The x-ray sources can be controlled for sequential activation one at a time for a predetermined dwell time and predetermined x-ray dose.

After passing through object **O** at position **P**, x-ray beams **XB** can be detected by x-ray detector **XD**. X-ray detector **XD** can be a high frame rate, digital area x-ray detector configured to continuously capture x-ray beams **XB**. After all or at least a portion of x-ray beams **XB** are collected and stored as x-ray signal data in a memory, a projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of the object **O**.

The tomography images can be constructed by using a suitable technique to obtain multi-projection images of an object from multiple x-ray sources using a single detector. Common techniques include shift-and-add, filtered back projection, ordered subsets convex maximum likelihood, etc.

According to another aspect of the subject matter disclosed herein, x-ray sources can be positioned along an arc defined by the x-ray generating device. The arc can define a plane that can be substantially perpendicular to an imaging plane of the x-ray detector. Figure 2 is a schematic diagram of a MBFEX system generally designated **200** having x-ray sources **XS** positioned along a linear line according to an embodiment of the subject matter described herein. Referring to Figure 2, x-ray sources **XS** can be positioned at least substantially along a linear line formed by x-ray generating device **XGD**. X-ray sources **XS** can be positioned for directing x-ray beams **XB** towards and through position **P** at which object **O** can be placed. The x-ray beams can be directed to position **P** from several different angles. Further, x-ray sources **XS**, x-ray detector **XD**, and position **P** can be positioned such that the generated projection images are detected by x-ray detector **XD**. After passing through object **O** at position **P**, x-ray beams **XB** can be detected by x-ray detector **XD**. After all or at least a portion of x-ray beams **XB** are collected and stored as x-ray signal data in a memory, a projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of the object **O**.

According to another aspect of the subject matter disclosed herein, x-ray sources can include focal spots positioned along a two-dimensional plane or matrix on an x-ray anode. Figure 3 is a schematic diagram of a MBFEX system generally designated **300** having x-ray sources **XS** positioned along a two-dimensional plane according to an embodiment of the subject matter described herein. Referring to Figure 3, x-ray sources **XS** can be positioned substantially along a two-dimensional plane formed by x-ray generating device **XGD**. X-ray sources **XS** can be positioned for directing x-ray beams **XB** towards and through position **P** at which object **O** can be placed. The x-ray beams can be directed to position **P** from several different angles. Further, x-ray sources **XS**, x-ray detector **XD**, and position **P** are positioned such that the generated projection images are detected by x-ray detector **XD**. After passing through object **O** at position **P**, x-ray beams **XB** can be detected by x-ray detector **XD**. After all or at least a portion of x-ray beams **XB** are collected and stored as x-ray signal data in a memory, a projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of the object **O**.

According to another aspect of the subject matter disclosed herein, x-ray sources can be positioned along a line, evenly spaced, and angled for directing x-ray beams towards an object. Figure 4A is a schematic diagram of a MBFEX system generally designated **400** having x-ray sources **XS** positioned along a line, evenly spaced, and angled for directing x-ray beams towards an object according to an embodiment of the subject matter described herein. Referring to Figure 4A, x-ray sources **XS** can be gated carbon nanotube emitting pixels positioned substantially along a two-dimensional plane formed by x-ray generating device **XGD**. In this example, the x-ray generating device includes twenty-five (25) x-ray sources in total, although the x-ray generating device can alternatively include any suitable number of x-ray sources, more or less even that twenty-five in number.

The x-ray source can be housed in a vacuum chamber having a 30 μm thick molybdenum (Mo) window. The window can function as a radiation filter. Each pixel can comprise a carbon nanotube cathode, a gate electrode to extract the electrons, and a set of electron focusing lenses (e.g., Einzel-type electrostatic focusing lenses) to focus the field emitted electrons to a small area (focal spot) on the target. The focal spots can be substantially the same size. The sizes of the focal spots and/or the x-ray flux generated by the x-ray sources can be adjusted by the controller. Alternatively, the focal spots can range between about 0.05 mm and 2 mm in size. The system is designed for an isotropic 0.2 x 0.2 mm effective focal spot size for each x-ray source pixel. The individual focal spot size can be adjusted by adjusting the electrical potentials of the focusing electrodes. To minimize the current fluctuation and delay and to reduce pixel to pixel variation, an electrical compensation loop can be incorporated to automatically adjust the gate voltage to maintain a constant pre-set emission current. The area of the carbon nanotube cathode can be selected such that a peak x-ray tube current of about 10 mA can be obtained with the effective focal spot size of 0.2 x 0.2 mm. A higher x-ray peak current of 50-100 mA can be obtained by increasing the carbon nanotube area and the focal spot size.

X-ray sources **XS** can be positioned for directing x-ray beams **XB** towards position **P** at which object **O** is placed. The x-ray beams can be directed towards and through position **P** from several different angles. Further, x-ray sources **XS**, x-ray detector **XD**, and position **P** are positioned such that the generated projection images are detected by x-ray detector **XD**. To collect the projection images of object **O** from different angles for tomosynthesis, controller **CTR** can sequentially activate an array of electron emitting pixels, as described in further detail below, which are spatially distributed over a relatively large area. X-ray sources **XS** are positioned such that the generated x-ray beams are directed at least substantially to position **P**. Each x-ray source **XS** can include a field emitter operable to generate an electron beam and operable to direct the electron beam to a focal point of a target. The emitted electron beam can be accelerated to the target where a scanning x-ray beam originates for different points over a large area of the target. The controller **CTR** can further vary the intensity of the x-ray radiation based on the distance between x-ray source **XS** and object **O** such that the x-ray dose delivered to object **O** from every viewing angle is the same.

X-ray sources **XS** can be positioned such that x-ray generating device **XGD** provides a substantially even 2 degree angular spacing between the x-ray focal spots at a source-to-detector distance of about 64.52 cm. The position and the orientation of the individual x-ray target can be such that the center axis of a generated x-ray cone beam goes through an iso-center **OC**, which can either be a location on object **O** to be imaged or a point on x-ray detector **XD**. The cone-shaped x-ray beams can have substantially the same x-ray intensity distribution on the object. Further, x-ray sources can produce x-ray radiation having different energy spectra.

After passing through object **O** at position **P**, x-ray beams **XB** can be detected by x-ray detector **XD**. After all or at least a portion of x-ray beams **XB** are collected and stored as x-ray signal data in a memory, a projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of the object **O**.

Figure 4B is a schematic diagram of two x-ray source pixels of system **400** shown in Figure 4A according to an embodiment of the subject matter described herein. Referring to Figure 4B, x-ray sources **XS1** and **XS2** are angled towards an iso-center **OC** of object **O**. The line formed by center **OC** of object **O** and an x-ray focal spot of an electron field emitter of each x-ray source is on a symmetric plane of the x-ray source. The x-ray source pixels can be tilted such that the central line of the x-ray beams generated by the x-ray source pixels are directed towards iso-center **OC**. In one example, the central lines of x-ray sources **XS1** and **XS2** form a tilted angle. The x-ray sources can be tilted with respect to one another to achieve a desired tilted angle.

In system **400**, the x-ray beam originating from a focal point can be generated by the electron beam from a corresponding pixel on a cathode. A scanning x-ray beam can be generated by sequentially activating individual pixels. A constant high DC voltage (about 0 – 100 KVp) can be applied between the x-ray anode and the gate electrode. A variable DC voltage (about 0-2 kV) can be applied on the gate electrode. Alternatively, the x-ray anodes can be configured at different voltages to produce x-ray radiation with multiple energies. For example, for a system having 25 x-ray sources, 12 anodes can be configured at low voltage, and 13 anodes can be configured at high voltage. Such a configuration enables the system for dual-energy imaging.

Switching on and off the individual emitting pixel can be effected by an electronic circuit (e.g., a MOSFET circuit) connected to the cathode. The electronic circuit can be used to individually control the x-ray intensities from the different x-ray focal spots **XS** (e.g., x-ray sources **XS1** and **XS2**) such that they can either be the same or be modulated to deliver a desired intensity or intensity distribution on object **O** to be imaged. An x-ray beam can be produced from a corresponding focal point when the electron beam bombards the anode surface of the target. To generate a scanning beam, a pulsed voltage with a predetermined pulse width can be scanned across the individual MOSFETs. At each point, the channel can be "opened" to generate an electron beam from the pixel, which can lead to the generation of an x-ray beam from the corresponding focal point on the target. To minimize the fluctuation of the x-ray flux, the cathode can be operated at a constant current mode. The gate voltage can be adjusted automatically to maintain the emission current and thus x-ray flux from each pixel to within a desired level.

The 25 x-ray source pixels of x-ray generating device **XGD** can span a distance of 57.45 cm from end-to-end. At a source-to-object distance of 64.52 cm, the device provides 48 degree coverage with a substantially even 2 degree angular spacing between adjacent pixels. The linear spacing between adjacent x-ray source pixels can vary to provide even angular spacing. The x-ray beams can be collimated to a 23.04 cm field-of-view (FOV) at the phantom plane. If the x-ray source pixels are arranged in a linear line parallel to the detector plane rather than an arc, the pixel-to-source distance can vary from pixel to pixel. In one option to compensate this variation in x-ray beam traveling distance, the x-ray tube current from each pixel can be individually adjusted such that the flux at the phantom surface remains the same. In another solution, the image intensities can be normalized in the reconstruction process. The phantom can be placed on a stage for positioning with a 2.54 cm air gap between the detector and the phantom.

Figure 5 is a flow chart illustrating an exemplary process of acquiring object images according to an embodiment of the subject matter disclosed herein. System **100** is referenced in this example, although any other system described herein may utilize the process for acquiring object images. Referring to Figures 1 and 5, controller **CTR** can activate x-ray sources **XS** to irradiate object **O** with x-ray beams for generating projection images of object **O** (block **500**). At block **502**, controller **CTR** can control x-ray detector **XD** to detect the projection images of object **O**. At block **504**, projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of object **O**. Any suitable technique can be utilized by function **PIRF** for reconstructing the tomography images of object **O**.

Figure 6 is a flow chart illustrating an exemplary process of sequentially acquiring object images utilizing a MBFEX system according to an embodiment of the subject matter disclosed herein. System **100** shown in Figure 1 is referenced in this example, although any other system described herein can utilize the process for acquiring object images. Referring to Figures 1 and 6, at block **600**, controller **CTR** of system **100** can initiate the process and set variable i to 1. Variable i represents the iteration number of the process. At block **602**, controller **CTR** can turn on x-ray sources **XS** corresponding to the i^{th} pixels. Particularly, one or more x-ray sources **XS** can correspond to an i^{th} group of x-ray sources. As described in further detail below, the process sequences through groups of i x-ray sources until the entirety of x-ray sources have been turned on and their x-ray beams **XB** detected.

At block **604**, controller **CTR** can control x-ray detector **XD** to acquire the i^{th} image. Particularly, x-ray detector **XD** can acquire the projection image of object **O** generated by the i^{th} x-ray source(s). Controller **CTR** can determine whether acquisition of images from all i groups of x-ray sources has been completed (block **606**). If it is determined that images have not been acquired from all i groups of x-ray sources, controller **CTR** can increment variable i by 1 (block **608**) and the process can proceed to block **502** to acquire images from the remaining groups of x-ray sources.

If it is determined that images have been acquired from all i groups of x-ray sources, projection image reconstruction function **PIRF** can reconstruct tomography images of object **O** based on the projection images of the object (block **610**). At block **612**, a display of computer **COM** can display the reconstructed slice images of object **O**.

Figure 7 is a flow chart illustrating an exemplary process of sequentially acquiring object images utilizing a MBFEX system according to an embodiment of the subject matter disclosed herein. System **100** shown in Figure 1 is referenced in this example, although any other system described herein can utilize the process for acquiring object images. A LABVIEW™ (available from National Instruments Corporation) based software application can be utilized to generate a function for electronically controlling the triggering and switching of x-ray beam pixels and to synchronize the x-ray exposure with detector data collection.

Referring to Figures 1 and 7, at block **700**, controller **CTR** of system **100** can initiate the process and set variable i to 1. Variable i represents the iteration number of the process. At block **702**, controller **CTR** can turn on x-ray sources **XS** corresponding to the i^{th} pixels. Particularly, one or more x-ray sources **XS** can correspond to an i^{th} group of x-ray sources. As described in further detail below, the process sequences through groups of i x-ray sources until the entirety of x-ray sources have been turned on and their x-ray beams **XB** detected. In one embodiment, controller **CTR** can control the x-ray sources to generate multiplexed x-ray beams, which can be demultiplexed for image reconstruction as described below.

At block **704**, controller **CTR** can control x-ray detector **XD** to acquire the i^{th} image. Particularly, x-ray detector **XD** can acquire the projection image of object **O** generated by the i^{th} x-ray source(s). Controller **CTR** can determine whether acquisition of images from all i groups of x-ray sources has been completed (block **706**). If it is determined that images have not been acquired from all i groups of x-ray sources, controller **CTR** can increment variable i by 1 (block **708**) and the process can proceed to block **702** to acquire images from the remaining groups of x-ray sources.

After image acquisition, projection image reconstruction function **PIRF** can apply tomosynthesis reconstruction (block **710**) and display the reconstructed images via the display of computer **COM** (block **712**). Alternatively, if the x-ray beams were multiplexed, projection image reconstruction function **PIRF** can demultiplex the images (block **714**), apply tomosynthesis reconstruction (block **716**) and display the reconstructed images via the display of computer **COM** (block **718**).

Any suitable multiplexing imaging technique can be utilized in a system in accordance with the subject matter described herein. In this imaging mode, all or a sub-group of x-ray source pixels can be switched on simultaneously to illuminate the object. One example of a multiplexing technique includes a frequency division multiplexing. By using a multiplexing technique, the total image collection time can be significantly increased.

In one example of a multiplexing technique, an orthogonal frequency division multiplexing technique can be utilized. In this example, pulsed x-ray signals are generated and each x-ray beam can have a unique pulse width and repetition rate. Further, in this example, the detector records the transmitted x-ray intensity from the "on" x-ray pixels as a function of time. The recorded image is then de-multiplexed in the frequency domain to obtain the projection images from the individual pixels.

In another example of a multiplexing technique, a binary multiplexing technique can be utilized. An example of a binary multiplexing technique is described in U.S. Patent Application Serial No. 11/804,897, titled "Methods, Systems, and Computer Program Products for Binary Multiplexing X-Ray Radiography, the disclosure of which is incorporated herein by reference in its entirety and which is commonly assigned to the same entities as the present patent application. In this example, a sub-set of the x-ray beams is switched on sequentially. The individual projection images are obtained by a linear combination of the composite images from the sub-sets.

X-ray sources may be triggered sequentially and projection images acquired accordingly. Figure 8 is a timing diagram of a detector trigger and x-ray source pixel triggers according to an embodiment of the subject matter described herein. In this example, the system includes 25 x-ray source pixels.

The signals represent control signals generated by a controller for controlling the x-ray sources and detector. X-ray radiation is on when a trigger signal is at 5 V. The exposure time is T_{exp} for each pixel, which is the same as the integration time T_{int} of the detector. The detector readout is triggered by the rising edge of the signal. The time to capture one image is represented by T_r. The total scan time is 25 * (T_{exp} + T_r).

In accordance with the subject matter described herein, the field emission x-ray sources can each include a field emission cathode, a gate electrode that extracts electrons from the cathode when an electrical field is applied between the gate and the cathode, a focusing unit that focuses the field emitted electrons to a defined focal area on an anode, and the anode that produces the x-ray radiation when it is bombarded by the electron beam. The field emission cathode can includes carbon nanotubes, nanowires, and/or microfabricated tips. The gate electrodes can be either controlled individually or connected electrically.

For purposes of experimentation, one embodiment of a system in accordance with the subject matter disclosed herein was constructed. Figure 9 is an image of a MBFEX system generally designated **900** in accordance with the subject matter disclosed herein. Referring to Figure 9, system **900** includes x-ray detector **XD**, x-ray generating device **XGD**, and a stage **S** for positioning of a phantom **PH** to be imaged. X-ray generating device **XGD** includes a carbon nanotube MBFEX source. X-ray detector **XD** is a flat panel x-ray detector. System **900** includes a control unit and a computer work station. X-ray detector **XD** can be a flat panel detector. The field of view can be about 19.5 cm x 24.4 cm, which can ensure a full image of a breast. With a 127 μm pixel pitch, the total array size is 1536 x 1920. The detector can run under non-binning mode and 2 x 2 binning mode. Referring again to Figure 7, in the user synchronization mode, the rising edge of a continuous TTL signal can trigger readout of the detector. The imaging time is determined by the integration window T_{int} and the detector readout time T_r. T_{int} is controllable through the triggering signal. X-ray radiation is delivered within the integration window T_{int}, and the radiation period is denoted as T_{exp}. The readout time T_r depends on the acquisition mode. For the normal 2 x 2 binning modes, the readout time is

128 ms and 32 ms, respectively. In this example, the projection images are taken sequentially.

Figure 10 is a schematic diagram showing the spatial relationship between x-ray detector **XD**, phantom **PH**, and x-ray sources **XS** of x-ray generating device **XGD**. The distance between the center of phantom **PH** and x-ray generating device **XGD** is about 64.5 cm. The x-ray generating device **XGD** to detector **XD** distance is about 69.6 cm, which leaves about a 2.5 cm air gap for a normal 5-cm breast phantom. The x-ray sources are arranged linearly to reduce the system complexity, with even-angular distribution and about a 2-degree step or increment. The total angular coverage of the x-ray generating device is about 48 degrees. In this system, the distance between the nearest x-ray focal spots varies from 2.5 cm to 2.7 cm, and the total span of the x-ray generating device array is about 57.5 cm.

System **900** includes a field emission x-ray source array. The construction of the 25 x-ray source pixels is substantially identical. Figures 11A and 11B are perspective and schematic diagram views, respectively, of an x-ray source **XS** according to an embodiment of the subject matter disclosed herein. Referring to Figures 11A and 11B, x-ray source **XS** can include an electron field emitter **FE** for emitting electrons. Electron field emitter **FE** can comprise one or more carbon nanotubes and/or other suitable electron field emission materials. Electron field emitter **FE** can be attached to a surface of a cathode **C**, conductive or contact line, or other suitable conductive material.

Electron field emitter **FE** can be controlled by a suitable controller (such as controller **CTR** shown in Figure 4A) including MOSFET circuitry. The controller can control voltage sources to apply voltage between electron field emitter **FE** and a gate electrode **GE** to generate electric fields for extracting electrons from electron field emitter **FE** to thereby produce an electron beam **EB**. The controller can operate MOSFET circuitry for individually controlling electron beam emission by x-ray sources. The drains of the MOSFETs can be connected to cathode **C** for controlling electron beam emission by emitter **FE**. The MOSFETs can be turned on and off by the application of high signal (e.g., 5 V) and a low signal (e.g., 0 V) to the gates of the MOSFET. When a high signal is applied to the gate of a MOSFET, a drain-to-source channel of the transistor is turned on to apply a voltage difference between cathode **C** and gate electrode **GE**. A voltage difference exceeding a threshold can generate an electric field between cathode **C** and gate electrode **GE** such that electrons are extracted from electron field emitter **FE**. Conversely, when a low voltage (e.g., 0 V) is applied to the gate of a MOSFET, a drain-to-source channel is turned off such that the voltage at emitter **FE** is electrically floating and the voltage difference between cathode **C** and gate electrode **GE** cannot generate an electric field of sufficient strength to extract electrons from emitter **FE**.

Cathode **C** can be grounded, and other electrodes maintained at constant voltages during imaging acquisition. The gate voltage determines the x-ray tube current. Below a threshold, there is no current, and the current increases exponentially with gate voltage above the threshold. In one example, each x-ray pixel can provide a tube current of between 0.1 and 1 mA at 40 kVp. The controller is operable to apply voltage pulses of different frequencies to the gates of the MOSFET.

Further, x-ray source **XS** can include an anode **A** having a focus spot for bombardment by electron beam **EB**. A voltage difference can be applied between anode **A** and gate electrode **GE** such that a field is generated for accelerating electrons emitted by electron field emitter **FE** toward a target structure **TS** of anode **A**. The target structure can produce x-ray beams having a predetermined signal upon bombardment by electron beam **EB**. X-ray source **XS** can include focusing electrodes **FEL1** and **FEL2** for focusing electrons extracted from electron field emitter **FE** on target structure **TS** and thus reduce the size of electron beam **EB**. Focusing electrodes **FEL1** and **FEL2** can be controlled by application of voltages to the focusing electrodes by a voltage source. The voltage applied to the focusing electrodes controls the electron trajectory. The gate voltage can be varied depending on required flux.

Electron field emitter **FE** and gate electrode **GE** can be contained within a vacuum chamber with a sealed interior at about 10^{-7} Torr pressure. The interior of the vacuum chamber can be evacuated to achieve a desired interior pressure. X-ray radiation can travel from the interior of the vacuum chamber to its exterior through an x-ray permeable portion or window. In one example, the x-ray permeable portion or window can be a beryllium (Be) or molybdenum

(Mo) window. The molybdenum anode and filter combination can be used for breast imaging among other applications. Up to a 40 keV high voltage can be applied on anode **A**. Anode **A** can be suitably shaped and/or angled such that the generated x-ray beams are transmitted toward an object from a plurality of different viewing angles. The targeted performance for the source is that each x-ray source pixel can provide a 10 mA peak current at 200 μm x 200 μm effective focal spot size. Alternatively, the energy filter can comprise Cerium, and the voltage applied on anode **A** can be in the range of 60-80 kV.

Figure 12 is a circuit diagram of controller **CTR** configured to control the emission of x-ray beams from a plurality of x-ray sources in accordance with the subject matter described herein. Referring to Figure 12, controller **CTR** can include a plurality of MOSFETs operable to individually switch on and off x-ray sources **XS**. The drains (D), gates (G), and sources (S) of the MOSFETs are connected to respective cathodes **C**, TTL trigger signals generated by a computer board **CB**, and a common ground **GND**. When a TTL trigger signal is at a low state, the conduction channel between the source and drain is closed. This causes the carbon nanotube cathode potential to float relatively to the common ground **GND**, and no electrons are emitted from cathode **C** and thus no x-ray beams are generated. When the TTL trigger signal is at a high state, cathode **C** is grounded because of the opened conduction channel. As a result, electrons are extracted by the electrical field between gate **G** and cathode **C**, and x-ray radiation is produced. The delay time (between switching of the TTL signal and the conduction channel) of the MOSFET is about 35 to 45 ns, which is sufficient considering the tens of milliseconds x-ray exposure period. The x-ray source pixels can be switched individually at any given time during the imaging acquisition process, which provides great flexibility. Variable resistors **R** are built-in for compensation of the variations in the individual cathode performance.

Figure 13 is an image of a MBFEX x-ray source array according to an embodiment of the subject matter described herein. The array includes 25 individually controllable x-ray source pixels **XS** which are tilted towards the iso-center of a position for placement of an object to be imaged.

To reconstruct slice images, an iterative ordered-subset convex (OSC) technique can be used by the reconstruction function based on a maximum-likelihood model. The reconstruction technique applies a sharing method to convert all projections images to a common frame of reference, and then uses a pre-computed cone-beam model to project and back-project in the common frame. To reduce the computation load, non-cubic voxels are reconstructed. This technique has been verified on both simulated data and breast phantom images measured from a field emission x-ray source array with a limited number of pixels.

Table 1 shows a comparison of system **900** shown in Figure 9 with commercially available systems.

	System **900** of Figure 9	GE: Senographe 2000D	Siemens: Mammomat Novation	Hologic: Selenia
X-Ray kVp, mA	25-35 kVp, 10 mA	25-30 kVp, ~130 mA	~28 kVp, ~180 mA	24-39 kVp, ~100 mA
Focal Spot Size	200 μm	300 μm	300 μm + blur*	300 μm + blur*
Target/Filer	Mo/Mo	Mo/Mo, Rh/Rh	W/Rh	(Mo, W)/(Rh, Al)
Angle Coverage	48 degrees	50 degrees	50 degrees	30 degrees
View Numbers	25	11	25/49	11
Gantry Motion	Stationary	Step and Shoot	Continuous Rotation	Continuous Rotation
Flat-Panel Detector	A-silicon	Cs:I a-silicon	Direct converter a-selenium	Direct converter a-selenium
Detector Size	19.5 x 24.4 cm pixel pitch: 127 μm	18.00 x 23.04 pixel pitch: 100 μm	23.9 x 30.5 cm pixel pitch: 85 μm	24 x 29 cm pixel pitch: 70 μm (140 μm for DBT)
Readout Time	0.128 s / 0.032 s	0.3 s	0.6 s / 0.3 s	0.6 s
Integration Time	0.32 s	0.4 s	0.2 s	1.0 s
Exposure Time	0.32 s	~0.1 s	~0.03 s	0.073 s
Total Scan Time**	11.2 s for 25 views	7 s for 11 views	20 s / 39.2 s for 25 / 49 views	18 s for 11 views

Reconstruction Technique	Ordered subsets convex (maximum likelihood)	ML-EM	FBP: filtered back projection	FBP: filtered back projection

*: Additional focal spot blur due to the gantry movement during exposure
**: Total scan time = (view number) x (cycle time); cycle time = (readout time) + (integration time)

Table 1: System Comparison

Advantages of a system in accordance with the subject matter disclosed herein over commercially available systems include: (1) the total spot size of system **900** is 200 μm while the values of other systems are 300 μm or larger; (2) the stationary design provides less gantry vibration by eliminating the mechanical movement; and (3) the exposure time matches the detector integration window. The targeted total scan time (8.8 s in binning mode and 11.2 s in full-resolution mode, 25 viewing angles) is shorter, which can be further reduced by increasing the x-ray tube current which requires relaxing of the focal spot size.

The energy spectrum of the x-ray source of system **900** was measured at 28 keV using a Si-pin photodiode detector. The energy filters used can be selected such that the x-ray radiation from each of the x-ray focal spots have the same energy spectrum. The spectrum is consistent when measured at different locations within the field of view and from different x-ray source pixels. The experimentally measured energy spectrum of system **900** (Figure 9) at 28 keV is shown in the graph of Figure 14. The results shown in Figure 14 agree well with a typically Mo/Mo x-ray spectrum. Two molybdenum characteristic peaks, one at 17.5 keV and the other at 19.6 keV, can be recognized in the graph. Alternatively, the energy filters can be varied such that energy spectra from the x-ray focal spots can be individually controlled.

Figure 15 is a graph of anode current as a function of gate voltage for system **900** shown in Figure 9. The threshold value for this x-ray source is about 650 V. The emission current from the carbon nanotube cathode depends on the electrical field between the gate and the cathode following the Fowler-Nordheim equation. In this particular x-ray source, 72% of the total current passes through the gate electrode and reaches the anode to produce x-ray radiation (also denoted as anode current). The graph of Figure 15 shows the typical anode current versus gate voltage data measured from one pixel. Due to the voltage limitation of the electronic control devices, the maximum gate voltage that can be applied in this experiment is 1500 V, which limits the anode current to ~4 mA, below the targeted value of 10 mA. This limitation can be overcome by changing the design and/or optimizing the carbon nanotube cathode (when measured in a separate setup, the cathodes fabricated under the same conditions can produce over 10 mA consistently at higher gate voltages).

In this experiment, nine of the pixels of system **900** shown in Figure 9 have been characterized. Due to the variation in the carbon nanotube cathodes, the gate voltages required to obtain the same current are different. As a reference, the voltages vary from 925 V to 1465 V for 1 mA tube current. Table 2 shows the gate voltage and standard deviation of current for the nine pixels as obtained by this experiment.

X-Ray Source Number	Gate Voltage (V)	Standard Deviation of Current (mA)
1	1230	0.02
2	925	0.01
3	1230	0.02
4	1015	0.01
5	1300	0.03
6	1070	0.01
7	1160	0.01
8	1465	0.02
9	1030	0.01

Table 2: Gate Voltage and Standard Deviation of Pixels

The voltage difference can be compensated by the variable resistors in the controller. With the improvement of fabrication technique and cathode quality control, the variation can be reduced. The current stability was determined by measuring the current of 100 pulsed x-rays at constant voltage. The standard deviation of the current is less than 0.03 mA for all pixels tested.

In this experiment, the designed x-ray focal spot size is about 200 x 200 μm for all 25 x-ray sources. The actual values were measured following the European standard EN12543-5. A customized cross wire phantom made of 1 mm tungsten wire was fabricated to measure the focal spot size along two orthogonal directions simultaneously. The phantom was placed close to the x-ray source to obtain the large magnification factor. The voltages applied to the two focusing electrodes were first varied to optimize the focal spot size. It was found that the optimal focal spot size is achieved when the two focusing electrodes are at 500 V and 1600 V, respectively. A typical projection image of the cross phantom is shown in Figure 16. Figure 17 is a graph showing the line profiles of the two wires, where the X axis is the direction of the x-ray source array, and the Y axis is perpendicular to the array.

Table 3 shows the focal spot size measurement of the nine x-ray source pixels.

X-Ray Source Number	F_x: Parallel to X-Ray Source Array	F_y: Perpendicular to X-Ray Source Array
1	0.20 mm	0.20 mm
2	0.20 mm	0.17 mm
3	0.18 mm	0.19 mm
4	0.19 mm	0.19 mm
5	0.20 mm	0.19 mm
6	0.19 mm	0.17 mm
7	0.18 mm	0.17 mm
8	0.19 mm	0.19 mm
9	0.18 mm	0.19 mm

Table 3: Focal Spot Size Measurement of Pixels

The results shown in Table 3 agree well with the designed specification of 0.20 x 0.20 mm. The x-ray sources have an isotropic focal spot with an average value of 0.19 mm. Measurements from different x-ray sources are also consistent.

Tomosynthesis reconstruction requires precise system geometry parameters. An analytic method was applied based on identification of ellipse parameters for the geometry calibration, which was first established for cone-beam CT calibration. A phantom with two point objects with known distance was machined. The geometry parameters of the 25 x-ray sources were individually calibrated. Six projection images of the phantom (60-degree rotation in-between) were acquired for each pixel. The traces of the two balls form two ellipses on the detector plane. The parameters, including the source-to-detector distance and x-ray source offset values on the detector plane, can be further calculated based on these elliptical curves. The source-to-detector distance is calculated to be 69.3 cm with 2 mm uncertainty. The distances between the x-ray sources are also calculated. The results agree with the design values within 1 mm uncertainty.

In one embodiment, an anti-scattering component can be positioned between the x-ray detector and the location for positioning the object. In particular, one-dimensional or two-dimensional anti-scattering grids can be made to utilize the advantage of linear MBFEX. For instance, in the case of a two-dimensional grid, the anti-scattering component can be adjusted based on a position of one or more of the x-ray sources being activated. Alternatively, in the case of a one-dimensional anti-scattering grid, the grid lines can run in either parallel or perpendicular to the linear/arc direction of the MBFEX. The grid geometry can be tailored to enable fan-beam reconstruction to enhance the tomographic image quality and increase the reconstruction speed.

Using such an anti-scattering component, a cone beam x-ray source can be used to produce fan-beam reconstructed tomography images of an object. For example, referring again to Figure 4A, a plurality of stationary field emission x-ray sources **XS** can be provided. X-ray sources **XS** can be spatially distributed in a substantially linear array (e.g., x-ray generating device **XGD**). The object **O** can then be irradiated with x-ray cone beams produced by x-ray sources **XS** to generate two-dimensional projection images of object **O**. A linear anti-scattering grid **AS** can be placed between object **O** and detector **XD** to reduce scatter of the x-ray cone beams. Two-dimensional projection images of object **O** can be detected, and the two-dimensional projection images can be divided into groups of one-dimensional data. These groups of one-dimensional data can be collected from all of the different x-ray sources **XS**, and a fan beam reconstruction can be used to reconstruct slice images of object **O** from the groups of one-dimensional data. The slice images can be merged together to form a three-dimensional image of object **O**.

It will be understood that various details of the presently disclosed subject matter may be changed without departing from the scope of the presently disclosed subject matter. Furthermore, the foregoing description is for the purpose of illustration only, and not for the purpose of limitation.

CLAIMS

What is claimed is:

1. A stationary x-ray digital breast tomosynthesis system comprising:

an x-ray source that generates x-ray radiation from an array of spatially distributed x-ray focal spots configured to image a human breast from different viewing angles for reconstruction without moving any of the source, the object, or the detector;

an area x-ray detector configured to detect the projection images of the breast;

an electronic controller for activating the x-ray radiation from the different x-ray focal spots in the x-ray source in a sequence and for synchronizing x-ray exposure from a given focal spot with image collection by the x-ray detector; and

wherein tomography images of the breast can be reconstructed using a plurality of projection images of the breast collected from different viewing angles.

2. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source includes a plurality of focal spots arranged in a substantially straight line parallel to an imaging plane of the x-ray detector.

3. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source includes a plurality of focal spots arranged substantially along an arc, wherein the focal spots define a plane that is substantially perpendicular to an imaging plane of the x-ray detector, wherein all the x-ray focal spots are equal distance to an iso-center.

4. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source includes a plurality of focal spots arranged in a two-dimensional matrix on an x-ray anode.

5. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source includes focal spots of substantially the same size.

6. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source includes focal spots having sizes ranging between about 0.05 mm and about 2 mm.

7. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source is configured to generate cone-shaped x-ray beams, and wherein central axes of the x-ray beams are substantially directed towards an iso-center.

8. The x-ray digital breast tomosynthesis system of claim 1, further comprising an electronic circuit to individually control the x-ray intensities from the different x-ray focal spots such that they can either be the same or be modulated to deliver a desired intensity or intensity distribution on the object to be imaged.

9. The x-ray digital breast tomosynthesis system of claim 1, wherein the array of spatially distributed x-ray focal spots in the x-ray source comprises between about 10 and 100 x-ray focal spots covering viewing angles of between 10° and 100° degrees.

10. The x-ray digital breast tomosynthesis system of claim 1, comprising a plurality of individually controllable electron field emission cathodes, one or more x-ray anodes, and an electron focusing lens that focuses the electron beam from a cathode to a desired area on an x-ray anode.

11. The x-ray digital breast tomosynthesis system of claim 1, wherein the electron focusing lens is a modified Einzel type lens with a plurality of electrostatic focusing electrodes.

12. The x-ray digital breast tomosynthesis system of claim 10, wherein the electron field emission cathodes include at least one of the following: nanowires, nanotubes, and carbon nanotubes.

13. The x-ray digital breast tomosynthesis system of claim 1, comprising a controller configured to adjust focal spot sizes of different x-ray beams.

14. The x-ray digital breast tomosynthesis system of claim 1, comprising a controller including a field-effect-transistor based electronic circuit configured to activate the x-ray sources.

15. The x-ray digital breast tomosynthesis system of claim 1, comprising an anti-scattering component positioned between the x-ray detector and the location for positioning the object.

16. The x-ray digital breast tomosynthesis system of claim 15, wherein the anti-scattering component is adjustable based on a position of one or more of the x-ray sources being activated.

17. The x-ray digital breast tomosynthesis system of claim 1, wherein the x-ray source comprises x-ray anodes configured at different voltages to produce x-ray radiation with two energies.

18. The x-ray digital breast tomosynthesis system of claim 17, wherein the x-ray source comprises 12 anodes configured at low voltage and 13 anodes configured at high voltage to enable the system for dual-energy imaging.

19. A stationary x-ray digital breast tomosynthesis system configured to image a human breast from different viewing angles for reconstruction without moving any of the source, the object, or the detector, the system comprising:

an x-ray source that generates x-ray radiation from an array of x-ray focal spots that are spatially distributed in a substantially straight line parallel to the imaging plane of the x-ray detector;

an area x-ray detector configured to detect the projection images of the breast;

an electronic controller for sequentially activating the x-ray radiation from the different x-ray focal spots and varying the intensity of the x-ray radiation based on the distance between the x-ray focal spot and the object to be imaged such that the x-ray dose delivered to the object from every viewing angle is substantially the same; and

wherein tomography images of the object can be reconstructed using a plurality of projection images of the object collected from different viewing angles.

20. The x-ray digital breast tomosynthesis system of claim 19, wherein the x-ray source is configured to generate cone-shaped x-ray beams, and wherein a central axis of each of the x-ray beams is substantially directed towards an iso-center.

21. The x-ray digital breast tomosynthesis system of claim 19, wherein adjacent focal spots of the x-ray source have substantially the same angular spacing with respect to an iso-center.

22. The x-ray digital breast tomosynthesis system of claim 19, wherein the adjacent focal spots of the x-ray source have substantially the same linear spacing.

23. A stationary x-ray digital breast tomosynthesis system configured to image a human breast from different viewing angles for reconstruction without moving any of the source, the object, or the detector, the system comprising:

an x-ray source that generates x-ray radiation from an array of x-ray focal spots that are spatially distributed along an arc substantially perpendicular to an imaging plane of the x-ray detector, wherein the focal spots are equal distance to the object to be imaged;

an area x-ray detector configured to detect the projection images of the breast;

an electronic controller for sequentially activating the x-ray beam from the different x-ray focal spots and delivering the same x-ray tube current to each focal spot; and

wherein tomography images of the object can be reconstructed using a plurality of projection images of the object collected from different viewing angles.

24. A quasi-monochromatic x-ray digital breast tomosynthesis system configured to image a human breast, the system comprising:

an x-ray source that generates scanning cone-beam x-ray radiation from an array of spatially distributed x-ray focal spots;

an energy filter;

an x-ray detector configured to collect projection images of an object;

an electronic control circuit that allows imaging of the object by simultaneously activating a plurality of x-ray beams at a given time based on a multiplexing imaging scheme, and synchronizes x-ray exposure with data collection by the x-ray detector; and

a computer program to de-multiplex the projection images;

wherein tomography images of the object can be reconstructed using a plurality of projection images of the object collected from different viewing angles.

25. The x-ray digital breast tomosynthesis system of claim 24, wherein the energy filters are the same such that the x-ray radiations from each focal spots have substantially the same energy spectrum.

26. The device of claim 24, wherein energy filters are varied such that energy spectra from the x-ray focal spots can be individually controlled.

27. The x-ray digital breast tomosynthesis system of claim 24, wherein the projection images are collected by the binary multiplexing scheme.

28. The x-ray digital breast tomosynthesis system of claim 24, wherein the projection images are collected by the frequency division multiplexing scheme.

29. The x-ray digital breast tomosynthesis system of claim 24, wherein the energy filter comprises Cerium and wherein the x-ray source comprises an x-ray anode that operates on a voltage in the range of 60-80kV.

30. The x-ray digital breast tomosynthesis system of claim 24, wherein the projection images from the different viewing angles are generated by electronically switching the x-ray beams from different focal spots without moving any of the x-ray source, the detector, or the patient.

31. A x-ray digital tomosynthesis system comprising:

a field emission x-ray source that generates a scanning x-ray beam from an array of spatially distributed x-ray focal spots configured to image an object from different viewing angles for tomosynthesis reconstruction;

an area x-ray detector configured to detect projection images of the object;

an electronic controller for activating the x-ray beam from different x-ray focal spots in a sequence, either one beam or a plurality of the beams simultaneously, and for synchronizing x-ray exposure with image collection by the x-ray detector; and

wherein tomography images of the object can be reconstructed using a plurality of projection images of the object from different viewing angles.

32. A multiplexing x-ray digital tomosynthesis system comprising:

a field emission x-ray source for generating a scanning x-ray beam from an array of spatially distributed x-ray focal spots configured to image an object from different viewing angles for tomosynthesis reconstruction;

an area x-ray detector configured to detect the projection images;

an electronic controller for imaging of the object by simultaneously activating a plurality of the x-ray beams at a given time based on a multiplexing imaging scheme, and for synchronizing x-ray exposure with data collection by the x-ray detector;

wherein the projection images can be de-multiplexed; and

wherein tomography images of the object can be reconstructed using a plurality of projection images of the object from different viewing angles.

33. The x-ray digital tomosynthesis system of claim 32, further comprising a controller for varying the x-ray intensity from each focal spot by controlling the field emission current from a field emission cathode and adjusting the focal spot size by varying the voltage applied to an electron focusing lenses.

34. A method of producing tomography images of an object, the method comprising:

providing a plurality of stationary field emission x-ray sources spatially distributed with respect to an object to be imaged;

irradiating the object with x-ray beams produced by the x-ray sources to generate projection images of the object;

detecting the projection images of the object; and

reconstructing tomography images of the object based on the projection images of the object.

35. The method of claim 34, wherein the x-ray sources include a plurality of focal spots arranged in a substantially straight line parallel to an imaging plane of the x-ray detector.

36. The method of claim 34, wherein the x-ray sources include a plurality of focal spots arranged substantially along an arc, wherein the focal spots define a plane that is substantially perpendicular to an imaging plane of the x-ray detector.

37. The method of claim 34, wherein the x-ray sources include a plurality of focal spots arranged in a two-dimensional matrix on an anode.

38. A method of producing images of an object using at least one of monochromatic and quasi-monochromatic x-ray beams, the method comprising:

providing a plurality of stationary field emission x-ray sources spatially distributed with respect to an object to be imaged;

irradiating the object with at least one of monochromatic and quasi-monochromatic x-ray beams produced by the x-ray sources to generate projection images of the object;

detecting the projection images of the object; and

reconstructing displayable images of the object based on the projection images of the object.

39. The method of claim 38, wherein the x-ray sources include a plurality of focal spots arranged in a substantially straight line parallel to an imaging plane of the x-ray detector.

40. The method of claim 38, wherein the x-ray sources include a plurality of focal spots arranged substantially along an arc, wherein the focal spots define a plane that is substantially perpendicular to an imaging plane of the x-ray detector.

41. The method of claim 38, wherein the x-ray sources include a plurality of focal spots arranged in a two-dimensional matrix on an anode.

42. The method of claim 38, wherein the x-ray sources include a plurality of individually controllable electron field emission cathodes and one or more x-ray anodes.

43. The method of claim 42, wherein the electron field emission cathodes include at least one of the following: nanowires, nanotubes, and carbon nanotubes.

44. A method of producing fan-beam reconstructed tomography images of an object using a cone beam x-ray source, the method comprising:

providing a plurality of stationary field emission x-ray sources spatially distributed in a substantially linear array;

irradiating the object with x-ray cone beams produced by the x-ray sources to generate two-dimensional projection images of the object;

placing a linear anti-scattering grid between the object and the detector to reduce scatter of the x-ray cone beams;

detecting the two-dimensional projection images of the object;

dividing the two-dimensional projection images into groups of one-dimensional data;

reconstructing slice images of the object from the groups of one-dimensional data; and

merging the slice images of the object to form a three-dimensional image of the object.

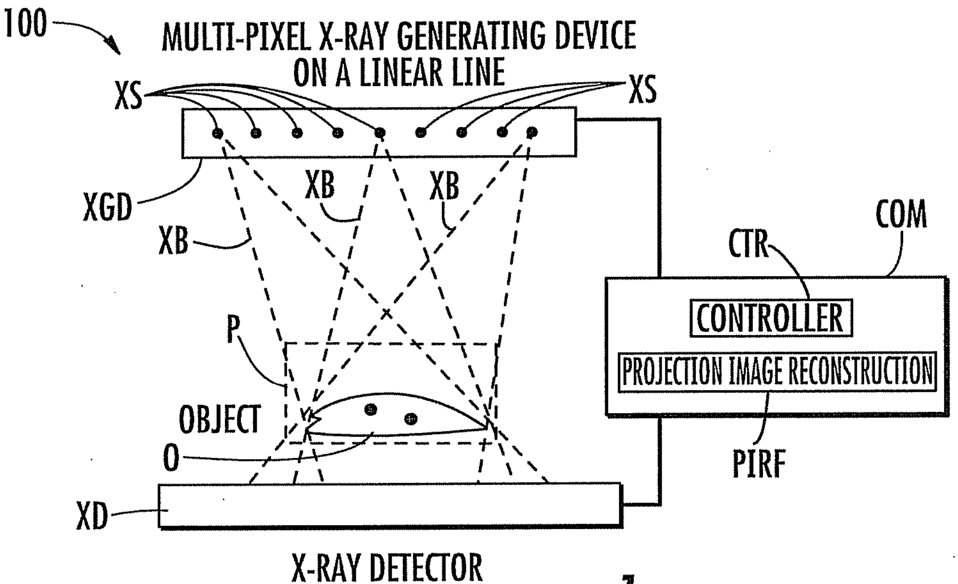
100
MULTI-PIXEL X-RAY GENERATING DEVICE ON A LINEAR LINE
XS
XS
XGD
XB
XB
XB
COM
CTR
CONTROLLER
PROJECTION IMAGE RECONSTRUCTION
PIRF
P
OBJECT
O
XD
X-RAY DETECTOR

FIG. 1

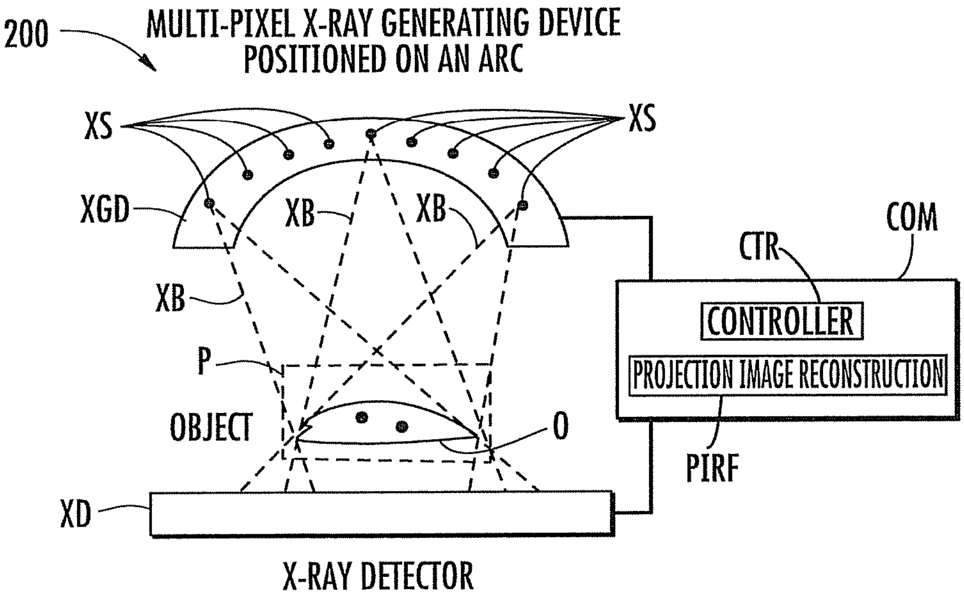
200
MULTI-PIXEL X-RAY GENERATING DEVICE POSITIONED ON AN ARC
XS
XS
XGD
XB
XB
XB
COM
CTR
CONTROLLER
PROJECTION IMAGE RECONSTRUCTION
PIRF
P
OBJECT
O
XD
X-RAY DETECTOR

FIG. 2

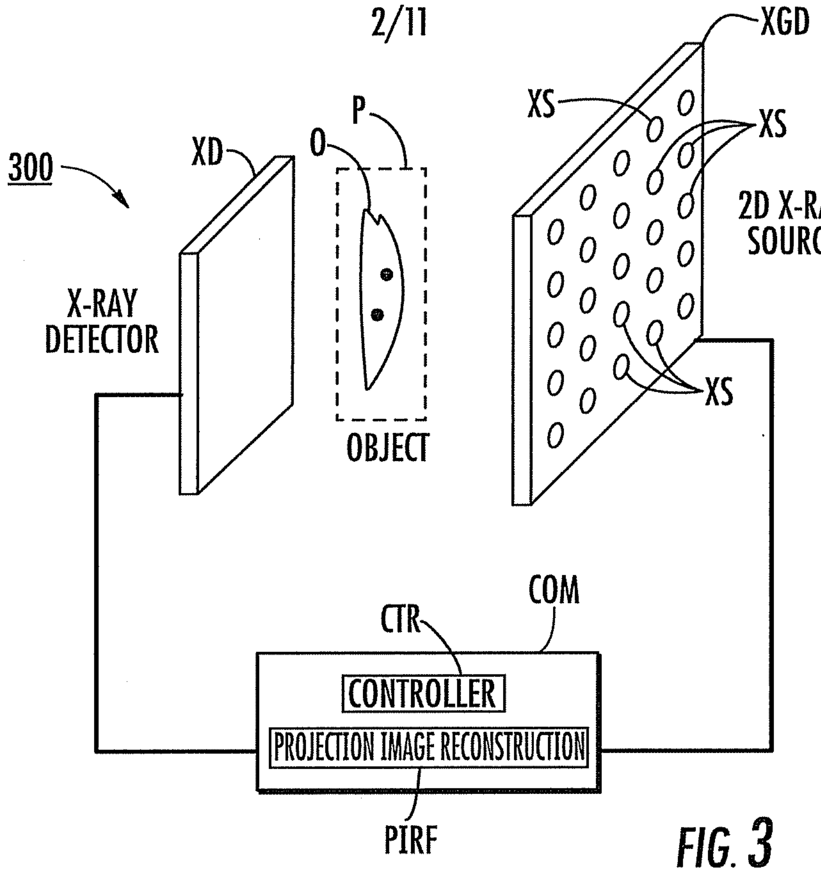
300
XD
O
P
XS
XGD
XS
2D X-RAY
SOURCE
X-RAY
DETECTOR
OBJECT
XS
CTR
COM
CONTROLLER
PROJECTION IMAGE RECONSTRUCTION
PIRF

FIG. 3

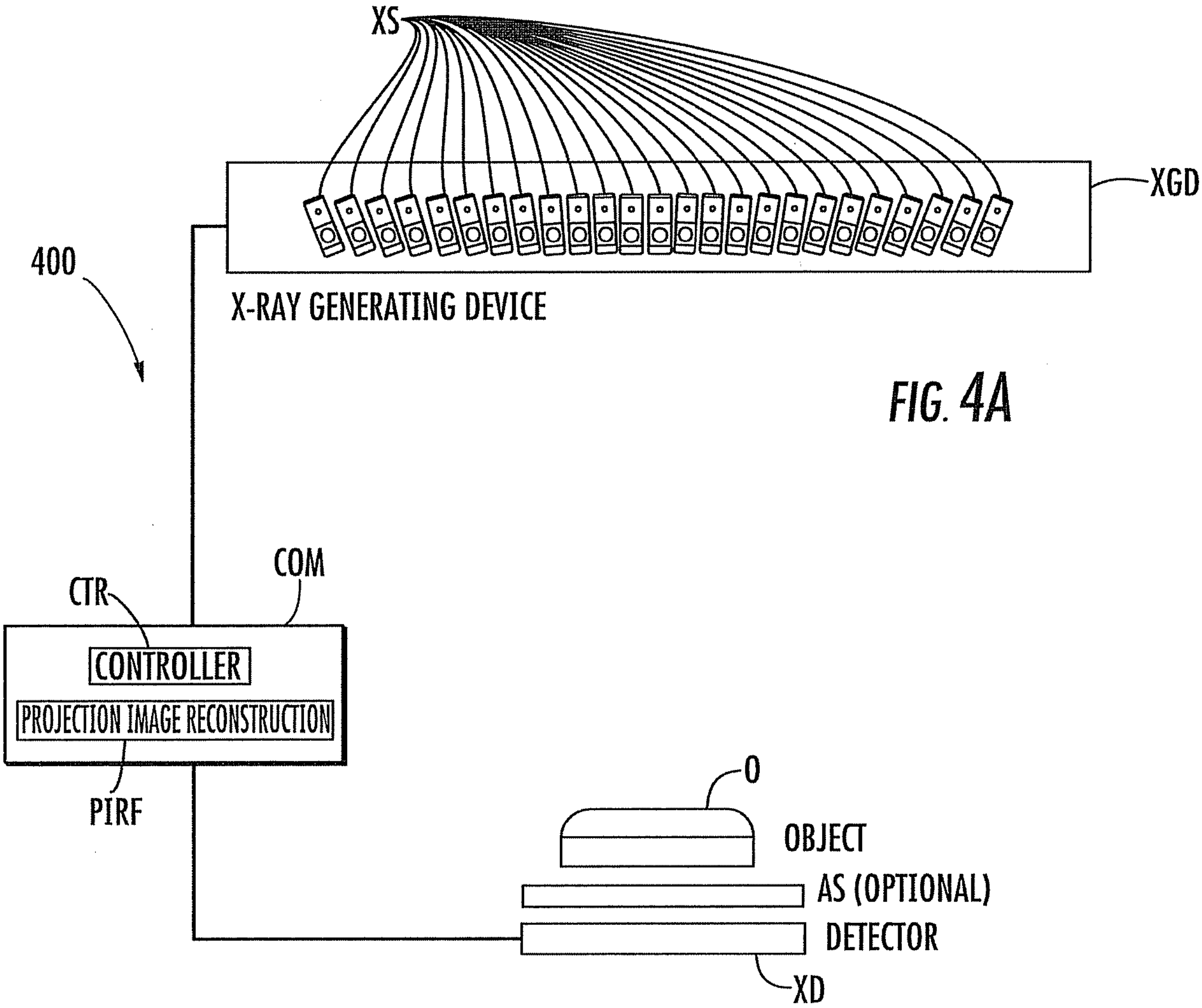
XS
XGD
400
X-RAY GENERATING DEVICE
CTR
COM
CONTROLLER
PROJECTION IMAGE RECONSTRUCTION
PIRF
O
OBJECT
AS (OPTIONAL)
DETECTOR
XD

FIG. 4A

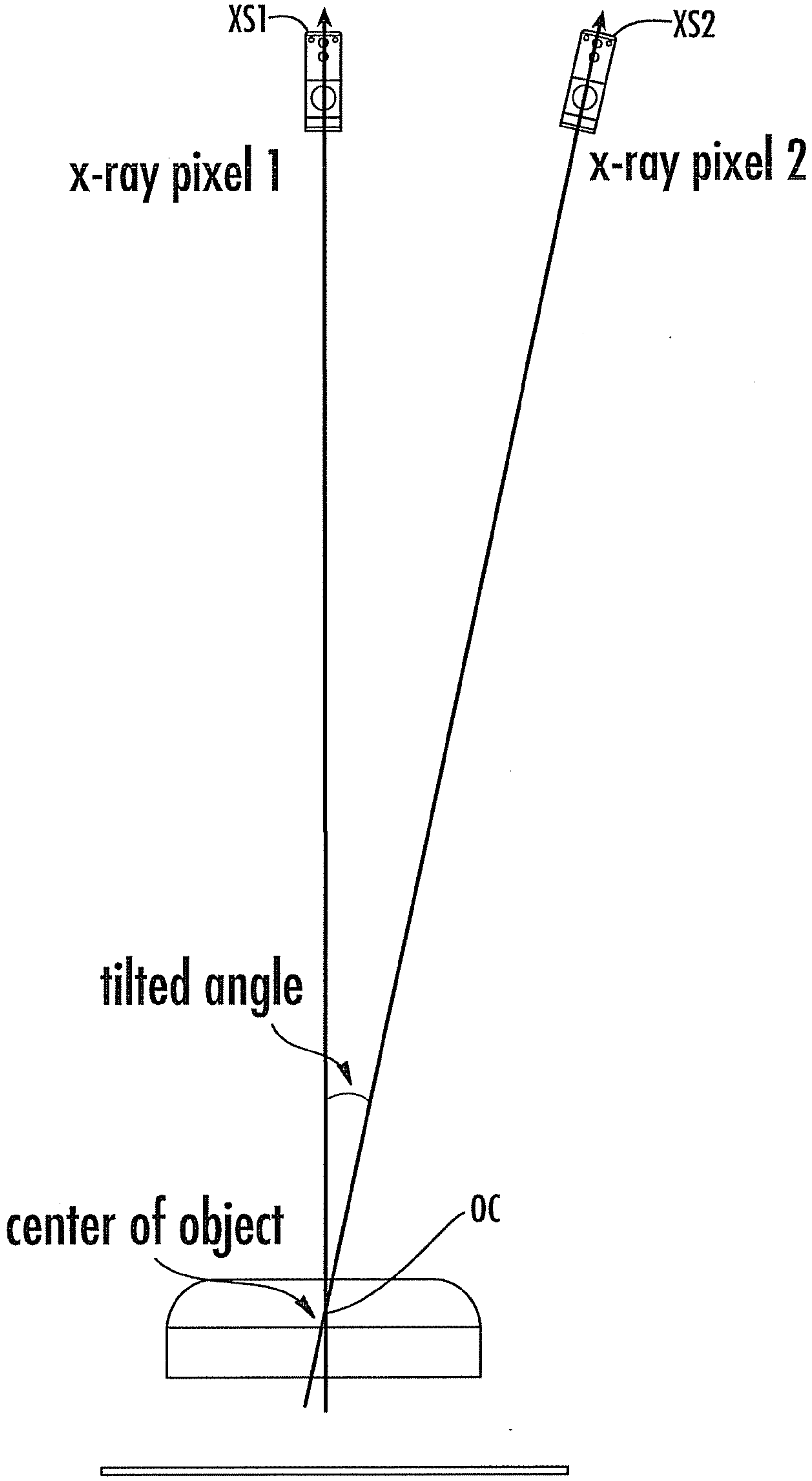
XS1
XS2
x-ray pixel 1
x-ray pixel 2
tilted angle
center of object
OC

FIG. 4B

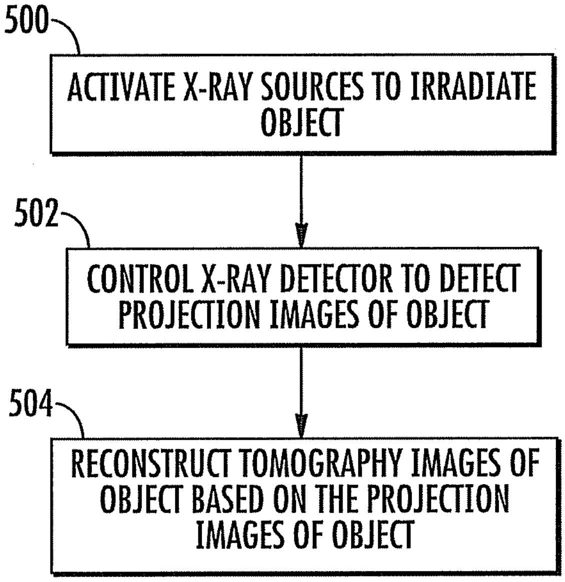
500
ACTIVATE X-RAY SOURCES TO IRRADIATE OBJECT
502
CONTROL X-RAY DETECTOR TO DETECT PROJECTION IMAGES OF OBJECT
504
RECONSTRUCT TOMOGRAPHY IMAGES OF OBJECT BASED ON THE PROJECTION IMAGES OF OBJECT

FIG. 5

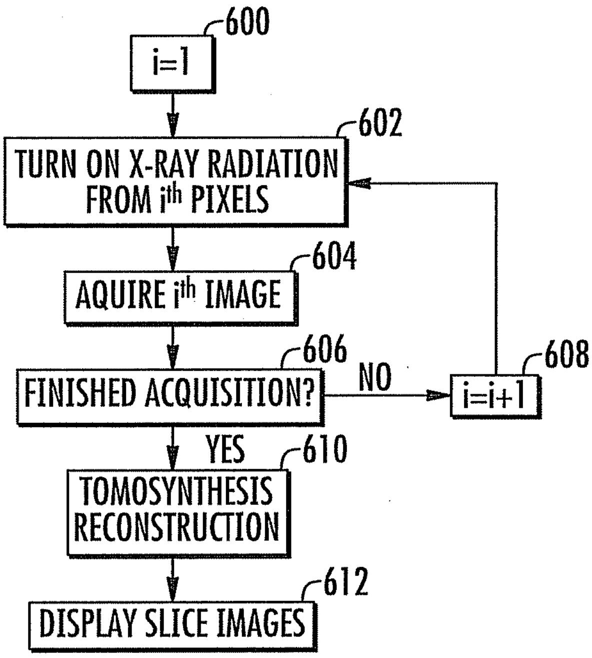
600
i=1
602
TURN ON X-RAY RADIATION FROM i^th PIXELS
604
AQUIRE i^th IMAGE
606
FINISHED ACQUISITION?
NO
608
i=i+1
YES
610
TOMOSYNTHESIS RECONSTRUCTION
612
DISPLAY SLICE IMAGES

FIG. 6

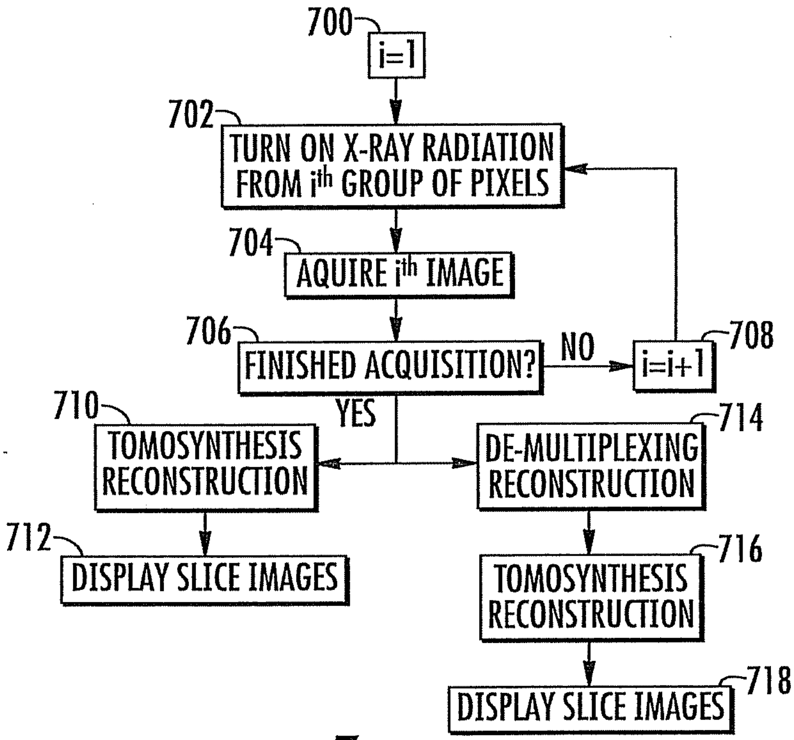
700
i=1
702
TURN ON X-RAY RADIATION FROM i^th GROUP OF PIXELS
704
AQUIRE i^th IMAGE
706
FINISHED ACQUISITION?
NO
i=i+1
708
YES
710
TOMOSYNTHESIS RECONSTRUCTION
712
DISPLAY SLICE IMAGES
714
DE-MULTIPLEXING RECONSTRUCTION
716
TOMOSYNTHESIS RECONSTRUCTION
718
DISPLAY SLICE IMAGES

FIG. 7

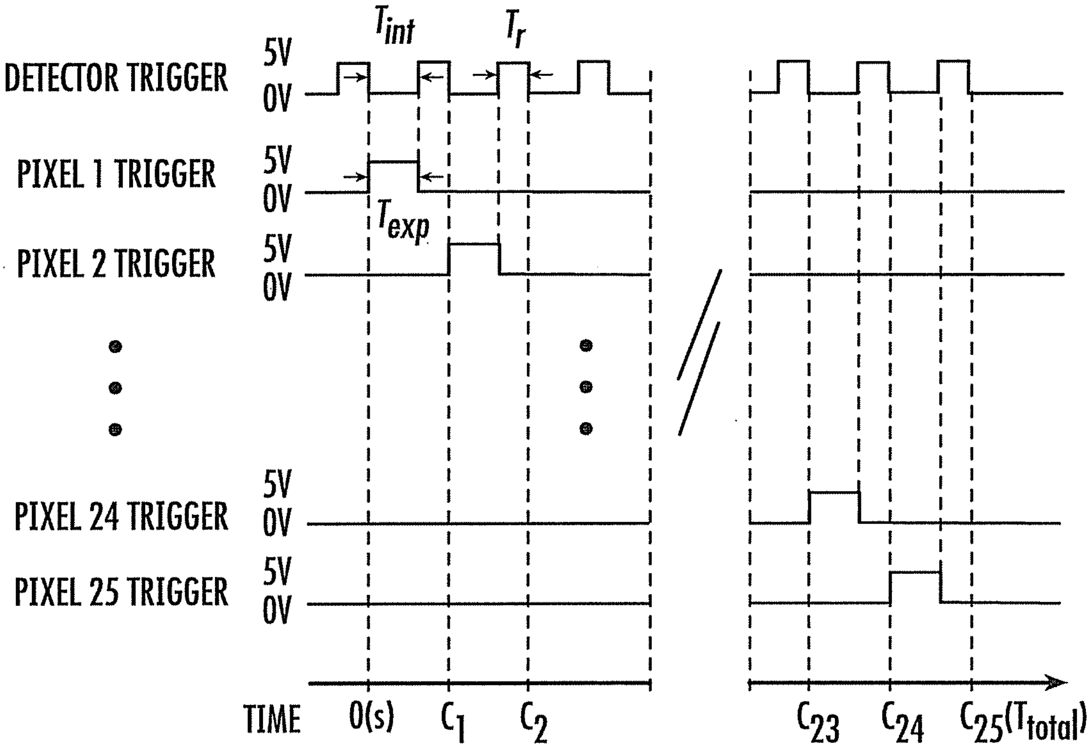
T_{int}
T_r
DETECTOR TRIGGER
5V
0V
PIXEL 1 TRIGGER
5V
0V
T_{exp}
PIXEL 2 TRIGGER
5V
0V
PIXEL 24 TRIGGER
5V
0V
PIXEL 25 TRIGGER
5V
0V
TIME
0(s)
C_1
C_2
C_{23}
C_{24}
$C_{25}(T_{total})$

FIG. 8

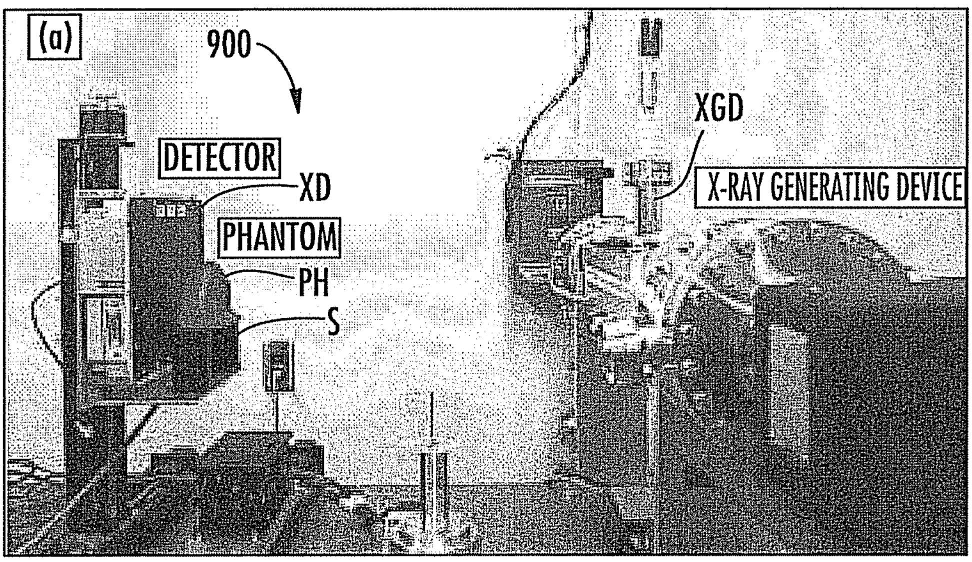
(a)
900
DETECTOR
XD
PHANTOM
PH
S
XGD
X-RAY GENERATING DEVICE

FIG. 9

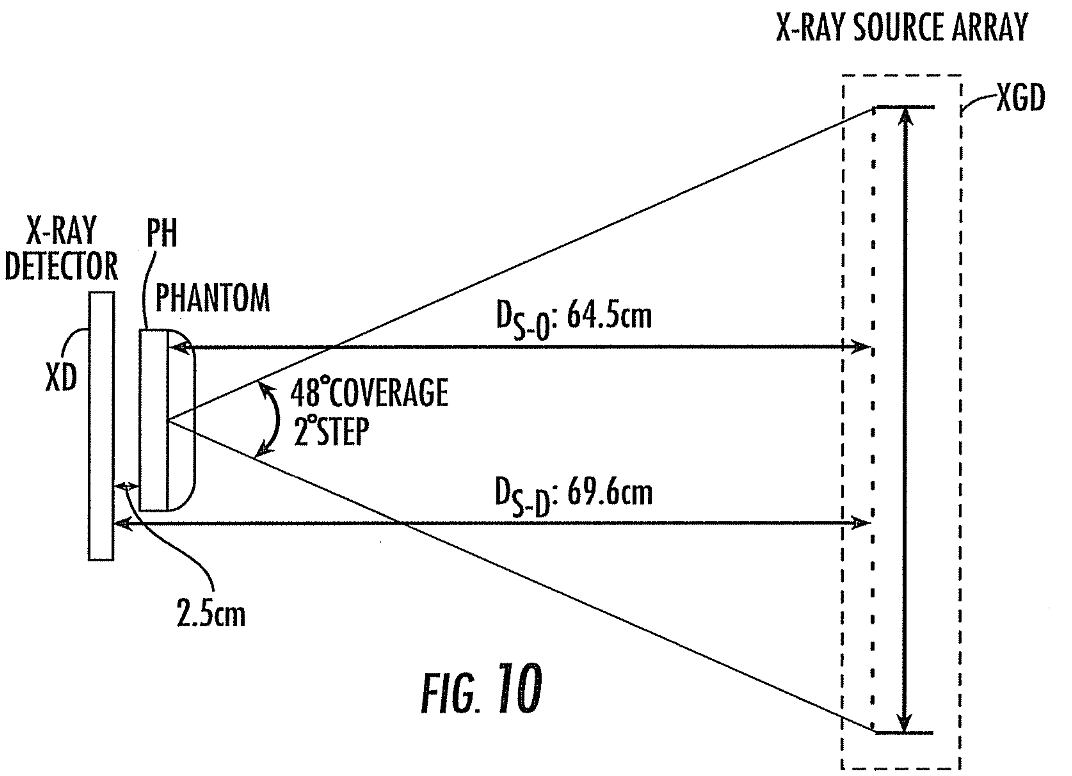

FIG. 10

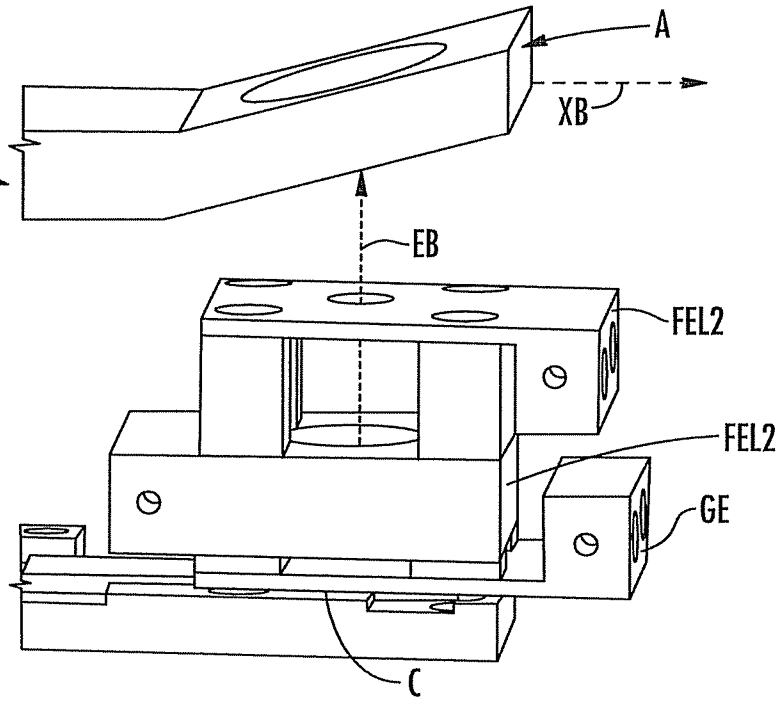

FIG. 11A

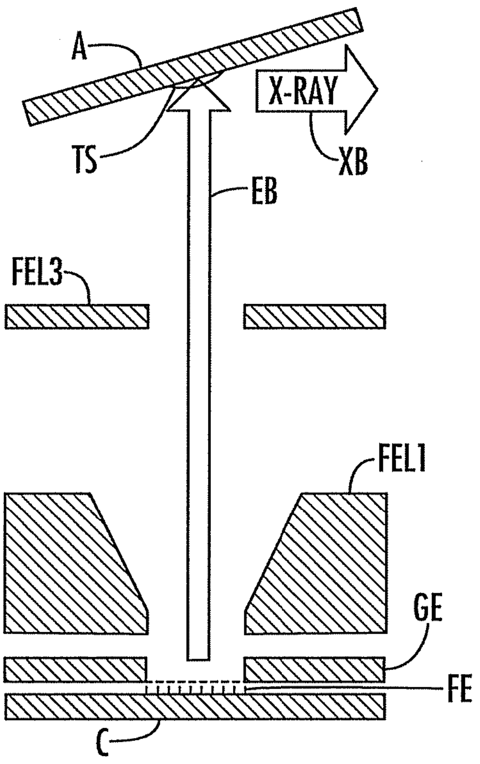

FIG. 11B

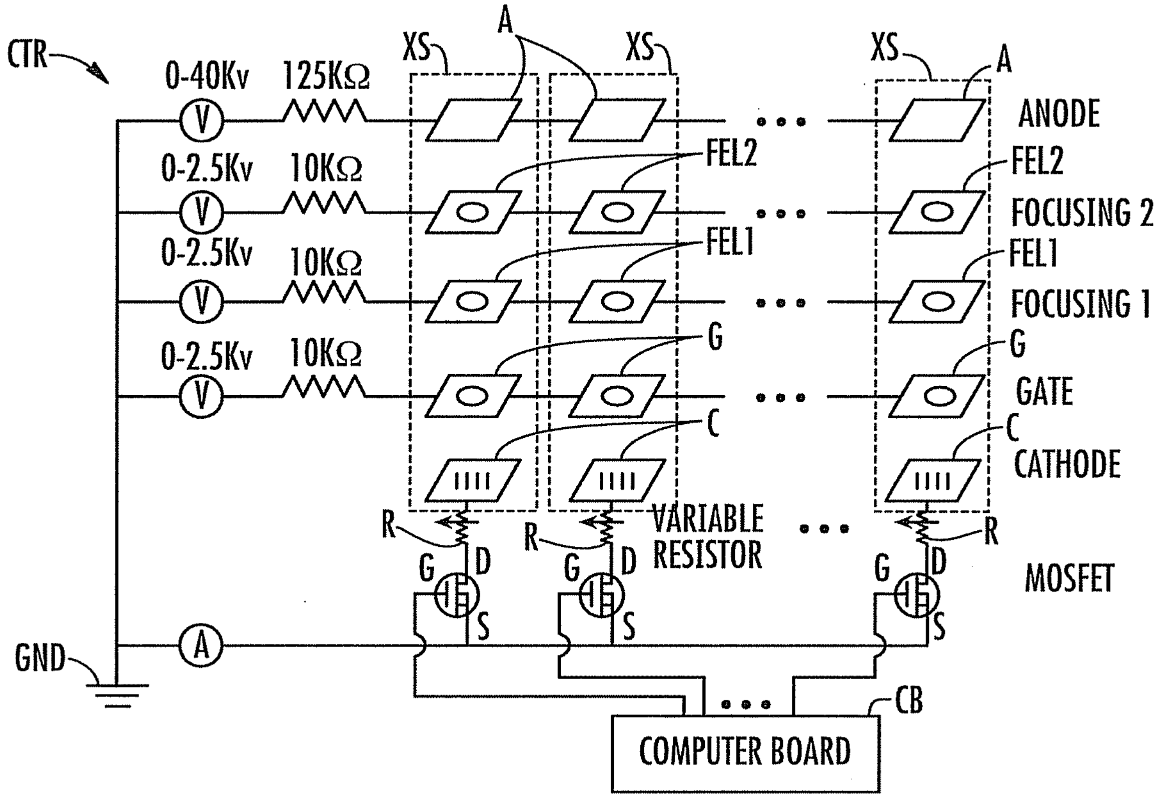

FIG. 12

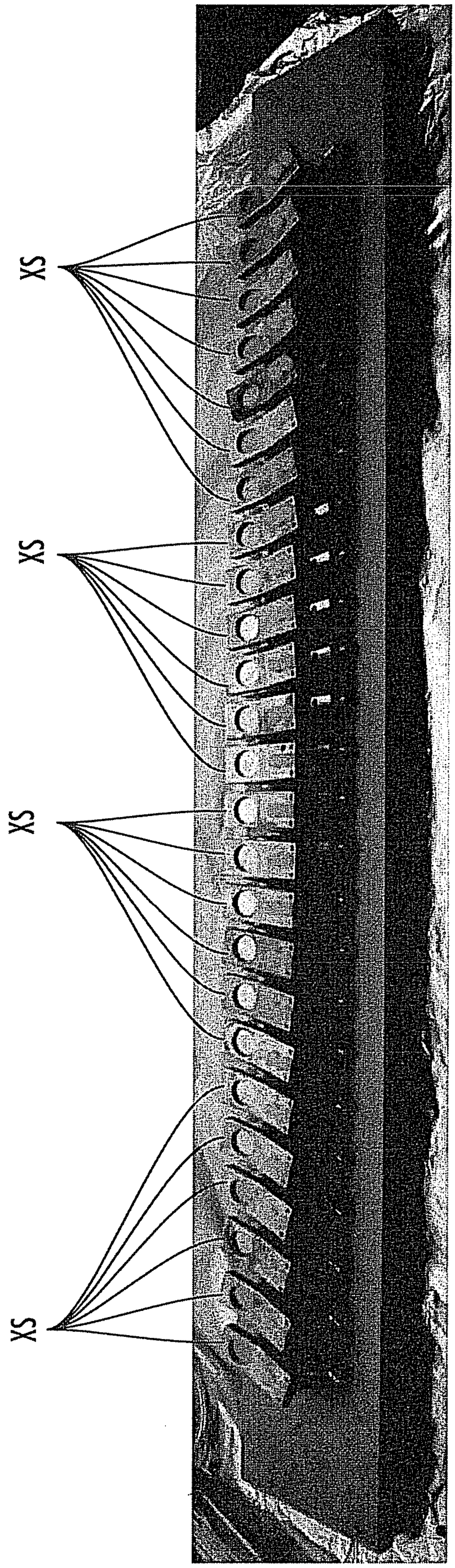

FIG. 13

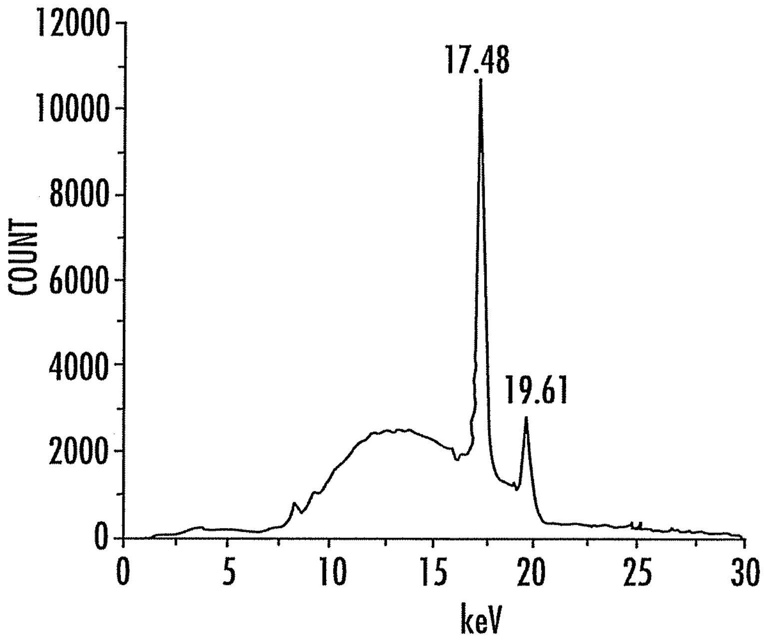
12000
10000
8000
6000
4000
2000
0
COUNT
17.48
19.61
0
5
10
15
20
25
30
keV

FIG. 14

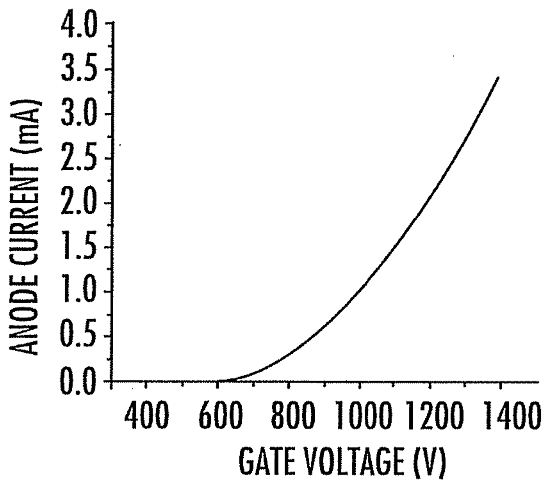
4.0
3.5
3.0
2.5
2.0
1.5
1.0
0.5
0.0
ANODE CURRENT (mA)
400
600
800
1000
1200
1400
GATE VOLTAGE (V)

FIG. 15

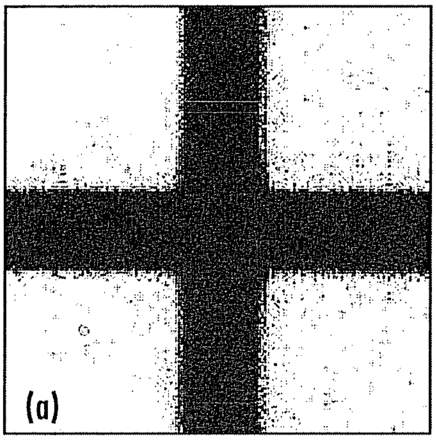
(a)

FIG. 16

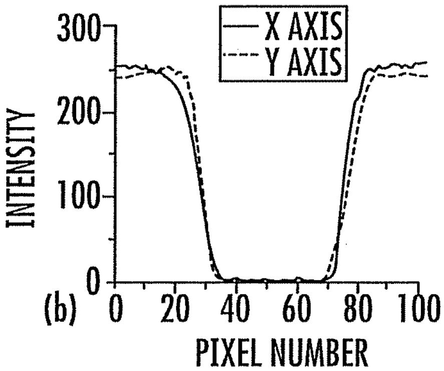
X AXIS
Y AXIS
300
200
100
0
INTENSITY
(b)
0
20
40
60
80
100
PIXEL NUMBER

FIG. 17

INTERNATIONAL SEARCH REPORT

International application No.
PCT/US 08/70477

A. CLASSIFICATION OF SUBJECT MATTER
IPC(8) - A61B 6/04 (2008.04)
USPC - 378/37
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)
USPC: 378/37

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched
USPC: 378/4, 21, 23, 24; 382/128, 154; 345/419

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)
PubWEST(USPT,PGPB,EPAB,JPAB); DialogPRO(Engineering); Google Scholar
Search Terms Used: x-ray, tomosynthesis, breast, focal, spot, array, spatial, area, detector, viewing, angle, controller, straight, line, matrix, anode, Einzel, lens, electrostat, FET, multiplexing, anti-scatter, frequency, division, Cerium, KV, 12, 13

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X -- Y	US 2007/0009081 A1 (Zhou et al.) 11 January 2007 (11.01.2007), entire document especially abstract; Figs. 4-6; para [0015], [0020], [0030], [0032], [0035], [0041-[0042], [0046]-[0057], [0063], [0070], [0075]	1-10, 12, 23, 31, 32, 34-43 -------------------- 11, 13-22, 24-30, 33, 44
Y	US 2005/0285541 A1 (LeChevalier) 29 December 2005 (29.12.2005), para [0033], [0225], [0226], [0296], [0305], [0306], [0324], [0551], [0711], [0750]-[0751]	11, 13, 14, 17-22, 24-30, 33
Y	US 3,617,285 A (Staudenmayer) 02 November 1971 (02.11.1971), col 1, ln 33-36; col 4, ln 9-11, ln 58-60; col 6, ln 5-7; col 8, ln 17-18	15, 16, 18, 29, 44

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

* 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 application or patent 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	Date of mailing of the international search report
22 September 2008 (22.09.2008)	01 OCT 2008

Name and mailing address of the ISA/US	Authorized officer:
Mail Stop PCT, Attn: ISA/US, Commissioner for Patents P.O. Box 1450, Alexandria, Virginia 22313-1450 Facsimile No. 571-273-3201	Lee W. Young PCT Helpdesk: 571-272-4300 PCT OSP: 571-272-7774