(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
19 February 2004 (19.02.2004)

PCT

(10) International Publication Number
WO 2004/015632 A1

(51) International Patent Classification[7]: **G06T 11/00**, A61B 6/03

(21) International Application Number: PCT/IB2003/003105

(22) International Filing Date: 11 July 2003 (11.07.2003)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
10/213,467 6 August 2002 (06.08.2002) US

(71) Applicant *(for all designated States except US)*: **KONINKLIJKE PHILIPS ELECTRONICS N.V.** [NL/NL]; Groenewoudseweg 1, NL-5621 BA Eindhoven (NL).

(71) Applicant *(for AE only)*: **U.S. PHILIPS CORPORATION** [US/US]; 1251 Avenue of the Americas, New York, NY 10510-8001 (US).

(72) Inventor; and
(75) Inventor/Applicant *(for US only)*: **VAN DE HAAR, Peter, G.** [NL/NL]; P.O. Box 220, NL-5600 AE Eindhoven (NL).

(74) Common Representative: KONINKLIJKE PHILIPS ELECTRONICS N.V.; c/o LUNDIN, Thomas, M., 595 Miner Road, Cleveland, OH 44143 (US).

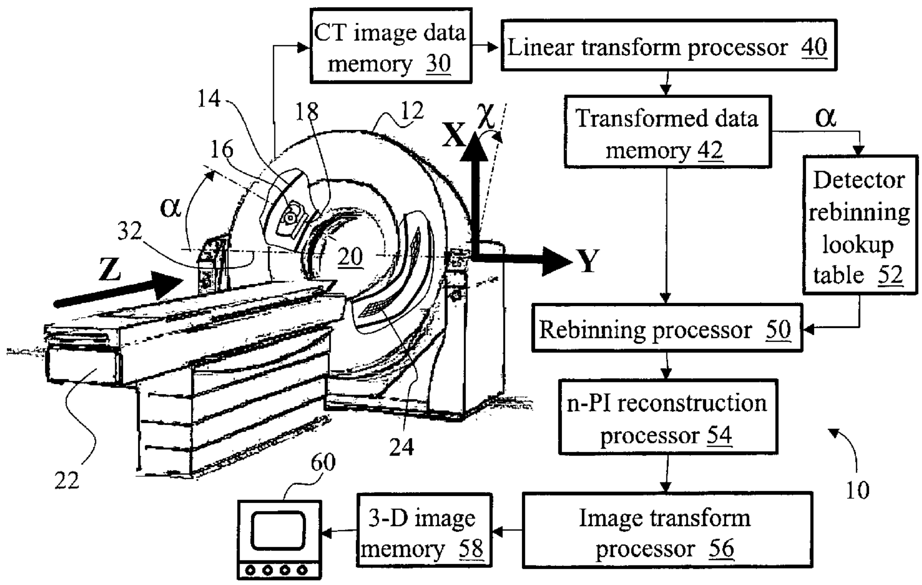
(81) Designated States *(national)*: AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BY, BZ, CA, CH, CN, CO, CR, CU, CZ, DE, DK, DM, DZ, EC, EE, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KP, KR, KZ, LC, LK, LR, LS, LT, LU, LV, MA, MD, MG, MK, MN, MW, MX, MZ, NI, NO, NZ, OM, PG, PH, PL, PT, RO, RU, SC, SD, SE, SG, SK, SL, SY, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, YU, ZA, ZM, ZW.

(84) Designated States *(regional)*: ARIPO patent (GH, GM, KE, LS, MW, MZ, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HU, IE, IT, LU, MC, NL, PT, RO, SE, SI, SK, TR), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, ML, MR, NE, SN, TD, TG).

Published:
— *with international search report*

[Continued on next page]

(54) Title: RECONSTRUCTION METHOD FOR TILTED-GANTRY COMPUTED TOMOGRAPHY



(57) Abstract: A computed tomography apparatus (10) for performing volumetric helical imaging acquires helical computed tomography imaging data (72) using a tilted gantry geometry. In the tilted gantry geometry, a rotational plane of a rotating radiation source (16) is tilted at an angle (?) with respect to a direction (Z) of linear motion of a subject. A transform processor (40) transforms (74) the imaging data (72) to a zero tilt geometry. A rebinning processor (50) rebins (110) the transformed imaging data (76) to a non sheared detector window. A reconstruction processor (54) reconstructs (130) the transformed and rebinned imaging data (112) to generate a three dimensional image representation (132). Optionally, an image transform processor (56) transforms (134) the reconstructed image representation (132) with an inverse of the zero tilt geometry transformation (74).

WO 2004/015632 A1

For two-letter codes and other abbreviations, refer to the "Guidance Notes on Codes and Abbreviations" appearing at the beginning of each regular issue of the PCT Gazette.

RECONSTRUCTION METHOD FOR TILTED-GANTRY COMPUTED TOMOGRAPHY

The present invention relates to the diagnostic imaging arts. It particularly relates to helical volumetric computed tomography (CT) imaging employing a geometric configuration in which the gantry is tilted relative to its axis of rotation, and will be described with particular reference thereto. However, the invention will also find application in other types of tomographic volumetric imaging using tilted geometric configurations.

Computed tomography (CT) imaging employs a radiation source, typically an x-ray source, that generates a fan-beam or cone-beam of x-rays that traverse an examination region. A subject arranged in the examination region interacts with and absorbs a portion of the traversing x-rays. In volume imaging, a two-dimensional detector array is arranged opposite the x-ray source to detect and measure intensities of the transmitted x-rays. Typically, the x-ray source and the detector array are mounted at opposite sides of a rotating gantry and rotate together as the gantry is rotated to acquire data over an angular range of projection views.

In helical CT imaging, the subject is advanced linearly through the examination region along a direction that is perpendicular to the gantry rotation plane such that the x-ray source traverses a helical trajectory relative to the subject. X-ray absorption data acquired during the helical orbiting is reconstructed using any of several known three-dimensional reconstruction methods such as an approximate n-PI filtered backprojection reconstruction method, an exact n-PI reconstruction method, or the like. The selected reconstruction generates a three-dimensional image representation of the subject or of a selected portion thereof.

In certain medical diagnostic applications of helical CT imaging, it is desirable to use a tilted gantry geometry in which the gantry rotation plane is tilted respective to the linear patient advancement direction. For example, in imaging of the head a substantial gantry tilt of up to 30° beneficially reduces radiation exposure of radiation-sensitive eye tissues. For imaging curvilinear anatomical structures such as the spine the gantry is beneficially dynamically tilted during imaging to keep the scanned structure generally perpendicular to the rotating gantry.

A problem arises because the tilted geometry results in a sheared helical trajectory of the x-ray source. The shearing is not accounted for in conventional helical computed tomography reconstruction techniques, and leads to substantial image degradation.

The present invention contemplates an improved apparatus and method that overcomes the aforementioned limitations and others.

According to one aspect of the invention, a method is provided for generating an image representation of an imaged subject from volumetric helical computed tomography imaging data acquired using a tilted gantry geometry. The imaging data is transformed to a zero tilt geometry. The transformed imaging data is rebinned to a non-sheared detector window. The transformed and rebinned imaging data is reconstructed to generate a three-dimensional image representation.

According to another aspect of the invention, an apparatus is disclosed for generating an image representation of an imaged subject from volumetric helical computed tomography imaging data acquired using a tilted gantry geometry. A means is provided for transforming the imaging data to a zero tilt geometry. A means is provided for rebinning the transformed imaging data to a non-sheared detector window. A means is provided for reconstructing the transformed and rebinned imaging data to generate a three-dimensional image representation.

One advantage of the present invention resides in simplified reconstruction of computed tomography imaging data acquired using a tilted gantry configuration.

Another advantage of the present invention resides in transforming tilted gantry computed tomography imaging data to a zero tilt geometry which is readily reconstructed by any of a variety of reconstruction techniques.

Numerous additional advantages and benefits of the present invention will become apparent to those of ordinary skill in the art upon reading the following detailed description of the preferred embodiment.

The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 schematically shows an exemplary computed tomography (CT) imaging apparatus according to one embodiment of the invention. Portions of the CT gantry are shown partially cut-out to reveal selected radiation source and detector components mounted on the gantry.

FIGURE 2A shows a perspective view of a pure helical radiation source path in a zero tilt helical volumetric computed tomography geometry imaging scan.

FIGURE 2B shows a perspective view of a sheared helical radiation source path in a tilted gantry helical volumetric computed tomography geometry imaging scan.

FIGURE 3 shows a view looking along the Z-direction of the sheared helical radiation source path of FIGURE 2B.

FIGURE 4 shows a preferred method for reconstructing tilted gantry computed tomography data.

FIGURE 5 schematically illustrates a linear transform of tilted gantry computed tomography imaging data to a zero tilt geometry.

FIGURE 6 diagrammatically shows a determination of the position of a focus-centered detector surface in a non-tilted geometry.

FIGURE 7 diagrammatically illustrates an evolution of a focus-centered detector window during data acquisition, linear transformation, and rebinning.

With reference to FIGURE 1, a computed tomography (CT) imaging apparatus or scanner **10** includes a tiltable stationary gantry **12** which houses a rotating gantry **14** that supports an x-ray source **16** and a collimator **18** (shown in partial cut-away). The x-ray source **16** and collimator **18** cooperate to produce a fan-shaped, cone-shaped, wedge-shaped, or otherwise-shaped x-ray beam directed across an examination region **20**. It is to be appreciated that other types of radiation sources besides an x-ray source can also be used. An angular orientation of the x-ray source **16** around the stationary gantry **12** is designated by an angular coordinate or position α. A patient or other subject is arranged on a subject support **22** which is linearly movable in a Z-direction.

A two-dimensional detector array **24** (shown in partial cut-away) is arranged across the examination region **20** opposite the x-ray source **16** to receive x-rays produced by the source **16** after traversing the examination region **20**. The x-ray source **16** and the detector array **24** are arranged in fixed relative position and rotate together with the rotating gantry **14**.

The detector array **24** detects transmitted x-ray intensities by converting the received x-rays into electrical signals, optical signals, or the like. In one suitable detector arrangement, a scintillator converts x-rays to scintillation events whose position and intensity are measured by an array of photodiodes, photodetectors, or the like. The detector signals are ported off the rotating gantry **14** using a slip ring arrangement, a radio frequency electromagnetic transmitter, or the like (not shown). In one suitable arrangement, electrical photodetector signals are converted to optical signals which are transmitted off the gantry via one or more fiber optical couplings of a slip ring. An electrical slip ring coupling can also be used. The x-ray data along with gantry angle, patient support, and detector array coordinates are formatted and stored in a CT image data memory **30**.

With continuing reference to FIGURE 1, the stationary gantry **12**, and hence the rotating gantry **14**, is tiltable about a tilt axis **32**, corresponding to a **Y**-direction in FIGURE 2 which is orthogonal to the **Z**-direction. The gantry tilt is designated by a tilt angle χ referenced to an **X**-direction which is orthogonal to the **Y**- and **Z**-directions. In a preferred helical CT imaging configuration, the x-ray source **16** and the detector array **24** rotate together in fixed relative position at a selected rotation rate. Alternatively, the x-ray source **16** rotates with the rotating gantry **14** and a band of detector arrays is mounted in stationary fashion to the stationary gantry circumscribing the rotating gantry. The subject support **22** simultaneously advances the patient or other subject linearly in the **Z**-direction as the x-ray source **16** rotates to define the helical trajectory.

With continuing reference to FIGURE 1 and with further reference to FIGURES 2A, 2B, and 3, a zero gantry tilt configuration $\chi=0°$, with uniform rotation and linear advancement rates (i.e., constant $d\alpha/dt$ and $d\mathbf{Z}/dt$ rates) results in a pure or right helical trajectory **35** of the x-ray source **16** relative to the subject, with the helical axis parallel to the **Z**-direction. For a non-zero gantry tilt, $\chi\neq0°$, and uniform gantry rotation and subject advancement rates, a sheared helical trajectory **37** of the x-ray source **16** results, in which the helix is sheared in the **X**-direction by the tilt angle χ. For this reason, the tilt angle χ is also referred to herein as shear angle χ. In presently existing tilted gantry CT imaging scanners, the gantry can typically be tilted over an angular range of about $\chi=-30°$ to $\chi=+30°$. However, larger tilt angles χ are also contemplated.

With continuing reference to FIGURES 1, 2A, 2B, and 3, the pure and sheared helical trajectories **35, 37** have a pitch P determined by an interrelation between the rotation rate of the x-ray source **16** and the rate of linear advancement of the subject support **22**. The pitch P is given by:

$$P = 2\pi p \qquad (1),$$

where p is a constant and the pitch P is the linear distance the source moves along the **Z**-direction for 360° of rotation.

For the purpose of describing the helical CT geometry, coordinates of the form $(x\ y\ z)^T$ are used, in which x corresponds to a coordinate in the **X**-direction, y corresponds to a coordinate in the **Y**-direction, z corresponds to a coordinate in the **Z**-direction, and T is the matrix transpose operator. In the case of a pure helical trajectory **35** of the radiation source **16**, a position of the x-ray source **16** in the coordinate system of the **X**-, **Y**-, and **Z**-directions is given by:

$$\vec{S}(\alpha)=\begin{pmatrix} S\cos\alpha \\ S\sin\alpha \\ p\alpha \end{pmatrix} \qquad (2),$$

where α is the previously defined angular position of the x-ray source **16**, S is a distance between the source **16** and a center of rotation which generally corresponds with a center of the examination region **20**, and p is the helical constant previously defined with reference to equation (1).

In the case of a sheared helical trajectory **37**, a position of the x-ray source **16** in the coordinate system of the **X**-, **Y**-, and **Z**-directions is given by:

$$\vec{S}(\alpha)=\begin{pmatrix} S\cos\alpha + p\alpha\tan\chi \\ S\sin\alpha \\ p\alpha \end{pmatrix} \qquad (3),$$

where it will be noticed that only the coordinate in the **X**-direction is modified by the gantry tilt χ. Hence, the x coordinate is referred to herein as a sheared coordinate.

With continuing reference to FIGURE 1, the tilted geometry imaging data is processed by a linear transform processor **40** that transforms the imaging data into a zero tilt geometry as described in greater detail below. The transformed imaging data is stored in a transformed data memory **42**. The transformed data is processed by a rebinning processor **50** to rebin the transformed data to a focus-centered (FC) detector window defined in the zero tilt geometry. In a preferred embodiment, the rebinning processor **50** accesses a detector rebinning lookup table **52** to obtain source angle α-dependent detector coordinates for the rebinning.

The transformed and detector-rebinned imaging data is suitably processed by a reconstruction processor **54**, which preferably implements an n-PI reconstruction, such as an exact n-PI reconstruction or an approximate n-PI filtered backprojection reconstruction, to produce an image representation.

As is known in the art, an n-PI reconstruction is suitable for reconstructing helical computed tomography imaging data acquired using pure helical orbiting **35** of the radiation source **16**. As used herein, n-PI reconstruction methods include methods with n=1 which are also commonly called PI reconstruction methods. In the apparatus **10**, the n-PI reconstruction is suitably applied by the reconstruction processor **54** to tilted gantry imaging data which has been transformed to a zero tilt geometry by the linear transform processor **40** and rebinned to a focus-centered detector in the zero tilt geometry by the rebinning processor **50**.

Although an n-PI reconstruction method is preferred, other volumetric helical computed tomography reconstruction methods that suitably reconstruct purely helical CT imaging data can instead be performed by the reconstruction processor **54**.

With continuing reference to FIGURE 1, it will be recognized that the image representation produced by the reconstruction processor **54** will be sheared by an angle $-\chi$ in the **x**-direction due to the action of the linear transform processor **40**. Hence, an image transform processor **56** preferably transforms the image representation in the zero tilt geometry back to the tilted gantry geometry. The image representation in the tilted gantry geometry is stored in a three-dimensional image memory **58**, and is optionally processed to construct a three-dimensional rendering, to extract selected slices, to compute a maximum intensity projection, or the like, which is displayed on a video, active matrix, CCD, monitor **60** or other display device.

With continuing reference to FIGURE 1 and with further reference to FIGURE 4, an image reconstruction method **70** suitably implemented by the combination of the linear transform processor **40**, rebinning processor **50**, reconstruction processor **54**, and image transform processor **56** is described. Tilted gantry CT imaging data **72** stored in the CT image data memory **30** is processed by the linear transform processor **40** to linearly transform the imaging data **72** to a non-sheared coordinate system in a step **74** to produce non-sheared imaging data **76** in a zero tilt geometry which is stored in the transformed data memory **42**. Comparison of equations (2) and (3) above shows that the tilted gantry coordinates $(x'\ y'\ z')^T$ are related to a zero tilt coordinate system $(x\ y\ z)^T$ by a linear transform:

$$\begin{pmatrix} x' \\ y' \\ z' \end{pmatrix} = \begin{pmatrix} x + p\alpha \tan\chi \\ y \\ z \end{pmatrix} = \begin{pmatrix} x + z \tan\chi \\ y \\ z \end{pmatrix} \qquad (4)$$

It will be noted that only the coordinate in the **X**-direction is linearly transformed by the coordinates transformation of equation (4). The coordinates y and z in the **Y**- and **Z**-directions, respectively, are unchanged. Hence, the sheared coordinate in the **X**-direction of the tilted gantry image data **72** is linearly transformed according to:

$$x_{zt} = x_{sh} - p\alpha \tan(\chi) = x_{sh} - z \tan(\chi) \qquad (5)$$

where x_{sh} is the sheared coordinate in the **X**-direction in the image data **72**, z the coordinate in the **Z**-direction (z has the same value in both the sheared and the zero tilt coordinate systems), χ is the shear angle, and x_{zt} is the coordinate in the **X**-direction transformed into the zero tilt geometry.

The transformation of equations (4) and (5) is diagrammatically illustrated in FIGURE 5. A source path **81** followed by the x-ray source **16** is transformed by equations (4) and (5) into a purely helical path **83** in the zero tilt geometry. Furthermore, because the coordinate system transform of equations (4) and (5) is purely linear, it follows that the PI lines, PI surfaces, and PI and n-PI detectors used in deriving PI and n-PI reconstructions also exist in the transformed zero tilt geometry. In other words, a transformed ray remains a ray.

With continuing reference to FIGURES 1, 4, and 5, the transformations of equations (4) and (5) distort the detector window. Two positions **85**, **87** of a physical focus-centered detector are shown in FIGURE 5, corresponding to two positions of the x-ray source **16**. For simplicity in physical construction and typically reduced complexity of image reconstruction, physical radiation detectors in computed tomography typically include straight sides, i.e. are rectangular. However, the corresponding detector positions **95**, **97** in the zero tilt geometry are sheared.

With continuing reference to FIGURE 5 and with further reference to FIGURE 6 which shows a projection **100** of a focus-centered detector in a pure helix orbit (χ=0) onto a plane defined by the **X**- and **Y**-directions, the rectangular focus-centered detector surface is given by:

$$\vec{x}_{FC}(\alpha, \beta, g, \chi = 0) = \begin{pmatrix} S\cos\alpha - (S+D)\cos(\alpha+\beta) \\ S\sin\alpha - (S+D)\sin(\alpha+\beta) \\ g + p\alpha \end{pmatrix} \quad (6),$$

where: D is a distance between the center of rotation (the intersection of the **x**- and **y**-directions in FIGURE 6, generally corresponding with a center of the examination region **20**) and a center of the focus-centered detector **100**; S is the distance between the x-ray source **16** and the center of rotation as defined previously; α is the rotational coordinate of the x-ray source **16** as defined previously, β is an angular position of a ray which impinged upon the detector **100** at coordinates (-x, -y); and g is a detector coordinate in the **z**-direction.

The fan angle of the x-ray beam is $2\beta_{max}$ where β_{max} and $-\beta_{max}$ are the angular positions of the outermost rays in the x-ray beam fan. The focus-centered detector is bounded by the fan angle range $[-\beta_{max}, \beta_{max}]$ and by boundaries $[-g_{max}, g_{max}]$ in the **z**-direction.

In the case of a tilted gantry, the detector surface of equation (6) is modified by the position of the x-ray source **16** which is given by equation (3) for a tilted gantry. The rectangular focus-centered detector surfaces **85**, **87** are given by:

$$\vec{x}_{FC}(\alpha, \beta, g, \chi) = \begin{pmatrix} S\cos\alpha - (S+D)\cos(\alpha+\beta) + p\alpha\tan\chi \\ S\sin\alpha - (S+D)\sin(\alpha+\beta) \\ g + p\alpha \end{pmatrix} \quad (7),$$

where the limits on the parameters β and g are given by the fan angle range $[-\beta_{max}, \beta_{max}]$ and by $[-g_{max}, g_{max}]$, respectively. The transformation:

$$\begin{pmatrix} x' \\ y' \\ z' \end{pmatrix} = \begin{pmatrix} x - p\alpha \tan\chi \\ y \\ z \end{pmatrix} = \begin{pmatrix} x - z \tan\chi \\ y \\ z \end{pmatrix} \quad (8)$$

transforms a sheared helix back to a non-sheared helix. For the sheared focus-centered detector the coordinate $z=g+p\alpha$ and so the transformation becomes:

$$\begin{pmatrix} x' \\ y' \\ z' \end{pmatrix} = \begin{pmatrix} x - (g + p\alpha) \tan\chi \\ y \\ z \end{pmatrix} \quad (9),$$

so that the transformed focus-centered detector surfaces **95, 97** have the shape:

$$\vec{x}_{FC}(\alpha, \beta, g, \chi) = \begin{pmatrix} S\cos\alpha - (S + D)\cos(\alpha+\beta) - g\tan\chi \\ S\sin\alpha - (S+D)\sin(\alpha+\beta) \\ g + p\alpha \end{pmatrix} \quad (10),$$

where again the limits on the parameters β and g are given by the fan angle range $[-\beta_{max}, \beta_{max}]$ and by $[-g_{max}, g_{max}]$, respectively. It is seen in equation (10) that the columns of the focus-centered detector surfaces **95, 97** have become sheared, as also seen in FIGURE 5. That is, a column given by a selected (α, β) has a g-dependent coordinate in the x-direction. The columns are sheared at the tilt angle χ.

With continuing reference to FIGURES 1 and 4 and with further reference to FIGURE 7, the projection data **76** which has been transformed into the zero tilt coordinates is rebinned to a non-sheared focus-centered detector in a step **110** by the rebinning processor **50** to produce rebinned image data **112** in the zero tilt coordinate system. Computation of a suitable non-sheared focus-centered detector in the zero tilt coordinate system is described with reference to FIGURE 7. Boundaries **113** of the physical focus-centered detector are transformed according to equation (9) to the sheared detector coordinate boundaries **115** in accordance with equation (10). A suitable window for rebinning has boundaries **117** shown in FIGURE 7 which are obtained by removing the non-rectangular shear extensions **119**, indicated by a cross-hatching in FIGURE 7.

Comparison of the rebinning window boundaries **117** with the physical detector window boundaries **113** shows that the rebinning window has a reduced available fan angle, reduced by an amount **121** indicated in FIGURE 7. However, for a typical focus-centered detector with a small height/width ratio (e.g., a detector with 20 rows along the **Z**-direction and 960 columns along the β direction has a height/width ratio of about 0.04) the non-rectangular shear extensions **119** and the corresponding reduction **121** in fan angle will be small. The fan beam produced by the cooperating x-ray source **16** and collimator **18** is preferably reduced to conform with the reduced available fan angle of the rebinning window **117** to reduce radiation exposure of the subject.

With returning reference to FIGURE 5, the transformed, sheared focus-centered detector windows **95**, **97** are seen to have different amounts of shearing. In general, the shearing of a detector window will depend upon the rotational position **α** of the x-ray source **16**.

Rebinning parameters are preferably stored in the rebinning coordinates lookup table **52**. For a number N_α of projection views per helical rotation, N_α sets of rebinning coefficients are generally used. This number of rebinning coefficients is substantially larger than the number of coefficients used in common types of rebinning such as fan-to-parallel, equidistant, or height rebinning. Hence, in a preferred embodiment, the rebinning transform coefficients are functionally parameterized with respect to **α** so that the number of parameters stored in the rebinning coordinates lookup table **52** is substantially reduced. Because the detector shearing varies in a continuous fashion with the x-ray source angle coordinate **α**, a suitable functional parameterization using a fitted continuous function is typically readily computable.

With reference to FIGURES 1 and 4, the transformed and rebinned imaging data **112** is processed by the reconstruction processor **54** in a step **130** to produce a sheared image reconstruction **132**. A reconstruction method designed for purely helical computed tomography data is suitably employed. In the preferred embodiment, an n-PI reconstruction method is employed, where n is an odd integer such as n=1, n=3, or the like.

As is known in the art, a PI reconstruction employs a Tam-Danielsson detector window, also called a PI detector window, which extends between two neighboring turns of a helical source orbit. In the n-PI reconstruction framework, the Tam-Danielsson window is also called an n-PI window with n=1. In the n-PI reconstruction, the Tam-Danielsson window encompasses an odd number of turns, e.g. a 1-PI window (same as a PI window) extends from a first turn and terminates at a neighboring turn, while a 3-PI window extends from a first turn through neighboring and second-neighboring turns and terminates at a third-neighboring turn.

Those skilled in the art recognize that the n-PI window has beneficial characteristics for helical computed tomography image reconstruction. As the n-PI detector sweeps along a helical orbit, each point within a region of interest is sampled over 180° without redundant samplings. The 180° sampling is sufficient to perform an exact reconstruction, and the elimination of redundant samplings in the n-PI reconstruction improves reconstruction speed and computational efficiency for both exact and approximate reconstruction methods.

Various n-PI reconstruction methods are known which take advantage of the n-PI detector window, including both exact n-PI reconstructions and approximate n-PI filtered backprojection reconstruction methods. Because the transformed and rebinned imaging data **112** were obtained through processing by purely linear transformations, the PI lines, PI surfaces, and n-PI detector windows operating in the zero tilt geometry on the rebinned data **112** retain their beneficial properties.

The reconstruction step **130** produces the sheared image representation **132**. Because the data was acquired with a tilted gantry geometry and transformed in the step **74** to the zero tilt geometry, it follows that the reconstructed image representation **132** exhibits a shear with a magnitude of the tilt or shear angle χ. Hence, the image transform processor **56** in a step **134** transforms the sheared image representation **132** back to the original tilted gantry coordinate system to produce an unsheared image representation **136**.

The transforming step **134** suitably employs a linear transform which is the inverse of the linear transform of equations (4) and (5). Only the coordinate in the sheared **x**-direction is transformed, according to:

$$x_{sh} = x_{zt} + p\alpha \tan(\chi) = x_{zt} + z \tan(\chi) \qquad (11)$$

which is the inverse of the transform of equation (5). The unsheared image representation **136** is preferably stored in the three-dimensional image memory **58** for further processing such as graphical display rendering. However, for small shear angles χ and diagnostic tasks for which a small image shear is acceptable, it is also contemplated to omit the inverse transforming step **134** and store the sheared image representation **132** in the image memory **58**.

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

1. A method for generating an image representation of an imaged subject from volumetric helical computed tomography imaging data **(72)** acquired using a tilted gantry geometry, the method comprising:

transforming **(74)** the imaging data **(72)** to a zero tilt geometry;

rebinning **(110)** the transformed imaging data **(76)** to a non-sheared detector window; and

reconstructing **(130)** the transformed and rebinned imaging data **(112)** to generate a three-dimensional image representation **(132)**.

2. The method as set forth in claim **1**, wherein the transforming step **(74)** includes:

linearly transforming a sheared coordinate.

3. The method as set forth in claim **2**, wherein the sheared coordinate is perpendicular to a linear motion of the imaged subject.

4. The method as set forth in either one of claim **2** and claim **3**, wherein the linear transform of the sheared coordinate includes:

$$x_{zt} = x_{sh} - z \tan(\chi)$$

where x_{sh} is the sheared coordinate, z is coordinate of a linear motion of the imaged subject, χ is a shear angle, and x_{zt} is the sheared coordinate transformed to a zero tilt geometry.

5. The method as set forth in claim **1**, wherein the transforming step **(74)** includes applying a three-dimensional transform:

$$\begin{pmatrix} x' \\ y' \\ z' \end{pmatrix} = \begin{pmatrix} x + z\tan\chi \\ y \\ z \end{pmatrix}$$

where x, y, and z are coordinates in a zero tilt geometry, x', y', and z' are the corresponding coordinates in the tilted gantry geometry, and χ is a shear angle of the tilted gantry geometry.

6. The method as set forth in any one of claims **1-5**, wherein the reconstructing step **(130)** includes:

performing an n-PI reconstruction of the transformed and rebinned imaging data **(112)**, where n is a positive odd integer greater than or equal to one.

7. The method as set forth in claim **6**, wherein n is selected from a group of integers consisting of one and three.

8. The method as set forth in either one of claim **6** and claim **7**, wherein the n-PI reconstruction is selected from a group consisting of:

an exact n-PI reconstruction, and

an approximate n-PI filtered backprojection reconstruction.

9. The method as set forth in any one of claims **1-8**, wherein a helical pitch of the volumetric helical computed tomography imaging is selected such that a point within the imaged subject is imaged on a physical detector **(24)** over a contiguous 180° rotation of a radiation source **(16)**.

10. The method as set forth in claim 9, wherein the physical detector (24) is a focus-centered detector.

11. The method as set forth in any one of claims 1-10, wherein the rebinning step (110) includes:

receiving coordinates of a physical detector (24) used in the imaging;

transforming the coordinates of the physical detector (24) to the zero-tilt geometry wherein the transformed coordinates of the physical detector (24) correspond to sheared detector coordinates; and

rebinning the transformed imaging data (76) to non-sheared detector coordinates based on the sheared detector coordinates.

12. The method as set forth in claim 11, wherein the rebinning step (110) is performed using a lookup table (52) containing transform coefficients.

13. The method as set forth in claim 12, wherein the transform coefficients include functional parameters that depend upon a radiation source angular coordinate (α).

14. The method as set forth in any one of claims 1-13, wherein the rebinning step (110) reduces a fan angle of the imaging data.

15. The method as set forth in any one of claims 1-14, further including:

transforming (134) the three-dimensional image representation (132) to the tilted gantry geometry.

16. The method as set forth in claim 15, wherein the transform used in transforming (134) the three-dimensional

image representation **(132)** to the tilted gantry geometry is an inverse transform of the transform **(74)** used in transforming the imaging data **(72)** to a zero tilt geometry.

17. An apparatus for generating an image representation of an imaged subject from volumetric helical computed tomography imaging data **(72)** acquired using a tilted gantry geometry, the apparatus comprising:

a means **(40)** for transforming **(74)** the imaging data **(72)** to a zero tilt geometry;

a means **(50)** for rebinning **(110)** the transformed imaging data **(76)** to a non-sheared detector window; and

a means **(54)** for reconstructing **(130)** the transformed and rebinned imaging data **(112)** to generate a three-dimensional image representation **(132)**.

18. The apparatus as set forth in claim **17**, wherein the means **(40)** for transforming **(74)** linearly transforms a sheared coordinate of the imaging data **(72)** to remove the shearing.

19. The apparatus as set forth in claim **18**, wherein the linear transform of the sheared coordinate includes:

$$x_{zt} = x_{sh} - p\alpha \tan(\chi)$$

where x_{sh} is the sheared coordinate, p is a constant associated with a linear motion of the imaged subject during the imaging data acquisition, α is an angular position of a radiation source **(16)** used in the imaging data acquisition, χ is a shear angle of the tilted gantry geometry, and x_{zt} is the sheared coordinate transformed to a zero tilt geometry.

20. The apparatus as set forth in claim **17**, wherein the means **(40)** for transforming **(74)** applies a three-dimensional transform:

$$\begin{pmatrix} x' \\ y' \\ z' \end{pmatrix} = \begin{pmatrix} x + p\alpha \tan\chi \\ y \\ z \end{pmatrix}$$

where x, y, and z are coordinates in a zero tilt geometry, x', y', and z' are the corresponding coordinates in the tilted gantry geometry, p is a constant associated with a linear motion of the subject during imaging data acquisition, α is an angular position of a radiation source **(16)** during imaging data acquisition, and χ is a shear angle of the tilted gantry geometry.

21. The apparatus as set forth in any one of claims **17-20**, wherein the means **(54)** for reconstructing **(130)** performs an n-PI reconstruction of the transformed and rebinned imaging data **(112)**, where n is a positive odd integer greater than or equal to one.

22. The apparatus as set forth in any one of claims **17-21**, further including:

a rebinning lookup table **(52)** in communication with the means **(50)** for rebinning **(110)**, the means **(50)** for rebinning **(110)** obtaining transform parameters from the rebinning lookup table **(52)**.

23. The apparatus as set forth in claim **22**, wherein the transform parameters include rebinning coefficients that depend upon a rotational angle **(α)** of a radiation source **(16)** used in imaging data acquisition.

24. The apparatus as set forth in any one of claims 17-23, further including:

an image transformation means (56) for transforming (134) the three-dimensional image representation (132) back to the tilted gantry geometry.

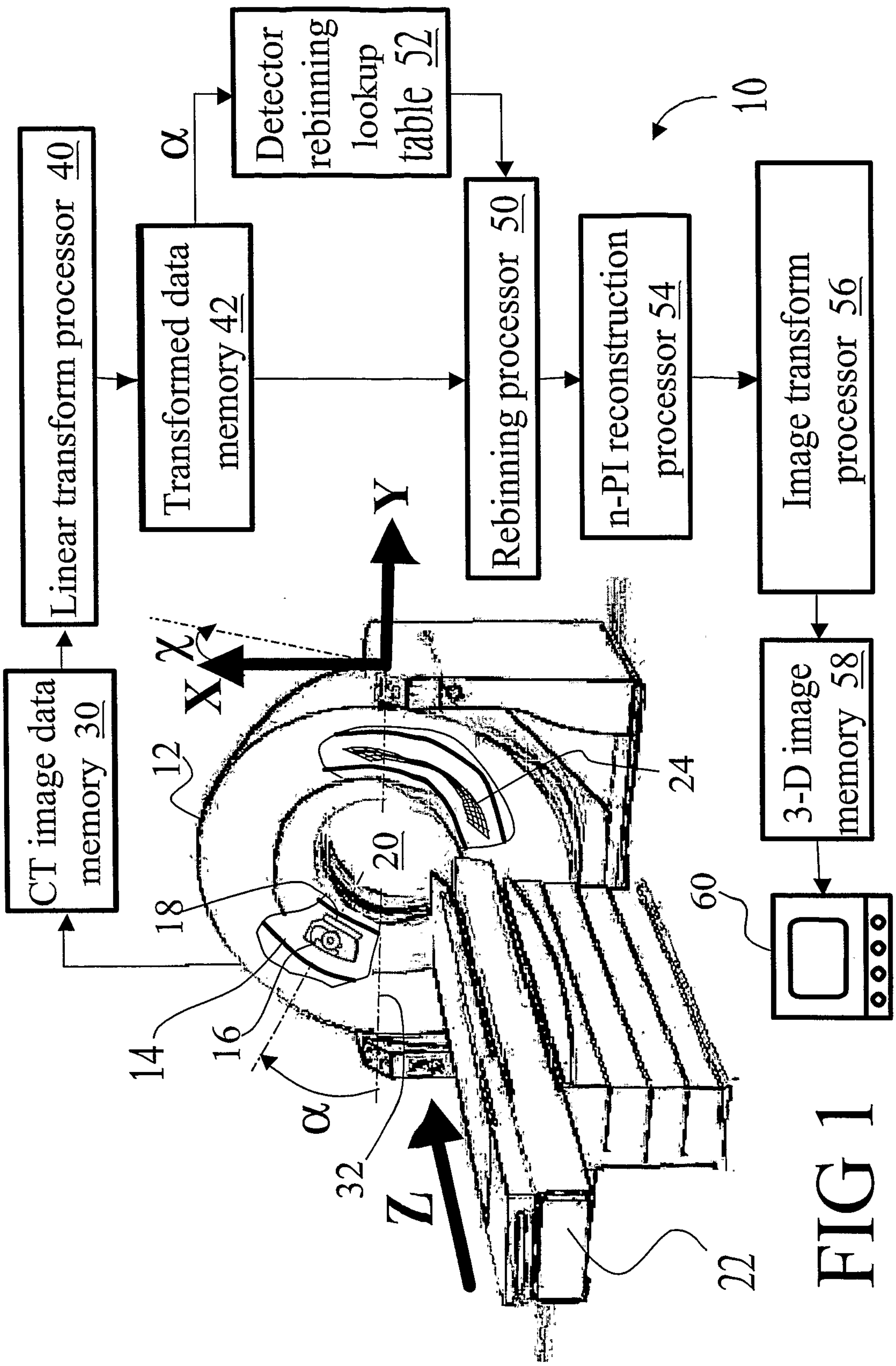

CT image data memory 30
Linear transform processor 40
Transformed data memory 42
α
Detector rebinning lookup table 52
Rebinning processor 50
n-PI reconstruction processor 54
Image transform processor 56
3-D image memory 58
10
12
14
16
18
20
22
24
32
60
X
Y
Z
χ
α

FIG 1

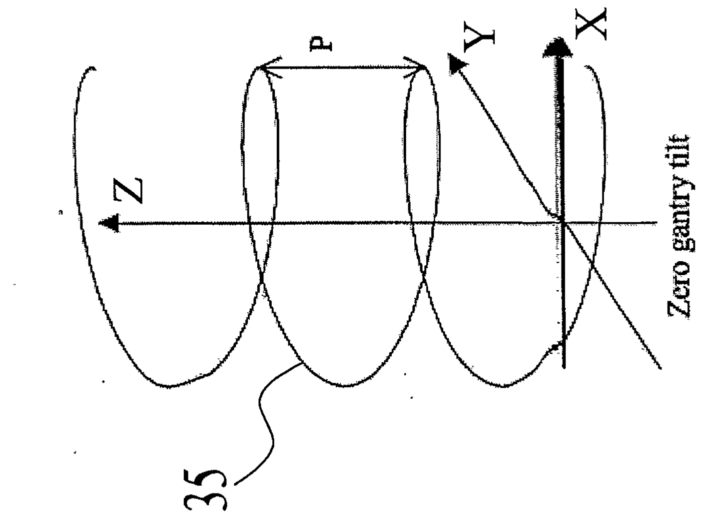


FIG 2A

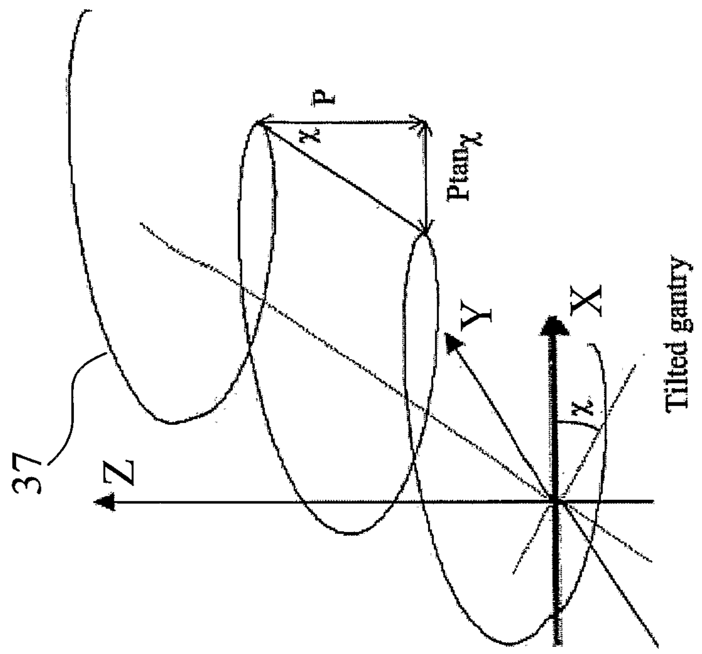


FIG 2B

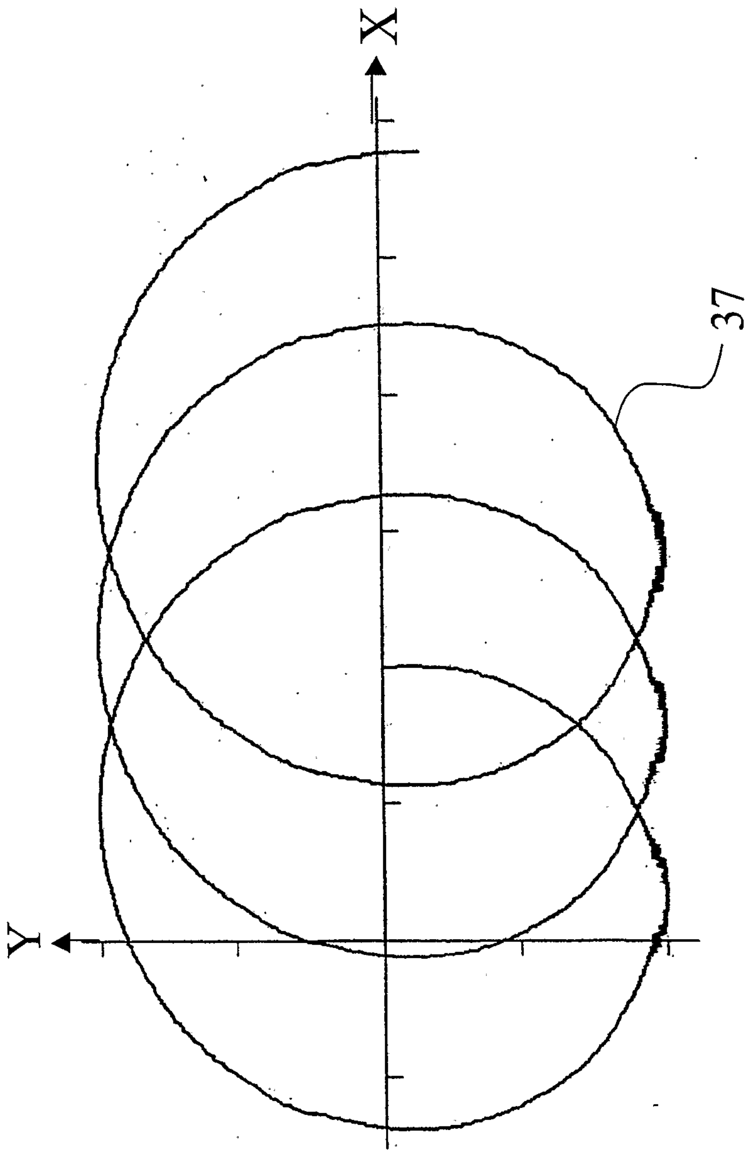


FIG 3

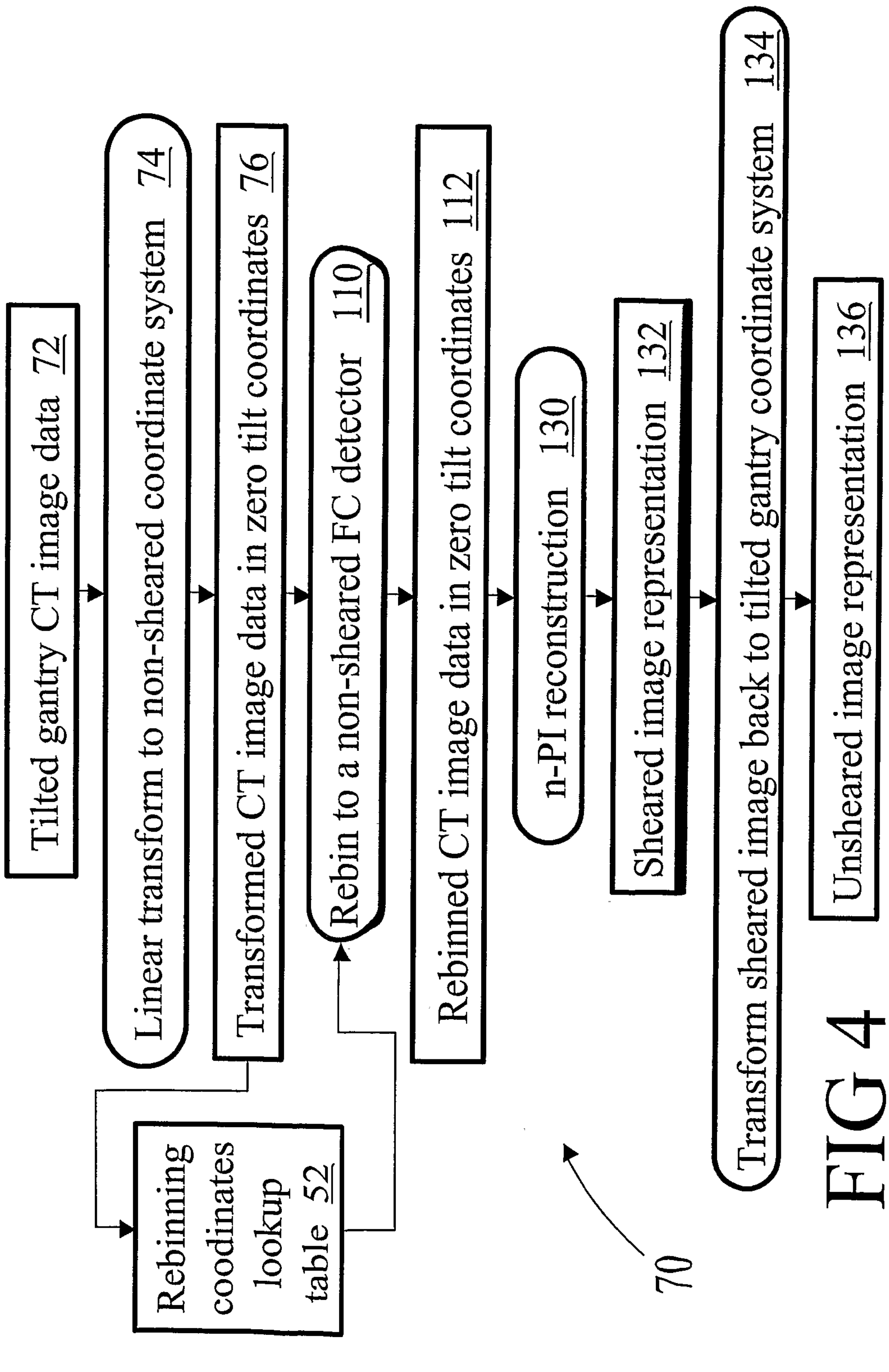

Tilted gantry CT image data 72
Linear transform to non-sheared coordinate system 74
Transformed CT image data in zero tilt coordinates 76
Rebinning coodinates lookup table 52
Rebin to a non-sheared FC detector 110
Rebinned CT image data in zero tilt coordinates 112
n-PI reconstruction 130
Sheared image representation 132
Transform sheared image back to tilted gantry coordinate system 134
Unsheared image representation 136
70

FIG 4

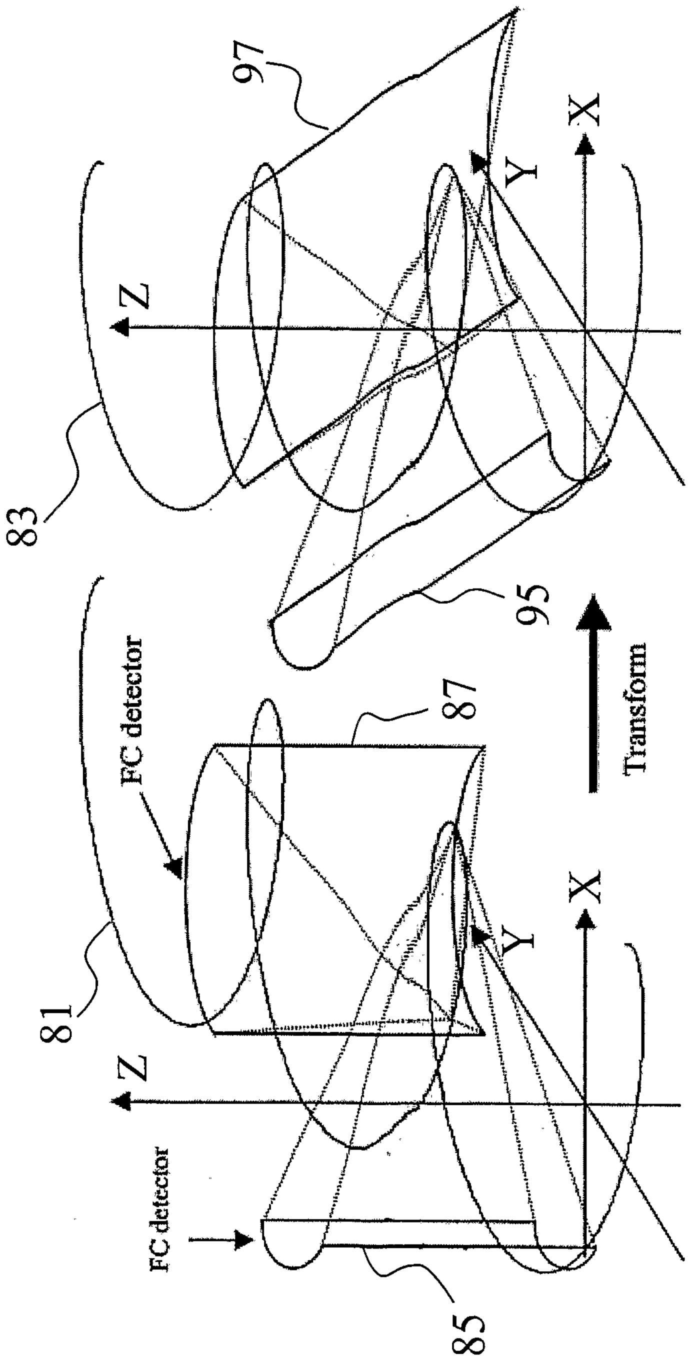


FIG 5

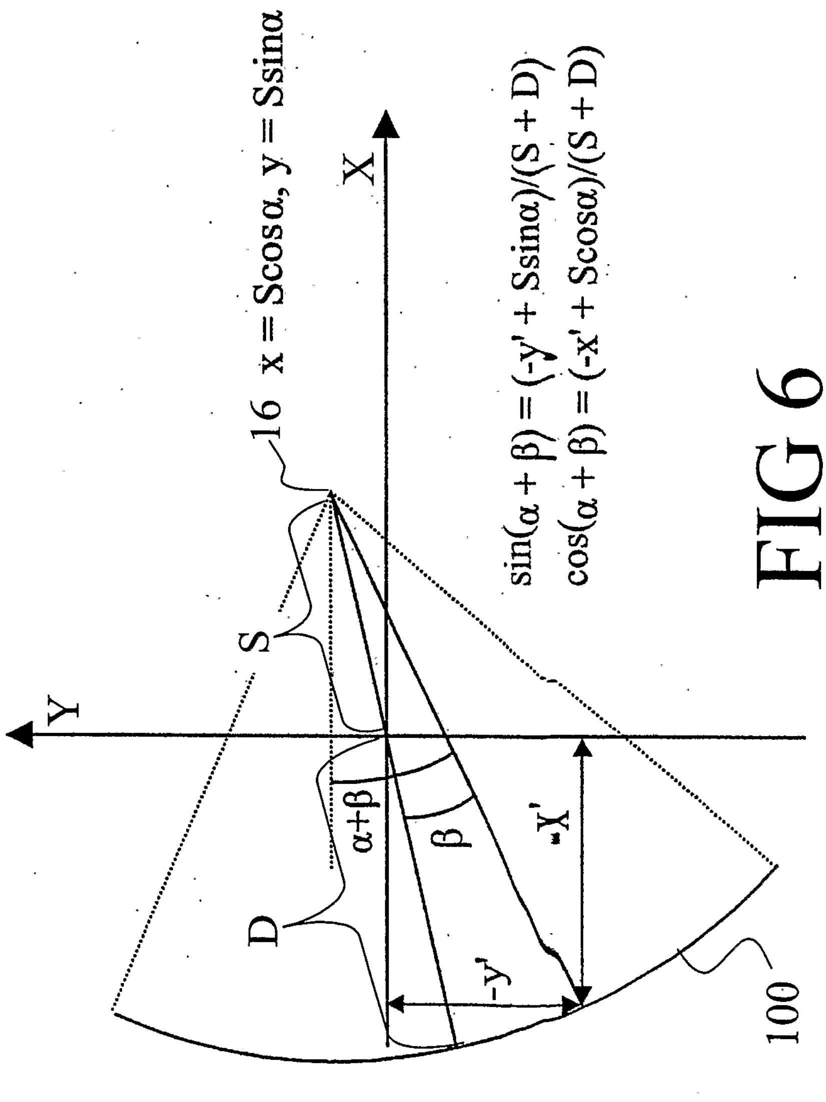

16 x = Scosα, y = Ssinα
X
Y
S
D
α+β
β
-X'
-y'
100
sin(α + β) = (-y' + Ssinα)/(S + D)
cos(α + β) = (-x' + Scosα)/(S + D)

FIG 6

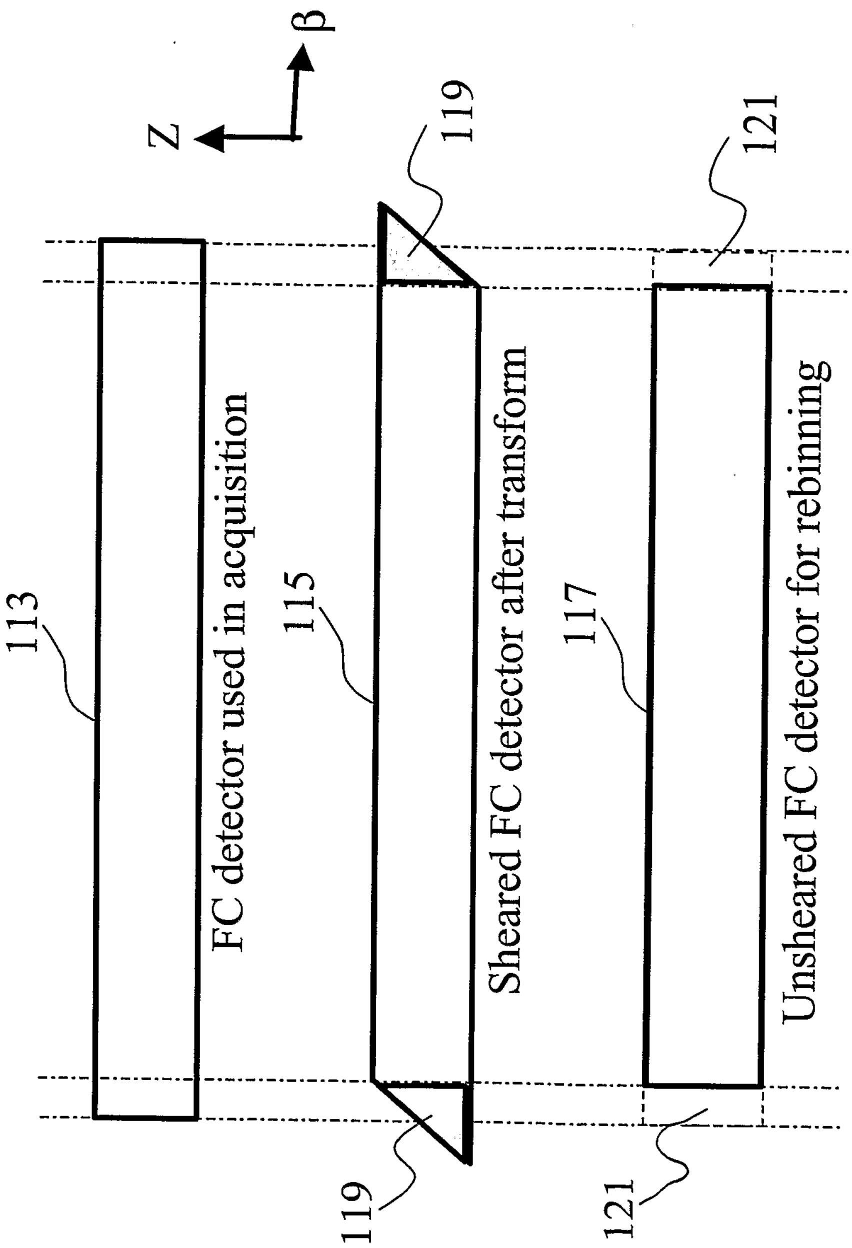

Z
β
113
FC detector used in acquisition
115
119
Sheared FC detector after transform
119
117
121
Unsheared FC detector for rebinning
121

FIG 7

INTERNATIONAL SEARCH REPORT

PCT/IB 03/03105

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 G06T11/00 A61B6/03

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)
IPC 7 G06T A61B

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

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

EPO-Internal, WPI Data, INSPEC

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	EP 0 981 996 A (GEN ELECTRIC) 1 March 2000 (2000-03-01) abstract page 2, line 28 - line 29 page 3, line 25 - line 27	1-24
A	EP 1 113 397 A (GEN ELECTRIC) 4 July 2001 (2001-07-04) abstract page 2, line 23 - line 24 page 3, line 31 - line 32	1-24
A	WO 02 30282 A (UNIV ROCHESTER) 18 April 2002 (2002-04-18) abstract	1-24

-/--

[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

° Special categories of cited documents :

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

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

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

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

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

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

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

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

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
28 October 2003	04/11/2003

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+31-70) 340-3016	González Arias, P

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication,where appropriate, of the relevant passages	Relevant to claim No.
A	US 6 061 420 A (BECKETT BOB L ET AL) 9 May 2000 (2000-05-09) abstract column 8, line 16 - line 18 ---	1-24
A	WO 02 43565 A (IMATRON INC) 6 June 2002 (2002-06-06) abstract page 3, line 23 - line 26 page 4, line 33 - line 37 -----	1-24

INTERNATIONAL SEARCH REPORT

PCT/IB 03/03105

Patent document cited in search report		Publication date	Patent family member(s)			Publication date
EP 0981996	A	01-03-2000	US	6229869	B1	08-05-2001
			DE	69902326	D1	05-09-2002
			DE	69902326	T2	20-03-2003
			EP	0981996	A1	01-03-2000
			IL	131401	A	10-11-2002
			JP	2000083947	A	28-03-2000
EP 1113397	A	04-07-2001	US	6332013	B1	18-12-2001
			CN	1304036	A	18-07-2001
			EP	1113397	A2	04-07-2001
			JP	2001218764	A	14-08-2001
WO 0230282	A	18-04-2002	US	6504892	B1	07-01-2003
			AU	1534002	A	22-04-2002
			CA	2425323	A1	18-04-2002
			EP	1324696	A2	09-07-2003
			WO	0230282	A2	18-04-2002
US 6061420	A	09-05-2000	NONE			
WO 0243565	A	06-06-2002	AU	3649602	A	11-06-2002
			WO	0243565	A1	06-06-2002
			US	2003161434	A1	28-08-2003