United States Patent [19]

Toth et al.

[54] **HELICAL SCANNING COMPUTED TOMOGRAPHY WITH TRACKING X-RAY SOURCE**

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[73] Assignee: **General Electric Company**, Milwaukee, Wis.

[56] **References Cited**

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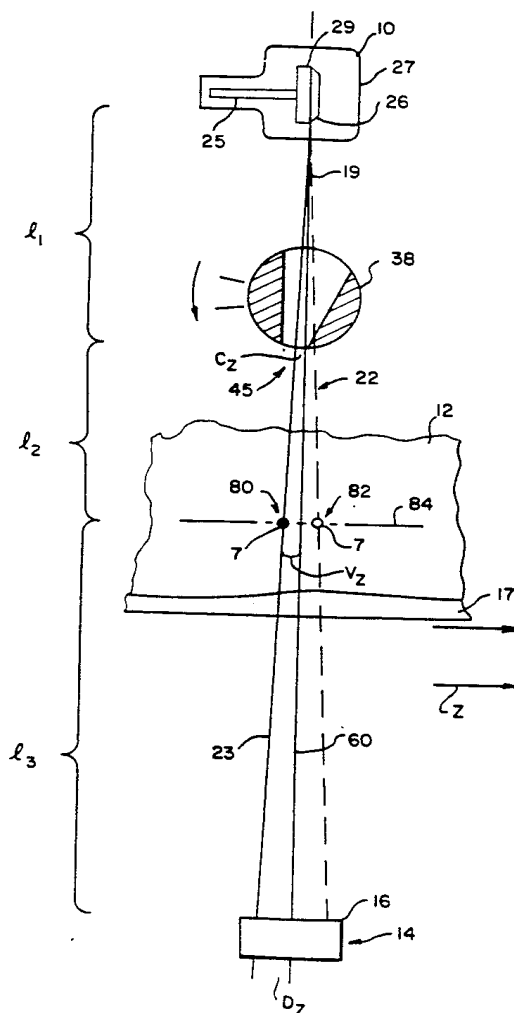|			
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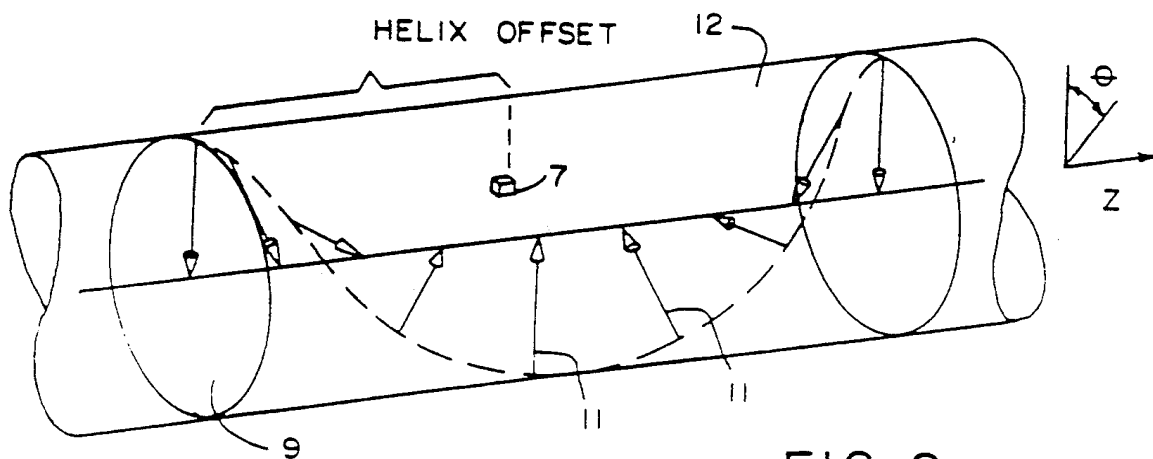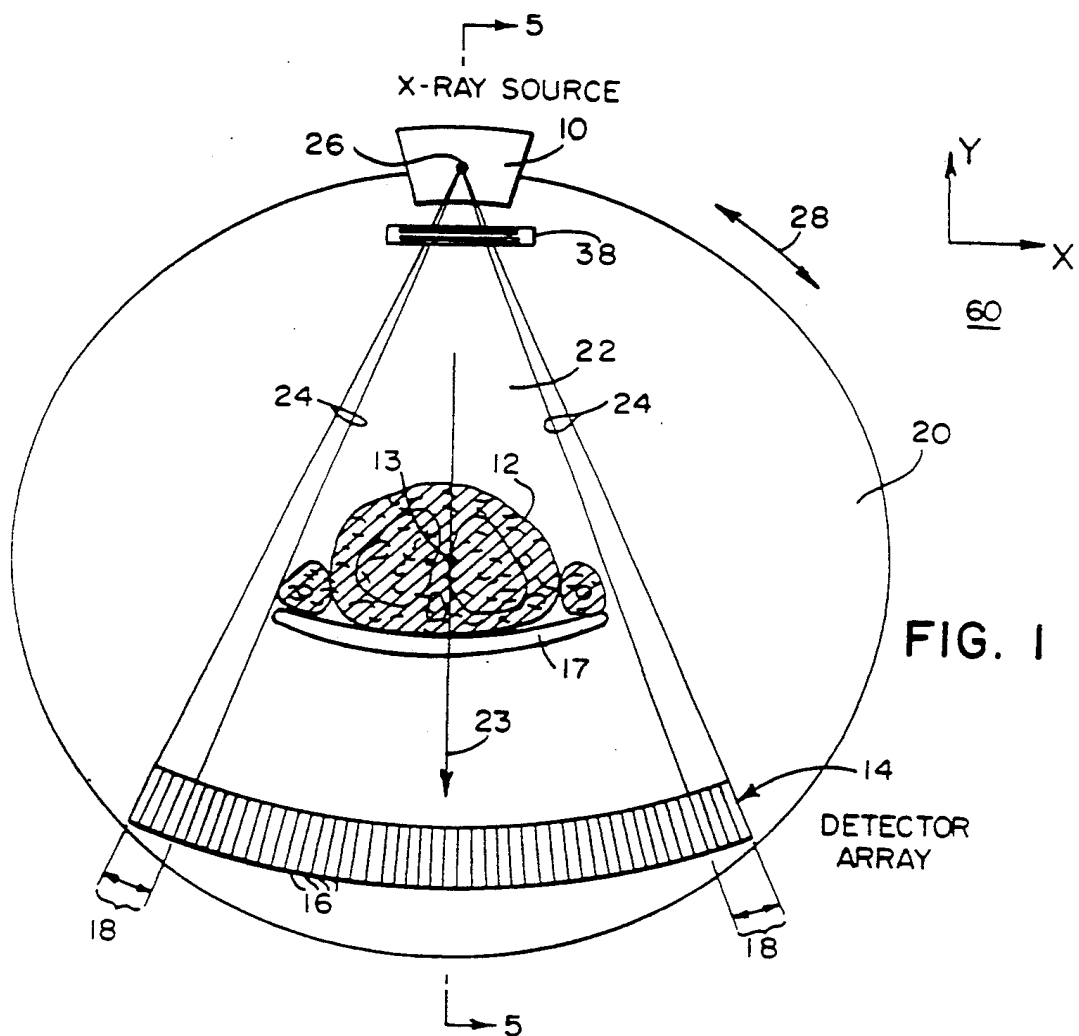*Primary Examiner*—Janice A. Howell
Assistant Examiner—Don Wong
Attorney, Agent, or Firm—Quarles & Brady

[57] **ABSTRACT**

A CT apparatus reduces errors in projection data acquired in helical scanning. The imaged object moves concurrently along a translation axis and the x-ray beam is periodically translated with the imaged object so as to subtend a single predetermined volume element during the acquisition of one projection set of data for a first slice. The x-ray beam then returns to its starting position and tracks a second predetermined volume element within a next slice. The x-ray beam may be translated by moving the focal point or a collimator or a combination of both.

14 Claims, 8 Drawing Sheets

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X-RAY SOURCE

26   10

38

Y

28

X

60

24   22   24

20

13   12

17

23

FIG. 1

14

DETECTOR
ARRAY

18   16   18

5

HELIX OFFSET   12

φ

7
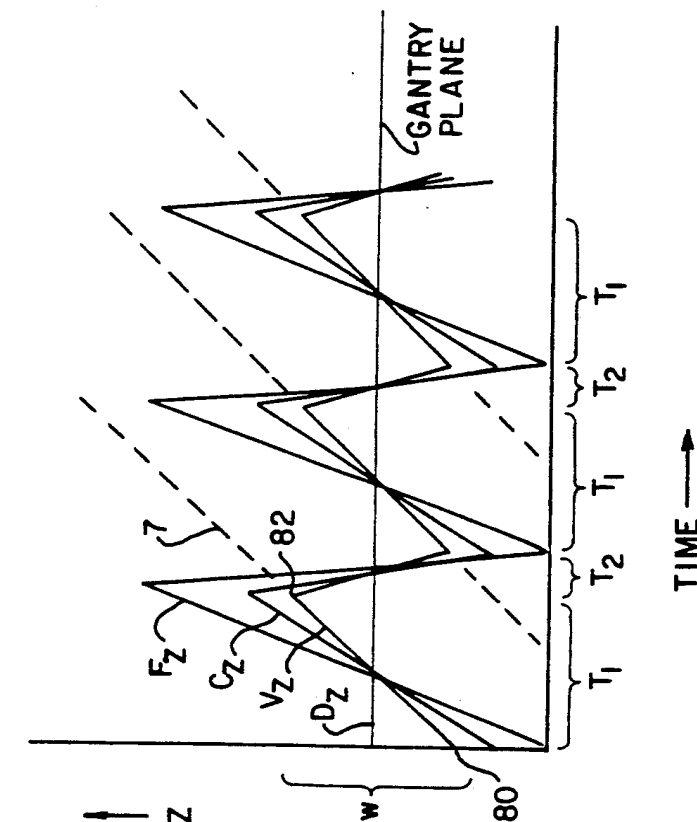
Z

9   11   11

FIG. 2

FIG. 3

FIG. 4a

FIG. 4b

FIG. 5

FIG. 6
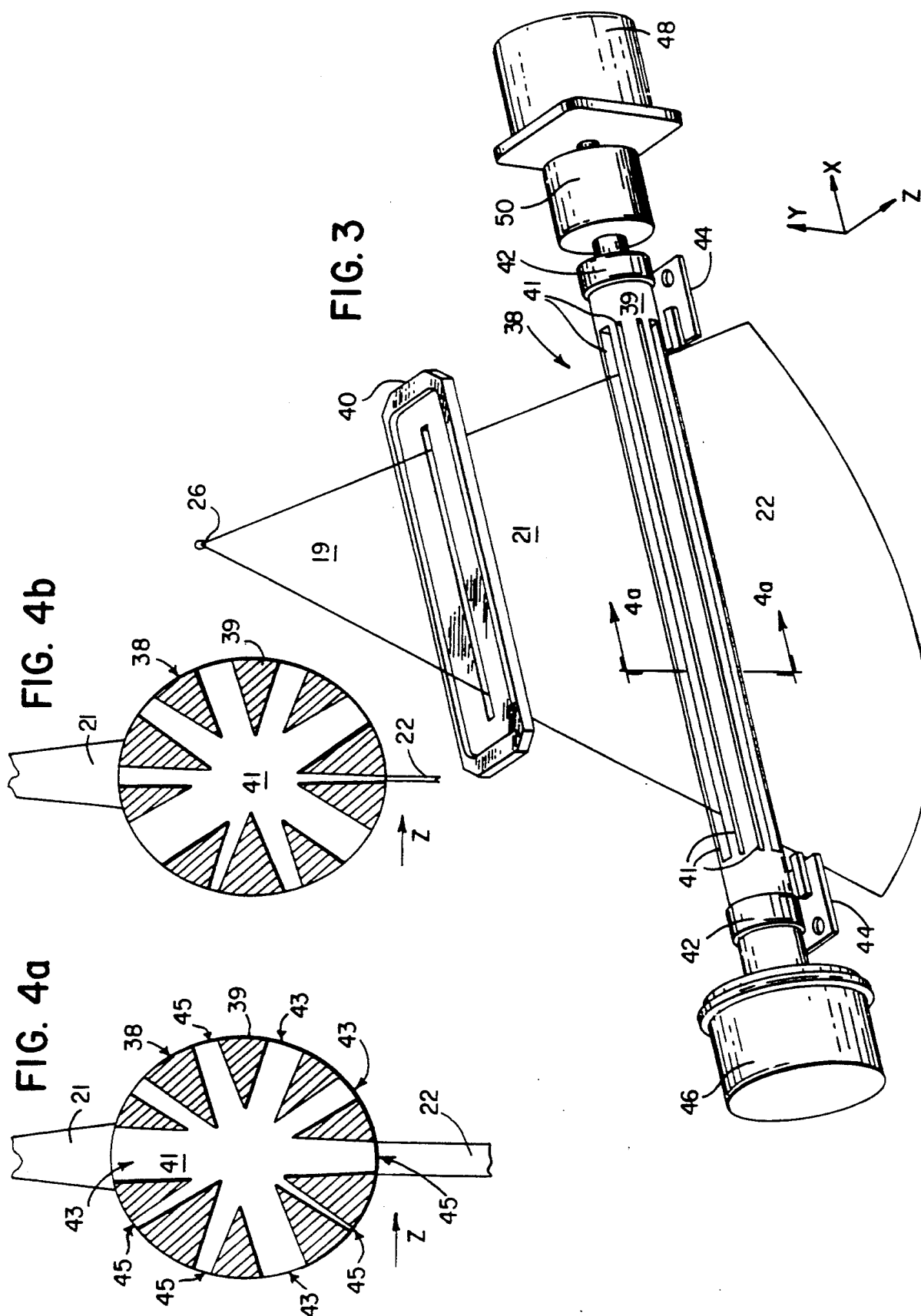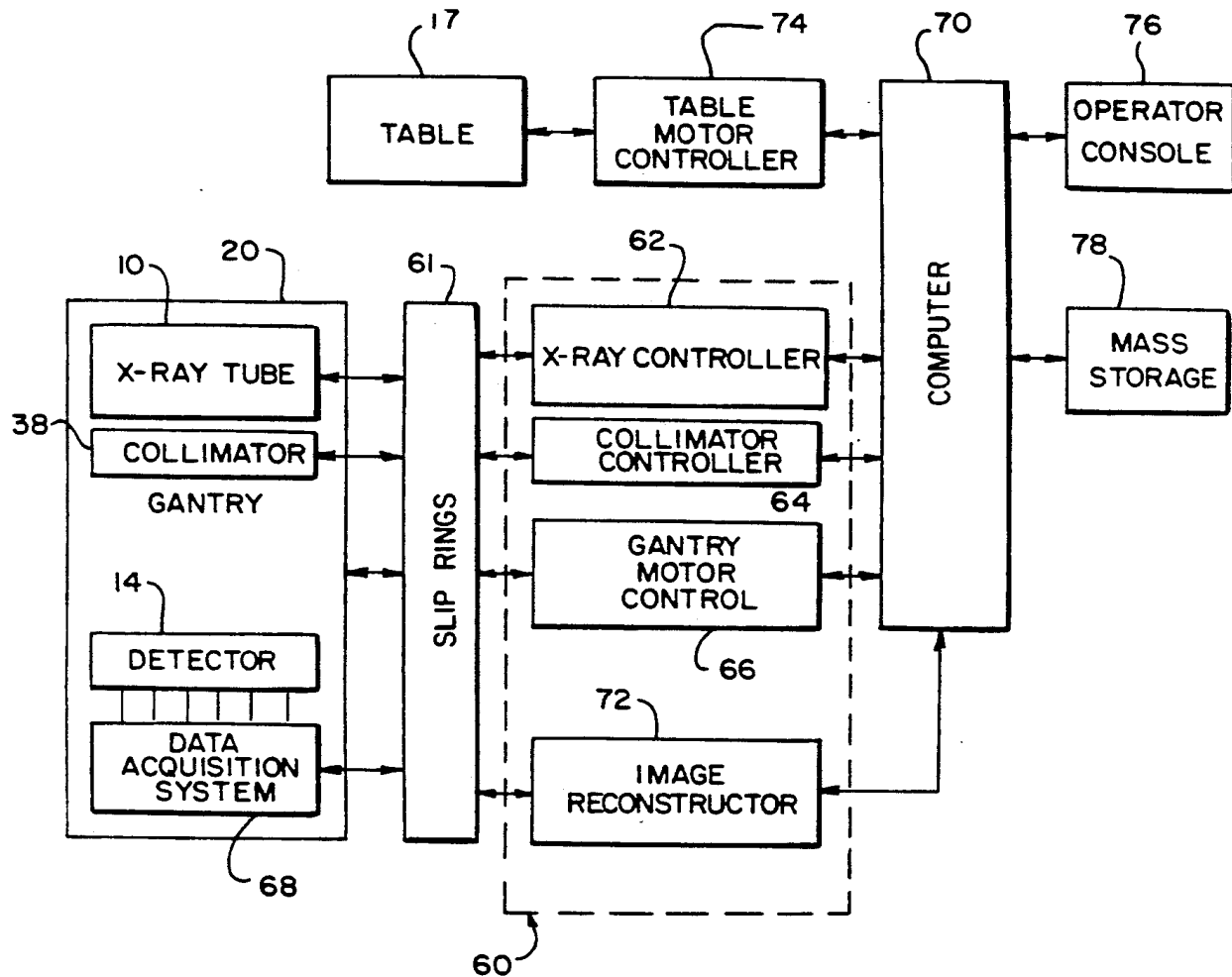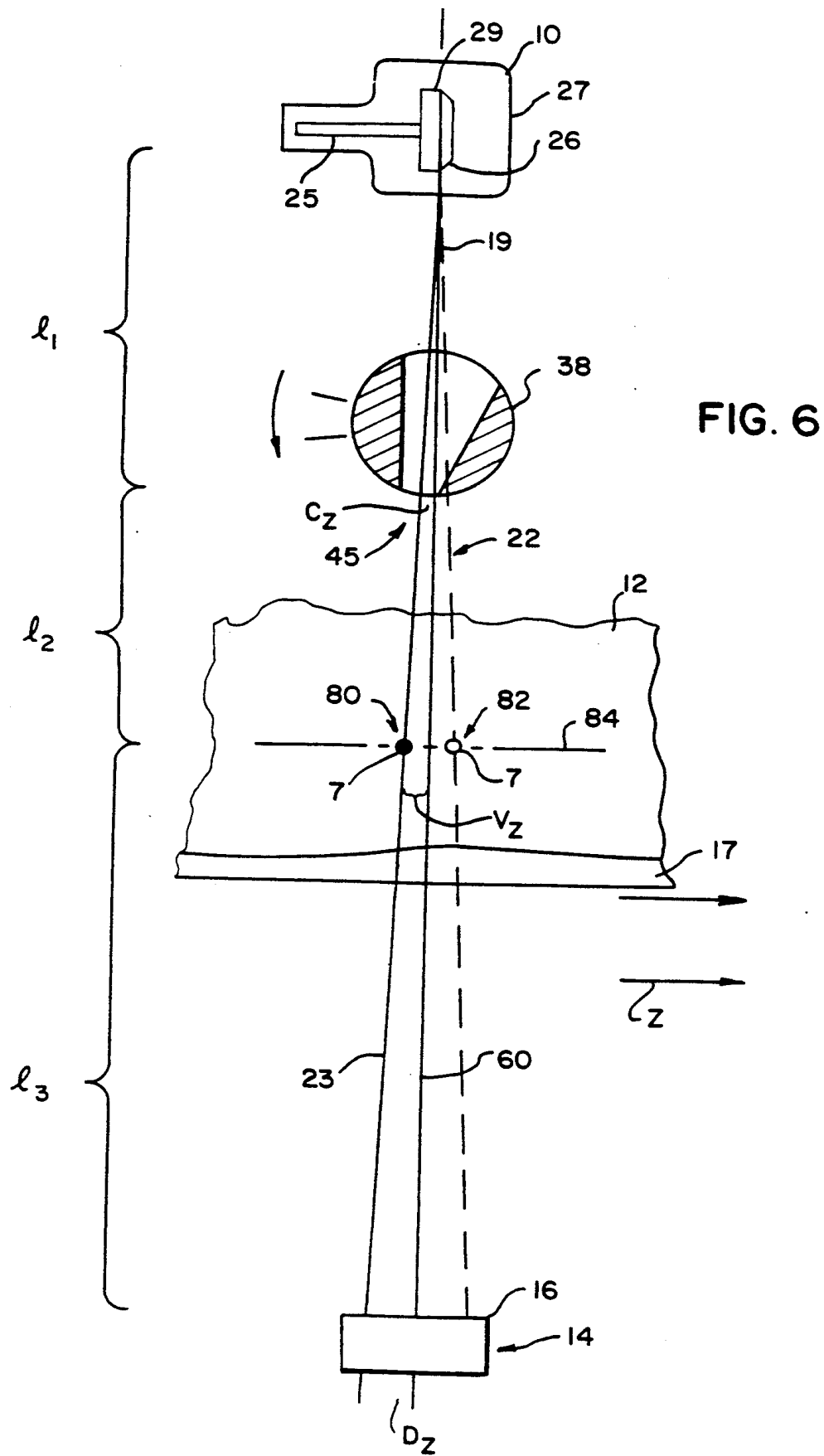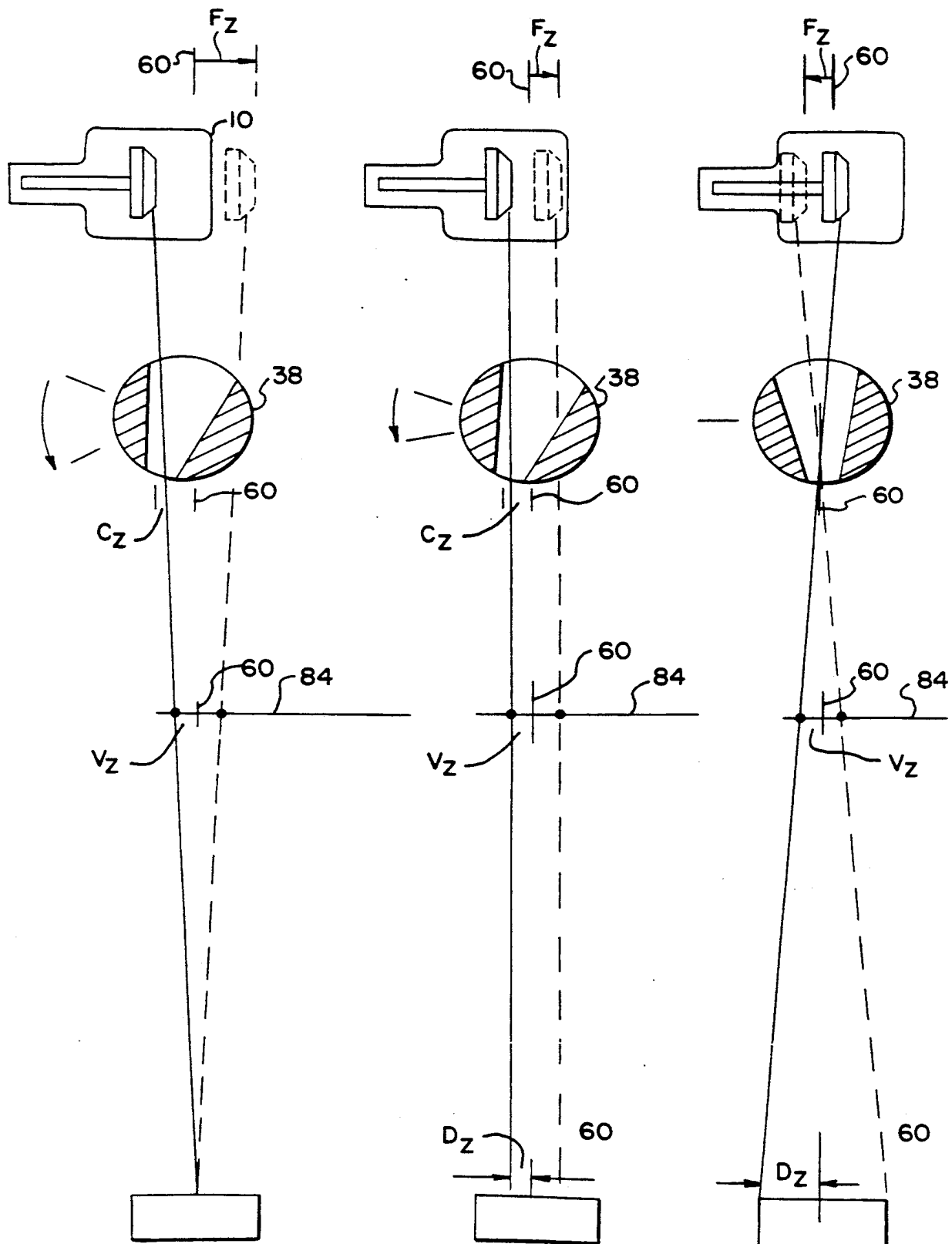
FIG. 7

FIG. 8
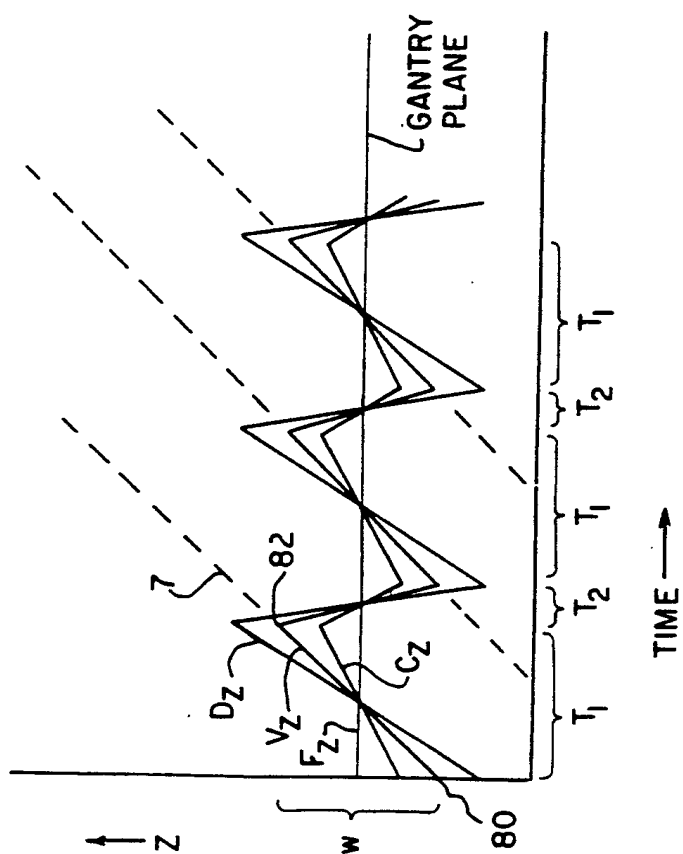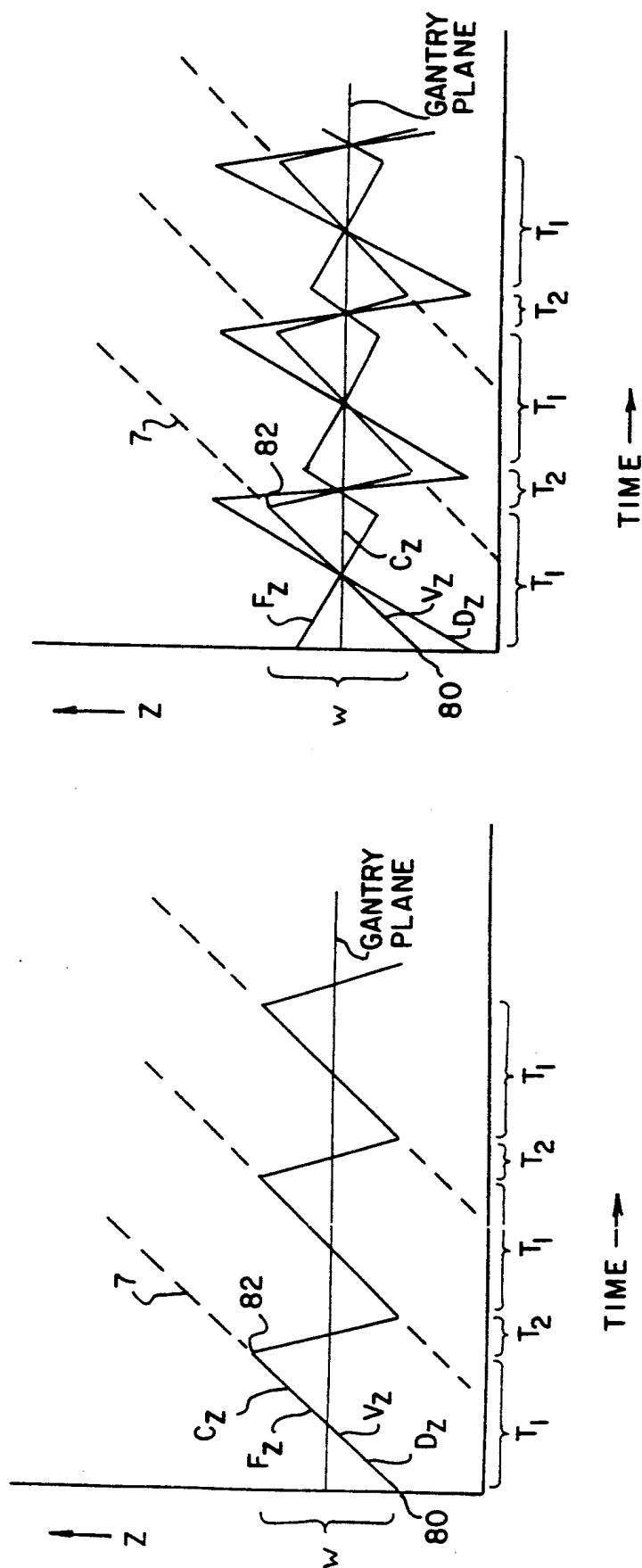
FIG. 9
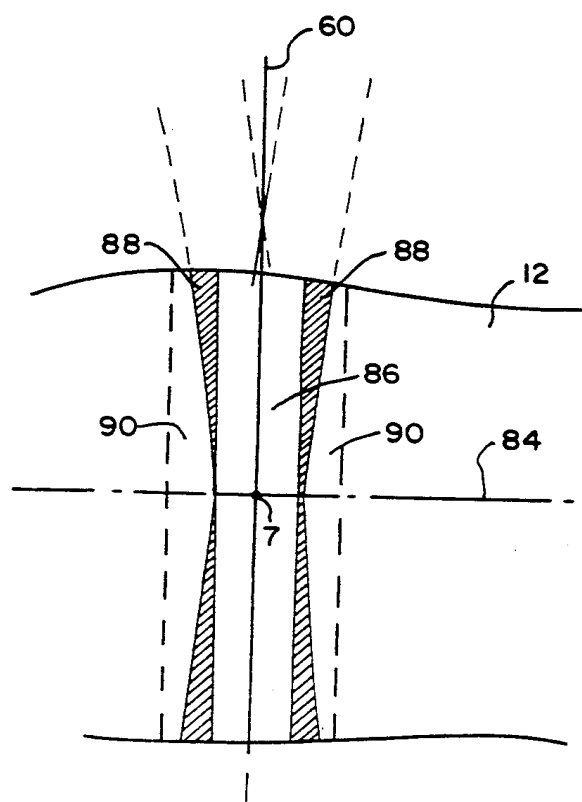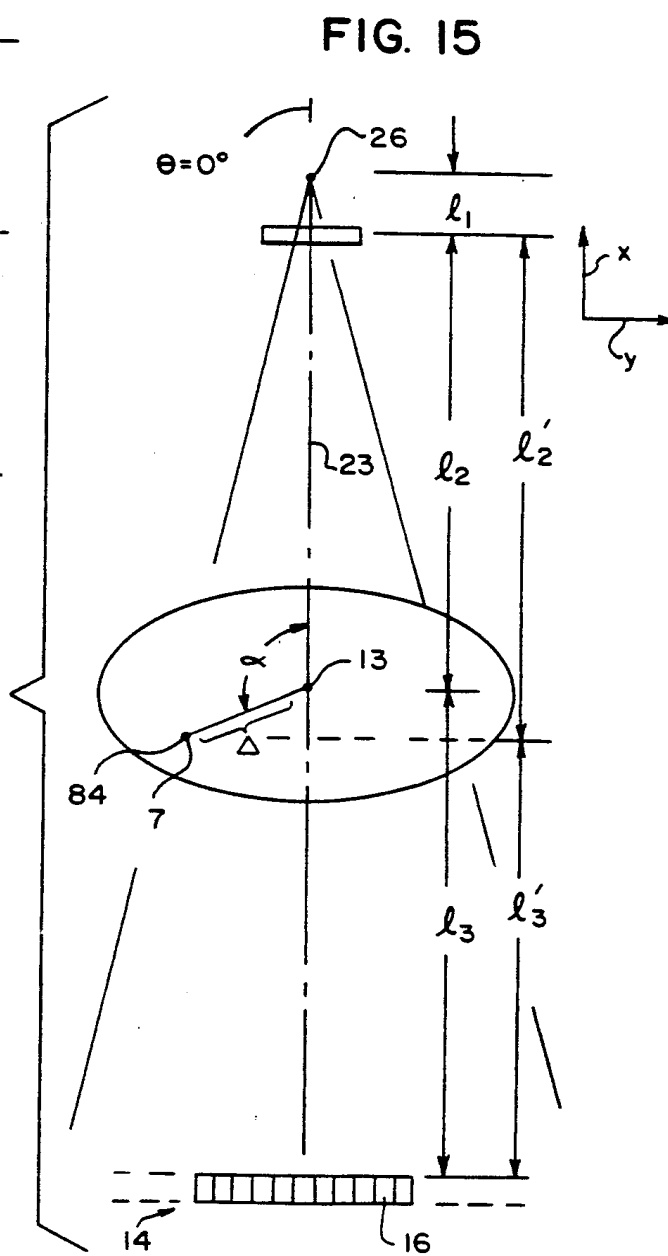
FIG. 11

FIG. 10

FIG. 13

FIG. 12

FIG. 14

FIG. 15

HELICAL SCANNING COMPUTED TOMOGRAPHY WITH TRACKING X-RAY SOURCE

BACKGROUND OF THE INVENTION

This invention relates to computed tomography (CT) systems and specifically to a helical scanning CT system in which the imaged object is concurrently translated during the acquisition of tomographic projections.

In a computed tomography system, an x-ray source is collimated to form a fan beam with a defined fan beam angle. The fan beam is typically oriented to lie within the x-y plane of a Cartesian coordinate system, termed the "gantry plane", and is transmitted through an imaged object to an x-ray detector array oriented within the gantry plane. The detector array is comprised of an array of detector elements each of which measures the intensity of transmitted radiation along a ray projected from the x-ray source to the particular detector element. The intensity of the transmitted radiation is dependent on the attenuation of the x-ray beam along that ray by the imaged object.

The center of the fan beam and its direction of the fan beam is identified by a fan beam axis.

The x-ray source and detector array may be rotated on a gantry within the gantry plane and around a center of rotation within the imaged object so that the angle at which the fan beam axis intersects the imaged object may be changed. At each gantry angle, a projection is acquired comprised of the intensity signals from each detector element. The gantry is then rotated to a new angle and the process is repeated to collect a number of projections along a number of gantry angles to form a tomographic projection set. The acquired tomographic projection sets are typically stored in numerical form for later computer processing to "reconstruct" a slice image according to reconstruction algorithms known in the art. A projection set of fan beam projections may be reconstructed directly into an image by means of fan beam reconstruction techniques, or the intensity data of the projections may be sorted into parallel beams and reconstructed according to parallel beam reconstruction techniques. The reconstructed tomographic images may be displayed on a conventional CRT tube or may be converted to a film record by means of a computer controlled camera.

A typical computed tomographic study involves the acquisition of a series of "slices" of an imaged object, each slice parallel to the gantry plane and having a slice thickness dictated by the width of the detector array, the size of the focal spot, the collimation and the geometry of the system. Each successive slice is displaced incrementally along a z-axis, perpendicular to the x and y axes, so as to provide a third spatial dimension of information. A radiologist may visualize this third dimension by viewing the slice images in order of position along the z-axis, or the numerical data comprising the set of reconstructed slices may be compiled by computer programs to produce shaded, perspective representations of the imaged object in three dimensions.

As the resolving power of computed tomography methods increases, additional slices are required in the z-dimension. The time and expense of a tomographic study increases with the number of slices required. Also, the longer scan times necessary to acquire more slices increases the discomfort to the patient who must remain nearly motionless to preserve the fidelity of the

tomographic reconstructions. Accordingly, there is considerable interest in reducing the time required to obtain a slice series.

The time required to collect the data for a series of slices depends in part on four components: a) the time required to accelerate the gantry to scanning speed, b) the time required to obtain a complete tomographic projection set, c) the time required to decelerate the gantry, and d) the time required to reposition the patient in the z-axis for the next slice. Reducing the time required to obtain a full slice series may be accomplished by reducing the time required to complete any of these four steps.

The time required for acceleration and deceleration of the gantry (a and c) may be avoided in tomographic systems that use slip rings rather than cables to communicate with the gantry. The slip rings permit continuous rotation of the gantry and avoid the need for acceleration and deceleration. Hereafter, it will be assumed that the CT systems discussed are equipped with slip rings or the equivalent to permit continuous rotation.

The time required to acquire the tomographic data set (b) is more difficult to reduce. Present CT scanners require on the order of one to two seconds to acquire the projection set for one slice. This scan time may be reduced by rotating the gantry at a faster speed. However, a higher gantry speed, in general, will decrease the signal-to-noise ratio of the acquired data by the square root of the factor of rotational rate increase. This may be overcome to some extent by increasing the radiation output of the x-ray tube, but is subject to the power limits of such devices.

Finally, a reduction in patient repositioning time (d) may be accomplished by translating the patient in the z-axis concurrently with the rotation of the gantry. The combination of continuous patient translation along the z-axis during the rotation of the gantry and acquisition of projection data has been termed "helical scanning" and refers to the apparent path of a point on the gantry with respect to a reference point on the imaged body. As used herein, "helical scanning" shall refer generally to the use of continuous translation of the patient or imaged object during the acquisition of tomographic imaging data, and "constant z-axis scanning" shall refer to the acquisition of the tomographic data set without translation of the patient or imaged object during the acquisition period.

Continuous translation of the imaged object during scanning shortens the total scanning time required for the acquisition of a given number of slices by eliminating the length of time normally required for repositioning the patient between scans. However, helical scanning introduces certain errors in the acquired tomographic projection sets. The mathematics of tomographic reconstruction assumes that the tomographic projection set is acquired along a constant z-axis slice plane. The helical scan path clearly deviates from this condition and this deviation results in image artifacts in the reconstructed slice image if there is any significant change in the object in the z-axis. The severity of the image artifacts depends generally on the "helix offset" in the projection data, measured as the z-axis difference between the scanned volume elements of the imaged object and the z axis value of the desired slice plane. Errors resulting from helical scanning will be referred to collectively as "skew" errors.

Several methods have been used to reduce skew errors in helical scanning. A first approach disclosed in U.S. Pat. No. 4,630,202 issued Dec. 16, 1986, reduces the pitch of the helical scan and then averages the projection data of consecutive 360° tomographic projection sets. The effect is equivalent to using a detector array with a larger width along the z axis, which also moves less in the z direction during a rotation of the gantry, i.e. with a lesser scanning pitch. Skew errors are reduced using this method, but at the expense of additional scanning time necessitated by the lower scanning pitch. Thus, this method reduces, to some extent, the advantages to be gained by helical scanning.

Skew errors at the ends of the tomographic projection set may be reduced in conjunction with this approach by changing the weighting of the last and first projections of the consecutive 360° tomographic projection sets in the "averaging" process to give greater weight to the projection closest to the slice plane.

A second approach disclosed in U.S. Pat. No. 4,789,929 issued Dec. 6, 1988, also applies weighting to the projections of combined, consecutive 360° tomographic projection sets, but the weighting is a function of the helix offset of each projection at the given gantry angle. This approach of interpolating over 720° generally increases partial volume artifacts. Partial volume artifacts are image artifacts arising when certain volume elements of the imaged object contribute to only some of the projections of the projection set.

A third approach, described in co-pending U.S. Patent application Ser. No. 07/435,980, entitled: "Extrapolative Reconstruction Method for Helical Scanning" and assigned to the same assignee as the present invention, uses a half-scanning technique to reduce the table motion during the acquisition of each slice. Projection data is acquired over 360° of gantry rotation and interpolated to a slice plane. The reduced gantry motion corresponds to reduced table motion and hence certain helical scanning artifacts are reduced.

SUMMARY OF THE INVENTION

The present invention reduces skew error by translating the x-ray beam with translation of the imaged object. Specifically, the imaged object is translated concurrently along a translation axis while an opposed x-ray generator and x-ray detector in opposition around the imaged body are rotated around the imaged object in a gantry plane substantially perpendicular to the translation axis. The x-ray generator projects a beam of x-rays through the imaged object and during a first period sweeps the beam along the translation axis following the translation of the imaged object. During a second period, the x-ray generator returns to its initial orientation, moving in a second direction along the translation axis counter to the translation of the imaged object. The sweeping of the x-ray beam may be such as to center the x-ray beam around a predetermined volume element in the imaged object and to track that volume element during the first period. The movement of the x-ray beam may be by moving a collimator or the x-ray source or both.

It is one object of the invention to reduce the helix offset of the projection data acquired in helical scanning without interrupting the continuous motion of the imaged object and gantry. The sweeping of the x-ray beam serves to counteract the effective motion of the imaged object during the acquisition of each projection set.

In one embodiment, the x-ray generator includes a stationary x-ray source and a movable collimator that may be rapidly repositioned in response to movement of the imaged object.

It is thus another object of the invention to provide a simple means of sweeping the x-ray beam with motion of the imaged object.

The foregoing and other objects and advantages of the invention will appear from the following description. In the description, reference is made to the accompanying drawings which form a part hereof and in which there is shown by way of illustration, a preferred embodiment of the invention. Such embodiment does not necessarily represent the full scope of the invention, however, and reference must be made therefore to the claims herein for interpreting the scope of the invention.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a schematic representation CT system gantry including an x-ray source and x-ray detector as may be used with the present invention;

FIG. 2 is a schematic illustration of the imaged object of FIG. 1 showing the relative orientation of the gantry and gantry axis with respect to the imaged object for helical scanning. The pitch of the helical scanning is exaggerated for clarity;

FIG. 3 is a perspective view of the collimator assembly of the present invention;

FIG. 4 (a) and (b) are cross-sectional views of the mandrel of the collimator of FIG. 3 showing orientation of the mandrel for thick and thin fan beams respectively;

FIG. 5 is a block diagram showing the control system for the collimator and x-ray focal point of FIG. 3 according to the present invention;

FIG. 6 is a simplified cross-sectional view of the path of the x-ray fan beam, taken along line 5—5 in FIG. 1, with the x-ray tube anode, the collimator and the detector array exaggerated for clarity and showing a first method of reducing helix offset requiring only movement of the collimator;

FIG. 7 is a cross-sectional view, similar to that of FIG. 6, of a second method of reducing helix offset requiring movement of the collimator and the x-ray focal point but reducing movement of the illuminated area of the detector;

FIG. 8 is a cross-sectional view, similar to that of FIG. 6, a third method of reducing helix offset requiring movement of the collimator and the x-ray focal point but further reducing skew error;

FIG. 9 is a cross sectional view, similar to that of FIG. 6, of a fourth method of reducing helix offset requiring only movement of the x-ray;

FIG. 10 is graph of the z-axis displacement with time for the collimator, the illumination area of the detector and the imaged element volume for the method of FIG. 6;

FIG. 11 is graph of the z-axis displacement with time for the collimator, the illumination area of the detector and the imaged element volume for the method of FIG. 7;

FIG. 12 is graph of the z-axis displacement with time for the collimator, the illumination area of the detector and the imaged element volume for the method of FIG. 8;

FIG. 13 is graph of the z-axis displacement with time for the collimator, the illumination area of the detector

and the imaged element volume for the method of FIG. 9; and

FIG. **14** is an exaggerated cross-sectional view of the imaged object taken along **5**—**5** in FIG. **1**, showing a single slice thickness, the effective thickness for helical scanning and for helical scanning with reduced helix offset per the methods of FIGS. **6–8**.

FIG. **15** is a schematic representation similar to FIG. **1** showing the determination of l_3' and l_2' for offset translation axes.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring to FIG. **1**, a gantry **20**, such as may be used in a "third generation" computed tomography (CT) scanner, includes an x-ray source **10** collimated by collimator **38** to project a fan beam of x-rays **22** through imaged object **12** to detector array **14**. The x-ray source **10** and detector array **14** rotate on the gantry **20** about center of rotation **13**. The rotation of the gantry **20**, as indicated by arrow **28** is within a gantry plane **60**, aligned with the x-y plane of a Cartesian coordinate system.

The imaged object **12** rests on table **17** which is radiotranslucent so as not to interfere with the imaging process. Table **17** may be controlled so that its upper surface translates along the z axis perpendicular to the x-y imaging plane, moving the imaged object **12** across the gantry plane **60**.

The detector array **14** is comprised of a number of detector elements **16**, organized within the gantry plane **60**, which together detect the projected image produced by the attenuated transmission of x-rays through the imaged object **12**.

The fan beam **22** emanates from a focal point **26** in the x-ray source **10** and is directed along a fan beam axis **23** centered within the fan beam **22**. The fan beam angle, measured along the broad face of the fan beam, is larger than the angle subtended by the imaged object **12** so that two peripheral beams **24** of the fan beam **22** are transmitted past the body without substantial attenuation. These peripheral beams **24** are received by peripheral detector elements **18** within the detector array **14**.

Referring to FIG. **6**, the x-ray source **10** includes an anode **29** position within an evacuated glass envelope and rotated about anode shaft **25** for heat dispersion. A stream of electrons from a cathode (not shown) is accelerated against the face of the anode **29** to produce the x-ray beam **19**. The face of the anode **29** is beveled with respect to the fan beam axis **23** so that radial displacement of the electron beam by focussing plates, (not shown) as is known in the art, will produce a z-axis displacement of the focal point **26**. The amount of this displacement may be controlled by x-ray controller **62**.

Referring to FIG. **2**, the angular position θ of the gantry **20** along the z-axis with respect to the imaged object **12** is shown by arrows **11**. The z-axis position of the imaged object **12** with respect to the imaging plane **60** changes constantly during the acquisition of each tomographic projection set. Accordingly, arrows **11** are shifted along a helix within the imaged object **12** along the z-axis. The pitch of the helix will be referred to as the scanning pitch. The z-axis distance from the center **9** of the slice being acquired to the volume elements **7** intercepting the fan beam **22** is termed the "helix offset" of that volume element. In the present invention the fan beam axis **23** may be shifted along the z-axis during the

helical scan to reduce the helix offset as will be described.

Referring to FIG. **3**, uncollimated x-rays **19** radiating from the focal point **26** in the x-ray source **10** (not shown in FIG. **3**) are formed into a coarse fan beam **21** by primary aperture **40**. As is understood in the art, the uncollimated x-rays **19** are produced by a high voltage x-ray tube typically including a rotating anode (not shown) receiving a high energy beam of electrons and re-emitting x-ray radiation. The coarse fan beam **21** is collimated into fan beam **22** by means of collimator **38**.

Referring generally to FIGS. **3**, **4**(a) and **4**(b), collimator **38** is comprised of a cylindrical x-ray absorbing molybdenum mandrel **39** held within the coarse fan beam **21** on bearings **42** allowing the mandrel **39** to rotate along its axis. A plurality of tapered slots **41** are cut through the mandrel's diameter and extend along the length of the mandrel **39**. The slots **41** are cut at varying angles about the mandrel's axis to permit rotation of the mandrel **39** to bring one such slot **41** into alignment with the coarse fan beam **21** so as to permit the passage of some rays of the coarse fan beam **21** through the slot **41** to form fan beam **22**.

Referring to FIG. **4**(a) and **4**(b), the tapered slots **41** are of varying width and hence the rotation of the mandrel **39** allows the width of the fan beam **22** to be varied between narrow (1 mm) as shown in FIG. **4**(b) and wide (10 mm) as shown in FIG. **4**(b). The slots **41** ensure dimensional accuracy and repeatability of the fan beam **22**.

The slots **41** are tapered so that the entrance aperture **43** of each slot **41**, when orientated with respect to the coarse fan beam **21**, is wider than the exit aperture **45**. The exit aperture **45** defines the width of the fan beam **22** and the extra width of the entrance aperture **43** prevents either edge of the entrance aperture **43** from blocking the coarse fan beam **21** during small angular rotation of the mandrel **39**. Such small rotations of the mandrel **39** are used to provide adjustment of the z-axis position of the fan beam **22** as will be discused in detail below.

Referring again to FIG. **3**, a positioning motor **48** is connected to one end of the mandrel **39** by flexible coupling **50**. The other end of the mandrel **39** is attached to a position encoder **46** which allows accurate positioning of the mandrel by motor **48**. Fan beam angle shutters **44** at either ends of the mandrel **39** control the fan beam angle.

Referring now to FIG. **5**, the control system of a CT scanner, suitable for use with the present invention, has gantry associated control modules **60** which include: x-ray controller **62** which provides power and timing signals to the x-ray source **10**, and which in certain embodiments of the invention, controls the position of the focal point **26**; collimator controller **64** which controls the rotation of the collimator **38**; gantry motor controller **66** which controls the rotational speed and position of the gantry **20**; and the data acquisition system **68** which receives projection data from the detector array **14** and converts the data to digital words for later computer processing.

The gantry associated control modules **60** communicate with the x-ray tube **10**, collimator **38** and detector **14** via slip rings **61**. It will be recognized that direct cabling using a take up reel may be substituted for the slip rings **61** for a limited gantry rotation system.

The x-ray controller **62**, the collimator controller **64** and the gantry motor **66** controller are connected to a

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computer 70. The computer 70 is a general purpose minicomputer such as the Data General Eclipse MV/7800C and may be programmed to synchronize the rotation of the gantry 20 with the position of the fan beam 22 per the present invention as will be described in detail below.

The data acquisition system 68 is connected to image reconstructor 72 which receives sampled and digitized signals from the detector array 14 via the data acquisition system 68 to perform high speed image reconstruction according to methods known in the art. The image reconstructor 72 may be an array processor such as is manufactured by Star Technologies of Virginia.

The speed and position of table 17 along the z-axis is communicated to and controlled by computer 70 through of table motor controller 74. The computer 70 receives commands and scanning parameters via operator console 76 which is generally a CRT display and keyboard which allows an operator to enter parameters for the scan and to display the reconstructed image and other information from the computer 70. A mass storage device 78 provides a means for storing operating programs for the CT imaging system, as well as image data for future reference by the operator.

Referring now to FIG. 6, the z-axis position of the exit aperture 45 of the collimator 38 may be adjusted so that the fan beam 22, as indicated by fan beam axis 23, diverges from the gantry plane 60 in the z-axis dimension during the acquisition of the first projection of a projection set. The amount of divergence of the fan beam axis 23 from the gantry plane 60 is such that a volume element 7 at position 80 within a slice and moving toward the gantry axis 60 with motion of table 17, is intersected by the fan beam axis 23.

The position of the table 17 during the acquisition of the projection set is determined by the table motor controller 74. The collimator 38 as controlled by the collimator controller 64 is coordinated by computer 70 with the position of table 17 so that during the movement of the table 17 and imaged object 12, the fan beam axis 23 is swept as to constantly intercept volume element 7.

As the projections of each projection set are acquired, during a period T_1, the imaged object 12 is translated along the z-axis with respect to the imaging plane 60 so that volume element ultimately moves to position 82 at the last projection of the projection set. Typically, the amount of translation will be equal to the slice thickness w.

At the completion of the acquisition of the projection set, the exit aperture 45 of the collimator 38 is returned to the position it had at the start of the projection set, moving in the opposite direction, during a period T_2, so that the fan beam axis 23 intercepts a new volume element in a new slice. The new volume element has the same relative position 80 with respect to the gantry plane 60 as did volume element 7 at the start of the acquisition of the previous projection set. Preferably, positions 80 and 82 are located symmetrically about the gantry plane 60 so as to reduce the maximum deviation of the fan beam axis 23 from the gantry plane 60 during any acquisition.

At the halfway point in the acquisition of the projection set, the focal point 26, the center line of the exit aperture 45 of the collimator 38, fan beam axis 23 and the center of illumination of the detector array 14 will be perfectly aligned with the gantry plane 60. At all other times, these various points may deviate from the

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gantry plane 60. The measures of the deviation of the center line of the exit aperture 45 of the collimator 38, the volume element intersected by fan beam axis 23, and the center of illumination of the detector array 14 from the gantry plane will be termed C_z, V_z, and D_z respectively.

For the first described embodiment shown in FIG. 6, F_z, the position of the focal point 26 with respect to the gantry plane 60 is constant and zero.

Referring to FIG. 10, during the first period T_1 of the acquisition of a projection set, the displacement of the collimator C_z will increase so that the fan beam axis 23 tracks the movement of the volume element 7. For large values of l_2 and l_3 and small values of slice thickness w, the relationship between the collimator displacement C_z and the displacement V_z of the fan beam axis 23 with axis of translation 84 of the volume element 7 is:

$$C_z = V_z \frac{l_1}{l_1 + l_2} . \tag{1}$$

where l_1 is the distance between the focal point 26 and the exit aperture 45 of the collimator 38 and l_2 is the distance between the exit aperture 45 and the translation axis 84 of the volume element 7. Accordingly, during the first period T_1, the position of table 17 as determined via the table motor controller 74, determines the position of the exit aperture 45 after suitable scaling by computer 70 as given in equation (1) above.

During a second time period T_2, immediately after the first time period T_1, the exit aperture 45 is returned to the position it had at the start of that acquisition of projections to prepare for acquisition of a second projection set. Preferably this period T_2 is made a short as possible by moving the collimator 38 at its maximum speed. During this return period T_z, no projection data is taken and the x-ray fan beam 22 may be decreased in intensity according to any of several methods known in the art such as decreasing current flow to the x-ray tube or shuttering the x-ray beam 19.

It will be noted that the displacement D_z of the fan beam axis 23 with respect to the surface of the detector array 14 will be larger than the displacement V_z according to the following ratio:

$$D_z = V_z \frac{l_1 + l_2 + l_3}{l_1 + l_2} \tag{2}$$

where l_3 is the distance between the axis of translation 84 of the volume element 7 and the exposed surface of detector array 14. Generally, the detector elements 16 of detector array 14 exhibit a change of sensitivity as a function of the z-axis position of their illumination. Hence a variation in D_z will introduce some variation into the projections measurements. This variation may be corrected by using the peripheral beams 24 and peripheral detector elements 18 to provide a reference for correcting sensitivity variation according to compensation methods understood in the art. One such method is given in U.S. Pat. No. 4,559,639 hereby incorporated by reference.

In a second embodiment shown in FIG. 7 and 11, both the x-ray focal point 26 and the exit aperture 45 of the collimator 38 are moved. Movement of the x-ray focal point 26 is accomplished by refocussing the electron beam on the anode 29 as has been previously described or by physical translation of the x-ray source 10

under the control of servo motors. The measure of the deviation of the focal point 26 from the gantry plane 60 will be termed: F_z. Referring to

FIG. 11, in this second embodiment, the intersection D_z of the fan beam axis 23 on the detector array 14 is maintained constant (at zero displacement) by controlling the displacement F_z of the focal point and the displacement C_z of the exit aperture 45 with respect to the displacement V_z of the volume element as follows:

$$F_z = V_z \frac{l_1 + l_2 + l_3}{l_3} \qquad (3)$$

and

$$C_z = V_z \frac{l_2 + l_3}{l_3} \qquad (4)$$

Referring to FIG. 14, the acquisition volume 86 within the imaged object 12 over which projection data is acquired in a non-helical scan will be approximately one half of acquisition volume of a helical scan: assuming that the scanning pitch times the rotation for one projection set is approximately equal to the slice thickness w. The present invention, as described in the above two embodiments, enlarges the acquisition volume over the non-helical acquisition volume 86 by flanking volumes 88 which are outwardly conically concave. This increase in acquisition volume represented by volumes 88 increases the helix offset of the projection data slightly but much less than that produced by helical scanning which adds areas 90 to effectively double the acquisition volume. In general, the greater the distance $l_1 + l_2$ in comparison to the radius of the image object 12 about the translation axis 84, the less the flanking volume 88 and thus the less the helix offset of the data.

Referring to FIGS. 14 and 8, a third embodiment of the invention eliminates the flanking volumes 88 and produces an acquisition volume 86 identical to that of non-helical scanning. Referring to FIG. 12, the displacement D_z of the collimator 38 and F_z of the focal point 26 are set equal to the displacement V_z of the volume element 7. The fan beam axis 23 is thus maintained parallel to the gantry plane 60 at all times.

In a fourth embodiment, shown in FIGS. 9 and 13, the displacement C_z of the exit aperture 45 of the collimator 38 is fixed (and equal to zero) and the displacement F_z of the focal point 26 is adjusted according to the following relationship:

$$F_z = -V_1 \frac{l_1}{l_2} \qquad (5)$$

The acquisition volume (not shown) for this method and the amount of displacement D_z of the fan beam axis 23 on the detector array 14 will be greater than the comparable quantities for the previously described method, for CT systems of similar dimensions as a result of the greater angular divergence of the fan beam axis 23 from the gantry plane 60 necessary to track a given volume element 7 without movement of the collimator 38.

For each of the above embodiments, the projection data for volume elements near volume element 7 on the translation axis 84, there will be little helix offset. To the contrary, the volume elements removed from volume element 7 and the translation axis 84 will have increasing amounts of helix offset for greater values of x and y

as dictated by the angle of the fan beam axis 23 with respect to the gantry plane 60.

For this reason, it may be desirable to position the volume element 7 and the translation axis 84 near internal structures of interest within imaged object 12.

The translation axis 84 will normally intersect the center of rotation 13 of the gantry 20. The center of rotation 13 and the translation axis 84 may both be moved within the imaged object simply by adjusting the height of table 17. Alternatively, the translation axis 84 may be moved independently from the center of rotation 13 by adjusting the fan beam angle 23 as a function of gantry rotation 28. This is most easily accomplished by modifying the apparent value of l_2 and l_3 used by computer 70 in the above embodiments as a function of gantry angle θ as follows:

$$l_2' = l_2 \cos(\theta + \alpha) (\Delta) \qquad (6)$$

$$l_3' = l_3 + \cos(\theta + \alpha) (\Delta) \qquad (7)$$

where α is the angle with respect to the center of rotation 13 between the volume of interest and gantry angle $\theta = O$, Δ is the distance between the volume of interest and the center of gantry rotation 13, and l_2' and l_3' are substituted into the above equations in place of l_2 and l_3 respectively.

For the embodiments shown in FIGS. 6, 7, and 9, it will be understood that the amount of helix offset, reduced as it is, also varies as a function of the order of the projection within the projection set. For example, when the starting and ending positions 80 and 82 of the volume of interest 7 are symmetrically displaced about the gantry plane 60, the centermost projections will have no helix offset and the starting and ending projections will have the most helix offset. For this reason, it is desirable to weight the projections so as to deemphasize the starting and ending projections and to emphasize the centermost projections of the projection set. Such weighting systems are disclosed in co-pending application 07/440,531 entitled:"Method for Reducing Patient Translation Artifacts in Tomographic Imaging" filed Nov. 22, 1989. Finally, for the first, second, and forth embodiments, where the center of illumination of the detector 14 changes during the acquisition of projections, it is important that the detector 14 be sufficiently wide so as to always receive the entire fan beam 22.

Many modifications and variations of the preferred embodiment which will still be within the spirit and scope of the invention will be apparent to those with ordinary skill in the art. For example, the collimator may be of a conventional bladed design. Further it will be apparent that this method is applicable to so called "forth generation" CT machines where the detector array 14 is stationary and may surround the imaged object 12. Clearly the the x-ray tube and collimator may be also mechanically translated and tipped as a single unit. Finally, the table motion need not be constant during the acquisition of successive projection sets but may be slowed, for example, during the period T_2 when the fan beam 22 repositions itself at a starting position. In order to apprise the public of the various embodiments that may fall within the scope of the invention, the following claims are made:

We claim:

1. A method of acquiring tomographic projection data of an imaged object comprising:

supporting and translating the imaged object concurrently along a translation axis;

projecting a beam of x-rays from an x-ray generator through the imaged object and alternately sweeping the beam along the translation axis in a first direction with translation of the imaged object during a first period, and in a second direction along the translation axis but counter to the translation of the imaged object during a second period;

receiving the beam from the x-ray generator with an x-ray detector array after it passes through the imaged object; and

holding the x-ray generator and x-ray detector in opposition around the imaged body and concurrently rotating the same around a center of rotation and the imaged object, in an gantry plane substantially perpendicular to the translation axis.

2. The method as recited in claim 1 wherein the x-ray exposure of the imaged object by the x-ray beam during the first period is greater than the x-ray exposure of the imaged object by the x-ray beam during the second period.

3. The method as recited in claim 2 wherein the intensity of the x-ray beam is reduced during the second period.

4. The method as recited in claim 1 wherein the sweeping of the x-ray beam during the first period is such as to maintain the beam centered on a predetermined volume element on the translation axis.

5. The method recited in claim 4 wherein the x-ray generator comprises:

a fixed x-ray source; and

a collimator having an aperture movable along the translation axis.

6. The method recited in claim 5 wherein the aperture of the collimator is controlled according to the following equation:

$$C_z = V_z \frac{l_1}{l_1 + l_2}$$

where V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

C_z is the distance along the translation axis between the center of the collimator and the gantry plane;

l_1 is the distance between the x-ray source and the collimator; and

l_2 is the distance between the collimator and the translation axis.

7. The method recited in claim 5 wherein the aperture of the collimator is controlled according to the following equation gantry angle θ as follows:

$$C_z = V_z \frac{l_1}{l_1 + l_2'}$$

where

$$l_2' = l\text{-}\cos(\theta + \alpha)\ (\Delta)$$

and where θ is the gantry angle;

V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

C_z is the distance along the translation axis between the center of the collimator and the gantry plane;

α is the angle, with respect to the center of rotation, between the volume of interest and gantry angle $\theta = O$;

Δ is the distance between the volume of interest and the center of gantry rotation;

l_1 is the distance between the x-ray source and the collimator; and

l_2 is the distance between the collimator and the translation axis.

8. The method recited in claim 4 wherein the x-ray generator comprises an x-ray source having a focal point movable along the translation axis and collimator having a aperture movable along the translation axis.

9. The method recited in claim 8 wherein the focal point and aperture of the collimator are controlled according to the following equations:

$$F_z = V_z \frac{l_1 + l_2 + l_3}{l_3}$$

and

$$C_z = V_z \frac{l_3 + l_2}{l_3}$$

where V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

F_z is the distance along the translation axis between x-ray source and the gantry plane;

C_z is the distance along the translation axis between the center of the collimator and the gantry plane;

l_1 is the distance between the x-ray source and the collimator;

l_2 is the distance between the collimator and the translation axis; and

l_3 is the distance between the translation axis and the x-ray detector.

10. The method recited in claim 8 wherein the focal point and the aperture of the collimator are controlled according to the following equations:

$$F_z = V_z \frac{l_1 + l_2' + l_3'}{l_3'}$$

and

$$C_z = V_z \frac{l_3' + l_2'}{l_3'}$$

where

$$l_2' = l_2\text{-}\cos(74 + \alpha)\ (\Delta)$$

$$l_3' = l_3 + \cos(\theta + \alpha)\ (\Delta)$$

and where θ is the gantry angle;

α is the angle, with respect to the center of rotation, between the volume of interest and gantry angle $\theta = O$;

Δ is the distance between the volume of interest and the center of gantry rotation;

V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

F_z is the distance along the translation axis between x-ray source and the gantry plane;

C_z is the distance along the translation axis between the center of the collimator and the gantry plane;

l_1 is the distance between the x-ray source and the collimator;

l_2 is the distance between the collimator and the translation axis; and

l_3 is the distance between the translation axis and the x-ray detector.

11. The method recited in claim **8** wherein the focal point and the aperture of the collimator are controlled according to the following equation:

$$F_z = C_z = V_z$$

where V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

F_z is the distance along the translation axis between x-ray source and the gantry plane; and

C_z is the distance along the translation axis between the center of the collimator and the gantry plane.

12. The method recited in claim **4** wherein the x-ray generator comprises an x-ray source having a focal point movable along translation axis and a fixed collimator aperture.

13. The method recited in claim **12** wherein the focal point is controlled according to the following equation:

$$F_z = -V_z \frac{l_1}{l_2}$$

where F_z is the distance along the translation axis between x-ray source and the gantry plane;

V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

l_1 is the distance between the x-ray source and the collimator; and

l_2 is the distance between the collimator and the translation axis.

14. The method recited in claim **10** wherein the focal point is controlled according to the following equation:

$$F_z = -V_z \frac{l_1}{l_2'}$$

where

$$l_2' = l_2 - \cos(\theta + \alpha)\,(\Delta)$$

and where θ is the gantry angle;

α is the angle, with respect to the center of rotation, between the volume of interest and gantry angle $\theta = 0$;

Δ is the distance between the volume of interest and the center of gantry rotation;

F_z is the distance along the translation axis between x-ray source and the gantry plane;

V_z is the distance along the translation axis between the predetermined volume element and the gantry plane;

l_1 is the distance between the x-ray source and the collimator; and

l_2 is the distance between the collimator and the translation axis.

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UNITED STATES PATENT AND TRADEMARK OFFICE
CERTIFICATE OF CORRECTION

PATENT NO. : 5,090,037

DATED : February 18, 1992

INVENTOR(S) : Toth et al.

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

Col. 5, line 9, $1_3'$ should be $1'_3$.

Col. 8, line 37, T_z should be T_2.

Col. 12, line 53   (74 + a) should be $(\theta + a)$.

Col. 13, line 23   "along translation" should be -- along the translation --.

Signed and Sealed this

Thirteenth Day of April, 1993

Attest:

STEPHEN G. KUNIN

Attesting Officer   *Acting Commissioner of Patents and Trademarks*