# PATENT SPECIFICATION

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## (54) X-RAY IMAGING SYSTEM

(71) We, GENERAL ELECTRIC COMPANY, a corporation organized and existing under the laws of the State of New York, United States of America, of 1 River Road, Schenectady 12305, State of New York, United States of America, do hereby declare the invention, for which we pray that a patent may be granted to us, and the method by which it is to be performed, to be particularly described in and by the following statement:-

This invention relates to computerized x-ray tomographic imaging apparatus.

Computerized x-ray tomography is a system for producing images of internal body organs which are free from the shadows of intervening structures. Prior art tomographic equipment, has generally, comprised an x-ray source disposed opposite one or more x-ray detectors on a movable structure. The source and detectors rotate and/or translate in a plane which passes through the body organs undergoing examination to produce electrical signals which are representative of x-ray transmission data along a plurality of ray paths. The signals are then combined, usually in digital computer equipment, to reconstruct shadow-free images of internal body sections. Tomography equipment of this type is described, for example, in U.S. Patent 3,778,614 to Hounsefield.

The rate of production of images in a tomography system which incorporates moving sources and detectors is necessarily limited by the time required to accomplish the physical translation or rotation of the mechanism and is, typically, limited to less than one image per second. Such equipment is, therefore, unsuited for viewing moving body organs, for example, a beating heart. Dr. Earl Wood of the Mayo Clinic has recently proposed a tomographic system for imaging moving body organs wherein a plurality of x-ray sources are sequentially pulsed to rapidly produce x-ray transmission data along a number of diverse ray paths.

The x-ray detectors utilized in prior art x-ray tomography apparatus have generally comprised scintillation crystals or phosphor screens coupled to optical detectors, for example, image orthicon or photomultiplier tubes. Such devices are rather large and must, generally, be utilized with collimation apparatus to achieve fine spatial resolution. Such scintillation detectors and collimation apparatus are, relatively inefficient detectors of x-ray energy. It is, therefore, necessary to expose a patient undergoing tomographic examination in such equipment to a relatively high dose ionizing radiation.

Our copending U.K. patent application No. 39713/76 (serial no 1561174) describes a high pressure, xenon filled ionization chamber array which is characterized by high detection efficiency and fine spatial resolution when utilized in x-ray tomography equipment. The detector comprises a large plurality of detector cells separated by substantially parallel metal collector plates which may be focused on a single source of diverging x-rays. X-ray photons entering the detector cells produce ion-electron pairs which drift under the influence of an electric field, to the collector plates. Detectors of this type are well suited for the efficient detection of diverging x-ray energy which, for example, may be produced from a single x-ray source and collimated to provide a planar, fan-like spatial distribution. The ion chamber array of that disclosure is, however, relatively inefficient for detecting x-ray energy which originates from an array of spatially separated x-ray sources of the type utilized in the above-described, high speed tomographic equipment.

The present invention provides a high

speed, tomographic, x-ray imaging system comprising an array of x-ray sources adapted to be disposed about a body undergoing examination; an array of x-ray detectors adapted to be disposed about said body opposite said array of x-ray sources; and means for pulsing said x-ray sources in a sequence of salvos, said system further comprising collimating means adapted to be disposed between each of said x-ray sources and said body, said collimating means being adapted to confine the emission of x-ray energy from each x-ray source into an x-ray beam which illuminates only a fraction of the x-ray detectors in said x-ray detector array; and said means for pulsing said x-ray sources being adapted to simultaneously pulse x-ray sources in each salvo, each of said salvos comprising two or more selected x-ray sources in said array, the particular x-ray sources in each of said salvos being selected so that each source in a salvo illuminates a separate and distinct fraction of said x-ray detectors and no x-ray detector is illuminated by more than one source in each salvo.

The invention further provides a method for generating tomographic image data from body structures, comprising the steps of: disposing an array of x-ray sources about said body structures; disposing an array of x-ray detectors about said body structures opposite said array of x-ray sources; collimating x-ray energy from each of said sources into a beam which illuminates only a fraction of said x-ray detector array; and simultaneously pulsing a subset of x-ray sources in said array, the sources in said subset being selected so that x-ray energy from each said simultaneously pulsed source illuminates a separate and distinct fraction of said x-ray detectors and no x-ray detector is illuminated by more than one source in each subset, whereby x-ray transmission data along a plurality of ray paths is obtained in a fast and unambiguous banner.

An ionization chamber array suitable for use in this tomographic equipment comprises a comb-like array of collector electrodes of a first polarity disposed equidistant between two parallel sheet electrodes of the opposite polarity and immersed in a high pressure, ionizable gas. X-ray energy enters the detector in a direction substantially parallel to the comb-like electrodes and interacts with the detector gas to produce electron-ion pairs. The electrons and ions drift under the influence of an electric field, in a direction sub stantially perpendicular to both the direction of the incident x-ray beam and the linear array direction, to the collection electrodes The detector cells of the present array are not focused on a single x-ray source, as were the cells of the array described in the above-referenced patent disclosure, and are therefore well suited for use in tomography systems having multiple, spatially distributed, x-ray sources.

In order that the invention may be clearly understood, a preferred embodiment thereof will be described by way of example only with reference to the accompanying drawings, in which:-

FIG. 1 is a high speed tomography system of the prior art;

FIG. 2 is a high speed tomography system according to the preferred embodiment of the present invention;

FIG. 3 is a single detector cell of the prior art; and

FIG. 4 is an ionization chamber array useful with the present invention.

FIG. 1 illustrates a high speed x-ray tomography system of the prior art. An array of pulsed x-ray sources 20 is disposed opposite an array of x-ray detectors 22. Each individual x-ray detector of the array 22 comprises a phosphor screen 24 adapted to emit light inproportion to incident x-ray intensity. Light from the screen 24 is focused by a lens 26 on a television camera type pick-up tube typically an image orthicon 28. Electrical signals, from each tube 28, which represent a linear distribution of x-ray intensities across the width of a screen 24, are transmitted to a digital computer for processing into x-ray tomographic images.

Body structures 30 undergoing examination are interposed between the source array 20 and the detector array 22. Each x-ray source in the array 20 is sequentially pulsed to produce a swath of ionizing radiation 32 which is attenuated in varying degreees by the body structure 30 and impinges on the detector array 22. The individual elements of the array 20 may be pulsed in rapid sequence to provide x-ray transmission information along a plurality of intersecting paths through the body structure 30 from which image information may be constructed. Each source in the array 20, however, necessarily illuminates substantially the entire detector array 22 and the rate of sequential pulsing of the individual sources is, therefore, necessarily limited by the speed at which data may be read from the detector elements (i.e., pick-up tubes 28). Information produced from a single x-ray pulse must be completely read out from a detector 28 before another source in the array 20 is pulsed in order to eliminate a redundancy of information which would occur if x-rays from two sources reached the same tube during a single readout period.

FIG. 2 is an improved high speed tomography system according to a preferred embodiment of the present invention. A substantially semicircular array of x-ray sources 20 comprises a plurality of individual x-ray anodes 40 separated by an array of collimators 42. The geometry of the collimators 42 is selected so that the x-ray beam from each anode 40 is restricted to a substantially planar, sectoral swath. X-ray energy in each swath passes through a body structure 30 and impinges on a curvilinear array of closely spaced, ionization chamber detectors 44 disposed in the plane of the x-ray swath. The dimensions and geometry of the collimators 42 are adapted to limit the width of each x-ray swath so that it illuminates only a relatively small sector of the array 44. Thus, in the illustration of FIG. 2, x-rays from anode 40*a* pass through collimator 42*a* to form a sectoral swath 46*a* which impinges on a small group 48*a* of detectors in the array 44. Likewise, x-rays from the anode 40*b* pass through the collimator 42*b* and impinge on a separate and distinct group 48*b* of detectors in the array 44.

The x-ray sources in the array 20 are pulsed in salvos, the grouping of sources in each salvo being chosen so that the individual sources 40 in the salvo illuminate separate and distinct groups of detectors in the array 44. After each salvo of sources is pulsed, data from the detectors is read out into a digital computer for processing and another salvo of sources, similarly chosen to illuminate other distinct detector groups, is pulsed. Each detector in the array 44 will, in general, receive x-ray data from a number of sequential salvos. The grouping of the individual detectors which are illuminated by the sources in each salvo will, however, change. The number of sources which may be simultaneously pulsed in each salvo is, of course, a function of the system and collimator geometry which, in turn, is determined by the size of the body structure examined, the spatial resolution required, and the desired image production time. Depending on this geometry and the number of detectors and sources in the arrays, the speed of image processing may be increased by a factor of two or more over prior art tomography systems.

The detector array 44 may comprise ionization chambers of the type described in the above-mentioned U.K. patent application No. 39713/76, (serial no. 1561174) which is incorporated herein, by reference, as background material. That detector comprises an array of detector cells defined between individual sheet collector electrodes which are disposed substantially parallel to the direction of the incident x-ray beam and perpendicular to the plane of the x-ray swath. The individual cells of such a detector are focused on a single source to provide high efficiency x-ray collection and detection, and such a detector is well suited for use in convential tomography apparatus which comprises a single x-ray source. When used in a multiple source tomography system, this array suffers from a substantial loss of detection efficiency for x-rays which originate off the focal point of its individual cells. The cause of this inefficiency may be noted by reference to FIG.2 and FIG. 3 which is an enlarged view of an individual detector cell illuminated by the x-rays from a source lying outside its focal region and incident on the plane of the cell at an angle $\beta$. If R is the radius of the detector arc and P is the radius of the field of view at the body 30, the maximum value of the angle $\beta$ occurs at the edge of each view such that $\sin \beta = P/R$. In a typical system of the type illustrated in FIG. 2 used, for example, for viewing a beating heart, P equals approximately 20 centimeters and R equals approximately 75 centimeters. yielding a maximum angle, $\beta$, of 16°. The efficiency of the cell for oblique detection angles is determined by the spacing of the collector electrodes *d*. The spacing *d* is determined, among other factors, by the degree of spatial resolution required by the system and by the time required for the electrons and ions produced within a cell to drift under the influence of an electric field to the individual electrodes. If, for example, the cell is filled with a xenon detector gas at a pressure of approximately 20 atmospheres, a spacing *d*, of approximately 1 mm is required to obtain a 1 millisecond response time. The response of such a cell, with 1 millimeter electrode spacing, for x-rays incident at an angle $\beta$ of 16° will be only approximately 14 percent of its efficiency for x-rays incident at an angle of 0°. This loss of efficiency introduces serious calibration problems into image reconstruction algorithms and necessarily increases the radiation dose which is required to produce an image of given resolution. The calibration problem in a multiple source array is, of course, greatly increased by the fact that the angle of incidence of x-rays on each cell is different for each detector and large numbers of calibration factors must, therefore, be stored and utilized.

FIG. 4 is an ion chamber array which has a substantially constant detection efficiency for varying angles of x-ray incidence. A pair of planar, conductive anodes 50 and 52 are disposed parallel to an incident sheet of x-ray radiation 54. A plurality of rod-like cathodes 56 are disposed, equi-distant between the anodes 50 and 52 and substantially parallel, one to the other, with their longest dimension generally parallel to the incident x-rays. One terminal of a voltage source 58 is connected to the anode sheets 50 and 52. Each of the cathodes 56 is connected through one of a purality of current detector circuits 60 to the other terminal of the voltage source 58. In a preferred embodiment of the invention, a common node of the voltage source and the current detectors represents ground potential.

It will be recognized, by those skilled in the art, that the polarity of the voltage source and the position of the ground connection may be varied without affecting the utility of the invention and that the designation of the collection electrodes 50, 52, and 56, as anodes and cathodes is for ease of description only.

A detector gas 62 fills the space between the anode sheets 50 and 52 and the cathodes 56. The gas type, gas pressure, and the spacing W between the electrodes are chosen using methods well known to the art so that a large fraction (typically more than 70 percent) of the incident x-ray photons are absorbed within the gas. The detector gas 62, typically comprises rare gas of high atomic number; for example, xenon, krypton, argon, or a molecular gas comprising atoms having an atomic weight greater than that of argon (i.e., 39.9); at a pressure in the range from approximately 10 atmospheres to approximately 100 atmospheres.

Incident x-rays 54 interact with the detector gas 62 between the anodes 50 and 52 to produce electron-ion pairs. The electrons drift under the influence of the electric field, imposed by the voltage source 58, to the anode plates 50 and 52 while the ions are similarly collected on the cathodes 56. Ion current flow to any individual cathode 56 is proportional to the number of interactions between photons and gas atoms in the region of that cathode so that the distribution of current flow among the individual current detector circuits 60 of the array is a function of the distribution of x-ray intensity along the detector array. The direction of electron and ion motion within the detector is substantially perpendicular to the array length and to the incident x-ray beam.

The cathodes 56 may be arrayed parallel one to the other to produce a linear detector array. Alternativey, the detectors may ie at small angles, one to the other, to define a curved or semicircular array of the type illustrated in Fig. 2.

The tomography system of the present invention allows fast and accurate imaging of internal body organs and is insensitive to the blurring effects which motion of those organs tends to produce in prior art systems. The system is also highly efficient for producing moving pictures of body organs, for example, of a beating heart.

WHAT WE CLAIM IS:—

1. A high speed, tomographic, x-ray imaging system comprising an array of x-ray sources adapted to be disposed about a body undergoing examination; an array of x-ray detector adapted to be disposed about said body opposite said array of x-ray sources; and means for pulsing said x-ray sources; in a sequence of salvos,

said system further comprising collimating means adapted to be disposed between each of said x-ray sources and said body, said collimating means being adapted to confine the emission of x-ray energy from each x-ray source into an x-ray beam which illuminates only a fraction of the x-ray detectors in said x-ray detector array; and

said means for pulsing said x-ray sources being adapted to simultaneously pulse x-ray sources in each salvo, each of said salvos comprising two or more selected x-ray sources in said array, the particular x-ray sources in each of said salvos being selected so that each source in a salvo *illuminates a separate and distinct fraction* of said x-ray detectors and no x-ray detector, is illuminated by more than one source in each salvo.

2. An imaging system according to claim 1 wherein said array of x-ray sources is a substantially semicircular array.

3. An imaging system according to claim 1 or claim 2, wherein said array of x-ray detectors is a substantially semicircular array.

4. An imaging system according to any one of claims 1 to 3, wherein said x-ray detectors are ionization chambers.

5. An imaging system according to any one of claims 1-4, wherein said collimating means are further adapted to confine the emission of x-rays emerging from each of said sources into a substantially planar, fan-like beam.

6. A method for generating tomographic image data from body structures, comprising the steps of:

disposing an array of x-ray sources about said body structure;

disposing an array of x-ray detectors about said body structures opposite said array of x-ray sources;

collimating x-ray energy from each of said sources into a beam which illuminates only a fraction of said x-ray detector array array; and

simultaneously pulsing a subset of x-ray sources in said array, the sources in said subset being selected so that x-ray energy from each said simultaneously pulsed source illuminates a separate and distinct fraction of said x-ray detectors and no x-ray detector is illuminated by more than one source in each subset, whereby x-ray transmission data along a plurality or ray paths is obtained in a fast and unambiguous manner.

7. A method according to claim 6, comprising the additional step of sequentially pulsing additional subsets of x-ray detectors in said array, the individual detectors in each of said additional subsets being selected so that x-ray energy from each of said detectors in a subset illuminates a separate and distinct fraction of the detectors in said array.

8. A high speed, tomographic x-ray imaging system according to claim 1, substantially as hereinbefore described with reference to and as shown in Figs. 2 and 4 of the accompanying drawings.

9. A method according to claim 6 for generating tomographic image data from body structures, substantially as hereinbefore described.

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1562768 COMPLETE SPECIFICATION

2 SHEETS *This drawing is a reproduction of the Original on a reduced scale*

Sheet 1

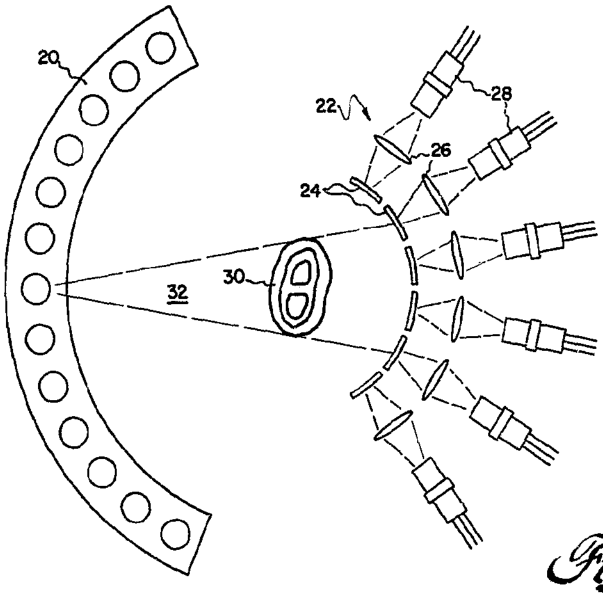



*Fig. 1*
(PRIOR ART)

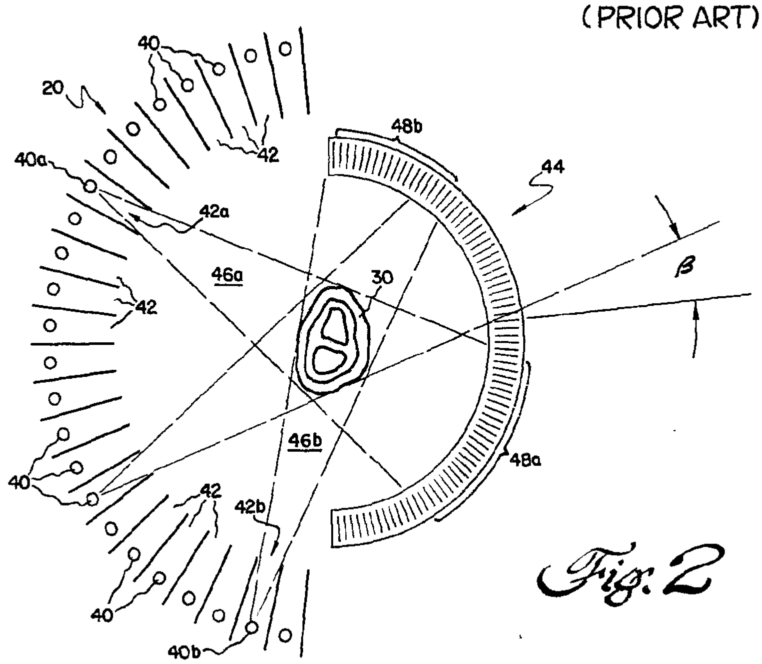



*Fig. 2*

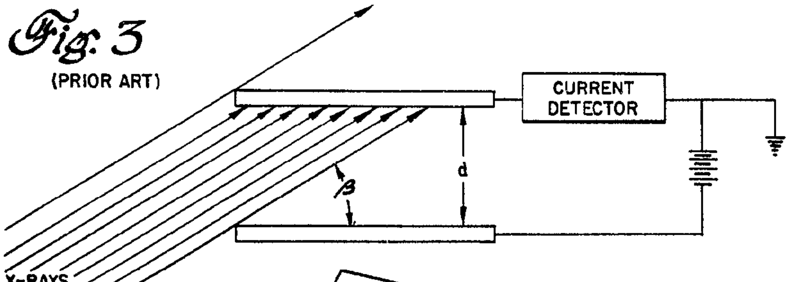
1562768 COMPLETE SPECIFICATION

2 SHEETS *This drawing is a reproduction of the Original on a reduced scale*

Sheet 2

Fig. 3

(PRIOR ART)



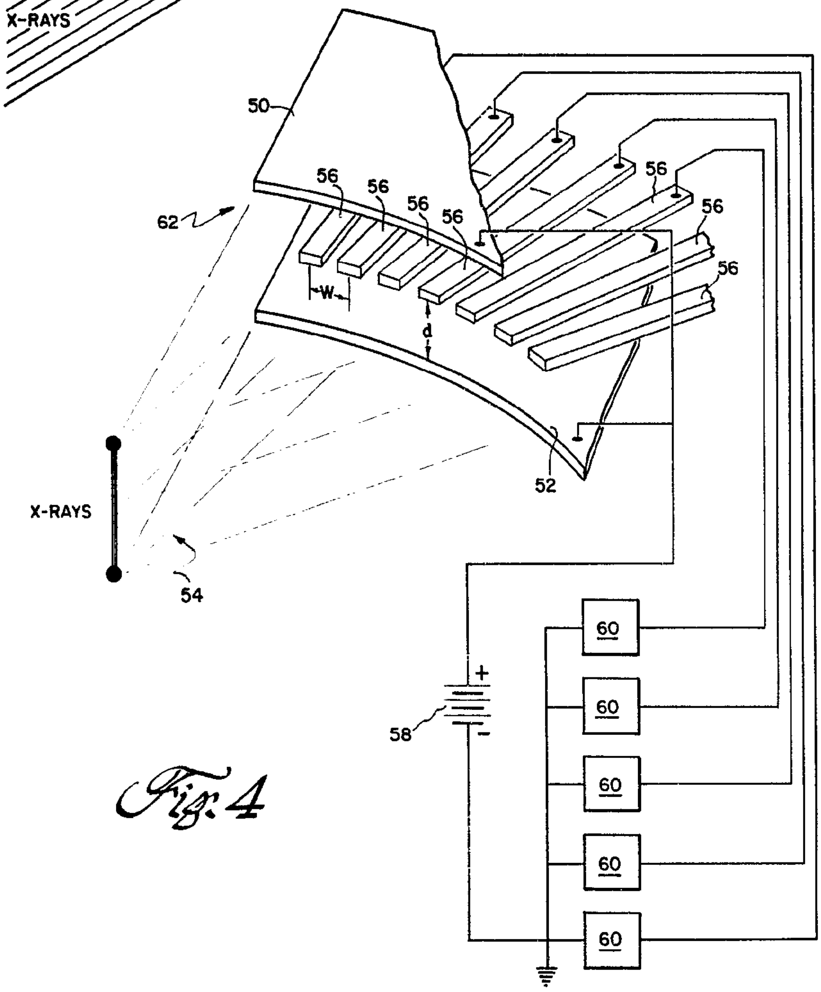




Fig. 4