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(19)

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(54) IMPROVEMENTS IN OR RELATING TO RADIOGRAPHY

(71) We, EMI LIMITED, a British company of Blyth Road, Hayes, Middlesex, 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—:

The present invention relates to radiography, and it relates more particularly to that branch of radiography which has become known as computerised tomography.

Computerised tomography enables the absorption coefficient, with respect to penetrating radiation projected through a body, to be evaluated at each of a number of elemental areas distributed over a cross-sectional slice of the body. Apparatus for performing computerised tomography is described and claimed in British Patent Specification No. 1,283,915.

Since the considerable value of computerised tomography as a diagnostic tool has become recognised, much effort has been expended in investigating ways in which the acquisition of the data, required for the abovementioned evaluation, can be accelerated. Commensurate increases in the rate at which the acquired data are processed to effect the evaluation are also sought.

A considerable increase in data acquisition rate can be achieved by following the procedures described and claimed in British Patent Specification No. 1,430,089, wherein a fan of radiation of sufficient breadth to span the body slice under examination is projected through the body and falls upon an extended array of detectors of sufficient breadth to accommodate the fan. In that case, the radiation source and the detectors are rotated synchronously around the body slice; the detector outputs being periodically sampled to permit the derivation of data indicative of the amounts of radiation emergent from the body slice along many substantial linear paths through the slice. Another technique permitting rapid data acquisition is disclosed and claimed in the Complete Specification of British Patent Application No. 8417/76 (equivalent to German Offenlegungsschrift, 'DT', 2709600 A1) in which the detector array is not merely sufficient to accommodate the fan of radiation but extends around the body slice to an angular extent of at least 180°. The detectors can then remain stationary while the source alone rotates around the body slice. This technique is, of course, more costly in terms of detectors and associated circuits but it has certain advantages over techniques in which the source and detectors rotate around the body.

The most expeditious processing technique known at present is that described and claimed in British Patent Specification No. 1,471,531. This involves a form of convolution of the acquired data.

It is clearly desirable to use the rapid acquisition techniques of the aforementioned British Patent No. 1,430,089 or Patent Application No. 8417/76 (equivalent to DT 2709600 A1) in conjunction with the expeditious convolution processing technique of the aforementioned British Patent No. 1,471,531. However the said processing technique is best applied to data presented thereto in groups relating to parallel paths followed by the radiation through the body slice.

It is known that data acquired in fan beam geometry can be sorted into groups relating to parallel sets of paths prior to convolution processing; such an operation of sorting being described and claimed in British Patent Application No. 4562/74 now Patent No. 1,493,594. When this is done, however, it is found that the paths of each parallel set exhibit non-uniform spacing unless steps are taken to the contrary. In British Patent Application No. 19681/76 (equivalent to U.S. Patent Specification 4,138,611) it is shown that errors introduced by the non-uniformity of spacing can be allowed for by the use of suitable multiplying factors generated by examination of a phantom body of known absorption characteristics. However it is desirable to avoid this sorting procedure and to effect convolution directly on data relating to divergent fans of beam paths, and some complex mathematical procedures have been devised to this end. By way of example, the following documents are referred to as describing such complex mathematical techniques:—

1. Reconstruction from Divergent Ray Data by A. V. Lakshminarayanan (Technical Report No. 92) State University of New York at Buffalo, Department of Computer Science, January 1975.

2. Tomographic Reconstruction from Fan Beam Geometry Using Radon's Integration Method by John W. Beattie (I.E.E.E. Transactions on Nuclear Science, Vol. NS-22, February 1975 pp. 359—363).

The present invention aims at permitting data acquired in fan beam geometry to be processed by convolution without the need for either sorting into groups relating to parallel paths or complex mathematical procedures.

According to the invention there is provided radiographic apparatus comprising a source of radiation projected towards a body, said radiation fanning out in one dimension as it proceeds away from said source to produce a flat, fan-shaped distribution which spans at least a substantial proportion of the extent of said body in said one dimension, scanning means for scanning said source angularly around said body to project radiation through said body from many different directions, detecting means for detecting the radiation emergent from said body along many groups of mutually inclined, substantially linear paths during the angular movement of said source and for producing corresponding groups of electrical signals indicative of the amounts of radiation emergent from the body along said groups of paths, processing means for processing the electrical signals corresponding to said groups of paths to produce a representation of the variation of absorption of said radiation with position over a cross-sectional slice of said body, and wherein said processing means includes means for combining each output signal with weighted components of other output signals of the same group in proportions appropriate to parallel beam paths, and compensating means, including a store of compensating factors each applicable to a respective region of such slice, evaluated during a preliminary opration of the apparatus with a body of known absorption characteristics, and means for combining evaluated absorption values for a body of unknown absorption characteristics with appropriate compensating factors prior to the production of said representation.

In order that the invention may be clearly understood and readily carried into effect, one embodiment thereof will now be described, by way of example only, with reference to the drawings accompanying the Provisional Specification, of which:—

Figure 1 shows, in plan view and partly in block diagrammatic form, a radiographic apparatus in accordance with one example of the invention, and

Figure 2 shows, in flow diagrammatic form, one way in which the invention can be carried out.

Referring now to Figure 1, an X-ray source, for example a rotating anode X-ray tube 1, is mounted upon an apertured turntable 2 which rotates, in a suitable bearing not shown, relative to a static main frame 3. The aperture in the turntable 2 is shown at 4 and is dimensioned to accommodate the body of a human patient (not shown) lying supine or prone on an elongated table; or platter, shown at 5. The table 5 is supported by the main frame 3 and the floor of the building in which the apparatus is located, but the supports for the table are not shown in the drawing as they may take any of a number of convenient and well known forms.

Also shown is a disc 6 of uniform, or at least known, absorption properties and the disc 6 can be placed in the aperture 4, in place of the human patient, for a purpose which will become clear later. The disc 6 will hereinafter be called a "phantom" which is the terminology used in the art for such devices.

Mounted on the turntable 2, and disposed at the opposite side of the aperture 4 to the source 1, is an array 7 of radiation sensitive detectors. These detectors, in this example, are closely packed together and extend across the whole breadth of a fan-shaped distribution 8 of X-radiation generated by the source tube 1. The detectors can conveniently comprise scintillator crystals, such as sodium iodide or caesium iodide crystals, optically coupled to respective devices for converting visible radiation into electrical signals. These converter devices can be, for example, photomultiplier tubes or semiconductive photodiodes. In any event, each detector produces electrical output signals indicative of the amount of radiation received.

The turntable is rotated, by means not shown, around the patient position defined by the aperture 4 so as to cause the source 1 to follow an arcuate locus around the patient position whereby the body disposed at the patient position can be irradiated from many different directions. The output signals produced by the detectors are effectively sectioned up, by an integration and resetting technique which will be described in more detail hereinafter, so that each output signal represents the amount of radiation projected through the patient position to a detector along a substantially linear beam path. The integration and resetting occurs a great many times during the rotation of the turntable around the patient position, and the integrators associated with all of the detectors in the array 7 are reset synchronously, so the effect is that, at each resetting instant, the array of detectors produces, in total, a group of output signals (one from each detector) relating to a group of divergent beam paths spanning the distribution 8 of radiation. The effect is substantially that of stopping the turntable 2 in a sequence of angular positions and of obtaining output signals while the turntable is stopped.

It will be appreciated that, since each output signal is required to relate to a substantially linear beam path through the patient position, it must not be contaminated by radiation which could, due to scatter of radiation within a body disposed at the patient position, reach the detector by other routes. For this reason, each detector views the source through a respective collimator, and an array of collimators is shown at 9. A source collimator 10 is also provided to ensure that the radiation conforms to the required fan-shaped distribution, and this collimator 10 preferably contains a beam splitting arrangement of the kind described and claimed in British Patent Application No. 41,582/75 (Serial No. 1,561,964) to reduce spreading of the radiation in the direction perpendicular to the plane of the fan.

The progress of the rotation of the turntable 2 around the patient position has to be monitored, and to this end a graticule 11 of reflective or transmissive form and containing equi-angularly spaced line markings is formed on the turntable 2 and co-operates in known manner with a photocell and light source assembly, generally shown at 12, mounted on the main frame 3, to produce electrical timing signals indicative of the motion of the turntable.

The timing signals derived from the assembly 12 are applied to a master timing circuit 13 which controls the operation of many of the circuit components now to be described.

An output signal connection 14r is shown as originating from the r'th detector in the array 7 and it will be appreciated that each detector in the array 7 has a separate output signal connection and that each such connection feeds a respective pre-processing circuit block, such as that shown at 15r, containing, in series connection, an amplifier such as 16r, an integrator circuit such as 17r, an analogue-to-digital converter such as 18r and a logarithmic converter circuit such as 19r. It will be appreciated that the integrator circuits such as 17r effect the sectioning up of output signals from the respective detectors which was referred to earlier, and that the resetting of the integrator circuits is effected in response to signal sgenerated by the master timing circuit 13 under the control of the timing signals produced by the photocell/light source assembly 12.

All of the pre-processing circuits such as 15r feed a main convolution processing circuit 20 which can take any convenient form but preferably takes one of the forms described in the aforementioned British Patent Specification No. 1,471,531. The processing circuit 20 produces, in known manner, values appropriate to each of a large number of elemental regions of a cross-sectional slice of the body disposed at the patient position and indicative of the absorption, of the radiation generated by the source 1, at those regions. These regions are commonly referred to as "pixels" and the values produced by the circuit 20 are applied to respective storage locations of a pixel value store 21. The store 21 has at least as many storage locations as there are individual pixels.

Successive groups of output signals from the array 7 of detectors are applied, as generated, to the processing circuit 20 and although, as mentioned previously, each group of output signals relates to a group of divergent beam paths through the patient position, the processing is effected as though the group related to a set of parallel beam paths, at a suitable spacing, through the patient position. In essence, the processing involves modifying each output signal of each group by combining, in a negative sense therewith, variously weighted components of the other output signals of the same group; the weighting being in accordance with a law, or function, which is monotonic and decreases in amplitude with increasing distance of the beam path giving rise to the output signal being weighted from the beam path giving rise to the output signal being modified. The full procedure is described in the aforementioned British Patent No. 1,471,531 which is incorporated herein by reference.

Components of the modified output signals for each path are then distributed among the locations of store 21 appropriate to the pixels actually intersected by the beam path in question. Due account is taken, in known manner, of the fact that the beam paths intersect different pixels to different extents.

It will be appreciated that each pixel is intersected by a large number of beam paths and thus, as successive groups of output signals are applied to the circuit 20, values will accumulate in the storage locations of the store 21 which successively move closely approximate to the actual absorption values of the respective pixels. However because, as mentioned previously, the convolution circuit 20 operates as if each group of output signals related to a set of parallel beam paths, and not the divergent paths followed in practice by the radiation (although as pointed out, the distribution of signal components to the pixel value store 21 — known in the art as "back-projection" — is done in accordance with the actual divergent beam paths) some errors will exist in the pixel absorption values as finally evaluated.

In accordance with this example of the invention, these errors are eliminated, or at least reduced, by producing a series of multiplying factors, one for each pixel. These multiplying factors are derived by operating the apparatus in a calibration mode, selected by a switch 22, while the phantom 6 is disposed in the patient position. In the calibration mode, the apparatus works exactly in the manner described hereinbefore, but once the convolution processing is complete and the pixel value store 21 contains the best approximations to the pixel absorption values, these absorption values are compared, on a pixel-by-pixel basis, with the actual absorption values which are known to exist at the relevant pixels of the phantom.

This comparison stage can be effected automatically, as shown in the drawing, by providing a store 23, of identical form to the store 21; the store 23 containing the correct (known) absorption values for each pixel. Values for the same pixel are derived simultaneously from the two stores 23 and 21 and the former value divided by the latter in a dividing circuit 24 to obtain a multiplying factor for that pixel which is routed to the appropriate storage location of a multiplying factor store 25 which also has a storage location for each pixel. Multiplying factors for all of the pixels are obtained in sequence in the same way and it will be appreciated that each multiplying factor is dimensioned so that, when multiplied with the evaluated absorption value for a pixel, as stored in store 21, it produces the correct absorption value for that pixel, as stored in store 23.

Returning now to operation of the apparatus with a patient's body in the patient position, the values stored in store 21 are multiplied, in a multiplying circuit 26, with the appropriate multiplying factors and each corrected pixel value is then inserted into store 21 at the location previously occupied for the uncorrected value for the same pixel. Once this correction has been effected for all pixels, then the store 21 is connected to a video display unit which produces the aforementioned visual representation of the absorption values held in the store 21. The video display unit can be a complex console which includes a cathode ray tube display with photographic facilities, a numerical print-out and possibly other forms of display. Window-width (i.e. dynamic range) and window level (adjustment of centre value of range) can also be provided if required. Such controls are known in the art.

Hitherto, the extent of the rotation effected by the turntable 2 has not been specified. There are special advantages to be had in rotating through 360°, thus irradiating the patient position along groups of beam paths symmetrically distributed around the body. If required, however, lesser amounts of rotation (such as 180°) can be used, provided that some residual errors can be tolerated due to the fact that (owing to the divergent nature of the beam paths in each group) the resolution will be higher on one side of the patient position (where the paths are closer together) than it will be on the other side of said position. Rotation through 360° overcomes that problem.

The angle of the distribution 8 of radiation can typically be 30° or more and a typical number of detectors is about nine per degree. Alternatively, of course, the detectors need not rotate with the source, in which case a large number of detectors may be arranged in a circle around the patient position. In that latter case, it will be appreciated that, as the source rotates relative to the detectors, each detector views in sequence a number of inclined beam paths. These beam paths intersect at the detector, of course, and are distributed over a fan of angle determined by the angle of the distribution of X-rays produced by the source. In this case, it is possible to process the data in terms of fans originating from the source (as in the case described above) or in terms of fans notionally originating (but in fact terminating) at the detectors.

In practice it is found that many of the multiplying factors are equal, or substantially so, and therefore the number of storage locations required in the multiplying factor store 24 can be substantially less than the number of pixels.

Instead of effecting the correction on a pixel-by-pixel basis after evaluation of the absorption values, it can be effected during the aforementioned stage of back-projection.

Figure 2 shows a schematic flow diagram which may conveniently be used to summarise the operation of one example of the invention and which is believed to be self explanatory.

As the multiplication of the evaluated pixel values by the appropriate multiplying factors is effected, in at least one example of the invention, at a late stage in the processing, it follows that the correction not only accommodates the errors caused by processing data relating to divergent beam paths as if they were parallel, but also corrects for other invariant sources of error in the apparatus. Examples of such other sources of error are hardness variation across the distribution of radiation generated by the source 1 and hardness variations introduced by any shaped attenuators which may be disposed between the source and the patient position, and possibly also between the patient position and the detector array, to tend to equalise the absorption suffered by radiation over the whole distribution and despite the varying lengths of the different beam paths through the patient position.

It will be understood that the absorption values for the various pixels are measured, in each case, from a preselected datum level. That datum level is preferably the same for all pixel absorption values used or derived during operation of apparatus in accordance with the invention, as well as for the convolution processing operations.

WHAT WE CLAIM IS:—

1. Radiographic apparatus comprising a source of radiation projected towards a body, said radiation fanning out in one dimension as it proceeds away from said source to produce a flat, fan-shaped distribution of radiation which spans at least a substantial proportion of the extent of said body in said one dimension, scanning means for scanning said source angularly around said body to project radiation through said body from many different directions, detecting means for detecting the radiation emergent from said body along many groups of mutually inclined, substantially linear paths during the angular movement of said source, and for producing corresponding groups of electrical signals indicative of the amounts of radiation emergent from the body along said groups of paths, processing means for processing the electrical signals corresponding to said groups of paths to produce a representation of the variation of absorption of said radiation with position over a cross-sectional slice of said body, and wherein said processing means includes means for combining each output signal with weighted components of other output signals of the same group in proportions appropriate to parallel beam paths, and compensating means, including a store of compensating factors each applicable to a respective region of such slice, evaluated during a preliminary operation of the apparatus with a body of known absorption characteristics, and means for combining evaluated absorption values for a body of unknown absorption characteristics with appropriate compensating factors prior to the production of said representation.

2. Apparatus according to claim 1 wherein said scanning means is arranged to cause said source to execute a 360° scan around said body.

3. Apparatus according to any preceding claim wherein said compensating means includes a dividing circuit arranged to receive evaluated absorption values for said body of known absorption characteristics and corresponding known absorption values and to provide said compensating factors.

4. Radiographic apparatus substantially as herein described with reference to the drawings accompanying the Provisional Specification.

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Chartered Patent Agent.

Agents for the Applicants

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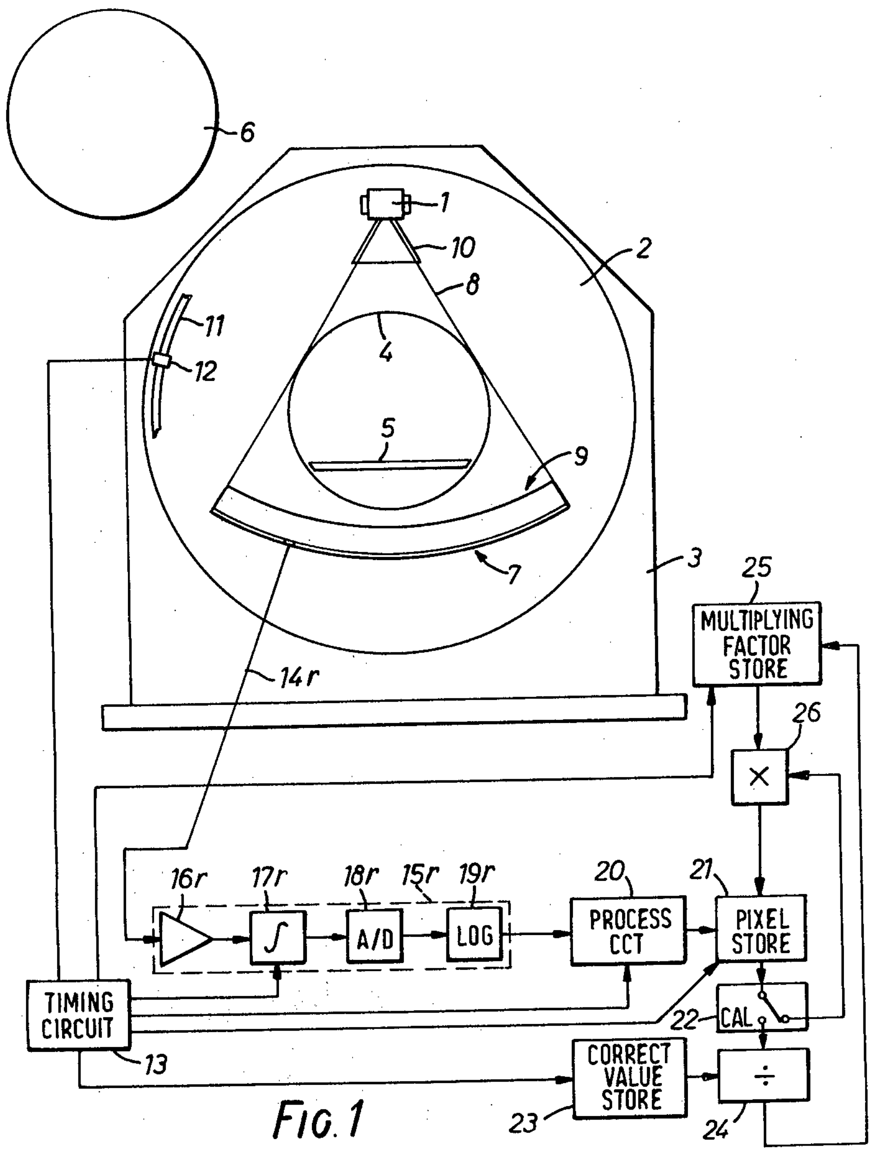
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FIG. 1

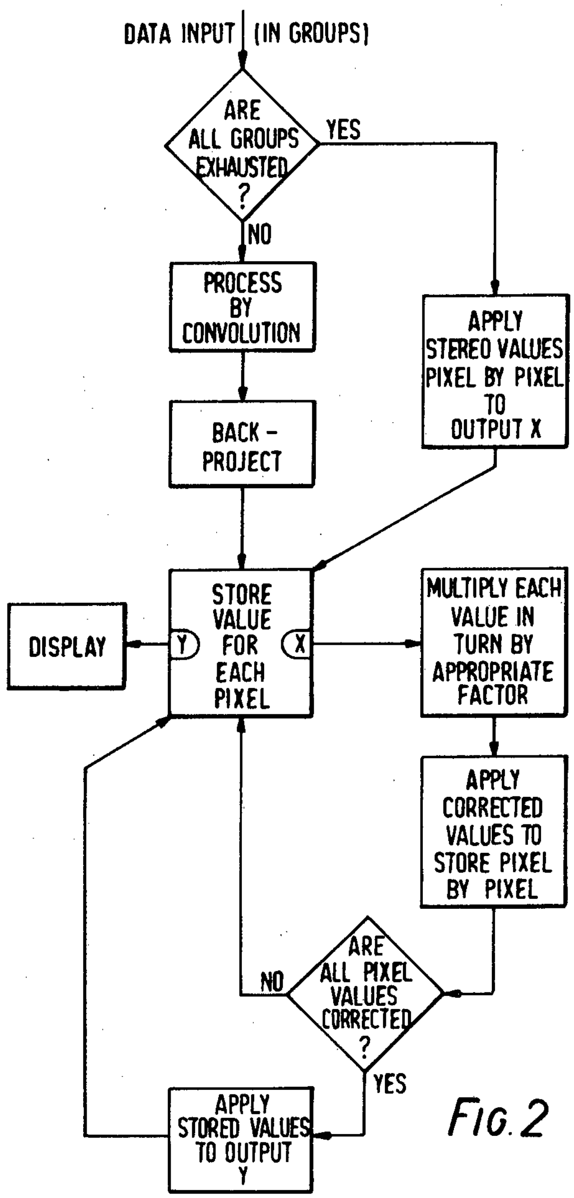
DATA INPUT (IN GROUPS)
ARE ALL GROUPS EXHAUSTED ?
YES
NO
PROCESS BY CONVOLUTION
APPLY STEREO VALUES PIXEL BY PIXEL TO OUTPUT X
BACK - PROJECT
STORE VALUE FOR EACH PIXEL
Y
X
DISPLAY
MULTIPLY EACH VALUE IN TURN BY APPROPRIATE FACTOR
APPLY CORRECTED VALUES TO STORE PIXEL BY PIXEL
NO
ARE ALL PIXEL VALUES CORRECTED ?
YES
APPLY STORED VALUES TO OUTPUT Y

FIG. 2