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

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
2 November 2006 (02.11.2006)

PCT

(10) International Publication Number
WO 2006/114716 A2

(51) International Patent Classification:
A61B 6/03 (2006.01)

(21) International Application Number:
PCT/IB2006/051091

(22) International Filing Date: 10 April 2006 (10.04.2006)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
60/674,900 26 April 2005 (26.04.2005) 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, New York 10020 (US).

(72) Inventors; and

(75) Inventors/Applicants *(for US only)*: **ALTMAN, Ami** [IL/IL]; P.O. Box 325, 31004 Haifa (IL). **SHAPIRO, Olga** [IL/IL]; Tzufit Str., 17/app 3, 31004 Haifa (IL). **LEVENE, Simha** [IL/IL]; D.n. Hanegev, 85140 Saad (IL). **WAINER, Naor** [IL/IL]; Ha Zait 34, 30900 Zichron Yaakov (IL).

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

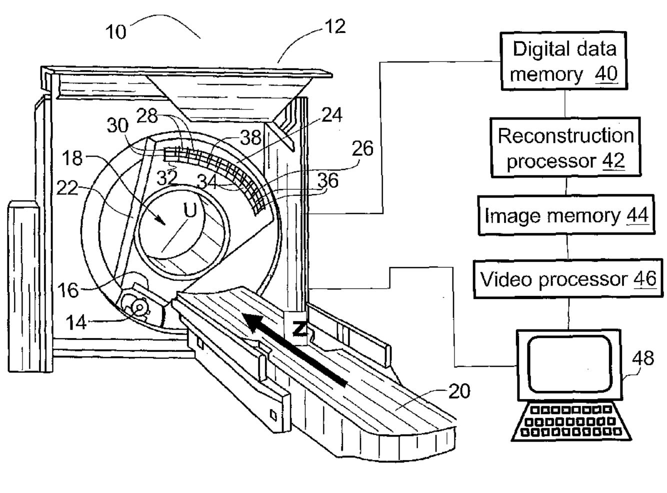
(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AT, AU, AZ, BA, BB, BG, BR, BW, BY, BZ, CA, CH, CN, CO, CR, CU, CZ, DE, DK, DM, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, HR, HU, ID, IL, IN, IS, JP, KE, KG, KM, KN, KP, KR, KZ, LC,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

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

Declaration under Rule 4.17:
— *as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))*

[Continued on next page]

(54) Title: DOUBLE DECKER DETECTOR FOR SPECTRAL CT



(57) Abstract: A radiation detector (24) includes a two-dimensional array of upper scintillators (30τ) which is disposed facing an x-ray source (14) to convert lower energy radiation into visible light and transmit higher energy radiation. A two-dimensional array of lower scintillators (30B) is disposed adjacent the upper scintillators (30τ) distally from the x-ray source (14) to convert the transmitted higher energy radiation into visible light. Respective active areas (94, 96) of each upper and lower photodetector arrays (38τ, 38B) are optically coupled to the respective upper and lower scintillators (30τ, 30B) at an inner side (60) of the scintillators (30τ, 30B) which inner side (60) is generally perpendicular to an axial direction (Z). Interference filters (110, 112) may be deposited on the active areas (94, 96) of the associated upper and lower photodetectors (38τ, 38B) to restrict radiation wavelengths received by the upper and lower photodetectors (38τ, 38B) to wavelengths emitted by the respective upper and lower scintillators (30τ, 30B). The upper scintillators (30τ) may include at least one of ZnSe(Te) and YAG(Ce).

WO 2006/114716 A2

Published:
— *without international search report and to be republished upon receipt of that report*

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.

DOUBLE DECKER DETECTOR FOR SPECTRAL CT

DESCRIPTION

The present application relates to the use of imaging systems. The subject matter finds particular application in spectral computed tomography (CT) scanners and will be described with particular reference thereto. However, the invention finds use in connection with DF and RF imaging, x-ray fluoroscopy, radiography, and other imaging systems for medical and non-medical examinations.

Computed tomography (CT) imaging typically employs an x-ray source that generates a fan-beam, wedge-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. A two-dimensional radiation detector including an array of detector elements is arranged opposite the x-ray source. The radiation detector includes the scintillator layer and an underlying photodetector layer which measure intensities of the transmitted x-rays. In a dual energy CT system, scintillation crystals are coupled to each of respective two photomultiplier tubes, e.g. a calcium fluoride (CaF) and sodium iodide (NaI). The two scintillators can be placed side by side, or, as shown in U.S. 4,247,774, the scintillators can be shaped to be partially overlapped such that some of the x-rays pass through both scintillators. Lower energy x-rays are absorbed in and cause scintillations in an upper CaF scintillator, while higher energy x-rays pass through to scintillate in the NaI scintillator. The scintillations give rise to electrical currents in the corresponding photomultipliers.

Typically, the x-ray source and the radiation detectors are mounted at opposite sides of a rotating gantry such that the gantry is rotated to obtain an angular range of projection views of the subject. In some configurations the x-ray source is mounted on the rotating gantry while the radiation detector is mounted on a stationary gantry. In either configuration, the projection views are reconstructed from the electrical signals using filtered backprojection or another reconstruction method to produce a three-dimensional image representation of the subject or of a selected portion thereof.

In dual energy CT systems, electrical signals corresponding to the higher and lower energy x-rays can be collected simultaneously and reconstructed into separate images that are inherently registered. The dual energy slice data can also be used to provide beam hardening corrections.

The present invention contemplates an improved method and apparatus which overcomes the above-referenced problems and others.

In accordance with one aspect of the present application, a radiation detector is disclosed. A two-dimensional array of upper scintillators is disposed facing an x-ray source to receive radiation therefrom and convert lower energy radiation into visible light, and transmit higher energy radiation. A two-dimensional array of lower scintillators is disposed adjacent the upper scintillators distally from the x-ray source to convert the transmitted higher energy radiation into visible light. An array of light- sensitive elements, which are in optical communication with the upper and lower scintillators, views the visible light and converts the visible light into electrical signals.

In accordance with another aspect of the present application, a method of manufacturing a radiation detector is disclosed. A two-dimensional array of photodetectors is fabricated integrally in a chip. Upper and lower scintillators are fabricated on light-sensitive faces of the photodetectors.

One advantage of the present application resides in using a safe scintillator material.

Another advantage resides in a commercially viable spectral scanner.

Another advantage resides in providing inexpensive detectors of high QDE and high optical detection efficiency for spectral CT.

Yet another advantage resides in substantial improvement of the light collection efficiency.

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

The invention may take form in various components and arrangements of components, and in various process operations and arrangements of process operations.

The drawings are only for the purpose of illustrating preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 is a diagrammatic illustration of an imaging system;

FIGURE 2A diagrammatically illustrates a portion of a radiation detector;

FIGURE 2B diagrammatically illustrates a top view of a portion of a radiation detector with linear tiles extending in the **Z**-direction;

FIGURE 3 shows graphs of absorption of YAG scintillation layers of different thicknesses;

FIGURE 4 diagrammatically illustrates a portion of a radiation detector which includes side-mounted photodiodes with interference filters;

FIGURE 5 diagrammatically illustrates a portion of the radiation detector which includes back- mounted photodiodes with the interference filters;

FIGURE 6A diagrammatically illustrates a side view of the radiation detector with a grid; and

FIGURE 6B diagrammatically illustrates a top view of a grid.

With reference to FIGURE 1, a computed tomography (CT) imaging apparatus or CT scanner **10** includes a gantry **12**. An x-ray source **14** and a source collimator **16** cooperate to produce a fan-shaped, cone-shaped, wedge-shaped, or otherwise-shaped x-ray beam directed into an examination region **18** which contains a subject (not shown) such as a patient arranged on a subject support **20**. The subject support **20** is linearly movable in a **Z**-direction while the x-ray source **14** on a rotating gantry **22** rotates around the **Z**-axis.

Preferably, the rotating gantry **22** rotates simultaneously with linear advancement of the subject support **20** to produce a generally helical trajectory of the x-ray source **14** and collimator **16** about the examination region **18**. However, other imaging modes can also be employed, such as a single- or multi-slice imaging mode in which the gantry **22** rotates as the subject support **20** remains stationary to produce a generally circular trajectory of the x-ray source **14** over which an axial image is acquired. After the axial image is acquired, the subject support optionally steps a pre-determined distance in the **Z**-direction and the axial image acquisition is repeated to acquire volumetric data in discrete steps along the **Z**-direction.

A radiation detector or detector array **24** is arranged on the gantry **22** across from the x-ray source **14**. The radiation detector **24** includes a scintillation array **26** of scintillators or crystals **28**. The scintillation array **26** is arranged in layers **30** and spans a selected angular range that comports with a fan angle of the x-ray beam. The radiation scintillation array **26** also extends along the **Z**-direction to form a matrix of n x m scintillators, such as 16 x 16, 32 x 32, 16 x 32, or the like. The layers **30** of the scintillation array **26** are stacked in the direction generally perpendicular to the **Z**-direction. The radiation detector **24** acquires a series of projection views as the gantry **22** rotates. It is also contemplated to arrange the radiation detector **24** on a stationary portion of the gantry encircling the rotating gantry such that the x-rays continuously impinge upon a continuously shifting portion of the radiation detector during source rotation. In one embodiment, a grid **32**, such as an anti-scatter grid, is arranged on a radiation- receiving face of the scintillation array **26**. An array or arrays **36** of photodiodes or other photodetectors **38** is optically coupled to the respective scintillators **28** of the scintillator array **26** to form a detector element or dixel.

A reconstruction processor **42** reconstructs the acquired projection data, using filtered backprojection, an n-PI reconstruction method, or other reconstruction method, to generate a three-dimensional image representation of the subject, or of a selected portion thereof, which is stored in an image memory **44**. The image representation is rendered or otherwise manipulated by a video processor **46** to produce a human-viewable image that is displayed on a user interface **48** or another display device, printing device, or the like for viewing by an operator.

The user interface **48** is additionally programmed to interface a human operator with the CT scanner **12** to allow the operator to initialize, execute, and control CT imaging sessions. The user interface **48** is optionally interfaced with a communication network such as a hospital or clinic information network via which image reconstructions are transmitted to medical personnel, a patient information database is accessed, or the like.

With reference to FIGURE 2A, the scintillation array **26** includes a double decker array which includes a bottom scintillation layer $\mathbf{30_B}$ and a top scintillation layer $\mathbf{30_T}$, which are separated by a reflective layer **58**. The photodetector array **36** of the photodetectors **38**, such as silicon photodetectors, amorphous silicon, charge-coupled devices, CMOS, or other semiconductor photodetectors is in optical communication with the scintillation array **26**. More specifically, the photodetectors include a photosensitive layer with an array of active areas and, preferably, an analog second layer that forms a p-n junction with the photosensitive layer, integrally formed on a chip **50**.

X-rays, which have passed through the examination region **18**, strike the top scintillation layer $\mathbf{30_T}$ along a direction **U**. The top scintillation layer $\mathbf{30_T}$, which is closest to the X-ray source, converts the softest or lowest-energy x-rays in the beam, which has passed through the examination region **18**, into light. The bottom scintillation layer $\mathbf{30_B}$, which is furthest from the X-ray source, receives the hardest x-rays. Light signals from the dixels of each layer **30** are detected by the corresponding photodetectors **38** of the photodetector array **36**. The top layer $\mathbf{30_T}$ is selected and sized to convert substantially all x-ray photons of 50keV or less into light and pass substantially all photons 90keV or higher to the bottom layer $\mathbf{30_B}$.

The photodetector array **36** is arranged vertically along the direction **U** on the inner side **60** of each double-decker array **26**. Top and bottom surfaces **62, 64, 66, 68** and side surfaces **70, 72** of the top and bottom scintillation layers $\mathbf{30_T}$, $\mathbf{30_B}$ are painted or otherwise covered with a light-reflective coating or layer **80**. The inner side **60** of the top and bottom scintillation layers $\mathbf{30_T}$, $\mathbf{30_B}$, which is adjacent the photodetectors **38**, is left open to communicate light to the photodetector array **36**. The reflective coating can function as the separation layer **58**. Alternately, the separation layer can be a separate layer selected to control the minimum energy of x-ray photons reaching the bottom layer $\mathbf{30_B}$.

In one embodiment, the bottom scintillation layer $\mathbf{30_B}$ comprises gadolinium oxy sulfide (Gd_2O_2S, Pr, Ce or "GOS"), while the top scintillation layer $\mathbf{30_T}$ comprises zinc selenide (ZnSe), a material known for wide transmission range. Preferably, zinc selenide is doped with tellurium (Te). Alternatively, the top layer $\mathbf{30_B}$ comprises cadmium tungstate ($CdWO_4$ or "CWO").

It is also contemplated that the scintillation array **26** includes more than two scintillation layers. In this case, there is **n** scintillation layers disposed between the top and bottom scintillation layers $\mathbf{30_T}$, $\mathbf{30_B}$ where **n** is greater than 0 and less than **A** and **A** is an integer.

With continuing reference to FIGURE 2A and further reference to FIGURE 2B, the photodetector array **36** is preferably a 2D array including upper and lower photodetector arrays **82, 84**, both part of the vertical chip **50**. An active area **94** of each upper photodetector $\mathbf{38_T}$ is disposed opposite and coupled to the top scintillation layer $\mathbf{30_T}$, while an active area **96** of each lower photodetector $\mathbf{38_B}$ is disposed opposite and coupled to the bottom scintillation layer $\mathbf{30_B}$. Each silicon chip **50** includes a pair of respective upper and lower photodetectors $\mathbf{38_T}$, $\mathbf{38_B}$. The silicon chips **50** are mounted parallel each other, preferably in the **Z**-direction, between adjacent rows of the scintillation array **26**. In one embodiment, the silicon chips **50** are mounted parallel each other in the **X** direction or the direction transverse the axial direction **Z**. Each chip and the scintillators it carries form a linear tile **98**. The chips are protected from x-rays by the grid **32**, as discussed below. An optical adhesive **100** is disposed between the chip **50** and the scintillation layers $\mathbf{30_T}$, $\mathbf{30_B}$ to improve optical coupling between the photodetectors **38** and the scintillation layers $\mathbf{30_T}$, $\mathbf{30_B}$.

In one embodiment, the upper and lower photodetectors $\mathbf{38_T}$, $\mathbf{38_B}$ can be back-contact photodiodes and have respective active areas **94, 96** that are sensitive to the light radiation produced by scintillation. Electrical contacts **102** are preferably disposed on a front side **104** of the photodetectors $\mathbf{38_T}$, $\mathbf{38_B}$. Other detectors which convert light energy into electrical signals, such as front surface photodetectors and charge-coupled devices (CCDs), are also contemplated.

Electronics, such as an application-specific integrated circuits (ASICs) (not shown), produce electrical driving outputs for operating the photodetector array **36**, and receive detector signals produced by the photodetector array **36**. The ASICs perform selected detector signal processing which results in the conversion of photodetector currents to digital data.

The signals from the dixels of each layer **30** are weighted and combined to form spectrally- weighted image data. Alternatively, images are formed separately from each of the layers, and combined to form spectrally-weighted image data. The weighting may include zeroing one or more of the dixel layers. By selecting different relative weighting among the dixels, image data is generated which emphasizes and de-emphasizes selected portions of the energy spectrum, i.e. selected x-ray energy absorption ranges. By appropriately selecting the weighting, CT images are reconstructed of specific selected x-ray energy absorption ranges to emphasize tissues while other selected tissues are superseded or substantially erased in the reconstructed image. For example, calcium in mammary tissue, and iodine in a contrast medium can be emphasized by subtracting images weighted to emphasize either side of the respective absorption lines. Although two layers are illustrated, it should be appreciated that a larger number of layers can be provided to provide more levels of energy discrimination.

With continuing reference to FIGURE 2A and further reference to FIGURE 2B, the detector array **24** includes a plurality of rows of scintillation arrays. Each row includes the photodetector chip array **50** and the linear array of scintillators optically coupled to the chip **50**. The array of scintillators includes the top layer $\mathbf{30_T}$ and the bottom layer $\mathbf{30_B}$ (not shown in FIGURE 2B). In FIGURE 2B, the chips **50** are shown with exaggerated width for simplicity of illustration.

In one embodiment, the top layer $\mathbf{30_T}$ is Yttrium Aluminum Garnet (YAG). YAG material is comprised of low-Z elements and has a relatively low density of less than 5 g/ml. This low density has limited x-ray stopping power and primarily absorbs soft or lower energy x-rays in the beam. The YAG material has excellent (short) afterglow and light output properties, and emits in a region of the visible light spectrum where silicon photodiodes have adequate sensitivity.

With reference to FIGURE 3, the dependence of the X-ray absorption on YAG layer thickness and on X-ray photon energy is shown. For example, a layer of 0.71mm thick YAG scintillation layer absorbs about 70% of the 50keV x-rays while it passes over 75% of the x-rays of 90 keV and over.

With reference to FIGURE 4, the top layer $\mathbf{30_T}$ is thin compared to the bottom layer $\mathbf{30_B}$, to selectively sense lower energy x-rays and transmit higher energy x-rays. For example, the top layer $\mathbf{30_T}$ must preferably absorb x-rays of the energy below 50keV while transmitting 75% or more of the x-rays of the energy above 90keV. Typically, the photodiodes active areas **94, 96** are made to match respective thicknesses of the top and bottom layers $\mathbf{30_T}$, $\mathbf{30_B}$.

With continuing reference to FIGURE 4, in this embodiment the light passes freely from one scintillation layer to another as the reflecting coating **80** does not extend between the top layer bottom surface **66** and the lower layer top surface **64**. The active area **94** of the upper photodetector $\mathbf{38_T}$ is substantially increased in size to overlap the region associated with the bottom scintillation layer 30_B. A top interference filter **110** of a transmission wavelength *l***1** which matches an emission wavelength *l***2** of the material comprising the top scintillation layer 30_T is deposited, preferably during manufacture of the detector, upon the upper photodetector active surface **94**. The match of the top interference filter wavelength *l***1** with the top scintillation layer wavelength *l***2** ensures that only the light emitted by the top scintillation layer 30_T impinges upon the upper photodetector 38_T. This allows the active area **94** of the upper photodetector 38_T to be enlarged. For example, the wavelength *l***1** of the top interference filter **110** can be 550nm to match the wavelength of the YAG which comprises the top scintillation layer 30_T in this embodiment. Such interference filter restricts the light to impinging upon the upper photodetector 38_T to only the YAG emission.

Similarly, the active area **96** of the lower photodetector 38_B is protected against the wavelengths of the top scintillation layer 30_T by a bottom interference filter **112** which has a wavelength *l***3** to match a bottom scintillation layer emission wavelength *l***4**. For example, the bottom interference filter **112** can be a 540nm wavelength filter which passes the emission wavelengths of cadmium tungstate (CWO) only to the lower photodetector 38_B.

In one embodiment, depending on the scintillators used, a single bandpass filter is deposited on the active area of one of the upper and lower photodiodes. The signal is derived by difference.

With reference to FIGURE 5, the photodetector array **36** includes back-illuminated photodiodes (BIP) **38** and is a single, monolithic, semiconductor substrate **120** having functional integrated circuitry formed thereon. The functional integrated circuitry includes a matrix of photosensitive elements or "dixels," preferably photodiodes, formed on the light-receiving side. The integrated circuitry of the array **36** is generally manufactured from silicon or other semiconductor wafers using established integrated circuit fabrication processes, such as masking, evaporation, etching, and diffusion processes, and so forth.

The diode pair 38_T, 38_B is mounted underneath the bottom layer 30_B. In this case, the diffuse reflective coating **80** on the bottom surface **66** of the top layer 30_T and the top surface **64** of the bottom layer 30_B is omitted.

The top interference filter **110** is of the transmission wavelength ***l*1**, which matches the emission wavelength ***l*2** of the material comprising the top scintillation layer $\mathbf{30_T}$, and is deposited, preferably during manufacture of the photodetector, upon the upper photodetector active area **94**. The match of the top interference filter wavelength ***l*1** with the top scintillation layer wavelength ***l*2** ensures that only the light emitted by the top scintillation layer $\mathbf{30_T}$ is received by the upper photodetector $\mathbf{38_T}$.

Similarly, the active area **96** of the lower photodetector $\mathbf{38_B}$ is protected against the wavelengths of the top scintillation layer $\mathbf{30_T}$ by the bottom interference filter **112** which has the wavelength ***l*3** to match the bottom scintillation layer emission wavelength ***l*4**.

With reference to FIGURES 6A and 6B, the grid **32** includes legs or strips **120** which each preferably overlaps the thickness of each corresponding silicon chip **50**. In this manner, the grid **32** protects silicon chips **50** from x-ray radiation. For example, if the silicon chips are about 0.125mm thick, the legs **120** can be about 0.140mm thick.

The application has been described with reference to the preferred embodiments. Obviously, modifications and alterations will occur to others upon reading and understanding the preceding detailed description. It is intended that the application be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

CLAIMS

1. A radiation detector **(24)** comprising:

a two-dimensional array of upper scintillators **(30_T)** disposed facing an x-ray source **(14)** to receive radiation therefrom, convert lower- energy radiation into visible light and transmit higher energy radiation;

a two-dimensional array of lower scintillators **(30_B)** disposed adjacent the upper scintillators **(30_T)** distally from the x-ray source **(14)** to convert the transmitted higher energy radiation into visible light; and

an array **(36)** of light sensitive elements **(38_T, 38_B)**, which are optically coupled with the upper and lower scintillators **(30_T, 30_B)**, to view the visible light and convert the visible light into electrical signals.

2. The detector as set forth in claim 1, wherein the upper scintillator **(30_T)** includes:

doped zinc selenide.

3. The detector as set forth in claim 2, wherein the lower scintillator **(30_B)** includes doped gadolinium oxy sulfide (GOS).

4. The detector as set forth in claim 1, wherein the upper scintillator **(30_B)** includes:

doped yttrium aluminum garnet.

5. The detector as forth in claim 1, wherein the array **(36)** of light sensitive elements **(38_T, 38_B)** includes:

an upper array **(82)** of light sensitive elements **(38_T)**, each upper light sensitive element **(38_T)** being optically coupled to the upper scintillator **(30_T)**; and

a lower array **(84)** of light sensitive elements **(38_B)**, each lower light sensitive element **(38_B)** being optically coupled to the lower scintillator **(30_B)**.

6. The detector as set forth in claim 5, further including:

a first interference filter **(110)** between an active area **(94)** of each upper light sensitive element **(38_T)** and the scintillators **(30_T, 30_B)**, the first interference filter **(110)** having a light wavelength ***(l1)***, which matches a light wavelength ***(l2)*** emitted by the upper scintillators **(30_T)**, to restrict radiation wavelengths received by the upper light sensitive elements **(38_T)** to the wavelengths ***(l2)*** emitted by the upper scintillators **(30_T)**.

7. The detector as set forth in claim 5, further including:

a second interference filter **(112)** between an active area **(96)** of each lower light sensitive element **(38_B)** and the scintillators **(30_T, 30_B)**, the second interference filter **(112)** having a light wavelength ***(l3)***, which matches a light wavelength ***(l4)*** emitted by the lower scintillators **(30_B)**, to restrict radiation wavelengths received by the lower light sensitive elements **(38_B)** to the wavelengths emitted by the lower scintillators **(30_B)**.

8. The detector as set forth in claim 5, wherein the upper and lower light sensitive elements **(38_T, 38_B)** are coupled to an inner side **(60)** of respective upper and lower scintillators **(30_T, 30_B)**, which inner side **(60)** is generally perpendicular to an axial direction **(Z)**.

9. The detector as set forth in claim 8, wherein the scintillators **(30_T, 30_B)** are coated with a reflective coating material **(80)** on all sides excluding the inner side **(60)** adjacent the light sensitive elements **(38_T, 38_B)**.

10. The detector as set forth in claim 1, wherein the upper scintillator **(30_T)** includes one of:

ZnSe,

YAG,

CdWO4, and

GOS;

and the lower scintillator **(30_B)** includes:

GOS,

ZnSe;

YAG ; and

CdWO4.

11. The detector as set forth in claim 1, further including a plurality of tiles **(98)**, each tile **(98)** including:

a silicon chip **(50)** disposed in a plane parallel to a direction **(U)** of the radiation emitted by the x-ray source **(14)**, which chip **(50)** includes:

at least upper and lower rows **(82, 84)** of light sensitive elements **(38_T, 38_B)**;

an upper scintillator **(30_T)** of the upper scintillator array coupled to each of the upper light sensitive elements **(38_T)**;

a lower scintillator **(30_B)** of the lower scintillator array coupled to each of the lower light sensitive elements **(30_B)**; and

a reflective coating **(80)** on the upper and lower scintillators **(30_T, 30_B)**.

12. The detector as set forth in claim 11, wherein each upper light sensitive element **(38_T)** is optically coupled to the upper and lower scintillators **(30_T, 30_B)**, and further including:

a filter **(110)** mounted between each upper light sensitive element **(38_T)** and the scintillators **(30_T, 30_B)**, the filter **(110)** passing light of a wavelength **(*l*2)** emitted by the upper scintillator **(30_T)** and blocking light of a wavelength **(*l*4)** emitted by the lower scintillator **(30_B)**.

13. The detector as set forth in claim 11, wherein the silicon chip **(50)** is less than 0.15mm thick.

14. A computed tomography scanner for a use with the radiation detector of claim 1.

15. A method of manufacturing a radiation detector comprising:

fabricating a two-dimensional array of photodetectors **(38_B, 38_T)** integrally in a chip **(50)**; and fabricating upper and lower scintillators **(30_T, 30_B)** on light sensitive faces of the photodetectors.

16. The method as forth in claim 15, wherein the step of fabricating the two-dimensional array of photodetectors includes:

fabricating an upper row of photodetectors;

optically coupling the upper row photodetectors to the upper scintillators;

fabricating a lower row of photodetectors, adjacent the upper row; and

optically coupling the lower row photodetectors to the lower scintillators.

17. The method as set forth in claim 16, further including:

depositing a first interference filter on an active area of each photodetector of the upper row; and

restricting radiation wavelengths impinging on the upper row photodetectors to wavelengths of the upper scintillator.

18. The method as set forth in claim 17, further including:

depositing a second interference filter on an active area of each photodetector of the lower row; and

restricting radiation wavelengths impinging on the lower row photodetectors to wavelengths of the lower scintillator.

19. The method as set forth in claim 15, wherein the upper scintillator is at least one of:

doped zinc selenide;

doped gadolinium oxy sulfide;

cadmium tungstate; and

doped yttrium aluminum garnet.

20. The method as set forth in claim 15, wherein the upper scintillator is comprised of zinc selenide and the lower scintillator is comprised of gadolinium oxy sulfide.

21. A radiation detector manufactured by the method of claim 15.

22. A radiation detector comprising:

a plurality of tiles **(98)** disposed adjacently one another, each tile including:

an upper array of scintillators **(30_T)**, facing an x-ray source, for converting lower energy x-rays into visible light and transmitting higher energy x-rays;

a lower array of scintillators, disposed adjacent the upper row and distally from the x-ray source, for converting the transmitted higher energy x-rays into visible light;

an upper array of photodetectors, being optically coupled at least to the upper scintillators, for sensing visible light emitted by the upper scintillators and converting the light emitted by the upper scintillators into electrical signals; and

a lower array of photodetectors, being optically coupled to the lower scintillators, for sensing visible light emitted by the lower scintillators and converting the light emitted by the lower scintillators into electrical signals.

1

1/6

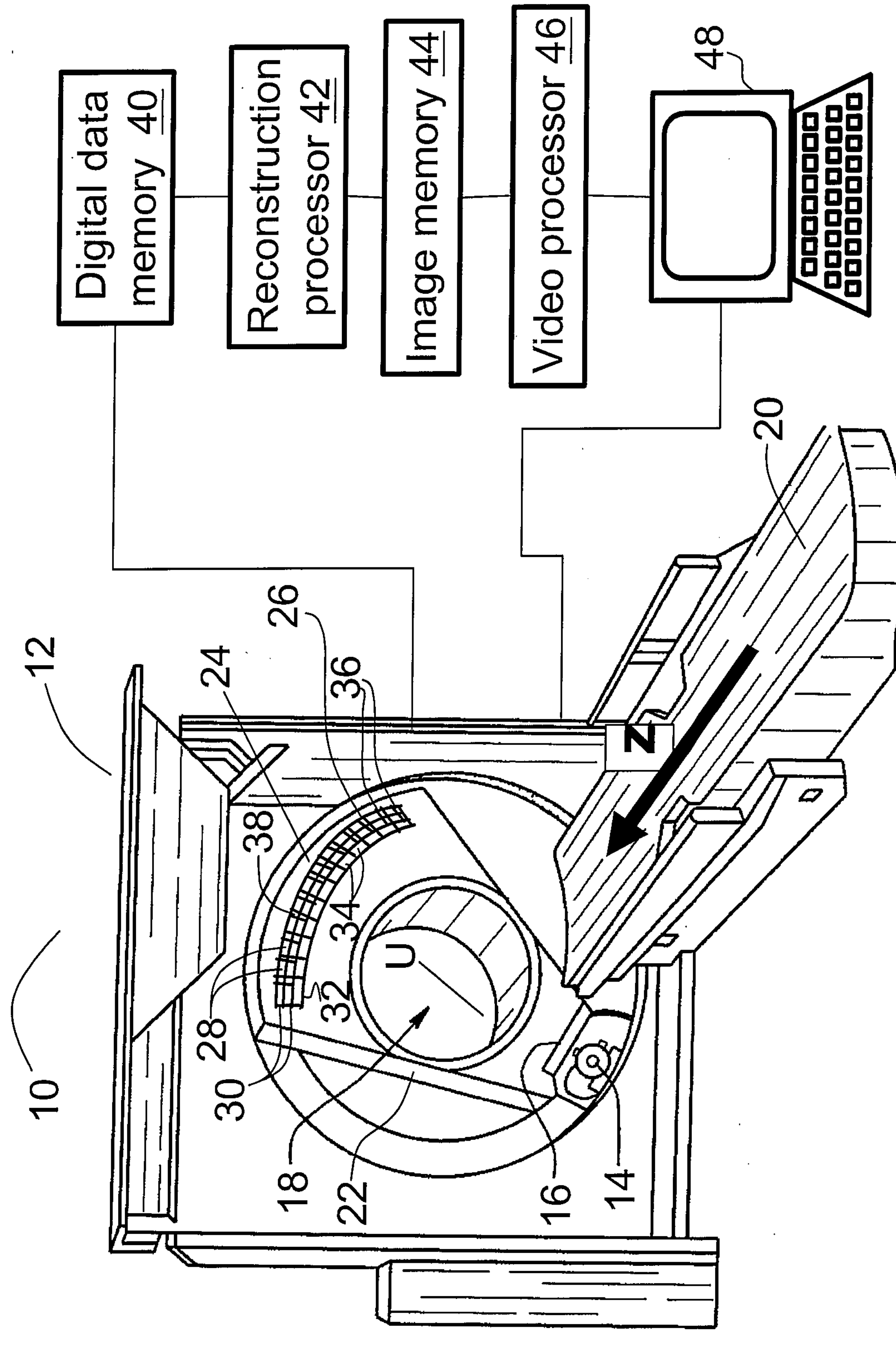

Digital data memory 40
Reconstruction processor 42
Image memory 44
Video processor 46
48
10
12
14
16
18
20
22
24
26
28
30
32
34
36
38
U
Z

FIG 1

2/6

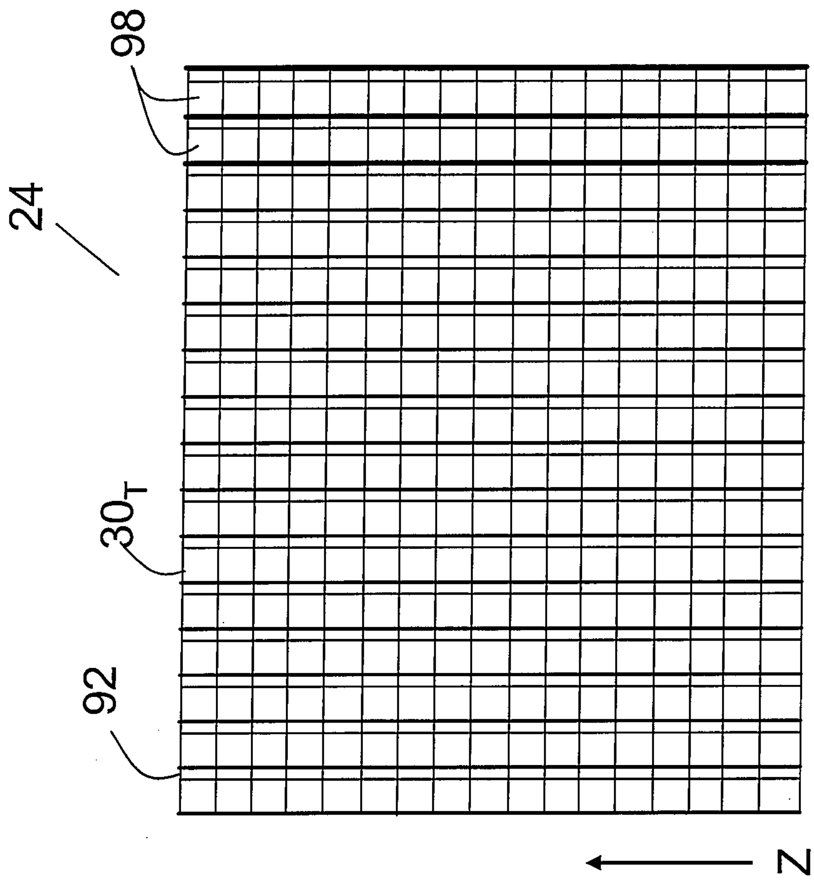


FIG 2B

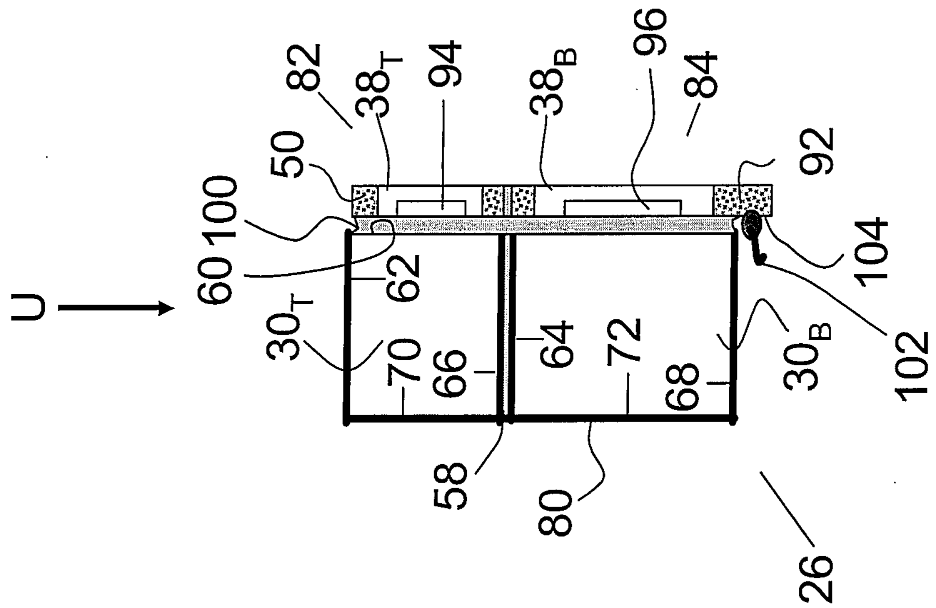


FIG 2A

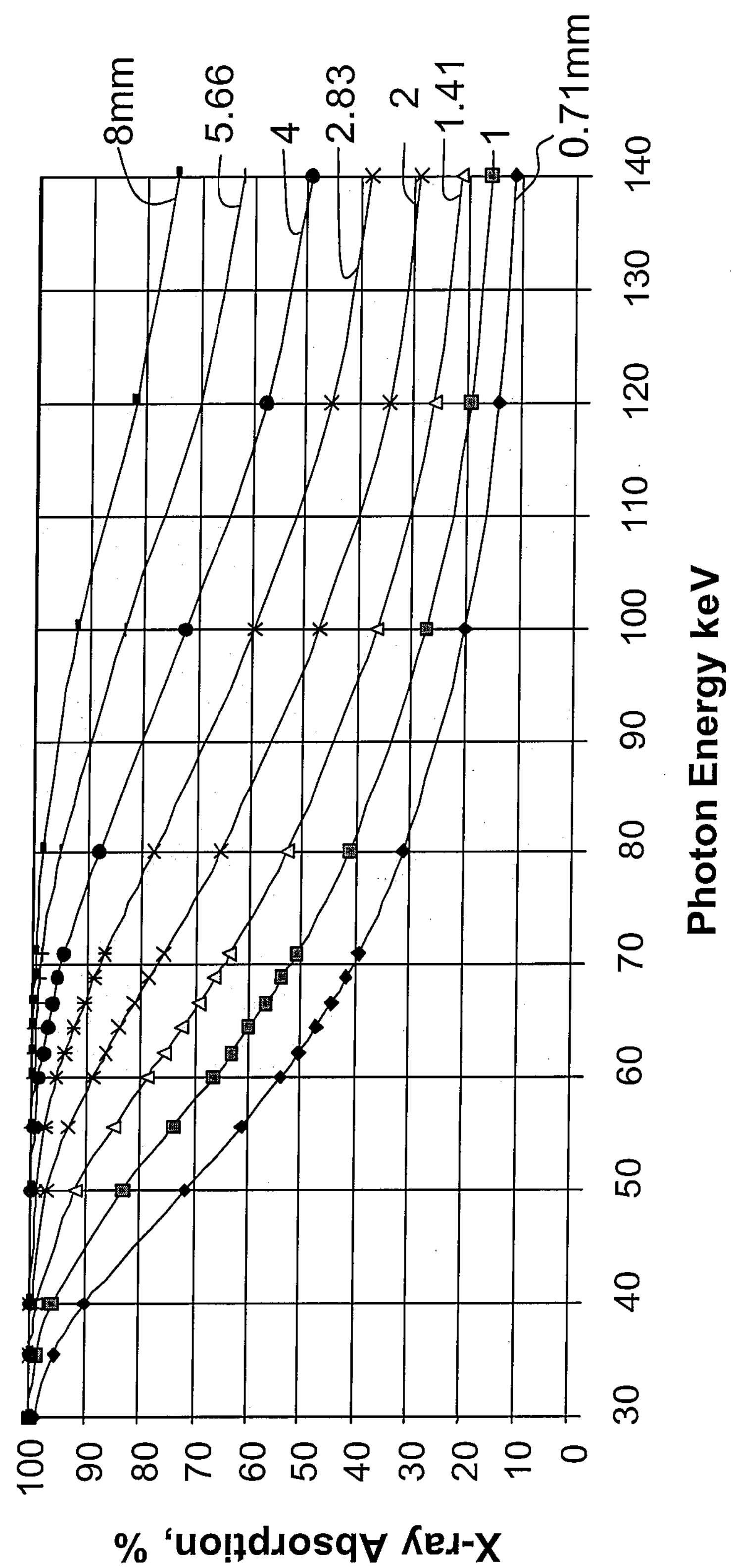

8mm
5.66
4
2.83
2
1.41
1
0.71mm
Photon Energy keV
X-ray Absorption, %
30
40
50
60
70
80
90
100
110
120
130
140
100
90
80
70
60
50
40
30
20
10
0

FIG 3

FIG 4

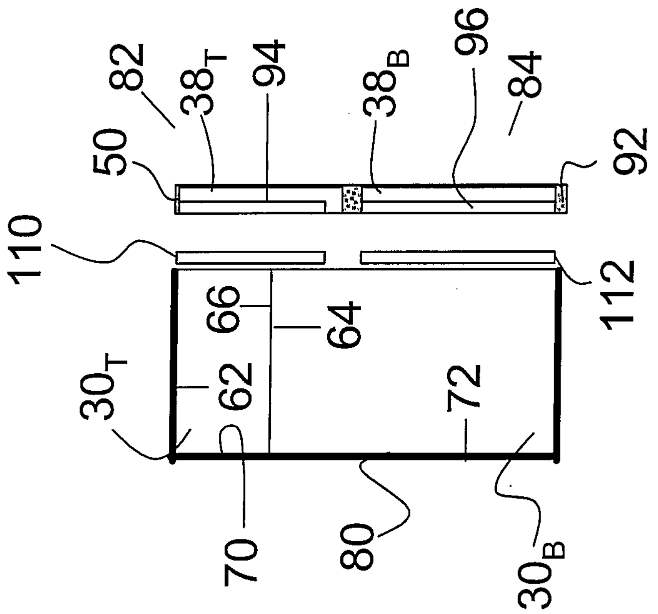
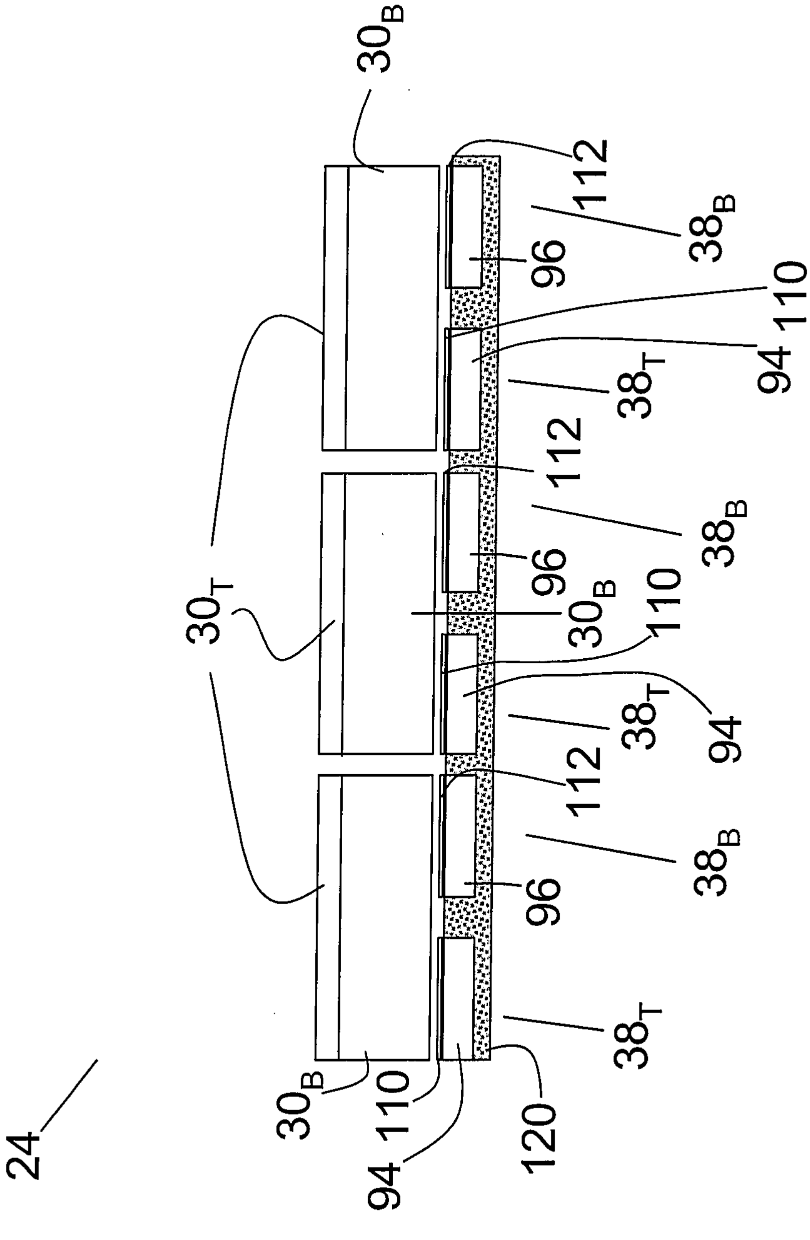


FIG 5



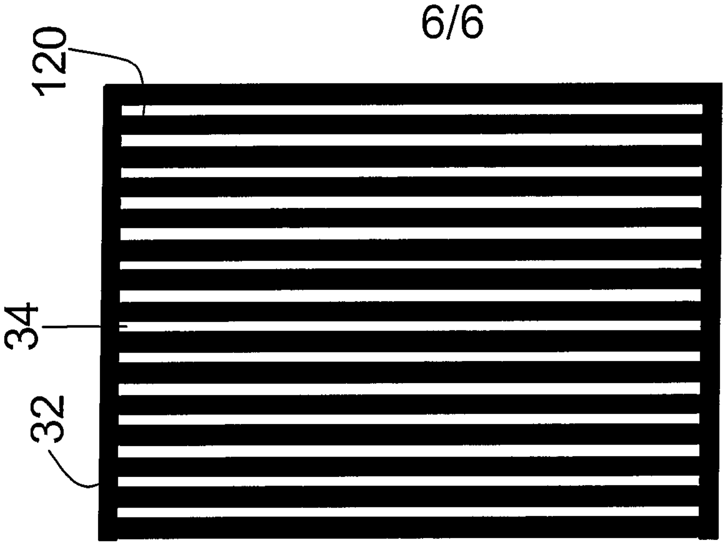


FIG 6B

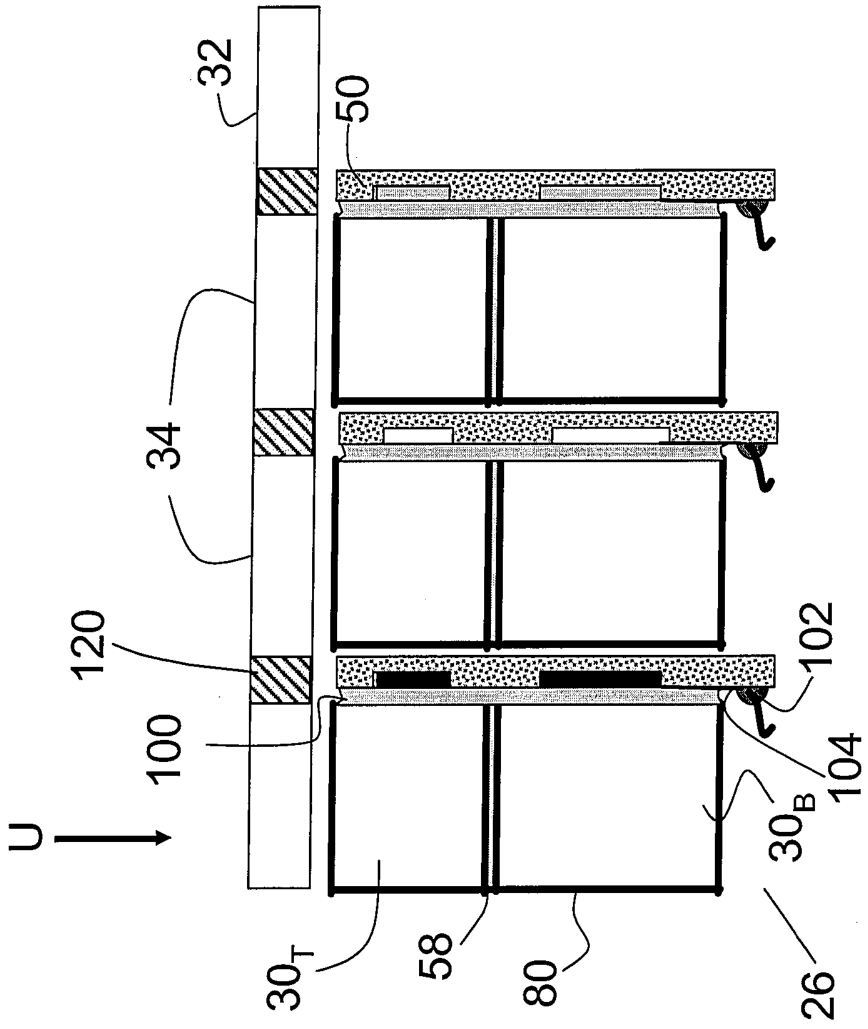


FIG 6A