(12) **UK Patent Application** (19) **GB** (11) **2 010 891** **A**

(54) **Scintillator Structure**

(57) A scintillator structure comprises at least one layer 5 of transparent fused quartz with a phosphor coating 3 on one or both sides adjacent at least one transparent layer 5 of epoxy resin which directs light from the phosphor to a detector. The phosphor layer may be formed from a powder optionally with a binder, a single crystal or a melt, or by evaporation or sintering. A plurality of multiple layers may be used (Fig. 4 not shown) or the structure tilted (Fig. 5 not shown) for greater absorption. The structure may be surrounded by another such structure (Fig. 3 not shown) optionally operating in cascade with the first. Many phosphors are specified.

A scintillator structure comprises phosphor particles dispersed in epoxy resin or copoly imide-silicone and cast in a multi-compartment box with long sides (21 Fig. 8 not shown) transparent to X-rays and dividers (20) opaque to X-rays.
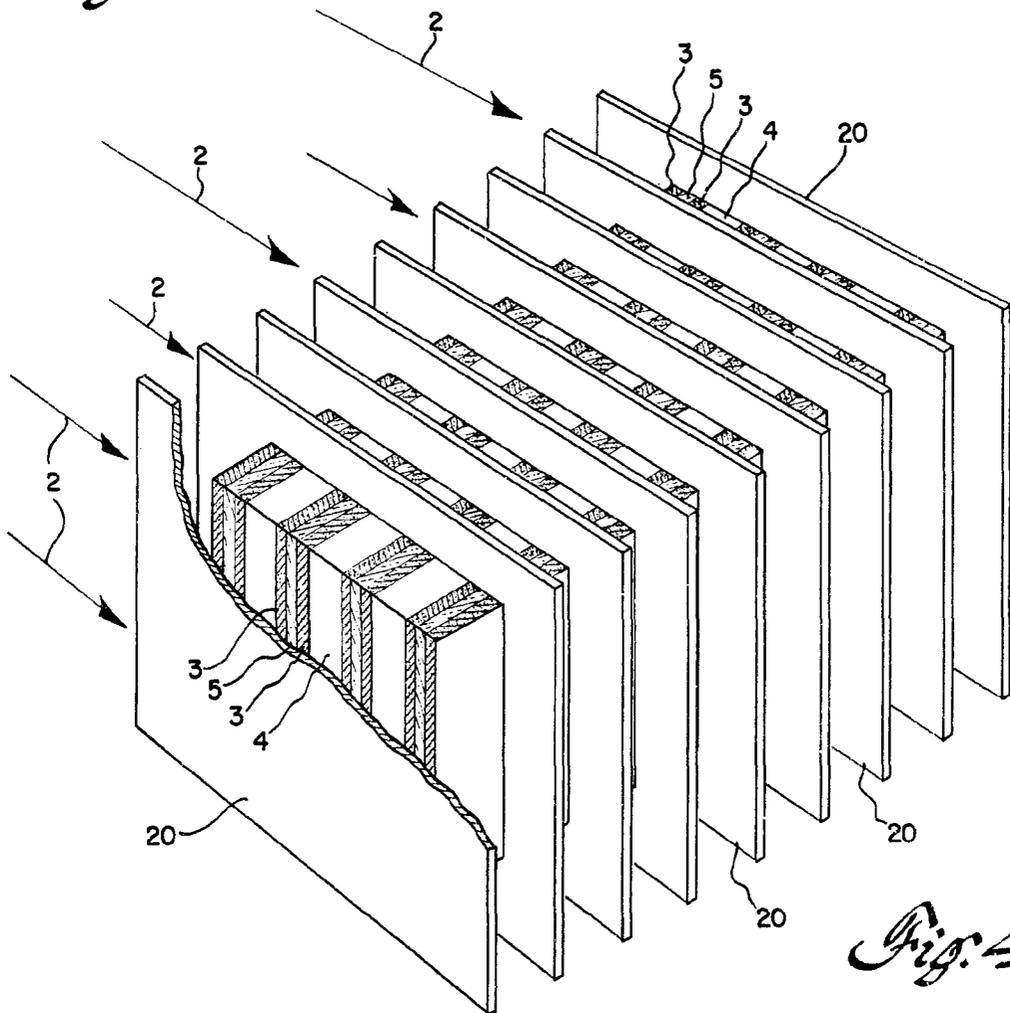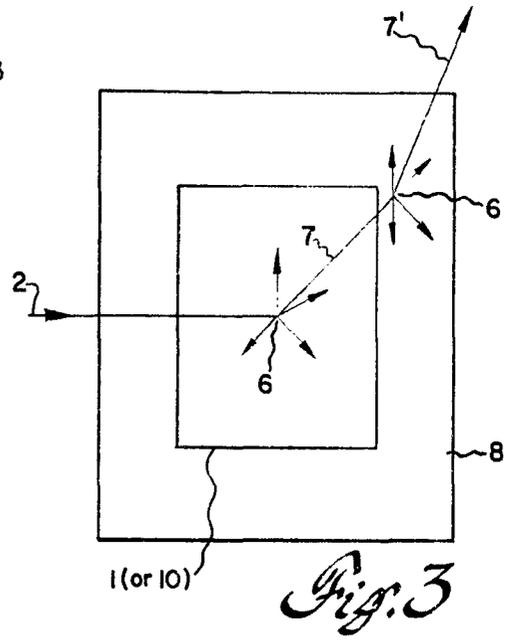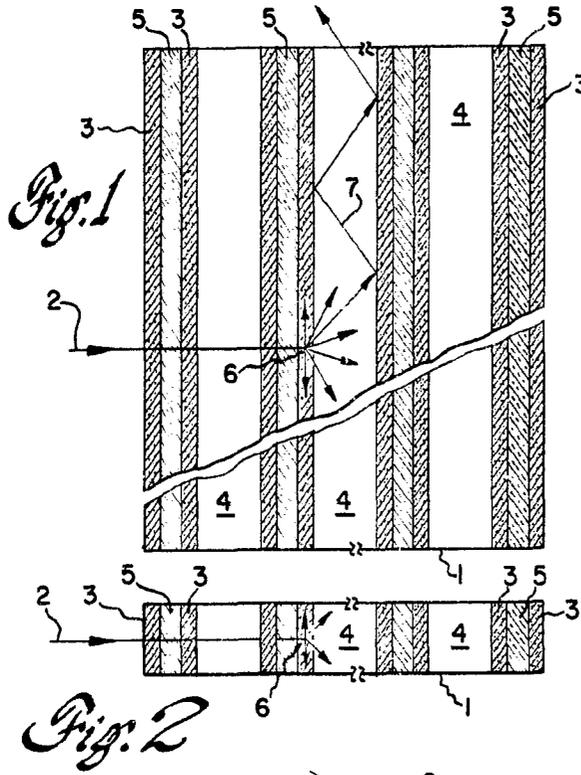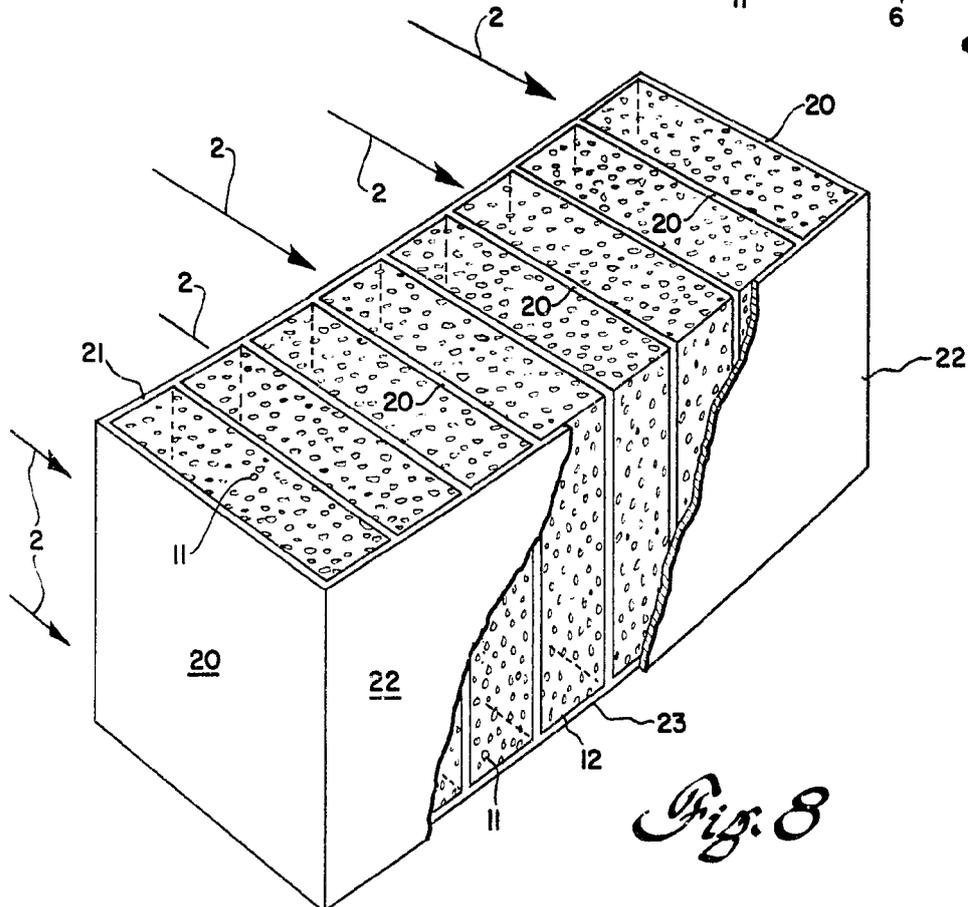
Fig. 1

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Fig.1

Fig.2

Fig.3

Fig.4

Fig. 5

Fig. 6

Fig. 7

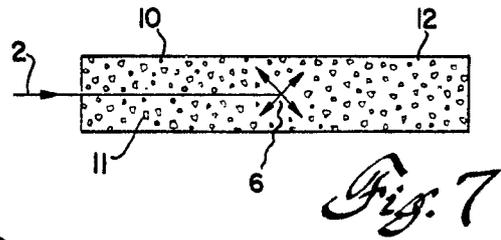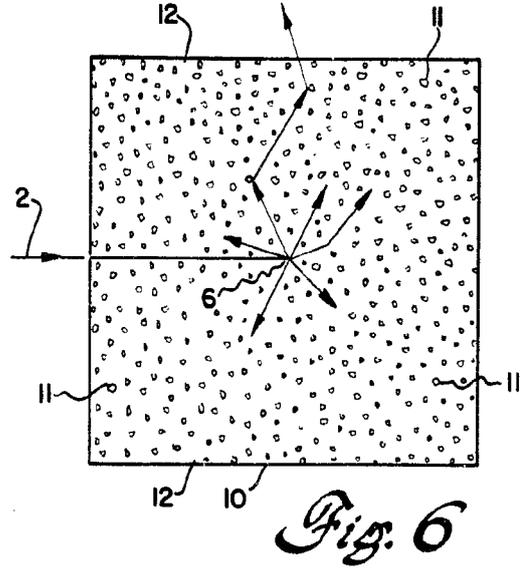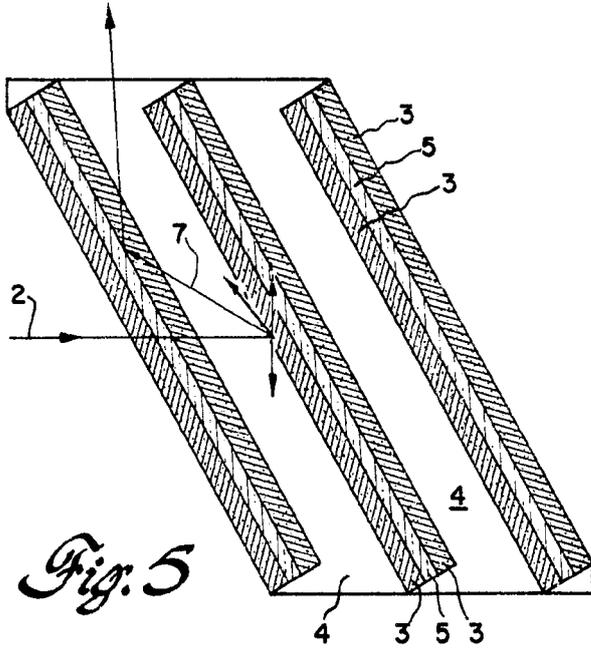*Fig. 8*
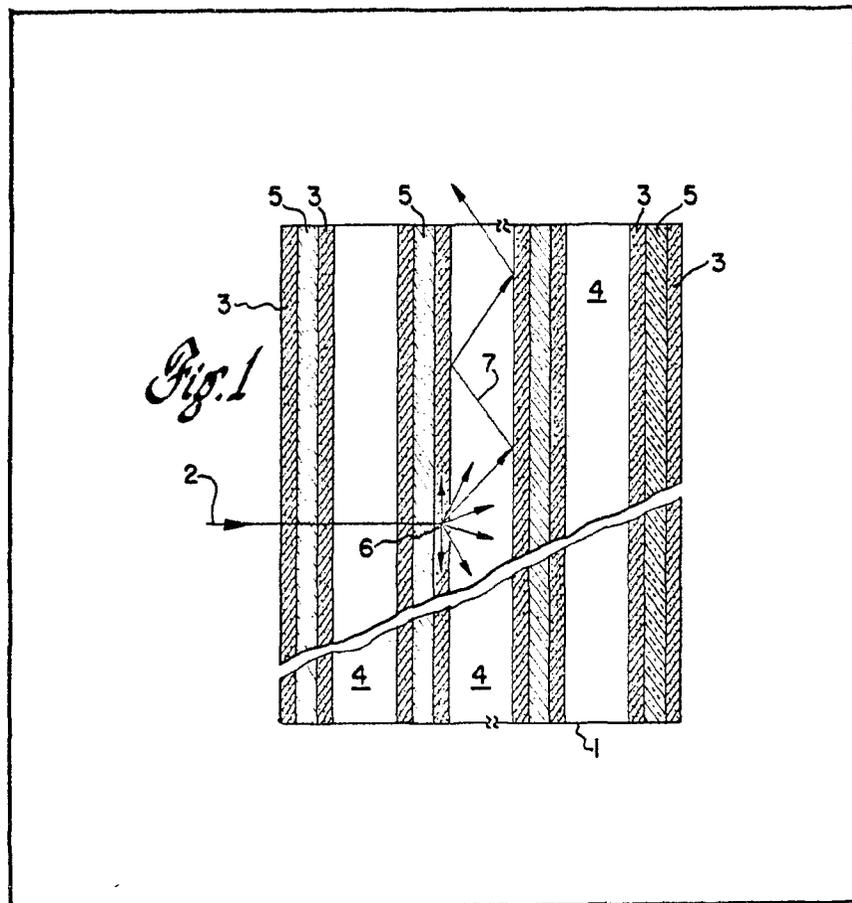
SPECIFICATION
Distributed Phosphor Scintillator Structures

This invention relates to scintillator structures
and methods for fabricating such structures. More
5 particularly, this invention relates to a method for
distributing the scintillator phosphor in such a
way as to enhance the escape of the visible
wavelength radiation that would otherwise be
dissipated within a scintillator body. Two
10 embodiments of the present invention are
disclosed: one in which the phosphor is
distributed in a layered structure and another in
which the phosphor is dispersed throughout a
transparent matrix.
15   In general, a scintillator is a material which
emits electro-magnetic radiation in the visible or
near-visible spectrum when excited by high
energy electromagnetic photons such as those in
the x-ray or gamma-ray regions of the spectrum,
20 hereinafter referred to as supra-optical
frequencies. Thus, these materials are excellent
choices for use as detectors in industrial or
medical x-ray or gamma-ray equipment. In most
typical applications, the output from a scintillator
25 materials is made to impinge upon
photoelectrically responsive materials in order to
produce an electrical output signal which is
directly related to the intensity of the initial x-ray
or gamma-ray bombardment.
30   Scintillator materials comprise a major portion
of those devices used to detect the presence and
intensity of incident high energy photons. Another
commonly used detector is the high pressure
noble gas ionization device. This other form of
35 high energy photon detector typically contains a
gas, such as xenon, at a high pressure (density),
which ionizes to a certain extent when subjected
to high energy x-ray or gamma-ray radiation. This
ionization causes a certain amount of current flow
40 between the cathode and the anode of these
detectors which are kept at relatively high and
opposite polarities from one another. The current
that flows is sensed by a current sensing circuit
whose output is reflective of the intensity of the
45 high energy radiation. Since this other form of
detector operates on an ionization path remains
open. Hence, these detectors are peculiarly
sensitive to their own form of "afterglow" which
results in the blurring in the time dimension, of
50 information contained in the irradiating signal as a
result of its passig through a body to be
examined, as in computerized tomography
applications.
    As used herein and in the appended claims, the
55 term "light" means those electromagnetic
radiations in the visible region of the spectrum
and also near-visible wavelengths given off by
certain fluorescent materials. Also, as used herein,
and in the appended claims, the term "optical"
60 encompasses the same spectrum region as the
term "light" does.
    In general, it is desirable that the amount of
light output from these scintillators be as large as
possible for a given amount of x-ray or gamma-

65 ray energy. This is particularly true in the medical
tomography area where it is desired that the
energy intensity of the x-ray be as small as
possible to minimize any danger to the patient.
    Another important property that scintillator
70 materials should possess is that of short afterglow
or persistance. This means that there should be a
relatively short period of time between the
termination of the high energy radiating excitation
and the cessation of light output from the
75 scintillator. If this is not the case, there is resultant
blurring, in time, of the information-bearing signal
generated, for example, when the scintillator is
used to produce tomographic imaging data.
Furthermore, if rapid tomographic scanning is
80 desired, the presence of the afterglow tends to
severely limit the scan rate, thereby rendering
difficult the viewing of moving bodily organs, such
as the heart or lungs.
    A scintillator body or substance, in order to be
85 effective, must be a good converter of high energy
radiation (that is, x-rays and gamma-rays).
Typically, present scintillator bodies consist of a
phosphor in a powder, polycrystalline, or
crystalline form. In these forms, the useful light
90 that is produced upon high energy excitation is
limited to that which can escape the interior of
the scintillator body and that generated in the
surface regions. The escape of light is difficult due
to the optical absorption resulting from multiple
95 internal reflections, each such reflection further
attenuating the amount of light available to
external detectors. Thus, it is necessary that not
only the phosphors themselves have a good
luminescent efficiency but it is also necessary that
100 the light output be available for detection.
    In the medical tomography area, where the
intensity of x-radiation is modulated by the body
through which is passes, and which modulated
radiation is then converted into electrical signals, it is
105 important to have x-ray detection devices which
have a good overall energy conversion efficiency.
For devices with low efficiency, a higher x-ray flux
radiation must be applied to produce the same
light and electrical output from the overall system.
110 In a medical tomographic context, this means that
such a system has a low signal-to-noise ratio.
    Typical scintillator phosphors which are used
include barium fluorochloride doped with a
europium activator (BaFCl:Eu). Other phosphors,
115 for example, include bismuth germanate
($Bi_4Ge_3O_{12}$), lanthanum oxybromide doped with
terbium (LaOBr:Tb), cesium iodide doped with
thalium (CsI:Ti), cesium iodide doped with sodium
(Csi:Na), calcium tungstate ($CaWO_4$), cadmium
120 tungstate ($CdWO_4$), zinc cadmium sulfide doped
with silver (ZnCdS:Ag), zinc cadmium sulfide
doped with silver and nickel (ZnCdS:Ag,Ni),
gadolinium oxysulfide doped with terbium
(Gd_2O_2S:Tb), and lanthanum oxybromide doped
125 with dysprosium (LaOBr:Dy). Other host-crystal
possibilities for phosphors include the
selenides of zinc and cadmium, the tellurides of
zinc and cadmium, sodium iodide (NAI), and the
oxysulfide of lanthanum (La_2O_2S).

In accordance with one embodiment of the present invention, a suitable phosphor is attached to both sides of an underlying substrate. The phosphor is attached either in the form of a
5 powder, in a more continuous form, or in a dispersed form in a transparent matrix. The resulting substrates with the phosphor material attached are further layered with one another and with a transparent material between the
10 phosphor-coated substrates. This structure exposes a greater area of phosphor material to a region from which the light output from the material can easily escape to be detected. The resulting scintillator bodies are useful in
15 tomographic detector arrays which require a high overall energy conversion efficiency for high resolution imagery and for patient safety.

In another embodiment of the present invention, the phosphor material is distributed in a
20 continuous, rather than layered, fashion throughout a transparent matrix. Thus, the light that is produced upon x-ray or gamma-ray absorption deep within the scintillator body escapes from the body with a minimum amount
25 of internal reflection and consequent light energy loss.

The phosphor material to be used in the above structure is inexpensive when compared to ionization type x-ray detectors which typically
30 consist of a highly pressurized (25 atmospheres) noble gas surroundng electrodes maintained at a high opposite polarities from one another.

The present invention will be further described, by way of example only, with reference to the
35 accompanying drawings, in which:—

Fig. 1 is a side elevation sectional view of one embodiment of the present invention showing the phosphor distributed in a layered fashion.

Fig. 2 is a top sectional view of the scintillator
40 body of Fig. 1 showing the phosphor distributed in a layered fashion.

Fig. 3 is a side elevation view of a scintillator body surrounded by a jacket which serves to convert the wave-length of the scintillator output
45 to a more convenient wavelength.

Fig. 4 is a perspective view showing scintillator bodies with phosphor distributed in a layered fashion arranged as a portion of a tomographic x-ray detector.
50 Fig. 5 is a side elevation sectional view of a layered scintillator structure tilted for greater absorption.

Fig. 6 is a side elevation sectional view of a scintillator body illustrating the continuous
55 dispersion of phosphor throughout the body.

Fig. 7 is a top sectional view of the scintillator body of Fig. 6 also showing a continuous dispersion of phosphor material throughout the body.
60 Fig. 8 shows a portion of a tomographic x-ray detector employing scintillator bodies with phosphor continuously distributed throughout.

The invention herein relates to a distributed phosphor scintillator structure which produces a
65 greater optical coupling between the scintillator

body and the photo-electrically responsive device, such as a photodiode, to which it is typically coupled. There are two primary embodiments of the present invention; one in which the phosphor
70 is distributed in a layered fashion, which will be discussed first, and the other embodiment in which the phosphor is distributed in a continuous fashion throughout the scintillator body. It is to be noted, however, that sections of scintillator
75 bodies with the phosphor distributed continuously throughout are in fact used in one embodiment of the multi-layered structure.

Fig. 1 shows a side elevation sectional view of a scintillator body 1 with a multi-layered structure.
80 In this embodiment, phosphor material 3 is attached to a substrate 5. The phosphored substrates are further layered with light channeling laminates 4 between layers of phosphored substrates. Fig. 1 further shows an
85 x-ray photon 2 being absorbed by a phosphor particle at an absorption site 6 in the fourth layer of the phosphor encountered. The absorption of the high energy, high frequency x-ray photon causes the production of a multitude of lower
90 energy, lower frequency optical wavelength photons. The path 7 of a typical optical wavelength photon is shown as it is reflected back and forth within the light channeling laminate 4 between the phosphor layers,
95 ultimately escaping the scintillator body whereby it may be more readily detected than if the absorption event had occurred deep withinside a dense and much less optically transparent body.

Fig. 2 shows a plan view of the same
100 scintillator structure as shown in Fig. 1. It is to be noted that both Fig. 1 and Fig. 2 suggest that the number of phosphored substrate layers need not be fixed at four as shown.

Fig. 1 and 2 both illustrate a phosphor layer 3.
105 In particular, this layer 3 applied to the substrate 5 is used in a variety of phosphor forms. The phosphor layer 3 consists of a phosphor either as a powder, as a single crystal, as a phosphor dispersed in a transparent matrix, or as a
110 phosphor applied in a continuous layer, as for example, a phosphor material formed by a quench from its liquid form.

Any of the phosphors mentioned above are applicable to the substrate material with an
115 appropriate adhesive and a suitable thickness. In particular, ZnCdS:Ag has a particularly large particle size which enhances the escape of light from the powder layer. The emission color of this particular phosphor is orange-red, which makes it
120 well suited to detection by silicon semi-conductor devices. A layer of 0.5 millimeters of this phosphor absorbs between 20 and 25 percent of the x-ray photons in a computerized tomography system in which the average x-ray energy is
125 approximately 65 kev. Five or six layers of this thickness, for this phosphor, suffices to absorb 90 percent or more of the x-ray photons. When other more x-ray absorptive rare earth phosphors such as LaOBr:Tm, LaOBr:Tb, Gd_2O_2S:Tb, or La_2O_2S:Tb
130 are employed, the number of layers to achieve the

same total absorption is lower and losses in the light guiding laminate layers are also lower.

There are numerous choices for the phosphor material. The three criteria to be used in selecting
5 a particular phosphor for computerized tomographic applications are its high luminescent efficiency, short afterglow duration, and low absorbance for the emitted light. The decay rate (afterglow) is particularly important in those
10 medical tomographic applications where repeated scans are made, as for example, when dealing with images of moving bodily organs. Choices for host-crystal lattices to be doped with an appropriate rare earth element (that is, those
15 elements with an atomic number between 58 and 71 inclusive) or other activator include the sulfides, selenides, and tellurides of zinc and cadmium, the iodides of sodium and cesium, the tungstates of calcium and cadmium, lanthanum
20 oxybromide, and the oxysulfides of lanthanum and gadolinium.

In addition to applying the phosphor material to the substrate in a powder form, the phosphor material is also applicable as a single crystal when
25 such crystals exist. For example, a single crystal of cesium iodide doped with thallium is attachable to the substrate with a suitable adhesive, such as an epoxy.

In another embodiment of the present
30 invention, the phosphor layer which is attached to the substrate consists of an appropriate phosphor continuously dispersed in a transparent matrix such as that which is more particularly described below as an embodiment of the entire scintillator
35 body. This particular layered embodiment combines features of both of the principal embodiments disclosed herein.

In another embodiment of the present invention, the phosphor layer 3 (Fig. 1) consists of
40 a phosphor layer which is either evaporated onto the substrate, is melt solidified, high pressure sintered, or hot-forged. Such methods for producing these continuous phosphor layers are described in the copending application of Dominic
45 A. Cusano et al, Serial No. 853,085 which is assigned to the same assignee of the present invention.

The substrate material itself must be of a material which is not absorptive of x-rays whose
50 frequencies are within the spectral region of interest. Typically, the substrate is composed of a clear fused quartz material. It is desirable that this substrate also be optically transparent although this is not essential since the majority of the
55 optical photons emitted find their way to the outside through the light channeling laminate layers. A typical thickness for this substrate in a tomographic application is 20 mils.

Between each phosphored substrate, there is a
60 layer of light channeling laminate material. Typically, this laminate is an epoxy. The principal criteria for selection of this laminate material is its optical transparency. Other criteria for its selection include its chemical compatibility with
65 the phosphor material, its structural rigidity, its low

x-ray absorptivity, and its ability to withstand prolonged x-ray bombardment.

The laminate layer also includes as a matter of choice, at least one wavelength conversion
70 material such as a fluorescent dye which emits a visible wavelength photon at a wavelength susceptible to detection by the photoelectrically responsive sensing device when the wavelength conversion material itself is stimulated by the
75 optical wavelength photons produced by the phosphor material. The intimate contact which the laminate layers have with the phosphored layers render them particularly useful and efficient for wavelength conversion.

80 If it is not desired that the wavelength conversion process take place within the scintillator body, then in another embodiment of the present invention, the scintillator body is surrounded by a jacket containing a wavelength
85 conversion substance. This embodiment is illustrated in Fig. 3 wherein multilayered scintillator body 1 (or continuously dispersed phosphor scintillator body 10, described below) is surrounded by a jacket 8 containing suitable
90 wavelength conversion material such as certain fluorescent dyes. In Fig. 3, x-ray photon 2 is absorbed at absorption site 6 within the scintillator body 1 as a result of which multiple photons at a first lower wavelength are emitted,
95 said photon paths being illustrated by a typical photon path 7. Said photon traverses a typical path 7 to a secondary absorptive site 6' within the jacket 8 surrounding the scintillator body 1. At this site 6', the first wavelength photon is
100 absorbed and another photon at a second wavelength is emitted traversing typical path 7'. In this fashion, the scintillator wavelength can be matched to a more sensitive spectral region of the photoelectrically responsive detector. If desired,
105 several wavelength conversions may be made through the use of multiply matched fluorescent materials.

By way of example, the fabrication of a typical multilayered scintillator body is described.
110 BaFCl:Eu phosphor is mixed with an equal weight of 1269A STY CAST® epoxy (supplied by Emersion and Cumming, Inc., of Canton, Mass.) containing rhodamine to the extent of 0.1 gm rhodamine per 30 ml of epoxy. The BaFCl:Eu
115 phosphor is typically doped with Eu to the extent of approximately 1 mole percent but doping to as low as 0.1 mole percent and as high as 5 mole percent is also used. The phosphor is suspended in epoxy by tumbling for 16 hours in a glass
120 container containing glass beads. The suspension is doctor bladed onto 20 mil CFQ (clear fused quartz) substrates to a thickness of 20 mils using a Gardner doctor blade. The film is cured for 18 hours at 88°C and the substrate is then coated on
125 a the other side and cured. A block scintillator material is made by laminating the phosphored substrate with 40 mil cured epoxy spacers using the same epoxy as a cement. In other embodiment of this invention, as described above,

the organic dye rhodamine is incorporated in the epoxy spacers rather than in the phosphor itself.

There are a number of desirable features of this particular scintillator structure. For instance, the
5 amount of x-ray radiation absorption is controlled by the number of layers of the phosphor. The number of layers is adjusted for the various absorptivity of the phosphors employed. If desired, certain substrates are coated with
10 phosphor material on one side only. Another advantage of this structure is that it permits the use of phosphors other than the single crystal phosphors such as CsI:Tl. This structure permits much more flexible choices in the selections of an
15 appropriate phosphor when such phosphor properties as light output wavelength, short afterglow duration, and luminescent efficiency must be balanced. In addition, there are certain phosphors such as CsI:Na, which while otherwise
20 being good phosphor materials, suffer from the problem that they are hygroscopic and when exposed in an atmospheric environment absorb water and suffer a resultant degradation in performance. This problem is much less severe in
25 a layered structure in which the phosphor material has a minimal exposure to the atmosphere than it is when the phosphor is applied to an exposed imaging screen. Another feature of the scintillator body stemming from the
30 structure of the present invention is its rigidity and ruggedness. A scintillator tomographic detector structure, such as that described in Fig. 4 does not suffer from the acoustic or microphonic noise pickup effects as do high pressure noble gas
35 ionization type detectors. Moreover, the scintillator bodies are manufactured to a high degree of accuracy which enables them to be precisely aligned in a detector array such as that of Fig. 4. Moreover, the phosphor substrates may
40 be angled as shown in Fig. 5 to produce a greater absorption without increasing the thickness of the phosphor on the substrate, thereby permitting, if desired, fewer layers to produce the same amount of absorption.
45 In accordance with the other principal preferred embodiment of the present invention, the phosphor material is dispersed in a continuous and uniform fashion throughout a transparent supportive matrix. Fig. 6 shows such a scintillator
50 body 10 being stimulated by a high energy x-ray photon 2. In this configuration, phosphor particles 11 are suspended in a rigid transparent matrix 12. The high energy x-ray photon 2 is converted to light by absorption at absorption site 6 from
55 which a multitude of lower energy optical wavelength photons are emitted. These photons are readily transmitted through the transparent matrix 12 being periodically reflected and scattered by the phosphor particles 11 which are
60 also present. However, most of the resultant light energy produced eventually reaches the exterior of the scintillator body where it is detected. Such a typical light path 7 is shown in Fig. 6.

Fig. 7 is a sectional plan view of Fig. 6 showing
65 the absorption event produced by x-ray photon 2

within scintillator body 10. The principal requirements for the transparent matrix 12 are: that it be a good transmitter of light at the wavelength produced by the phosphor; that it not
70 react with the phosphor and that it maintain the phosphor in a fixed suspension after the phosphor is thoroughly dispersed throughout it. A number of plastics are appropriate for this purpose, such as the polyimide silicone copolymers.
75 In the embodiment described in Fig. 6, the particular choice of phosphor material is a matter of design consideration to be based on such factors as absorptivity, afterglow duration, luminescent effiency, and output wavelength.
80 However, the concentration of phosphor within the transparent (typically plastic) matrix is controlled to effect changes in the overall absorption. Typically, in this embodiment, the dispersed phosphor scintillator body consists of
85 the phosphor in a concentration of 10 to 20 percent by volume.

Wavelength conversion is accomplished in the dispersed phosphor structure, if desired, in one of two ways. In accordance with one embodiment,
90 the scintillator body is surrounded by a jacket containing a suitable wavelength converter such as the organic dye rhodamine. Such a structure is illustrated in Fig. 3 which is equally applicable to the dispersed phosphor structure as it is for the
95 multilayered structure discussed above. In accordance with another embodiment of the invention, the wavelength conversion substance is mixed with the transparent matrix material.

The fabrication of dispersed phosphor
100 scintillator bodies is particularly easy. The phosphor chosen is thoroughly mixed with an epoxy, plastic, or other polymer whose optical and chemical properties are not seriously affected by x-radiation and the resultant mixture is then
105 solidified or if not actually solidified, the phosphor is fixed in suspension. The solidification is accomplished by a variety of processes including chemical activation, slight temperature elevation (curing), or ultraviolet irradiation. The scintillator
110 bodies are produced either individually or en masse in a prefabricated detector array structure. Fig. 8 describes such a scintillator structure comprising a front wall member 21 composed of a low Z (atomic number) material, such as
115 beryllium or aluminum, which is highly nonabsorptive for x-rays, collimator members 20 composed of a high Z material such as tungsten or tantalum which are relatively impervious to x-rays, a floor member 23, and a
120 rear wall member 22. Detector members 20, 21, 22 and 23 define a series of volumes into which the dispersed phosphor material is inserted. To insure an adequate and thorough fill of this collimator structure, the entire detector assembly
125 is agitated, typically at ultrasonic frequencies, during the insertion process. The material is then solidified either through chemical activation, curing, or by ultraviolet irradiation. This process results in an extremely rugged detector array
130 which is much less susceptible to acoustic noise

vibration during operation than is a typical
ionization detector array. Again, a wavelength
conversion material is added, if desired.

As with the multilayered structure, the
5 dispersed phosphor structure is also rigid, rugged,
and is easy to align accurately. The hydroscopic
problem suffered by certain phosphor materials is
also greatly reduced. In addition, the use of non-
single crystal phosphor material is possible.
10 Furthermore, the dispersed phosphor structure is
not very sensitive to changes in the x-ray
spectrum occurring as a result of any filtering
effects of the matrix material.

Thus, it can be appreciated from the above that
15 the principal preferred embodiments of the
present invention describe scintillator structures
with decided advantages over other scintillator
structures and which in particularly greatly
enhance the escape of light from the scintillator
20 body.

Claims

1. A method of manufacturing multilayered
scintillator bodies to increase detectable output at
optical wavelengths when excited by high energy
25 photons at supra-optical frequencies, said method
comprising the steps of:

A). applying a uniform layer of phosphor
material to at least one side of each of at least
two rigid substrates, said substrates being inert,
30 transparent to radiation at supra-optical
frequencies, and having substantially flat sides;
and then

B). forming a further layered structure by
applying an optically transparent laminate
35 material to alternating layers of the phosphored
substrates from step A, said laminate material
being transparent to radiation at supra-optical
frequencies, whereby optical wavelength output
produced within the scintillator body is channeled
40 to the exterior of the body for detection.

2. A method as claimed in claim 1 in which, in
step A, the phosphor material is applied to the
substrates by doctor blading.

3. A method as claimed in claim 1 or claim 2 in
45 which, in step B, the laminate material is applied
to the phosphored substrates by doctor blading,
said laminate material being in unhardened form,
and then after step B, causing the laminate to
harden.

50 4. A method as claimed in claim 3 in which at
least one wavelength conversion substance is
added to the unhardened laminate before said
laminate is applied to the phosphored substrate,
whereby the optical output of the phosphor is
55 matched to the spectral response of a
photoelectrically responsive detector.

5. A method as claimed in any one of claims 1
to 3 in which at least one wavelength conversion
substance is added to the phosphor before
60 application of the phosphor to the substrate,
whereby the optical output of the phosphor is
matched to the spectral response of a
photoelectrically responsive detector.

6. A method as claimed in any one of claims 1

65 to 5 in which the laminate material is an epoxy.

7. A method as claimed in any one of claims 1
to 6 in which the substrate is quartz.

8. A method as claimed in any one of claims 1
to 7 in which the phosphor layer applied is a
70 single crystal.

9. A method as claimed in any one of claims 1
to 7 in which the phosphor layer applied is in
powder form.

10. A method as claimed in any one of claims 1
75 to 7 in which the phosphor layer applied is
evaporated onto the substrate.

11. A method as claimed in any one of claims
1 to 7 in which the phosphor layer applied is
made by high pressure sintering.

80 12. A method as claimed in any one of claims
1 to 11 in which the number of layers formed is
controllably selected and angled to produce a
desired amount of absorption of said high energy
photons, whereby the absorption is controlled
85 independently of the absorption of the phosphor
material itself.

13. A method as claimed in any one of claims
1 to 12 in which the phosphor is BaFCl:Eu,
ZnCdS:Ag, ZnCdS:Ag,Ni, CsI:Tl, CsI:Na, CaF$_2$:Eu,
90 Gd$_2$O$_2$S:Tb, LaOBr:Dy, LaOBr:Tm, LaOBr:Tb,
La$_2$O$_2$S:Tb, Bi$_4$Ge$_3$O$_{12}$, CaWO$_4$, ZnS, ZnSe, ZnTe,
CdS, CdSe, CdTe, or NaI.

14. A method of making a scintillator body as
claimed in claim 1 substantially as hereinbefore
95 described.

15. A scintillator body produced in accordance
with the method of any one of claims 1 to 14.

16. A method of manufacturing dispersed
phosphor scintillator bodies to increase
100 detectable output at optical wavelengths when
excited by high energy photons at supra-optical
frequencies, said method comprising the steps of:

A). mixing a suitable phosphor with a
nonreactive liquid matrix substance to form a
105 suspension, said liquid matrix substance being
hardenable, and transparent to optical and
supraoptical radiation after hardening;

B). inserting the suspension from step A into a
mold of desired shape and then
110 C). causing the liquid matrix substance to
harden, whereby the phosphor particles are fixed
in position and whereby optical wavelength
output produced within the resulting scintillator
body is permitted to escape to the exterior of the
115 body for detection.

17. A method as claimed in claim 16 in which
the mold in step B is an array with a plurality of
substantially identical separate compartments,
said array comprising:
120 a front wall member, said front wall member
being permeable to radiation at supra-optical
frequencies;
side wall members, said side wall members
being opaque to radiation at supra-optical
125 frequencies;
interior wall members defining a plurality of
said components, said interior wall members
being oriented parallel to said side wall members

and being opaque to radiation at supra-optical frequencies;

    a floor member; and

    a rear wall member;

5   whereby the resultant structure upon hardening is usable as a portion of a detector system.

    18. A method as claimed in claim 17 in which during the insertion step B, the structure is 10 agitated, whereby a uniform distribution of suspension is achieved throughout each compartment.

    19. A method as claimed in claim 18 in which the agitation occurs at ultrasonic frequency.

15   20. A method as claimed in any one of claims 17 to 18 in which the side wall members and the interior members are tungsten or tantalum.

    21. A method as claimed in any one of claims 16 to 20 in which the phosphor is BaFCl:Eu, 20 ZnCDS:AG, ZnCdS:Ag, Ni, CsI:Tl, CsI:Na, CaF$_2$:Eu, Gd$_2$O$_2$S:Tb, LaOBr:Dy, LaOBr:Tm, LaOBr:Tb, La$_2$O$_2$S:Tb, Bi$_4$Ge$_3$O$_{12}$, ZnS, ZnSe, ZnTe, CdS, CdSe, CdTe, or NaI.

    22. A method as claimed in any one of claims

25 16 to 21 in which in step A, at least one wavelength conversion substance is also mixed in the liquid matrix substance, whereby the optical output of the phosphor is matched to the spectral response of a photoelectrically responsive 30 detector.

    23. A method as claimed in claim 22 in which the wavelength conversion substance is rhodamine.

    24. A method as claimed in any one of claims 16 35 to 23 in which the liquid matrix is an epoxy or a silicone polyimide copolymer.

    25. A method as claimed in claim 16 in which the hardening is caused by chemical activation, ultraviolet irradiation or by curing at a 40 temperature sufficient to harden the matrix material.

    26. A method of making a scintillator body as claimed in claim 16 substantially as hereinbefore described.

45   27. A scintillator body made in accordance with the method of claims 16 to 26.

    28. A scintillator detector array made in accordance with claim 17.