Apparatus for verifying radiation dose distributions may be used to verify a radiation dose distribution produced by a radiation beam from a medical radiation source, such as a linear accelerator. The apparatus has a tissue-equivalent scintillator located to intersect the radiation beam and an optical element that redirects light from the scintillator to a camera. A layer of transparent tissue-equivalent material adjacent to the scintillator generates backscattered radiation.

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METHOD AND APPARATUS FOR VERIFYING RADIATION DOSE DISTRIBUTIONS

Cross-Reference to Related Application

[0001] This application claims Paris convention priority from US patent application No. 60/576,591 filed on 4 June, 2004 and entitled METHOD AND APPARATUS FOR VERIFYING RADIATION DOSE DISTRIBUTIONS, which is hereby incorporated herein in its entirety. For purposes of the United States of America, this application claims the benefit of US application No. 60/576,591 under 35 U.S.C. §119.

Technical Field

[0002] The invention relates to the verification of radiation dose distributions. The invention may be applied to verifying the dose distributions to be used in intensity modulated radiation therapy (“IMRT”).

Background

[0003] Various medical conditions can be treated by way of radiation therapy. For example, some cancers can be treated by exposure to radiation. Modern methods attempt to deliver radiation dose distributions which are designed carefully to deliver radiation to desired locations while sparing surrounding tissues in a subject. For example, intensity modulated radiation therapy (IMRT) can provide improved target coverage and improved sparing of normal tissue as compared with conformal radiotherapy.

[0004] The radiation delivery apparatus can be complex. Even though such apparatus has various built-in safeguards it is desirable and, according to many protocols, mandatory to test the radiation delivery apparatus after it has been set up. Performing dose verification helps to ensure that the radiation delivery apparatus will deliver the expected distribution of radiation to a subject. The complex radiation fluence maps that can be produced by radiation delivery apparatus call for high resolution verification of the dose distribution in two or three dimensions prior to the beginning of each treatment course.
[0005] Verifying dose distributions produced by modern radiation delivery systems such as IMRT systems can be labour-intensive and time consuming. Available tools for verifying radiation dose distributions include portal imaging devices, radiographic film dosimetry and gel dosimetry. All of these tools have significant disadvantages.

[0006] Portal imaging devices are typically used during irradiation of a patient to verify the location at which radiation is delivered by localization of bony anatomy. Portal imaging devices can also be used to verify radiation dose distributions prior to treating a patient. While this can be an efficient way to perform dose verification, the inventors consider that the presence of high atomic number ("high Z") materials (such as copper intensifier plates and readout electronics) in most commercial portal imaging systems means that the dose detected is not an accurate representation of the dose delivered to the target. Common problems with such portal imaging systems include over response to low energy radiation compared to tissue. A discussion of portal imaging devices is provided in K A Langmack, Portal imaging, The British Journal of Radiology, 74 (2001), 789–804.

[0007] Ma et al. Quality assurance for dynamic multileaf collimator modulated fields using a fast beam imaging system, Med Phys. 24(8), Aug. 1997 discloses a dedicated system for verifying radiation dose distributions. The IMRT QA device available from Scanditronix Wellhofer North America of Bartlett, TN, USA is an example of a commercially available system for verifying radiation dose distributions. Each of these systems has a construction similar to a portal imaging device.


Phys. 28(12) December 2001 disclose dosimetry systems having scintillators oriented with their surfaces parallel to the direction of an incident radiation beam.

[0010] There is a need for convenient and effective methods and apparatus for verification of radiation dose distributions produced by therapeutic radiation sources.

Summary of the Invention

[0011] This invention has a number of aspects. These include, without limitation, apparatus and methods for measuring dose distributions produced by medical radiation devices such as linear accelerators.

[0012] One aspect of the invention provides apparatus for use in radiation dose measurement or verification. The apparatus comprises a tissue-equivalent scintillator which can be placed in a radiation beam and an optical element located behind the scintillator and configured to receive and redirect light emitted by the scintillator to a camera located outside of the radiation beam. A volume adjacent to the scintillator and between the scintillator and the optical element is filled with a tissue-equivalent filler material.

[0013] Further aspects of the invention and features of specific embodiments of the invention are described below.

25 **Brief Description of the Drawings**

[0014] In drawings which illustrate non-limiting embodiments of the invention,

- Figure 1 is a schematic diagram illustrating a dose measurement system;
- Figure 1A is a schematic view of the phantom of the dose measurement system of Figure 1;
Figure 2 is an image of a scintillator irradiated by a square light or radiation field which may be used to establish location of the phantom and scaling of the maps produced by the dose measurement system;

Figure 3 shows summed raw data acquired by a prototype dose measurement system;

Figure 4 shows a flood field correction map;

Figure 5 shows a flood field corrected image obtained using the flood field correction map of Figure 4;

Figures 6 and 6A are schematic cross section views of phantoms having alternative structures;

Figure 7 is a schematic top view of a phantom set up to acquire calibration data by a percentage depth dose measurement method;

Figure 8 is a map showing light intensity as a function of position on the scintillator for a percentage depth dose measurement;

Figure 9 is a calibration curve derived from a number of maps similar to the map shown in Figure 8 and the known radiation dose delivered at various depths;

Figure 10 is a dose distribution for an example field measured using apparatus according to a prototype embodiment of the invention;

Figure 11 is a map of expected dose calculated by a treatment planning system for the radiation fields used in Figure 10;

Figures 12A and 12B compare profiles of the measured doses of Figure 10 and the predicted doses of Figure 11 along selected axes;

Figure 13 is a map showing differences between the measured doses of Figure 10 and the predicted doses of Figure 11;

Figure 14 is a dose distribution for an example field measured using film dosimetry;

Figures 15A and 15B compare profiles of the film-measured doses of Figure 14 and the scintillator-system-measured doses of Figure 10 along selected axes;

Figure 16 is a map showing differences between the film-measured doses of Figure 14 and the scintillator-system-measured doses of Figure 10;
Figure 17 shows a number of comparisons of calculated doses at various points to doses measured at those points in various ways including through the use of a dose measurement system according to this invention;

Figure 18 shows a dose measurement system that incorporates an enlarged phantom body;

Figure 19 is a cross sectional schematic view of a dose measurement system having an alternative structure; and,

Figure 20 shows a phantom surrounded by tissue-equivalent material that provides sidescattered radiation.

Description

[0015] Throughout the following description, specific details are set forth in order to provide a more thorough understanding of the invention. However, the invention may be practiced without these particulars. In other instances, well known elements have not been shown or described in detail to avoid unnecessarily obscuring the invention. Accordingly, the specification and drawings are to be regarded in an illustrative, rather than a restrictive, sense.

[0016] This invention provides apparatus for measuring dose distributions. The apparatus may be applied to measuring complex dose distributions from intensity modulated radiation therapy plans, for example.

[0017] One embodiment of the invention comprises a phantom which is made of a tissue-equivalent material. A tissue-equivalent material is a material that, for the radiation in question, has a coefficient of energy absorption similar to that of living tissue and backscatters the radiation in an amount similar to living tissue. Tissue-equivalent materials are typically "low Z" materials (i.e. they are materials that have effective atomic numbers that are less than 12 and preferably less than 8).

[0018] For example, the phantom may comprise a housing of a tissue-equivalent plastic material, such as LUCITE™ filled with an optically transparent
tissue-equivalent material such as water. A scintillator screen is provided on or near one side of the phantom. The scintillator screen converts energy absorbed from the radiotherapy beam into light, typically visible light. The light emitted by the scintillator screen travels through the phantom to a mirror. The mirror reflects the light to a camera. The camera is located off-axis so that it is not damaged by radiation of the radiation beam and also so that its components do not disrupt dose measurements.

[0019] The scintillator screen and mirror are preferably of tissue-equivalent materials, such as suitable plastics. By making the portion of the phantom that is in the radiation beam tissue-equivalent, the phantom does not perturb the radiation beam during measurement. Thus the measured dose can accurately represent the dose that will be delivered to the target.

[0020] The scintillator may, for example, comprise a plastic scintillator or other tissue-equivalent scintillator. Various plastic scintillators are available. These include the model EJ-200 plastic scintillator available from Eljen Technology, U.S.A. The scintillator has a suitable thickness, for example about 6 mm. This scintillator has a polyvinyltoluene polymer base, and 64% anthracene for light output. The efficiency of such a scintillator can be 10,000 photons/MeV. The light emission peaks at a wavelength of 425 nm. The decay time is 2.1 ns. This material has a density of 1.032 g/cm³ and an effective Z of 5.76. This material has the advantages of being easy to shape and machine. Beddar AS et al. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: I. Physical characteristics and theoretical considerations, Phys. Med. Biol. 37(10):1883 and Beddar, A S et al. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: II. Properties and measurements, Phys. Med. Biol. 37(10):1901, describe various studies of the properties and response to radiation of scintillators. The mirror may comprise a Lucite™ mirror, which may be a first surface mirror, for example.
[0021] The camera may be a digital camera having a suitable imaging array, such as a
CCD array. The camera may be located outside of the phantom. The camera may be
shielded, for example with lead, to protect its components from stray radiation. In a
prototype dosimetry system the camera was a Hitachi™ model KP-M1 high-
resolution monochrome CCD video camera available from Hitachi-Denshi of USA.
The camera was interfaced to a computer by a Matrox Meteor II™ frame grabber
available from Matrox Electronic Systems Ltd. of Dorval, Quebec, Canada. Any other
suitable camera and computer interface compatible with the camera may be used.

[0022] Outside surfaces of the phantom may be opaque to reduce the contribution of
stray light to the camera image. For example, the housing may comprise an opaque
plastic such as opaque LUCITE™. In this case the phantom has an optically-
transparent viewing window in an optical path between the mirror and the camera.

[0023] The amount of light emitted by a portion of the scintillator screen increases
with the amount of radiation passing through that portion of the scintillator screen.
The relationship between emitted light and radiation can be measured and the
measurements used to calibrate the apparatus.

[0024] Images acquired by the camera are provided to a data processor, which may
comprise a suitably-programmed computer by way of a suitable interface. The
computer is linked to receive a signal from a monitor unit of the radiation delivery
apparatus (typically a linear accelerator). The camera integrates the emitted light and
produces a cumulative image for every monitor unit delivered.

[0025] Figure 1 shows apparatus 10 according to one embodiment of the invention.
Apparatus 10 includes a phantom 11. One face 12 of phantom 11 comprises a
scintillator sheet of a tissue-equivalent material. Another face 14 extending
substantially perpendicular to scintillator sheet 12 is transparent, or substantially so, to
light given off by scintillator sheet 12 when scintillator sheet 12 is irradiated by a
radiation beam from an irradiating apparatus 13 (such as a linear accelerator). Face 12
is preferably transparent or translucent. The remaining walls of phantom 11, including side walls 14A and 14B and rear wall 14C may be opaque to the light given off by scintillator sheet 12.

[0026] Within phantom 11 is a mirror 16. Mirror 16 is mounted on a tissue-equivalent wedge 17 and is oriented at an angle to scintillator sheet 12 so that it reflects light emitted by scintillator sheet 12 to a camera 20. Mirror 16 is conveniently oriented at an angle of approximately 45 degrees to scintillator sheet 12.

[0027] A light-tight housing 18 may be provided to prevent stray light from entering camera 20. Housing 18 may also hold camera 20 at a fixed position relative to mirror 16. An opaque layer of a tissue-equivalent material (not shown in Figure 1) may also be provided on top of scintillator sheet 12 so that apparatus 10 may be used in a room that is not dark.

[0028] Phantom 11 has a flat top surface 19. Blocks or sheets 19A of tissue-equivalent material may be placed on surface 19 to duplicate the effect of layers of tissue overlying scintillator sheet 12. For example, blocks of Solid Water™ or other suitable tissue-equivalent material may be stacked on top of scintillator sheet 12 to provide dose measurements at depth. In a prototype, phantom 11 has dimensions of about 17cm wide \times 17cm deep \times 20cm high.

[0029] In some embodiments, the radiation emitted by irradiating apparatus 13 is of a type that is backscattered significantly by high Z materials. For example, the radiation may be made up of photons having energies of several MeV. In such cases it is desirable that mirror 16 is made of a plastic or other tissue-equivalent material.

[0030] Light emitted by scintillator sheet 12 is turned through an angle so that it is directed out of the radiation beam (90 degrees in the illustrated embodiment) by mirror 16 and directed out through transparent face 14 to digital camera 20. The
volume 23 of phantom 11 between mirror 16 and scintillator sheet 12 is filled with a transparent tissue-equivalent material, for example, water.

[0031] In some applications, the radiation being measured is of a type that is backscattered significantly by tissue. For example, MeV X-ray photons are backscattered by tissue. In such cases it is desirable to provide a layer of tissue-equivalent material on the side of scintillator screen 12 away from the radiation source that is at least as thick as a characteristic distance of the backscattering. This permits the dose measured at scintillator screen 12 to include a contribution arising from backscattering from deeper layers of tissue. In apparatus 10 of Figure 1, the tissue-equivalent filler material in volume 23, the tissue-equivalent mirror 16 and the tissue-equivalent wedge 17 provide this function. The layer of tissue-equivalent material may be more than 5 cm thick. In some cases, the layer of tissue-equivalent material is in the range of about 5 cm to about 15 cm thick. Preferably, where the radiation consists primarily of MeV X-ray photons, the layer of tissue-equivalent material is at least 8 to 10 cm thick. The layer of tissue-equivalent material may comprise water or a tissue-equivalent plastic for example.

[0032] Images 21 from digital camera 20 and information regarding a current intensity of the radiation beam (e.g. monitor unit ("MU") pulse signals from apparatus 13) are delivered to a computer 22. Computer 22 may comprise one or more stand-alone data processors, or one or more data processors that are integrated with some other device. In some embodiments, computer 22 may also perform control functions for radiation source 13. Image 21 comprises image data that indicates the intensity of light observed by camera 20 to emanate from different locations on scintillator sheet 12.

[0033] It is desirable to determine the location of phantom 11 relative to a radiation beam. In systems which detect radiation by way of films this can be difficult. Apparatus as shown in Figure 1 can be used to determine the position of phantom 11 relative to a radiation beam by setting jaws of the radiation source (e.g. a linear accelerator) to produce a radiation beam having known dimensions. The image of the
light emitted by scintillator sheet 12 in response to the radiation beam indicates the location of phantom 11 relative to the radiation beam. A cross hair can be set to the centre or to edges of the illuminated area of the image.

[0034] As discussed further below, many medical radiation sources include a light projector that projects a beam of light having the same dimensions as a radiation beam. A radiation source equipped with such a light projector can project a beam of light onto scintillator sheet 12. The projected light can be imaged by camera 20. When the light projector is properly aligned, the image of the light beam indicates the location of phantom 11 relative to the location of a radiation beam.

[0035] Figure 2 is an image from camera 20 of scintillator 12 while phantom 11 of a prototype system is being irradiated by a square radiation field (or light beam). The location of the radiation field (or light beam) can be clearly identified in the image.

[0036] It is also important in many applications to know how the scale of features in the image obtained by camera 20 relates to the scale of the radiation field being measured. This can be determined by measuring the locations of the edges of the illuminated areas which correspond to edges of a radiation beam that has known dimensions.

[0037] Apparatus like that of Figure 1 can be used to measure a dose distribution produced by an IMRT sequence by capturing a sequence of images 21 over the time period while IMRT fields are being delivered to phantom 11, determining a radiation dose map for each of the images 21 and then summing the radiation dose maps to obtain the dose resulting from the IMRT sequence. The images are acquired at a rate sufficient to track changes in the IMRT fields. For example, in a prototype embodiment, images are acquired at a rate of approximately 25 Hz for IMRT fields that are applied for approximately 3 seconds each. Thus, apparatus 10 can be used to measure radiation doses for IMRT fields being delivered dynamically.
[0038] Providing a link between the radiation delivery apparatus (e.g. linear accelerator) and computer 22 permits temporal dose information to be obtained. Temporal dose information indicates how dose is deposited over time. Knowledge of how the dose is deposited throughout a treatment may help in identifying the causes of any dose discrepancies.

[0039] To achieve the best quantitative dose measurements with a system including a phantom 11 it is desirable to correct for scattering and, more generally, optical photon spread. Optical photon spread may be accounted for by deconvolving a glare kernel from the raw images. For example, a Gaussian glare kernel may be deconvolved using a fast Fourier transform ("FFT") or equivalent algorithm. Where construction of phantom 11 causes the optical photon spread to be asymmetrical, an asymmetrical glare kernel, such as an elliptical Gaussian glare kernel may be used.

[0040] Figure 3 shows a summation of raw data for a simple square field. If necessary, the raw data is processed to reduce or eliminate any artefacts (such as the intensity gradient described below) that may result from the way in which optical photons are generated in or propagate through phantom 11. The data is also processed to correct for the general blurring of optical photons. Blurring may be corrected by deconvolving the images using a suitable glare kernel.

[0041] An intensity gradient can be observed in Figure 3. The inventors believe that this intensity gradient is the result of mirror 16 being farther from scintillation screen 12 on one side of the image than the other. The higher intensity at the top of Figure 3 is believed to be the result of multiple reflections of photons between mirror 16 and scintillation screen 12.

[0042] One way to minimize the effect of the intensity gradient exhibited in Figure 3 is to create a flood field correction map. This may be done by irradiating phantom 11 with a large open field. For example, a 25cm by 25 cm open field at a depth of 10 cm has been used to determine a flood field correction map for a prototype phantom 11.
In addition to minimizing the effect of the intensity gradient of Figure 3, the flood field correction map may be used to correct for any intrinsic inhomogeneities in the response of scintillator sheet 12 to radiation. A flood field correction map may be used to correct for such intrinsic inhomogeneities even if an alternative method is used to avoid or cancel the gradient illustrated in Figure 3.

[0043] Figure 4 shows an example flood field correction map. The summed raw data can be corrected by dividing the summed raw data by the flood field correction map to yield a flood field corrected image of the scintillation screen as illustrated in Figure 5.

[0044] Another way to reduce the intensity gradient is to move mirror 16 so that it is farther from scintillator sheet 12. Figure 6 shows a phantom 11A wherein mirror 16 is displaced away from scintillator sheet 12. A layer 24 of tissue-equivalent material, such as water or a suitable plastic, is provided adjacent to scintillator sheet 12.

[0045] In phantom 11A, mirror 16 may be in air. Increasing the proportion of the path between scintillator screen 12 and camera 20 that is in air can reduce blurring. Reducing the amount of tissue-equivalent material in phantom 11 can reduce weight and cost. In the alternative, water or other tissue-equivalent material may fill the space 23 between mirror 16 and scintillator sheet 12.

[0046] Another way to reduce the intensity gradient is to provide a light collimator between scintillator sheet 12 and mirror 16. Figure 6A shows an example of a phantom 11B that includes a collimator 25. In the illustrated example, collimator 25 comprises a plurality of vanes 26 that extend in a direction transverse to mirror 16 (i.e. in a direction such that all parts of each vane are essentially the same distance from mirror 16). Vanes 26 are preferably of a low Z material and are preferably immersed in water or another optically-transparent and tissue equivalent medium. Collimator 25 may comprise, for example, a micro-louved screen such as a microlouvre film of the type used in 3M™ privacy filters for computer monitors. The screen may be directly
against scintillator sheet 12. It is optional and not necessary for collimator 25 to collimate light in the direction parallel to vanes 26.

[0047] The provision of collimator 25 reduces the intensity gradient and thereby simplifies deconvolution to reduce glare. A suitable asymmetrical glare kernel, such as an asymmetrical elliptical Gaussian kernel may be used to reduce glare.

[0048] To obtain quantitative measurements of dose it is necessary to calibrate the apparatus (i.e. to establish a relationship between the intensity of light at a point in the image 21 acquired by camera 20 and the radiation dose at a corresponding location on scintillator screen 12). System calibration may be performed using radiation fields of known dose distributions.

[0049] One way to obtain such a calibration relationship is to use a percent dose distribution strategy. Figure 7 is a schematic top view of a phantom 11 arranged to facilitate the acquisition of percent dose distribution curves across scintillator sheet 12. A radiation beam is incident in a direction parallel to scintillator sheet 12 so that the dose decreases with distance along scintillator sheet 12 in the direction of the beam (i.e. the dose decreases with increasing depth measured from the surface 27 of phantom 11 on which the radiation beam is incident. Figure 8 is an example percent depth dose measurement. Intensity to dose values may be obtained by measuring the light intensity at a set of points (for example the points marked with dark squares in Figure 8) and comparing those intensities to doses known to have been delivered at the locations of the points.

[0050] The result is a calibration curve. Figure 9 is an example calibration curve. The calibration curve of Figure 9 can be applied to data that has been flood field corrected, if necessary or desired, and corrected to reduce blurring to yield a 2-dimensional dose distribution map as shown in Figure 10.
[0051] The calibration of a dose measurement system can be checked by comparing the measured dose distribution to an expected dose distribution. Figure 11 shows the dose distribution expected for the radiation fields that were applied to produce Figure 10. The dose distribution map of Figure 11 was calculated by an ECLIPSE™ treatment planning system. Figures 12A and 12B show profiles which compare expected to measured doses along two axes. The axes along which the profiles of Figure 12A and 12B were measured are indicated respectively by the horizontal and vertical lines shown in Figure 11.

[0052] When apparatus 10 has been calibrated, the expected dose data and measured dose data can be compared by subtracting one from the other to yield a dose difference map as shown in Figure 13. In some embodiments of the invention this comparison is performed in computer 22. The dose difference map may be displayed on a display associated with computer 22 or stored for future reference. Computer 22 may also extract and inform a user of information such as one or more of:

- the magnitude of the maximum amount by which the measured dose exceeds the expected dose;
- the magnitude of the maximum amount by which the measured dose is less than the expected dose;
- a computed measure of the overall acceptability of the fit between the measured dose and the expected dose; and,
- the like.

[0053] Another calibration method involves delivering different doses of radiation using a radiation beam incident perpendicular to scintillator sheet 12. The radiation beam may be a square beam, such as a 10 cm × 10 cm square or a 5 cm × 5 cm square, for example. In an example calibration method, varying radiation doses are delivered with a 5 cm × 5 cm square radiation field at a depth of 3 cm and a source-to-axis distance (“SAD”) of 100 cm while monitoring the mean intensity of light emitted from a small region (e.g. 1 cm²) of scintillator sheet 12 surrounding the isocenter of the radiation beam.
IMRT dose distributions acquired using a prototype system according to the invention have been verified using film dosimetry and a high resolution mini-ionization chamber. Figure 14 shows the dose measured by film dosimetry of the same radiation fields used to produce the dose map of Figure 10. Figures 15A and 15B compare the doses measured with film dosimetry to the doses shown in Figure 10 as measured by the prototype scintillator system along the axes respectively indicated by horizontal and vertical lines in Figure 14. Figure 16 is a dose difference map showing differences between the doses measured by film dosimetry and the doses shown in Figure 10 as measured by the prototype scintillator system.

Figure 17 compares doses measured by the system of the invention (in the column labelled Scintillator Dosimetry Measurement) to the expected values indicated by the treatment planning system and corresponding values measured by a mini-ionization chamber and film dosimetry.

It can be appreciated from the foregoing that apparatus according to the invention may be configured to permit measurement of dose distributions in a beam's-eye view ("BEV"), that is perpendicular to the beam direction. The BEV is the preferred orientation for field-by-field dose verification.

A dose measurement apparatus which incorporates a digital camera, as described above, is much less labour-intensive and time consuming to operate than a film-based or a gel-based dosimetry system. In film dosimetry systems, films must be developed and scanned. In gel dosimetry systems gels must be scanned with an MRI or a CT. Additionally, use of a system as described above can result in cost saving over film-based systems since film and processor chemicals are not required.

With apparatus as described above, multiple measurements of the same field can be performed rapidly to check any questionable results. The apparatus is relatively pressure and temperature insensitive.
[0059] As noted above, linear accelerators and other irradiation apparatus often include light projectors. The light projectors project beams of light having edges that are intended to coincide with edges of the radiation beam. A technician can use the light beam to ensure that the radiation beam will be delivered in the correct location on a subject. It is typically necessary to check periodically to ensure that the boundaries of the light beam issued by such a light projector do coincide with the boundaries of the radiation beam. Typical protocols require that this alignment be performed to an accuracy of better than ±2 mm.

[0060] Where there is no opaque material between the light projector and scintillator sheet 12, an apparatus as described above may be used to verify correct alignment of the light projector by imaging the light produced by the light projector using camera 20 and also imaging light given off by scintillator sheet 12 while a radiation beam is incident on scintillator sheet 12. Where there is an optically opaque layer on scintillator sheet 12 to permit apparatus 10 to be used in a room that is not dark, the opaque layer may be removable (e.g. it may be provided by an opaque cover that can be taken off or opened) to permit apparatus 10 to be used for alignment of a light projector.

[0061] The apparatus may comprise a diffuser screen close to and extending parallel to a plane of the scintillator. The projected area of the light from the light projector on the diffuser screen is imaged by the camera. The diffuser screen may be integrated with the scintillator.

[0062] Image analysis software in computer 22 may be configured to compare a first image of the light produced by the light projector to a second image of the light produced by a radiation beam interacting with scintillator sheet 12. The software may, for example, determine whether, and in what direction, the light produced by the light projector should be translated so as to coincide with the radiation beam. This
determination may be made, for example, by maximizing a correlation between the first and second images.

[0063] Apparatus as described herein may be integrated with a larger phantom. The phantom may be anthropomorphic or nearly so. For example, the phantom may have rounded edges. Figure 18 shows an example system 50 that includes a body 52 of a tissue-equivalent material. Body 52 has an interior cavity 54 that receives a phantom 11 that may be constructed as described above, for example. The body may be aligned with a radiation beam so that scintillator sheet 12 is located at the isocenter of the radiation device (i.e. the point in space where radiation beams emitted at different gantry angles intersect).

[0064] System 50 may be used to measure the cumulative dose distribution in the plane of scintillator sheet 12 from several coplanar treatment beams. A film cassette or a cassette containing a dosimetric gel could be inserted into cavity 54 in place of phantom 11 to cross verify the results provided by phantom 11.

[0065] Figure 19 shows apparatus 60 according to another embodiment. Apparatus 60 has a base 62 housing mirror 16. Base 62 supports a scintillator assembly 64 comprising a scintillator sheet 12, a first layer 65 of an optically transparent tissue-equivalent material between scintillator sheet 12 and base 62 and a second layer 66 of tissue equivalent material on the side of scintillator sheet 12 away from base 62. The edges 67 of scintillator assembly 64 are radiused.

[0066] Camera 20 images light given off by scintillator sheet 12 while co-planar radiation beams are incident on apparatus 60. Scintillator sheet 12 may intersect the isocenter of the radiation device producing the radiation beams.

[0067] In apparatus 60 it is preferred that base 62 be essentially transparent to radiation. This can be achieved by making base 62 of one or more low Z materials and by making the walls and mirror 16 of base 62 thin.
[0068] In some cases, sidescattered radiation can contribute significantly to the dose that will be delivered to the tissues of a subject. To measure dose in a manner that ensures that sidescatter will be taken into account, a phantom may be provided with thick walls of tissue-equivalent material or the phantom may be nested among blocks of tissue-equivalent material. Figure 20 shows a phantom 70 that includes blocks 72 of tissue-equivalent material that surround scintillator sheet 12. Radiation will be sidescattered from blocks 72 in a way that simulates the side scattering of radiation from adjacent tissues when radiation is delivered to a subject. A block 72A that makes up or sits adjacent to optical window side 14 of the phantom is transparent to permit a camera (not shown in Figure 20) to image the light emitted by scintillator sheet 12.

[0069] Certain implementations of the invention comprise computer processors which execute software instructions which cause the processors to perform a method of the invention. For example, one or more processors in a dose measurement system according to this invention may implement the methods described above executing software instructions in a program memory accessible to the processors. The invention may also be provided in the form of a program product. The program product may comprise any medium which carries a set of computer-readable signals comprising instructions which, when executed by a computer processor, cause the data processor to execute a method of the invention. Program products according to the invention may be in any of a wide variety of forms. The program product may comprise, for example, physical media such as magnetic data storage media including floppy diskettes, hard disk drives, optical data storage media including CD ROMs, DVDs, electronic data storage media including ROMs, flash RAM, or the like or transmission-type media such as digital or analog communication links. The instructions may be optionally compressed and/or encrypted on the program product.

[0070] Where a component (e.g. a software module, processor, assembly, device, circuit, etc.) is referred to above, unless otherwise indicated, reference to that component (including a reference to a "means") should be interpreted as including as
equivalents of that component any component which performs the function of the described component (i.e., that is functionally equivalent), including components which are not structurally equivalent to the disclosed structure which performs the function in the illustrated exemplary embodiments of the invention.

[0071] As will be apparent to those skilled in the art in the light of the foregoing disclosure, many alterations and modifications are possible in the practice of this invention without departing from the spirit or scope thereof. For example:

- While it is preferred in many applications, it is not mandatory that the scintillator screen be oriented perpendicular to the radiation beam. The scintillator screen could be at a different angle to the radiation beam such that the projection of the radiation beam onto the scintillator sheet covers an area on the scintillator sheet.
- The mirror is not necessarily at an angle of 45 degrees to the radiation beam. Other geometries can be used in which the camera is outside of the radiation beam and images the scintillator screen which is in the radiation beam.
- The optical path between the scintillator screen and the camera may include multiple mirrors or other optical elements which permit the camera to image the scintillator screen while the camera is in a location outside of the radiation beam. For example, the optical element may comprise a bundle of optical fibers that carry light to the camera.
- The camera may be made especially sensitive to light at wavelengths given off predominantly by the scintillator screen.
- The apparatus may include a filter in the optical path between the scintillator and the camera to reduce the effect of any stray light on the images acquired by the camera.
- The phantom may be used to measure three dimensional dose distributions by moving it through the radiation field as described, for example, in Nishizawa, US patent No. 6,594,336.
- Light from a scintillator screen may be directed to a camera by an array of optical fibres instead of or in addition to a mirror. In some embodiments, the optical
fibres may be made of a tissue-equivalent material and may themselves provide a layer of tissue-equivalent material adjacent to the scintillator sheet. Accordingly, the scope of the invention is to be construed in accordance with the substance defined by the following claims.
WHAT IS CLAIMED IS:

1. Apparatus for use in radiation dose measurement or verification, the apparatus comprising:
   a tissue-equivalent scintillator which can be placed in a radiation beam;
   and,
   an optical element located behind the scintillator and configured to receive and redirect light emitted by the scintillator to a camera located outside of the radiation beam;
   wherein a volume adjacent to the scintillator and between the scintillator and the optical element is filled with a tissue-equivalent filler material.

2. Apparatus according to claim 1 wherein the optical element comprises a mirror.

3. Apparatus according to claim 2 wherein the mirror is a planar mirror oriented at an angle to the scintillator.

4. Apparatus according to claim 2 or 3 wherein the mirror is of a tissue-equivalent material.

5. Apparatus according to claim 3 or 4 comprising a wedge of a tissue-equivalent material supporting the mirror at the angle to the scintillator.

6. Apparatus according to any one of claims 1 to 5 wherein an optical path followed by redirected light from the optical element to the camera extends substantially at a right angle to a direction of the radiation beam.

7. Apparatus according to any one of claims 1 to 6 wherein the optical element is made of one or more tissue-equivalent materials.
8. Apparatus according to any one of claims 1 to 7 wherein the filler material comprises water.

9. Apparatus according to any one of claims 1 to 7 wherein the filler material comprises a solid material.

10. Apparatus according to one of claims 1 to 9 wherein the filler material is arranged in a layer having a thickness at least equal to a distance characterizing backscattering of the radiation.

11. Apparatus according to one of claims 1 to 9 wherein the filler material is arranged in a layer having a thickness of at least 5 cm.

12. Apparatus according to one of claims 1 to 9 wherein the filler material is arranged in a layer having a thickness in the range of 5 cm to 15 cm.

13. Apparatus according to any one of claims 1 to 12 comprising a light collimator between the scintillator sheet and the optical element.

14. Apparatus according to claim 13 wherein the light collimator comprises a plurality of vanes.

15. Apparatus according to claim 14 wherein the optical element is generally planar and is farther from the scintillator on a side closest to the camera than it is on a side farther from the camera and the vanes extend in a transverse direction such that all parts of each vane are essentially the same distance from the optical element.

16. Apparatus according to claim 14 or 15 wherein the light collimator comprises a microslouvre film.
17. Apparatus according to any one of claims 13 to 16 wherein the light collimator is made entirely of one or more materials having effective atomic numbers less than 12.

18. Apparatus according to any one of claims 13 to 16 wherein the light collimator is located directly against the scintillator.

19. Apparatus according to any one of claims 13 to 16 wherein the light collimator is embedded in an optically transparent tissue equivalent material.

20. Apparatus according to claim 19 wherein the optically transparent tissue equivalent material is a liquid material.

21. Apparatus according to claim 20 wherein the liquid material is water.

22. Apparatus according to any one of claims 1 to 21 wherein, upon exposure to the radiation, the scintillator emits light in a wavelength range and the apparatus comprises an optical bandpass filter between the scintillator and the camera, the optical bandpass filter passing light in the wavelength range and attenuating light having wavelengths outside of the wavelength range.

23. Apparatus according to any one of claims 1 to 22 wherein the scintillator comprises a sheet of a polymer material having an effective atomic number of less than 12.

24. Apparatus according to claim 23 wherein the scintillator comprises a polyvinyltoluene polymer material.

25. Apparatus according to any one of claims 1 to 24 in combination with a radiation source incorporating a light projector wherein a substantially
unobstructed optical path permits light from the light projector to be projected onto the scintillator and viewed by the camera.

26. Apparatus according to claim 25 comprising a diffuser screen extending parallel to a plane of the scintillator.

27. Apparatus according to claim 26 wherein the diffuser screen is integrated with the scintillator.

28. Apparatus according to any one of claims 24 to 28 comprising a removable optically opaque layer on a side of the scintillator away from the optical element.

29. Apparatus according to one of claims 24 to 28 comprising a data processor connected to receive image data from the camera, the data processor configured to compare a first image of light produced by the light projector to a second image of light produced by a radiation beam interacting with the scintillator to determine a direction in which the light produced by the light projector should be translated so as to coincide with the radiation beam.

30. Apparatus according to claim 29 wherein the data processor is configured to maximize a correlation between the first and second images.

31. Apparatus according to one of claims 1 to 28 comprising a data processor connected to receive image data from the camera, the data processor configured to apply a glare kernel to the image data to obtain image data corrected for light scattering.

32. Apparatus according to claim 31 wherein the glare kernel is an asymmetrical glare kernel.
33. Apparatus according to claim 31 or 32 wherein the glare kernel is a Gaussian glare kernel.

34. Apparatus according to claim 31, 32 or 33 wherein the data processor is connected to receive information regarding a current intensity of the radiation beam.

35. Apparatus according to one of claims 31 to 34 wherein the data processor is configured to sum information derived from image data for each of a plurality of images received from the camera to obtain a dose map indicating radiation dose as a function of position on the scintillator.

36. Apparatus according to any of claims 1 to 35 comprising a phantom of a tissue-equivalent material having a cavity disposed therein, wherein the scintillator, optical element and volume adjacent to the scintillator are removably disposed within the cavity.

37. Apparatus according to claim 36 wherein the scintillator, optical element and volume adjacent to the scintillator are removable from the cavity as a unit.

38. Apparatus according to any one of claims 1 to 4 wherein the volume comprises a layer of a solid tissue-equivalent material.

39. Apparatus according to claim 38 wherein the solid tissue-equivalent material is bonded to the scintillator.

40. Apparatus according to any one of claims 1, 38 or 39 wherein the scintillator is disposed between two layers of tissue equivalent material.

41. Apparatus according to claim 40 wherein the two layers of tissue-equivalent material are both solid materials.
42. Apparatus according to claim 40 or 41 wherein the scintillator and two layers of tissue equivalent material are incorporated in a removable scintillator assembly having substantially planar first and second faces extending substantially parallel to a plane of the scintillator and radiused edges.

43. Apparatus according to one of claims 40 to 42 wherein the optical element is disposed in a support structure that is substantially transparent to the radiation.

44. Apparatus according to claim 43 wherein the support structure comprises a pair of thin opposed side walls of a tissue equivalent material and the optical element is supported between the side walls.

45. Apparatus according to claim 1 wherein the scintillator is disposed between two volumes of tissue equivalent material.

46. Apparatus according to one of claims 1 to 28 comprising a data processor connected to receive image data from the camera, the data processor configured to compute a dose distribution map from the image data and to compute a difference map comprising information identifying differences between the dose distribution map and a map of expected dose distribution as a function of position.

47. Apparatus of claim 46 wherein the map of expected dose distribution is output by a radiation treatment planning system and the radiation treatment planning system is connected to provide control information to control a source of the radiation beam.

48. Apparatus comprising any new and inventive feature, means, combination of features and/or means or subcombination of features and/or means described herein.
49. A method comprising any new inventive step, act, combination of steps and/or acts or subcombination of steps and/or acts described herein.
FIGURE 15A

FIGURE 15B
<table>
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<tr>
<th>POSITION</th>
<th>TREATMENT PLANNING SYSTEM</th>
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<th>NAC MINI-IONIZATION CHAMBER</th>
<th>FILM DOSIMETRY</th>
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* ALL DOSES IN %

**FIGURE 17**
FIGURE 18
### INTERNATIONAL SEARCH REPORT

**A. CLASSIFICATION OF SUBJECT MATTER**

**IPC(7):** G01T 1/164, G01T 1/29, A61N 5/10

According to International Patent Classification (IPC) or to both national classification and IPC

**B. FIELDS SEARCHED**

Minimum documentation searched (classification system followed by classification symbols)

G01T-1/164, 1/29; A61N-5/10

CA Classes: 358/11.05, 167/47; US Classes: 350/363.01, 367, 368; 378/65; 382/65, 128

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic database(s) consulted during the international search (name of database(s) and, where practicable, search terms used)

Qpat, West, Delphicon, IEEE, Cassis, PubMed (MedLine)

**scintill+, tissue(adjl)equivalent or tissue(adjl)simil+, phantom, radiation, orthogonal**

### C. DOCUMENTS CONSIDERED TO BE RELEVANT

<table>
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<th>Category*</th>
<th>Citation of document, with indication, where appropriate, of the relevant passages</th>
<th>Relevant to claim No.</th>
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<tr>
<td>Y</td>
<td>US - 6,066,851 A 23 May 2000 (23-05-2000) Madono et al. Col. 2, lines 12 to 14; col. 6, line 56 to col. 7, line 22; col. 8, line 19 to col. 9, line 20; Figures 6 and 8</td>
<td>1 to 3, 6, 8, 9, 11, 12, 22, 38 to 42, and 45</td>
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<td>Y</td>
<td>US - 5,625,137 A 29 April 1997 (29-04-1997) Madsen et al. Col. 5, line 39 to col. 4 line 45</td>
<td>1 to 3, 6, 8, 9, 11, 12, 22, 25, 38 to 42, and 45</td>
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<tr>
<td>Y</td>
<td>US - 6,635,486 B2 21 October 2003 (21-10-2003) Madsen et al. Col. 7 line 40 to col. 8 line 52</td>
<td>1 to 3, 6, 8, 9, 11, 12, 22, 25, 38 to 42, and 45</td>
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<td>Y</td>
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<td>1 to 3, 6, 9, 11, 12, 22, 25, 38 to 42, and 45</td>
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<td>Y</td>
<td>Boon et al., Performance of a fluorescent screen and CCD camera as a two-dimensional dosimetry system for dynamic treatment techniques, October 2000, Medical Physics, Vol. 27, No. 10, pp. 2198 to 2208, Pages 2203 to 2204</td>
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</table>

[ ] Further documents are listed in the continuation of Box C.  

[ ] See patent family annex.

* Special categories of cited documents:
  
  "A" document defining the general state of the art which is not considered to be of particular relevance
  
  "E" earlier application or patent but published on or after the international filing date
  
  "L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
  
  "O" document referring to an oral disclaimer, use, exhibition or other means
  
  "P" document published prior to the international filing date but later than the priority date claimed

**T** later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

**X** document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

**Y** document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other cited documents, such combination being obvious to a person skilled in the art

**&** document number of the same patent family

Date of the actual completion of the international search: 23 September 2005 (23-09-2005)

Date of mailing of the international search report: 30 September 2005 (30-09-2005)

Name and mailing address of the ISA/CA:  

Authorized officer:  

Louis-Pierre Riel (819) 997-0232

Form PCT/ISA/210 (second sheet) (April 2005)
INTERNATIONAL SEARCH REPORT

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<td>1.</td>
<td>[ ] Claim Nos.: because they relate to subject matter not required to be searched by this Authority, namely:</td>
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<td>2.</td>
<td>[X] Claim Nos.: 48 and 49 because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out, specifically: These claims rely on references to the description, therefore are not complying with Rule 6.2(a) of the PCT.</td>
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<td>3.</td>
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<td>1.</td>
<td>[ ] As all required additional search fees were timely paid by the applicant, this international search report covers all searchable claims.</td>
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<td>[ ] As all searchable claims could be searched without effort justifying additional fees, this Authority did not invite payment of additional fees.</td>
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<td>[ ] No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claim Nos.:</td>
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**Remark on Protest**

[ ] The additional search fees were accompanied by the applicant's protest and, where applicable, the payment of a protest fee.

[ ] The additional search fees were accompanied by the applicant's protest but the applicable protest fee was not paid within the time limit specified in the invitation.

[ ] No protest accompanied the payment of additional search fees.
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