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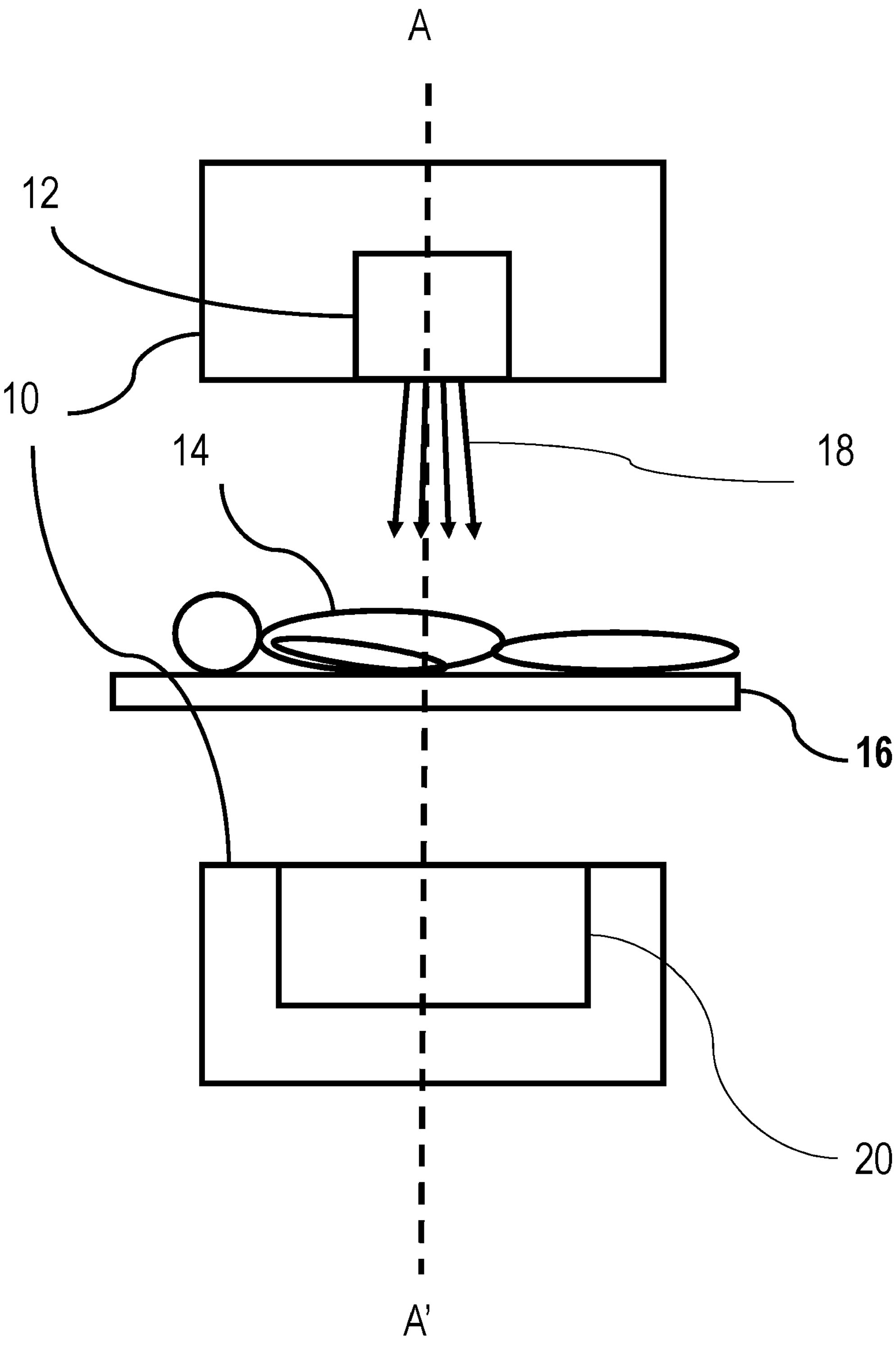


Figure 1a

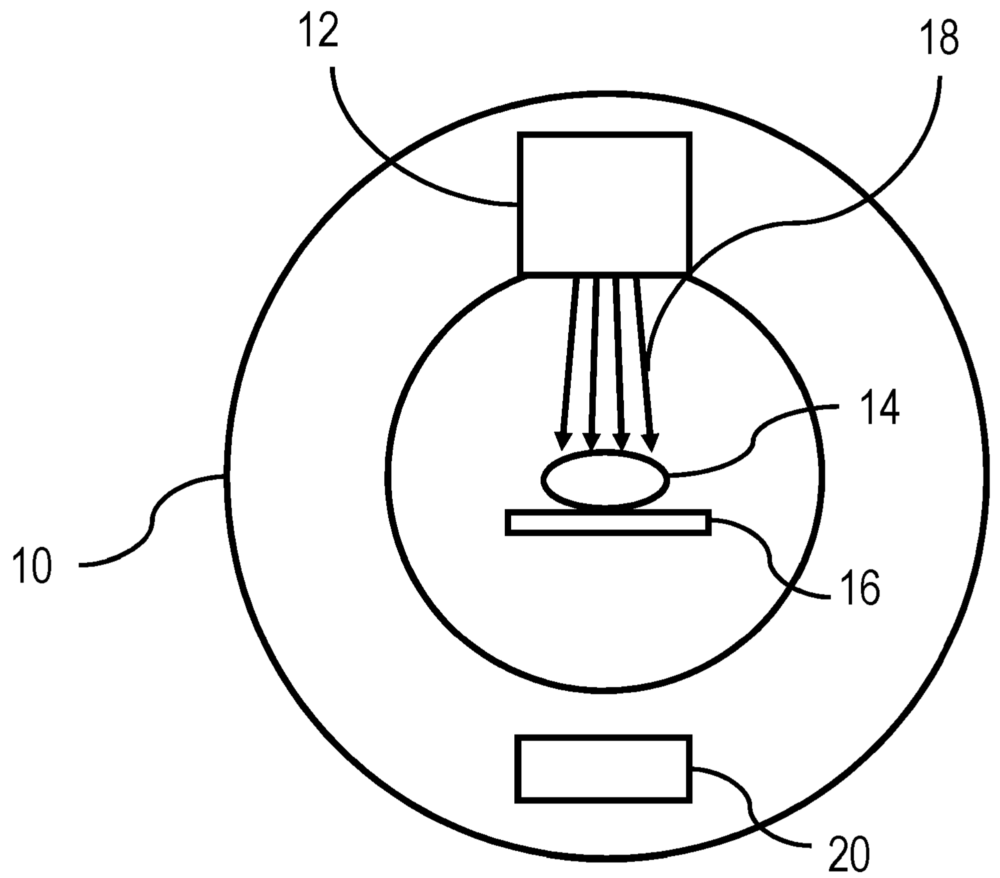
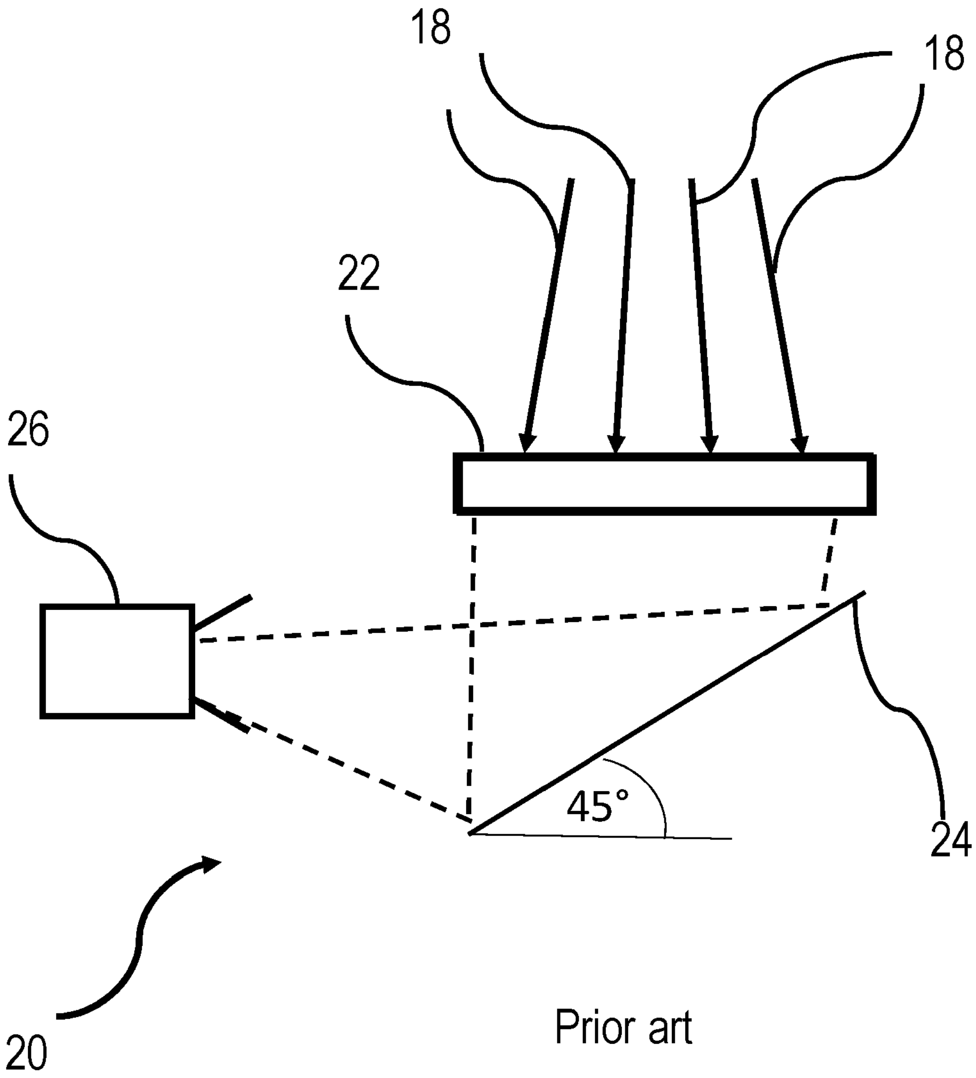


Figure 1b



Prior art

Figure 2

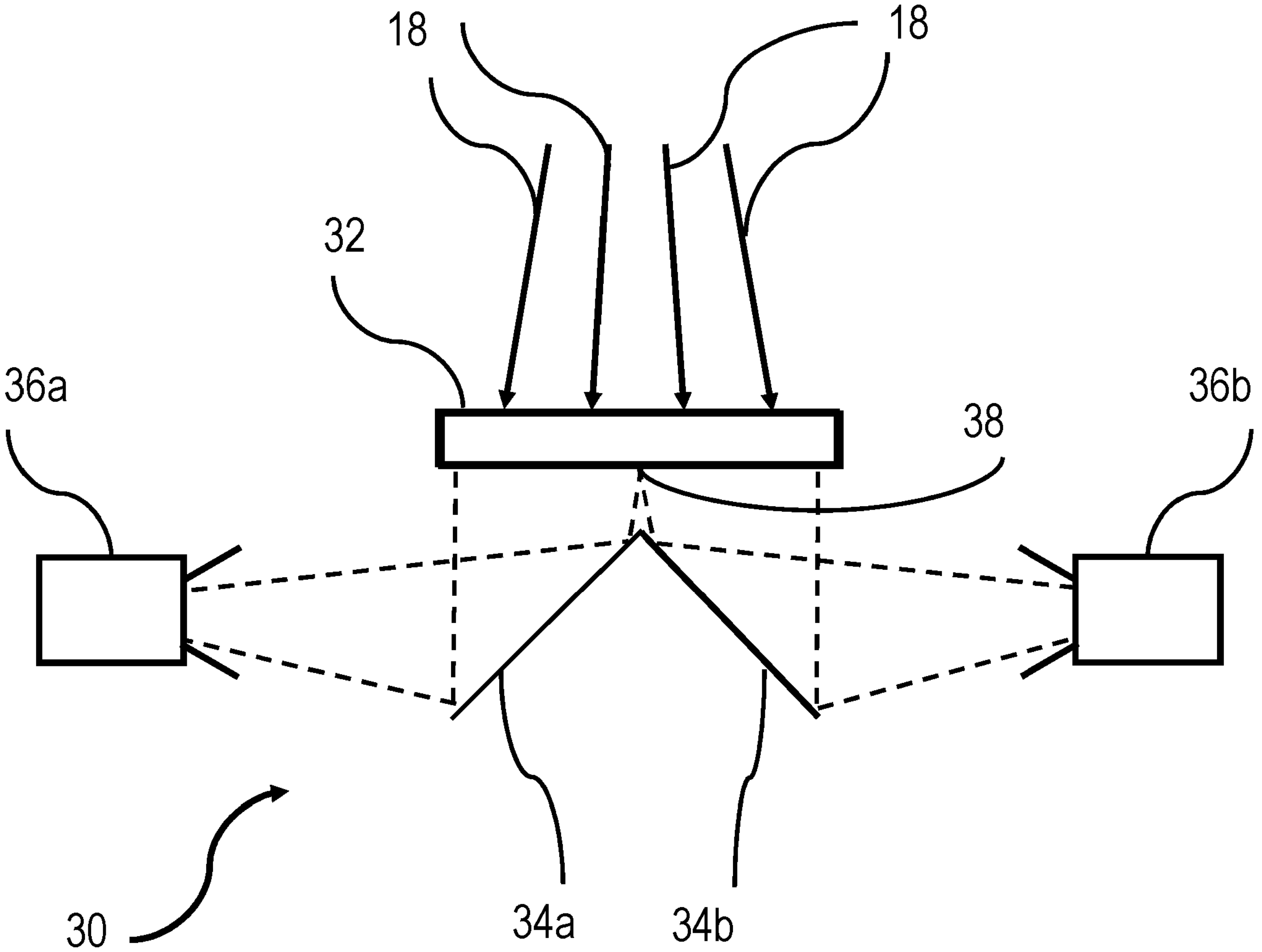


Figure 3

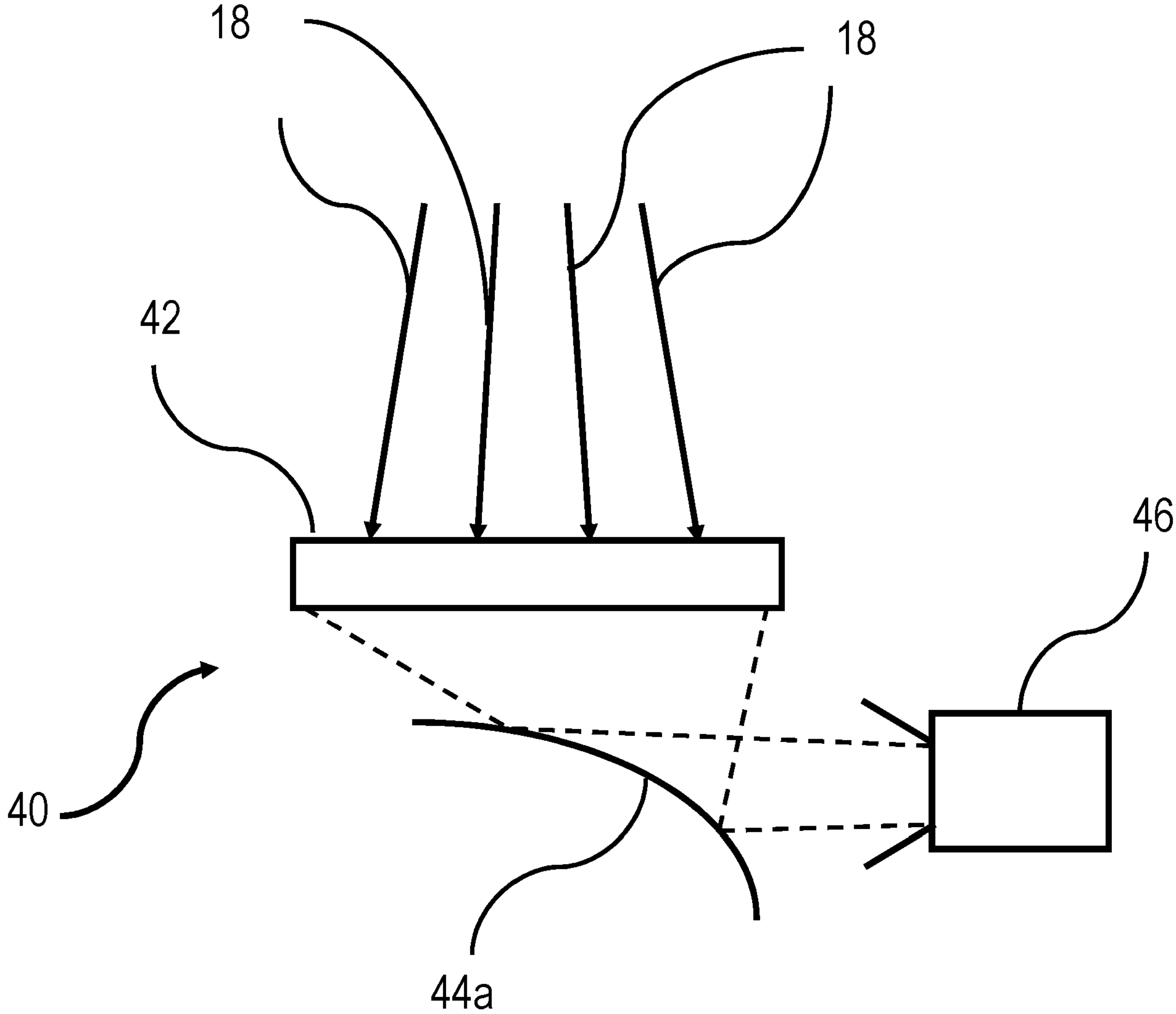


Figure 4a

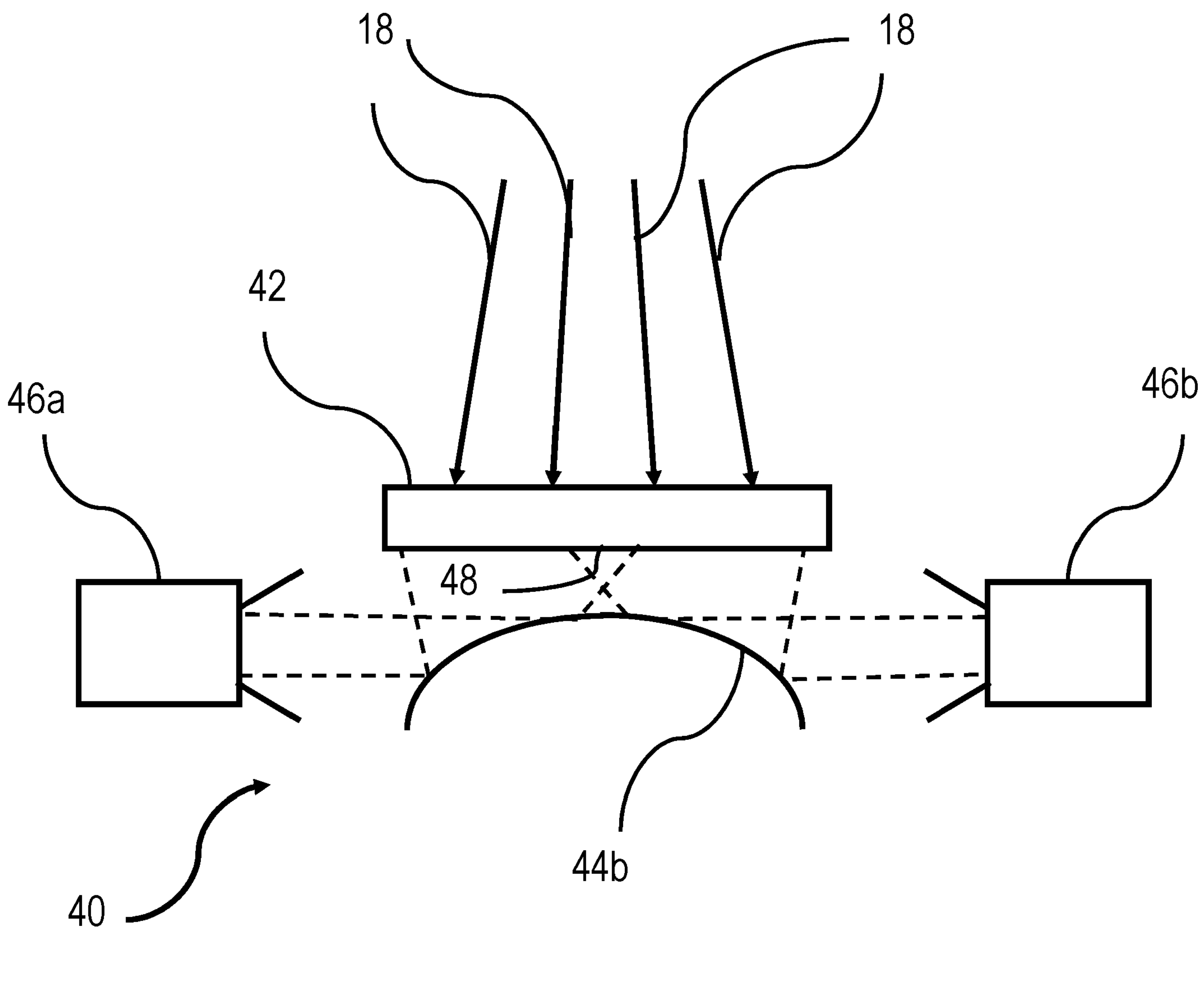


Figure 4b

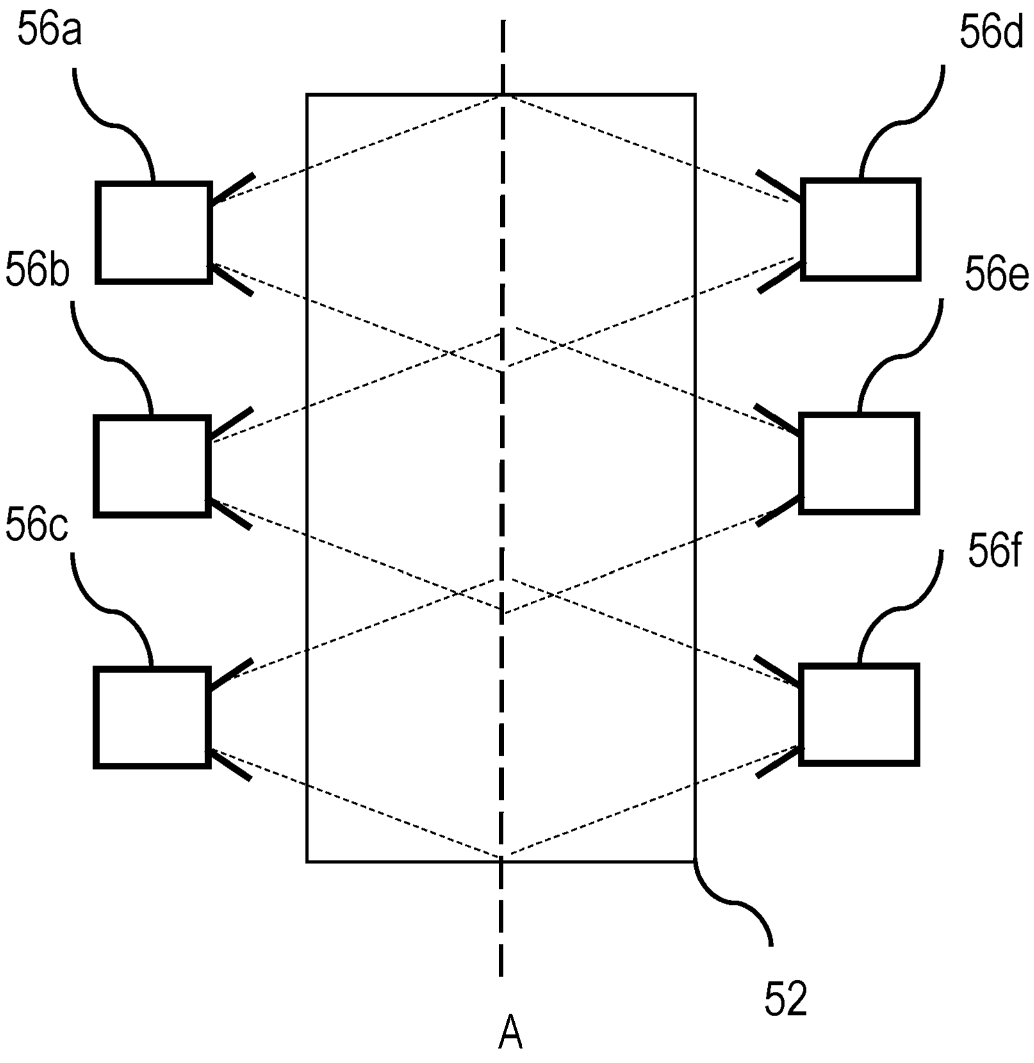


Figure 5

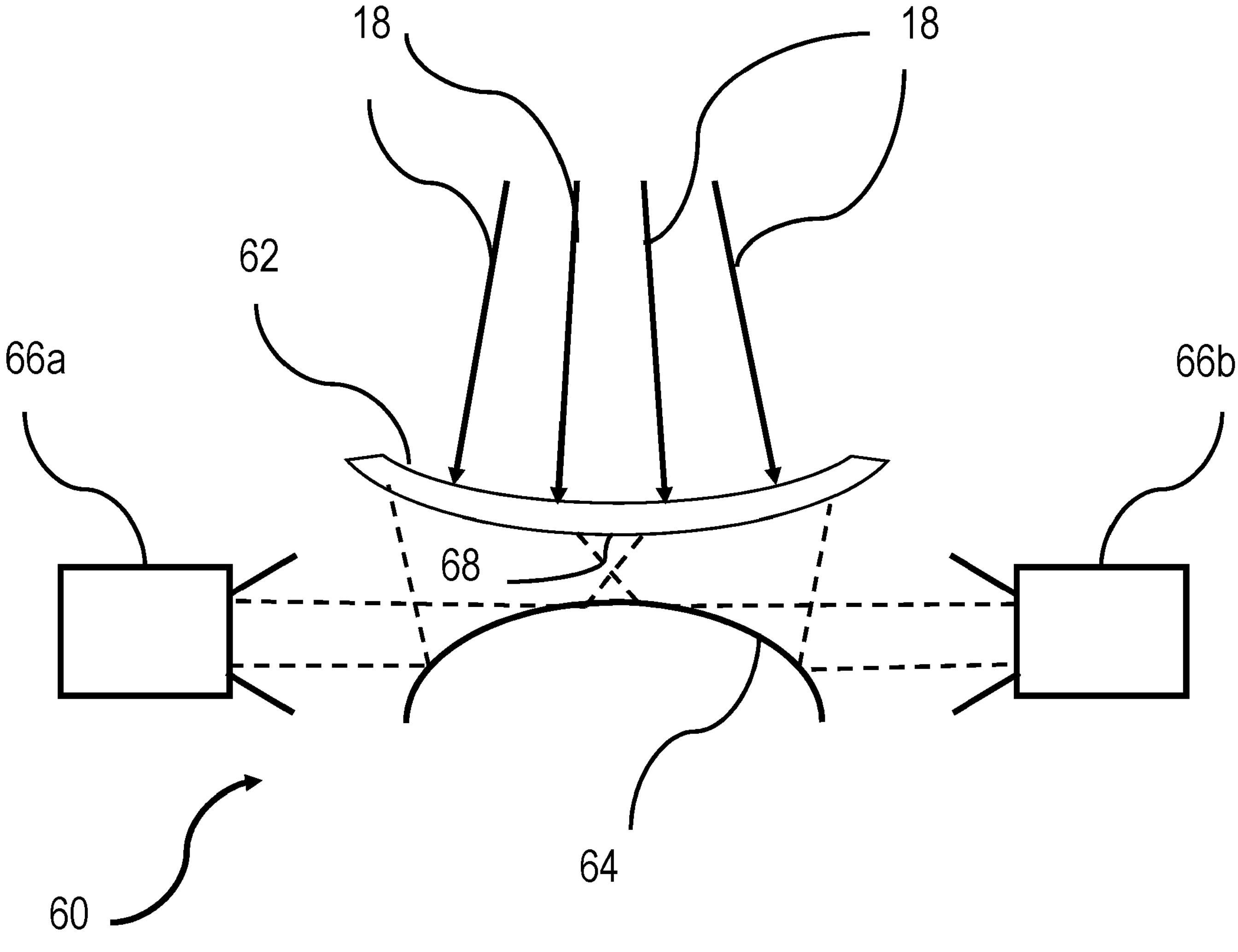


Figure 6

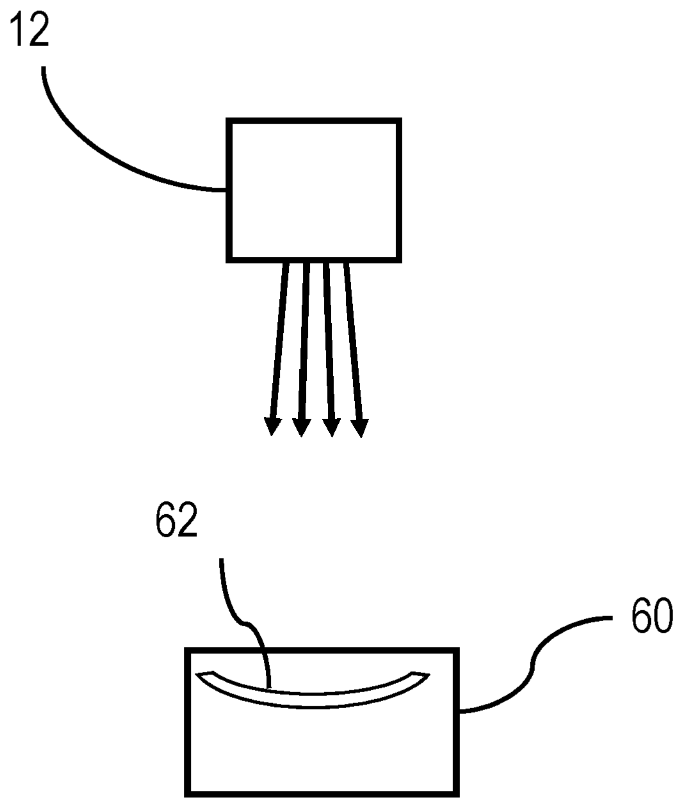


Figure 7

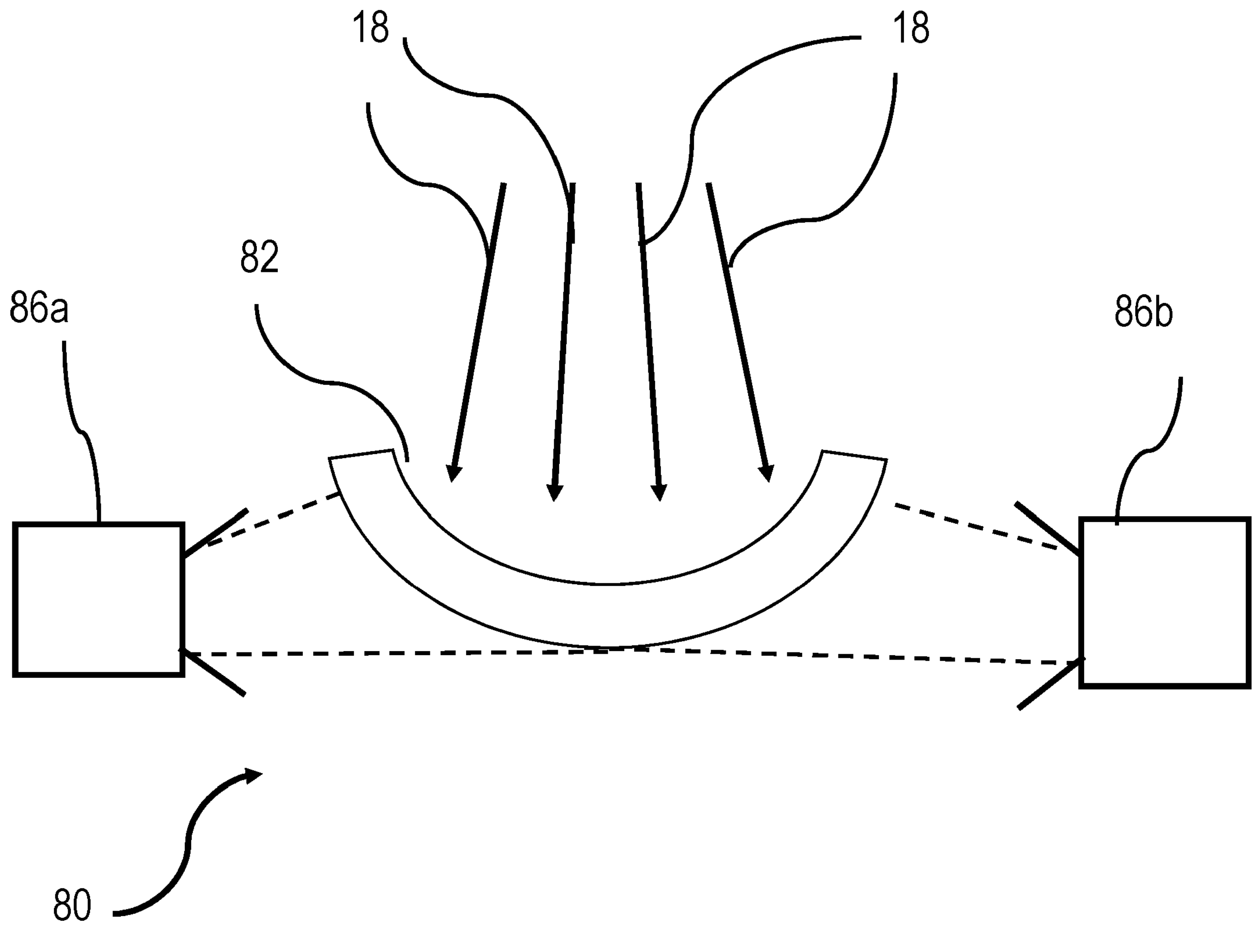


Figure 8

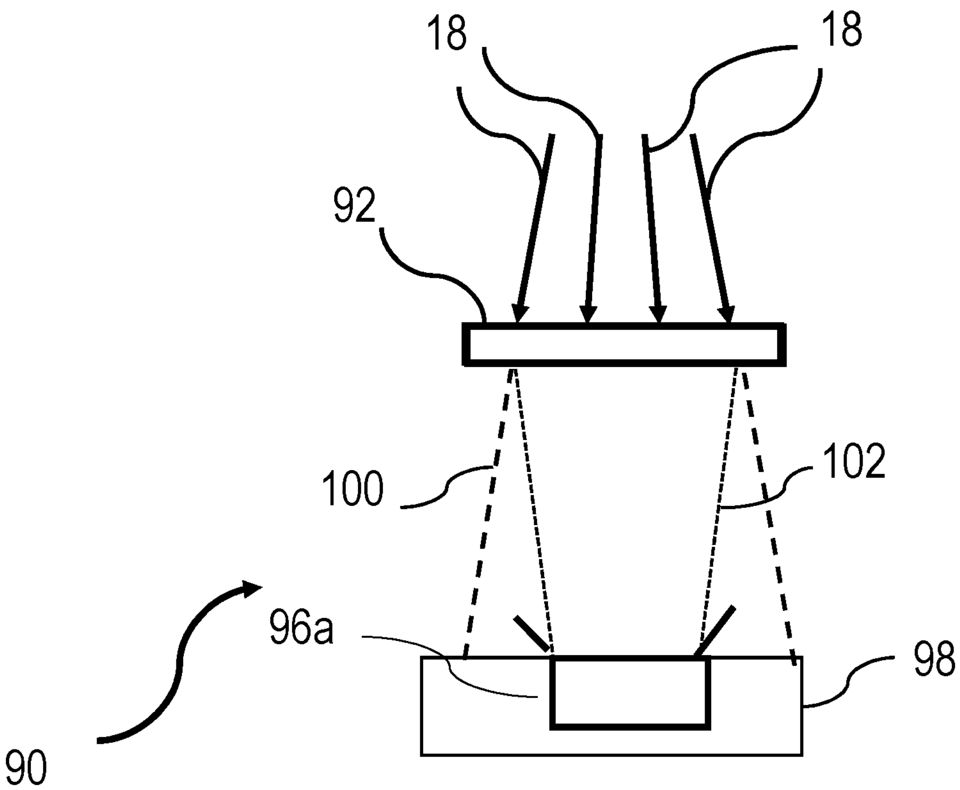


Figure 9a

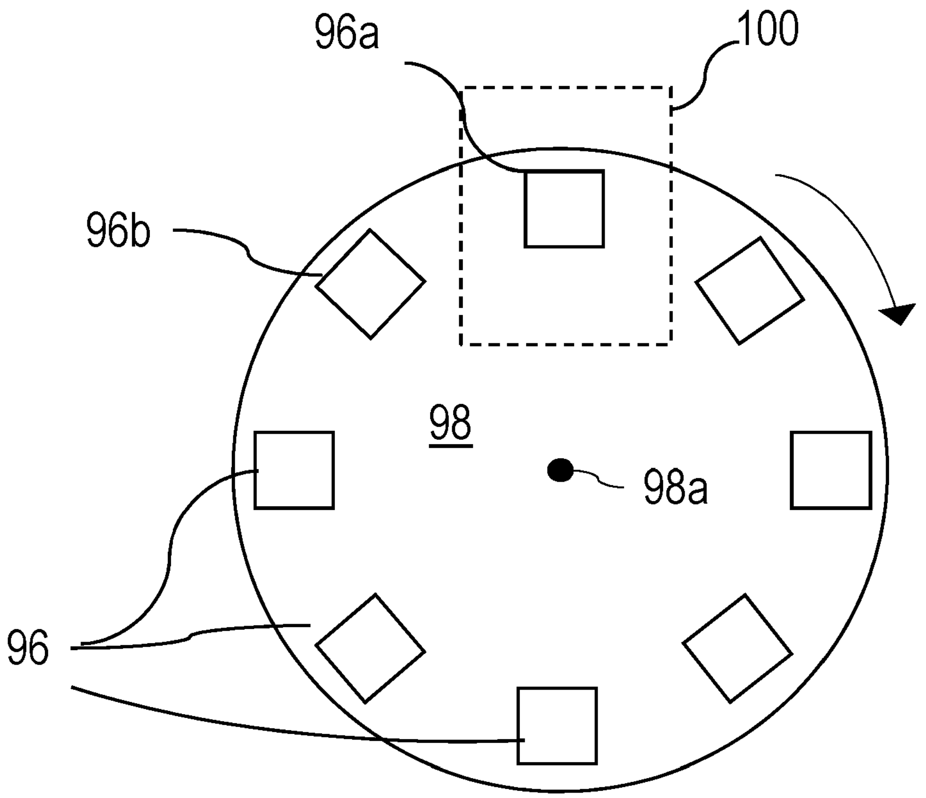


Figure 9b

Imager

FIELD

The present disclosure relates to an imager, in particular an imager suitable for portal imaging, and to a radiotherapy device comprising an imager.

BACKGROUND

X-ray imaging is widely used within the field of medicine. X-ray portal imaging in particular is often used during radiotherapy treatment of a patient.

Radiotherapy uses high-energy X-rays to destroy cancerous tissue within the body. As a means of positioning the patient, it is common to acquire X-ray images of the patient prior to treatment to ensure they have been set up in the correct position. As a means of assessing the radiation dose imparted to the patient, it is common practice to use portal imaging devices to measure the intensity of the radiation after it has passed through the patient, showing the shape of the radiation beam at any given time during the treatment.

Known imagers include: X-ray sensitive film which can provide a stationary image of the subject; a scintillator used to convert X-rays into light and a mirror to direct the light towards an optical camera to create an image representing the X-rays received at the scintillator; and Electronic Portal Imaging Devices (known as EPIDs), which use large area solid state detectors such as Charge-Coupled Detectors (CCDs) detectors.

SUMMARY

Aspects and features of the present invention are described in the accompanying claims.

BRIEF DESCRIPTION OF THE DRAWINGS

Specific embodiments are described below by way of example only and with reference to the accompanying drawings in which:

Figures 1a and 1b illustrate a known radiotherapy device;

Figure 2 illustrates a known X-ray imager;

Figure 3 illustrates an imager according to a first aspect of the present disclosure;

Figure 4a illustrates an imager according to a second aspect of the present disclosure;

Figure 4b illustrates an imager according to a third aspect of the present disclosure;

Figure 5 illustrates an imager according to a fourth aspect of the present disclosure;

Figure 6 illustrates an imager according to a fifth aspect of the present disclosure;

Figure 7 illustrates the imager of Figure 6 in a radiotherapy device;

Figure 8 illustrates an imager according to a sixth aspect of the present disclosure;

Figure 9a illustrates an imager according to a seventh aspect of the present disclosure;

Figure 9b illustrates a carousel used in the imager of Figure 9a.

OVERVIEW

The present invention relates to an imager for a radiotherapy device. According to an aspect of the present disclosure there is provided an imager as defined in claim 1.

SPECIFIC DESCRIPTION OF CERTAIN EXAMPLE EMBODIMENTS

Figure 1a shows a known radiotherapy device. Figure 1b is a cut-through of Figure 1a along line AA’. The radiotherapy device has a gantry 10 and a radiation source 12. A patient 14 lies on a patient support 16 to be treated. An imager 20 is positioned below the patient support in the gantry. X-rays 18 leave the radiation source 12 and pass through the patient 14 and the patient support 16. The X-rays 18 can be treatment X-rays to treat a tumour in a patient or can be imaging X-rays to image a patient. The X-rays will strike the imager 20, and a portion of the X-rays are imaged for dosage and/or positioning information. In some radiotherapy devices, the source 12 and imager 20 are held on arms and are rotatable around the gantry 10 to treat a tumour in the patient with radiation whilst minimising the dose of radiation applied to healthy tissue. The imager 20 rotates with the source 12 to remain on the opposite side of the patient 14 to the source. Imagers can also be used in radiotherapy devices which are not rotatable and which can only provide imaging and/or treatment radiation.

The radiation emitted by the source in the radiotherapy device described in the embodiments throughout this description emit X-rays which are used to image the patient. However, in other embodiments, the radiation used may be, for example, gamma rays, and the scintillator used is a gamma ray scintillator. Other types of radiation may additionally be envisaged.

An example of an imager known in the prior art is shown in Figure 2. The imager 20 includes a scintillator 22, a planar mirror 24 and a camera 26. The mirror is angled at 45 degrees to the plane of the scintillator. The imager can be used in the radiation device of Figures 1a and 1b. When used in a radiation device, X-rays 18 travel towards an upper face of the scintillator 22 and a portion are converted (i.e. scintillated) into visible light (not shown). The light is emitted in all directions from the scintillator. The majority of the X-rays are not converted into light and pass through the scintillator.

The camera 26 is focussed on the scintillator 22, enabling it to detect light which is emitted from a lower face of the scintillator 22. The camera creates an image of the scintillated light and by extension of the X-rays received at the scintillator. The portion of the scintillator which is imaged by the camera is the imaging area. The camera is a digital camera and providing it has a high enough refresh rate it can create moving video images of the X-rays received at the scintillator. The images can be used to determine how the position of the patient and/or the dose of radiation given to the patient changes during the course of the treatment.

X-rays have a shorter wavelength and therefore a higher energy than visible light. The X-rays can cause damage to components of the camera. To protect the camera, it is custom to position the camera outside the path of the X-rays passing through the scintillator. The mirror 24 is positioned in the path of the X-rays so that the camera 26, focussed on the mirror, can view visible light emitted from the scintillator. A standard mirror is used which will reflect light but will not reflect X-rays, which pass through the mirror. The camera 26 is placed at sufficient distance from the mirror 24 so that, with the appropriate lensing, it can view all of the imaging area of the scintillator 22 in the reflection of the mirror. With the mirror angled at 45 degrees, the camera can be placed facing along a plane parallel to the plane of the scintillator.

Figure 3 shows an imager 30 in accordance with a first aspect of the present disclosure. The imager has a scintillator 32, two planar mirrors 34a and 34b and two cameras 36a and 36b. The planar mirrors are positioned facing away from one another, back-to-back, and angled at 45 degrees relative to the plane of the scintillator 32, and are positioned below the scintillator relative to the direction of impinging X-rays 18. The mirrors are aligned substantially with the middle of the scintillator. As in the imager in Figure 2, the scintillator 32 converts a portion of X-rays 18 into light.

The first and second cameras 36a and 36b are focussed on the first and second mirrors 34a and 34b respectively. The first camera 36a is positioned to view a reflection in mirror 34a of a first portion of the scintillator. As the scintillator emits light in all directions, it will be understood that not all of the light emitted from the first portion of the scintillator will be reflected by the first mirror, only the light emitted from the first portion of the scintillator and in the direction of the first mirror will be reflected by the first mirror. The same understanding applies to reflectors throughout the disclosure.

The first camera 36a detects light reflected from the first mirror 34a and creates an image of the scintillated light from the first portion and by extension of the X-rays received at the first portion of the scintillator 32. The camera is a digital camera and sends the image as image data to a processor (not shown). The second camera 36b is positioned to view a reflection in mirror 34b of a second portion of the scintillator. The second camera detects the light reflected from the second mirror 34b and creates an image of the scintillated light from the second portion and by extension of the X-rays, received at the second portion of the scintillator 32. The camera is a digital camera and sends the image as image data to the processor (not shown). The images of the first portion from the first camera 36a and of the second portion from the second camera 36b are stitched together by the processor to create an image of the scintillated light as viewed by the two cameras.

The first and second portions are different, but there is a region 38 where the first and second portions overlap. This is useful when processing the image data from the cameras 36a and 36b. Partial overlap of the imaged sections allows for accurate reconstruction of the complete imaging area when stitching the images together. That is, images from the different cameras can be aligned correctly using the common imaged area.

The imaging area of the scintillator is divided into different fields of view which are imaged by two cameras, via the two back-to-back planar mirrors, each being angled at 45° to the scintillator. Because each camera now only has to image half of the scintillator, the depth of the mirror portion of the imager 30 is half the width of the total imaging area. Therefore, the depth of the imager is reduced by around 50% compared to the prior art imager illustrated in Figure 2. The depth might not be reduced by exactly 50%, as the depth of the scintillator, although small, contributes to the depth of the imager, and there may be a gap between the mirror(s) and the scintillator.

Figure 4a shows an imager 40 in accordance with a second aspect of the present disclosure. The imager 40 comprises a scintillator 42, a curved mirror 44a and a camera 46. As with the imager in Figure 3, the scintillator converts a portion of X-rays 18 into light. Light emitted from the scintillator 42 is reflected from the curved mirror 44a towards the camera 46. The curvature of the mirror results in infinitesimal reflective portions angled respective to one another. The tangents of different portions of the curved mirror are at different angles relative to the light received from the scintillator, and therefore the different portions reflect the light in different directions. The light is reflected towards the camera 46. The camera creates an image of the scintillated light and by extension of the X-rays received by the scintillator. The entire imaging area is imaged by the camera, but the image is distorted due to the curvature of the mirror. The camera sends the image to a processor (not shown). The distortion caused by the curved mirror and recorded by the camera can be corrected via known calibration and processing techniques.

The depth of the curved mirror is less than the width of the scintillator. The curvature of the mirror 44 allows the camera to image a greater area of the scintillator (i.e. it increases the divergence of the camera's view). The curvature of the mirror therefore reduces the depth of the imager compared to the prior art.

Figure 4b shows an imager 40 in accordance with a third aspect of the present disclosure. The imager 40 comprises a scintillator 42, a curved mirror 44b and two cameras 46a and 46b. As in the above imagers, the scintillator 42 converts a portion of the X-rays 18 into visible light.

The first and second cameras 46a and 46b are focussed on first and second sides of the curved mirror 44b, so that in the reflection of the mirror, they can image the first and second portions of the scintillator. The first camera 46a images a first portion of the scintillator via a reflection in the first side of the curved mirror 44b. The first camera detects the light and creates an image of the of the scintillated light from the first portion and by extension of the X-rays received at the first portion of the scintillator. The camera sends the image to a processor (not shown). The second camera 46b images a second portion of the scintillator via a reflection in the second side of the curved mirror 44b. The second camera detects the light and creates an image of the scintillated light from the second portion and by extension of the X-rays received at the second portion of the scintillator. The camera sends the image to the processor.

The images created by the first and second cameras will be distorted due to the curvature of the mirror. The distortion includes geometrical distortion and intensity distortion. The intensity distortion translates to a dose distortion, as the intensity of the light is proportional to the amount of radiation. The distortion caused by the curved mirror can be corrected via known calibration and processing techniques.

The images of the first portion from the first camera 46a and of the second portion from the second camera 46b are stitched together by the processor to create an image of the scintillator. The images can be stitched together after the correction of the distortion in the images, or before the correction of the distortion in the images.

There is a region 48 where the imaging areas of the first and second cameras 46a and 46b overlap. This is useful when processing the image data from the cameras, as the overlap allows for accurate reconstruction of the complete imaging area when stitching the images together. That is, images from the different cameras can be aligned correctly using the common imaged area.

The cameras 46a and 46b are positioned either side of the scintillator 42 and mirror 44b. This alters the layout of the apparatus compared to an imager with a single detector.

As in the imager Figure 3, in the imager 40 the imaging area of the scintillator is divided into different fields of view which are imaged by different cameras. As in the imager of Figure 4a, the depth of the curved mirror 44b is less than the width of scintillator. The curvature of the mirror causes the imaging area of each camera to be expanded (i.e. it increases the angular range over which the camera can capture an image), and the depth of the mirror portion of the imager is less than the width of the scintillator. Therefore the depth of the imager is reduced compared to the prior art.

The above aspects in Figures 3, 4a and 4b reduce the depth of the imager by providing a mirror with non-planar portions, which increases the respective fields of view of the cameras. The depth of the mirror portion of the imager in each aspect is less than the width of the imaging area.

It will be appreciated that various alterations can be made to the above aspects. For example, features can be combined from each of the above aspects.

A combination of planar and curved mirrors could be used in any suitable arrangement. A single mirror with angled planes could replace the two separate mirrors in Figure 2. Mirrors facing away from each other can be used. In an aspect, planar mirrors are angled at angles other than 45 degrees to the imaging area. In the above aspects, the cameras are all angled to detect light travelling parallel to the plane of the imaging area. However, cameras could be angled to detect light at other angles to the imaging area.

One advantageous arrangement may be to reduce the angle of the mirror relative to the horizontal, for example to 30º. The camera can then be angled such that top of the camera is in roughly the same plane as the top of the scintillator, being directed towards the mirror. The camera and mirror angle result in the camera imaging an area of the scintillator reflected from the mirror.

It will be appreciated that the imager is not limited to a certain angle of the mirrors relative to the horizontal. Any number of mirror angles and camera arrangements can be utilized.

Non-planar reflective portions include reflective portions which do not form a single plane. A reflector arrangement may also be referred to as a reflector.

In an aspect, instead of curved mirrors a plurality of planar portions approximating a curve are included in the imager.

The scintillators in any of the aspects could be curved or planar. In an aspect, a non-planar scintillator with a non-planar imaging area is used, for example a scintillator having planar portions angled relative to one another and approximating a curve is used.

In a further aspect an imager has n mirrors and n corresponding detectors, each detector positioned to image a respective portion of the imaging area via a mirror. The mirrors could be curved mirrors, or planar mirrors with the plane of the mirrors angled relative to the plane of at least one other mirror.

Such an example is shown in Figure 5, which illustrates a scintillator 52 and a plurality of cameras 56a-56f from above. In use radiation will travel into the page. The scintillator has a major axis A. Below the scintillator is a reflector arrangement aligned along the major axis. The arrangement could be an extension of the arrangement shown in Figures 3, 4a or 4b, with a series of non-planar mirrors and detectors aligned in a direction coming out of the page of these figures. As in Figures 3, 4a and 4b, cameras are positioned beneath the scintillator, spaced away from directly underneath the scintillator and therefore out of the path of the radiation. Each camera is angled towards the arrangement to view a reflection in the arrangement of a portion of the scintillator. The dashed lines show the field of view of each of the cameras. With the cameras spaced along the direction of the major axis of the scintillator, the area of imaging can be increased relative to the arrangements shown in Figures 3, 4a and 4b without requiring a large solid state imager or depth to accommodate a wide mirror. The imaging area can be divided up and standard cameras can be used. Therefore, a large imaging area can be imaged without requiring a large camera or a large mirror. This arrangement could be scaled up to any size of imaging area, and could be used with a large curved scintillator.

In Figure 5 cameras are positioned on both sides of the scintillator. However, it will be appreciated that in other arrangements, for example using the reflector arrangement of Figure 4a, cameras could all be positioned only on one side of the scintillator.

Figure 6 illustrates an imager 60 according to a fourth aspect of the present disclosure. The imager 60 comprises a curved scintillator 62, a curved mirror 64 and two cameras 66a and 66b. The scintillator 62 converts a portion of X-rays 18 to light as in the other imagers. The field of view of the first and second cameras 66a and 66b is reflected from the curved mirror 64 to image light emitted from first and second portions of the scintillator respectively. The first and second cameras create images of the scintillated light from the first and second portions respectively, which can be combined to create and image of the entire imaging area as in the imager in Figure 5. As in the imager of Figure 5, the images are distorted and are corrected using calibration techniques. There is overlap 68 of the first and second portions of the scintillator.

The imager 60 is shown in a radiotherapy device in Figure 7, with like reference numerals being used to denote like features. The camera and mirrors of the imager 60 are not shown in Figure 7. The radiotherapy device has a radiation source 12. The loci of the curvature of the scintillator 62 is positioned close to or on the source of radiation so that the distance between the source 12 and the scintillator 62 remains constant even over a wide-angle range. Because the intensity of the X-ray radiation drops off as $1/x^2$, the signal from a region of the scintillator further from the source will appear less intense than the signal from a region of the scintillator closer to the source. Changes in intensity reflect the density of the material the X-rays have passed through within the patient, or the distance between the radiation source and the imager, so it is important to understand where the variations in the signal intensity come from. By curving the scintillator around a focal point at the radiation source, all regions of the scintillator are at a common distance from the source, thus removing the $1/x^2$ intensity variation, making it easier to calibrate the images acquired from the scintillator, and therefore reducing the calibration required to obtain a clear image from the apparatus.

In Figures 6 and 7 the curvature of the scintillator runs across the page. However, the scintillator could instead be curved in the tangential plane (i.e. perpendicular to the plane of the image), or in both axes so as to subtend some portion of a curved cylinder.

As will be appreciated from the below description, a major advantage of providing a curved scintillator is to allow a larger area to be imaged without requiring the expense of a large solid-state imager, and is to the overall size of the mirror and camera arrangement is smaller compared to known imagers. Accordingly, a radiotherapy device using a curved scintillator can have a large imaging area whist maintaining a relatively small space around or behind the scintillator. This saves space in radiotherapy devices, where previously a large volume would be required to accommodate the imager of a large imaging area.

Figure 8 shows an imager 80 in accordance with a fifth aspect of the present disclosure. The imager 80 comprises a curved scintillator 82 and two cameras 86a and 86b. The scintillator 82 converts a portion of X-rays 18 to light as in the previous imagers, which is emitted in all directions. Cameras 86a and 86b are positioned to the side of the scintillator 82 such that the cameras are out of the path of the X-rays travelling through the scintillator. The cameras are directed at the scintillator to detect light from the scintillator. The curvature of the scintillator means that every portion of the underside of the scintillator can be viewed by at least one of the cameras. Camera 86a images a first portion of the imaging area and camera 86b images a second portion of the imaging area. The images from each of the cameras are stitched together to form a complete image of the imaging area. As with the previous aspects, there may be overlap in the portions.

The cameras 86a and 86b are positioned outside the path of the X-rays and therefore are not exposed to the damaging X-ray radiation. This increases the life span of the cameras and the imager. There is no mirror required, reducing the size of the imager. The imager is therefore easier and cheaper to manufacture.

Figure 9a shows an imager according a sixth aspect of the present disclosure. The imager has a scintillator 92, a plurality of cameras including camera 96a and a carousel 98. The carousel 98 is shown from above in Figure 9b.

The scintillator 92 converts a portion of X-rays 18 to light. Camera 96a is positioned below the scintillator 92 to directly image light emitted from the scintillator. The field of view of the light detected and imaged by the camera is shown by dotted lines 102. Light which travels from the scintillator to the camera 96a creates an image of the scintillated light and by extension of the X-rays received by the scintillator. The path of the X-rays 18 which pass through the scintillator and are not converted to light is shown by dashed lines 100. The camera 96a is positioned in an imaging position in the X-ray path 100.

The carousel 98 houses the plurality of cameras 96. The carousel is rotatable about an axis 98a. Camera 96a is posited in the imaging position. The imaging position is in the X-ray path 100. In this position the camera 96a can image light emitted from the scintillator 92. The remaining cameras 96 that are not in the imaging position are positioned outside of the X-ray path 100. The camera in imaging position will experience damage from the X-rays, and the cameras outside the X-ray path are not exposed to the damaging X-ray radiation. A radiation shield (not shown) can be placed above the cameras not in the X-ray path 100 to help shield them from scattered radiation.

The camera 96a images the scintillator 92 from which light is emitted. The camera 96a will deteriorate through exposure to the X-rays passing through the scintillator. When the camera has deteriorated beyond use or to a low quality, the carousel 98 can be rotated about axis 98a as shown by the arrow in Figure 9b and can move the camera 96a out of the imaging position and move the adjacent camera 96b into the imaging position. Camera 96b will not have previously been positioned in the X-ray path 100 and therefore will not have experienced the deterioration of camera 96a and will be useable to image light emitted from the scintillator 92. Camera 96b can then be used to image the scintillator 92 until it too deteriorates due to exposure to the X-rays. This may take happen over the course of a number of different patient treatments.

After camera 98b has deteriorated beyond use, the carousel is rotated again to position the next camera in the imaging position. This process can be repeated until all of the cameras 96 have been used for imaging. After this time, all of the cameras will have been exposed to the X-rays and therefore will be damaged, and the carousel is removed from the imager. The cameras 96 are removed from the carousel, a new set of cameras are housed in the carousel and the carousel is re-inserted into the imager. Alternatively, a new carousel housing new cameras is inserted into the imager.

Cameras are increasingly low in cost and are easily replaceable. No reflector is needed in the imager of Figures 9a and 9b, and the camera can be positioned directly beneath the scintillator such that the imager can be considerably more compact. Further, the carousel and/or the detector can be replaceable, resulting in an increased lifespan of the imager as a whole. The carousel provides a convenient and quick mechanism to move cameras into the imaging area.

A carousel can take the form of a belt comprising a plurality of camera, or a magazine comprising a plurality of cameras, or a rotatable wheel comprising a plurality of cameras. Other arrangements may be envisaged.

Any means for moving detectors into an imaging position one by one to directly image scintillated light from a scintillator can be used.

The imagers of any of the aspects illustrated in Figures 3 to 6, 8 and 9 can be used in a radiotherapy device. The imagers can be used in the position of the imager 20 illustrated in Figures 1a and 1b. There is also provided a radiotherapy device comprising an imager according to any of the above aspects. The radiotherapy device can treat or image a patient with radiotherapy from a source and the imager can image the patient using the imager as described above.

The above imagers use digital cameras, although other detectors could also be used. Digital cameras provide many advantages over other detectors including instant imaging, high quality of imaging and continuous imaging (i.e. video). The above aspects include mirrors, although any reflectors can be used. Cameras in the above embodiments could be replaced with any conceivable detectors.

The scintillators of the above imagers convert X-rays to light. However, in other aspects the scintillator can convert photons of a first wavelength of photos of a second, longer wavelength. Converting to light is beneficial, as mirrors and detectors (such as cameras) for optical photons are cheap and widely available. However, other wavelengths such as infrared (IR) or ultraviolet (UV) may also be suitable for such a purpose, and have the advantage that the container for the detector apparatus need not be so light tight, as stray optical light from the room will not affect IR or UV detectors, given appropriate filtering on the cameras.

Directly imaging the scintillator refers to imaging converted photos received at the camera without being reflected from a mirror.

Non-planar is used throughout to mean not in a flat plane. This could include a surface with two or more flat portions angled relative to one another, or a curved surface.

Thus there is provided an imager for a radiotherapy device, and a radiotherapy device comprising an imager.

Features of the above aspects can be combined in any suitable manner. It will be understood that the above description is of specific embodiments by way of aspect only and that many modifications and alterations will be within the skilled person's reach and are intended to be covered by the scope of the appendant claims.

Claims

1. An imager comprising:
a scintillator configured to convert photons of a first wavelength to photons of a second, longer wavelength; and
a plurality of detectors moveable to position each detector in turn in an imaging position to directly image the scintillator.

2. An imager according to claim 1, wherein the scintillator is configured to convert X-rays to photons of a longer wavelength.

3. An imager according to claim 1, wherein the scintillator is configured to convert X-rays to visible light.

4. A radiotherapy device comprising an imager according to any of claims 1 to 3.

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