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Development of a Micro-Size Dosimeter Using an Optical Fiber Probe Based on Photostimulable Phosphorescence

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Abstract

We have developed a novel micro-size dosimeter using an optical fiber probe coupled with a photostimulable storage phosphor, such as $BaFBr:Eu^{2+}$. Our micro-size dosimeter is based on the concept in which the radiation dose response is temporally compressed by using integrating-type dosimeter elements, such as a photostimulable storage phosphor, and the signal-to-noise ratio is improved to downsize the dosimeter probe. We fabricate the prototype dosimeter probe which has the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. The lower limit of detectable dose, which is defined as three times the noise level, is evaluated to be 0.9 mGy. This value is much smaller than the typical irradiation dose (a few Gy per treatment) in the radiotherapy. We, therefore, conclude that the proposed micro-size dosimeter can be a promising candidate of in-vivo and on-line dose monitor in the radiotherapy.

KEYWORDS: dosimeter, photostimulable phosphorescence, optical fiber probe, radiotherapy

1. Introduction

Radiotherapy utilizes a difference in the sensitivity between cancer cells and healthy tissue. Because cancer cells are generally more sensitive to radiation damage than healthy tissue, irradiation can preferentially cause damage to cancer cells. Since the difference in their sensitivities is generally quite slight, the irradiation should be spatially concentrated into a tumor in practical radiotherapy. Recently, advanced irradiation techniques, such as the intensity modulated radiation therapy (IMRT) and the respiration-synchronous irradiation, have been developed (Starkschall, 2007). At present, dose control and evaluation in radiation therapy are performed based on computer simulations. However, since the effect of radiotherapy has dose response with S-shaped curve and strongly depends on irradiation dose near the threshold dose, especially in these complex irradiation techniques, more accurate, in-situ and on-line dose monitoring techniques are required.

In order to accurately estimate the irradiation dose into affected regions, dose monitoring probes are required to be inserted into the affected region in a patient body. However, it is difficult to insert dosimeters into a patient body because general dosimeters have the size of a few millimeters at least and patients must have physical pain to insert dosimeters into their body. Therefore, in-vivo dose monitoring probes are desired to be miniaturized as small as possible. Recently, some small dosimeters, such as a MOS-FET dosimeter, were attempted for in-vivo dose monitoring (Kohno et al., 2008). When downsizing the detector sensitive volume, the signal intensity decreases compared with the noise causing in the signal line. When using small dosimeter probe of the successive signal readout type, such as a scintillator, we suffer from low signal to noise ratio in either the pulse mode or the current mode. Some researchers developed the optical fiber type dosimeters using a small scintillator, as a detection element (Mori et al., 1999, Lee te al., 2007 and Ishikawa et al., 2004). However, they suffer from noise signal produced in the optical fiber. Cherenkov radiation produced in the optical fiber is one of the noise sources. In the case of extremely small dosimeter probe, the noise produced in the optical fiber can submerge the true signal from the dosimeter probe. On the other hand, the integrating-type dosimeter probe can obtain larger signals than a noise level even if the dosimeter probe size is smaller than the charged particle range, because small energy depositions can temporally be accumulated in the detector element until exceeding a noise level.

We, therefore, have developed a novel micro-size dosimeter using an optical fiber probe based on photostimulable phosphorescence from a photostimulable storage phosphor (PSP) (Nakahashi et al., 2011). The PSP, such as $BaFBr:Eu^{2+}$, is one of the integrating-type dosimeter elements and is usually used as a detecting element of the imaging plate (Iwabuchi et al., 1994). In this paper, we fabricate the prototype dosimeter probe which has the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. We evaluate the basic performances, such as the lower limit of detectable dose, for the fabricated small size dosimeter through basic experiments and consider the feasibility of our dosimeter for in-situ and on-line dosimeter in radiation therapy.

2. Fabricated Dosimeter System

We fabricated the prototype dosimeter probe which has the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. **Figure 1** shows the fabricated dosimeter probe. The fabricated dosimeter probe consists of an optical fiber mounting the PSP, which is $BaFBr:Eu^{2+}$, at the fiber end. When ionizing radiation interacts with the PSP, many electrons are excited. Excited electrons are partially trapped and create color centers, which consist of anion-site vacancies and trapped elections. Trapped electrons are stable and can be accumulated unless the release process is performed. In the release process, the PSP are irradiated by stimulus laser light and then the radiation energy accumulated in the form of color centers is released as luminescence photons with the specific wavelength, which is shorter than that of a stimulus laser photon. In the case of $BaFBr:Eu^{2+}$, the wavelength of luminescence photons is about 400 nm corresponding to deexcitation of Eu^{2+} ion. This dose monitoring system, therefore, needs the laser light source to read out the accumulated dose information. We constructed the optical fiber based system with the core diameter of 400 μm using an optical fiber coupler, which divides stimulus laser light and photostimulable luminescence photons. **Figure 2** shows the micro-size dosimeter system. A laser diode

(BWT Beijing, K63S09F-0.40W) with an optical fiber output was used as the stimulus laser light source. The wavelength of the laser diode was 630 nm. The optical fiber system was constructed with the fibers with the same numerical aperture for effective photon transportation. The photomultiplier tube (PMT, Hamamatsu H6612) was used.as a photostimulable luminescence photon detector. The bandpass filter was mounted in front of the photocathode of the PMT to avoid the reflection light of the laser. The transmission wavelength of the band-pass filter was 400$\pm$10 nm, which match the wavelength of the photostimulable luminescence. The laser diode was operated in the pulse mode with the duration of 50 ms by modulating the operating current. The pulse duration and interval were controlled by the PC. Typical pulse interval was a few seconds. If the radiation is irradiated during the readout phase, the signal response becomes complicate. The duty ratio of readout laser pulse, however, is quite small so that we can neglect the influence of irradiation during the readout phase. The photostimulable luminescence signals were recorded through the digitizer into the PC. The trigger of the digitizer was synchronized with the stimulus laser pulse. The recorded signal was analyzed in the control PC.

3. Results and discussion

3.1 Fading Effect

The PSP suffer from the fading effect, in which the trapped electrons decrease temporally through the thermal excitation. Since the accumulated information in the PSP is read out with a constant interval in our system, our system has smaller influence of the fading effect than normal PSP applications. The fading effect, however, depends on temperature (Meadowcroft et al., 2008). We, therefore, evaluate the temperature dependence of the fading effect.

Figure 3 schematically shows the measurement cycle of experiments for the fading effect. The temporal decrease of the PSL intensity was measured by changing the waiting time. The temperature of the dosimeter probe was controlled at 30, 35, 40 and 45 degrees Celsius under X-ray irradiation. The X-ray tube was operated with the tube voltage of 65 keV. **Figure 4** shows the temperature dependence of the fading effect. From these results, the fading effect is almost independent of the temperature ranging from 30 to 45 degrees Celsius in our PSL read out system. In our dosimeter system, the temperature dependence of the fading effect lead little influence in the measured PSL intensity.

3.2 Angular Response

The X-rays are irradiated from all directions in the complex irradiation techniques, such as IMRT. The angular response of the dosimeter probe is one of the important characteristics. Co-60 gamma rays were irradiated with 500 mm distance between the source and the dosimeter probe. The irradiation angle is defined as the angle between the gamma-ray source direction and the normal direction of the end surface of the optical fiber, which is the same as the PSP surface. We measured the PSL intensity with changing the irradiation angle at the same tip position of the optical fiber probe. **Figure 5** shows the angular response of the dosimeter probe. The difference of the angular response is within 5%, although the response slightly changes with the irradiation angle.

3.3 Lower limit of detectable dose

As the fundamental performance of the fabricated dosimeter, we evaluate the lower limit of detectable dose. In order to evaluate the detectable lower limit dose, Co-60 gamma-ray irradiation experiments were performed Co-60 gamma rays were irradiated at the Co-60 irradiation facility (70 TBq) at Nagoya University. The dosimeter probe was set at 9 cm depth from the surface of the epoxy resin water-substitute phantom with 30x30x30 cm^3 size. The distance between the source and the phantom surface was variable to change the dose rate at the dosimeter probe position. The PSL intensity was measured with changing the dose, which was adjusted by changing the irradiation time and the dose rate or distance from the source. **Figure 6** shows the relationship between the PSL intensity and the dose. The good linearity is confirmed. The uncertainty of the PSL level was evaluated as the standard deviation of multiple measurements under the same condition. The error bar is almost within the marker size in this figure. The lower limit of detectable dose is defined as three times the noise level. The lower limit is evaluated to be 0.9 mGy by comparison with the noise level for the micro-size dosimeter probe with the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. Our minimum detectable dose is inferior to that of the conventional PSL dosimeter. The deterioration of the lower limit is caused by limited light collection efficiency in the optical fiber system. This value, however, is much smaller than the typical irradiation dose (a few Gy per treatment) in the radiotherapy. We can measure the dose in a few hundred subdivisions during a treatment. We, therefore, conclude that the proposed micro-size dosimeter can be a promising candidate of in-vivo and on-line dose monitor in the radiotherapy.

4. Conclusions

We have developed a novel micro-size dosimeter using an optical fiber probe coupled with a photostimulable storage phosphor, such as $BaFBr:Eu^{2+}$. The lower limit of detectable dose for the fabricated dosimeter was evaluated to be 0.9 mGy for the micro-size dosimeter probe with the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. We evaluated the fundamental performances, such as the fading effect and the angular response. We confirmed that the fading effect had little temperature dependence ranging from 30 to 45 degrees Celsius and the deviation in the angular response was 5%. We, therefore, conclude that the proposed micro-size dosimeter can be a promising candidate of in-vivo and on-line dose monitor in the radiotherapy.

In the future work, we will check the X-ray energy dependence of the dosimeter response and the response for other types of ionizing radiation, such as high energy protons.

Acknowledgements

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Figure Captions

Figure 1 The fabricated dosimeter probe which has the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm.

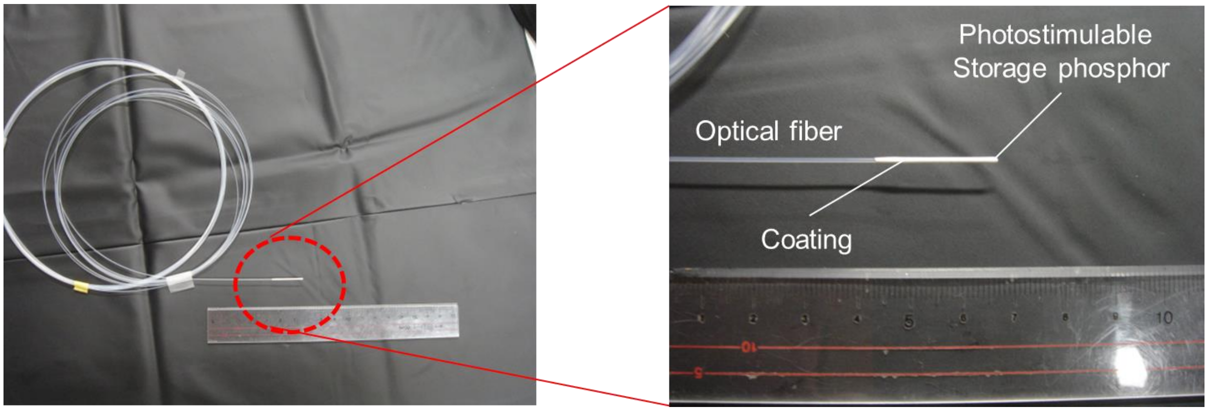
Figure 2 The micro-size dosimeter system. This system consists of a laser diode, a dosimeter probe, an optical fiber based light transmission system, a photo-detector and a data acquisition system.

Figure 3 The measurement cycle of experiments for the fading effect. The signal read-out time width was set to be 500 ms.

Figure 4 The temperature dependence of the fading effect. The temperature of the dosimeter probe was controlled at 30, 35, 40 and 45 degrees Celsius under X-ray irradiation.

Figure 5 The angular response of the dosimeter probe.

Figure 6 The relationship between the dose and the PSL intensity for the micro-size dosimeter probe with the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. The error bar is almost within the marker size in this figure. The lower limit is evaluated to be 0.9 mGy by comparison with the noise level.



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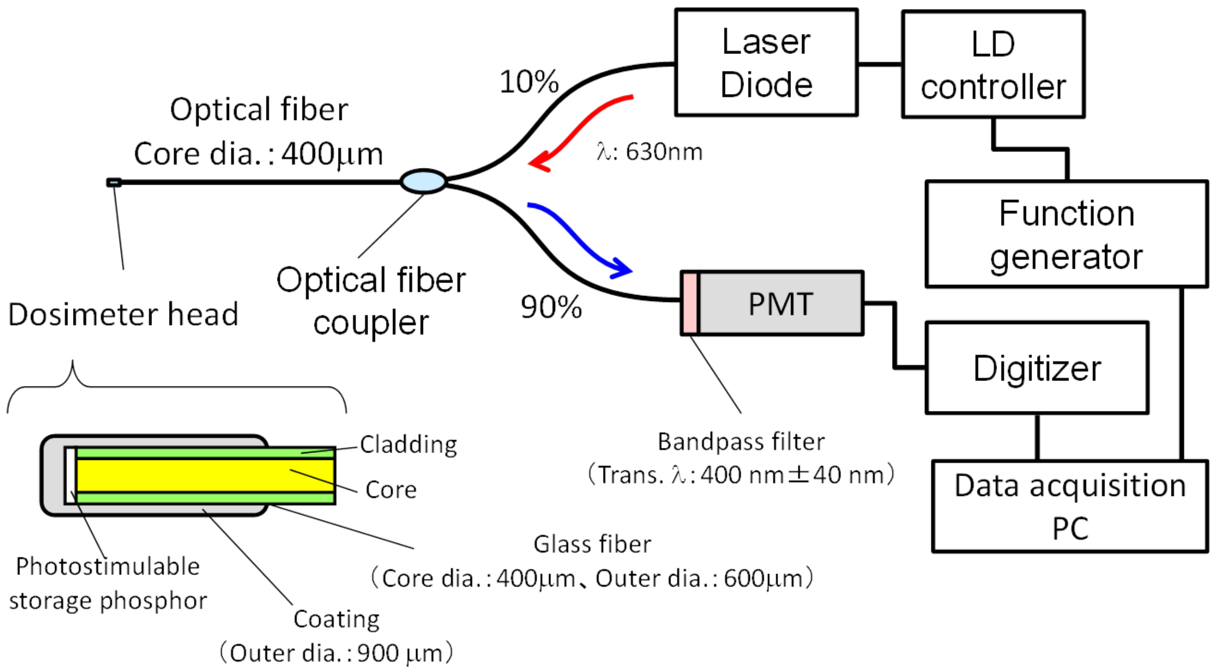

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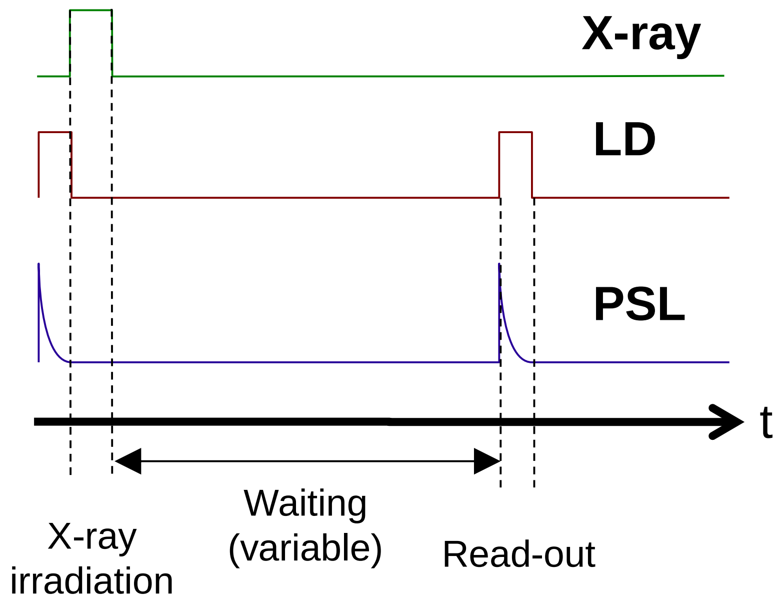

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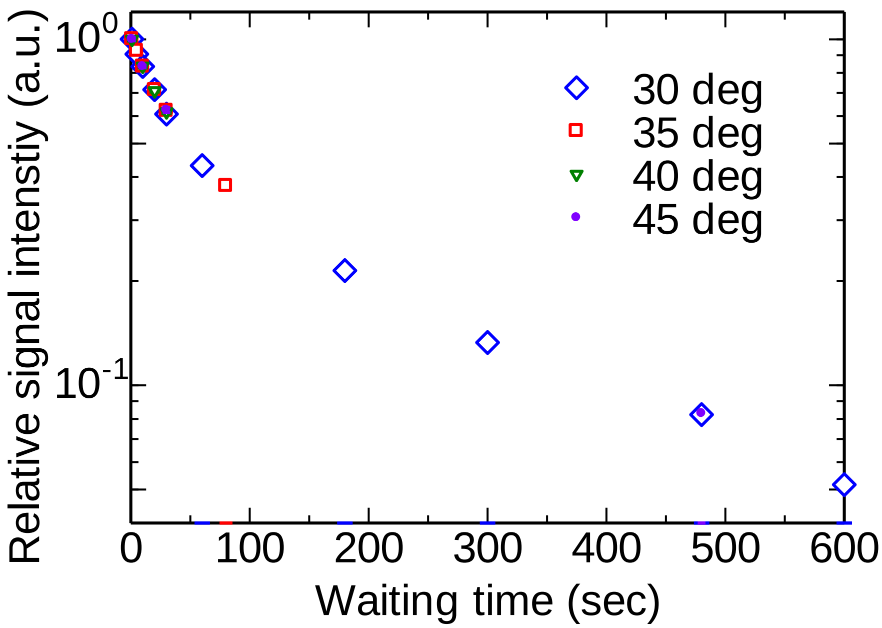

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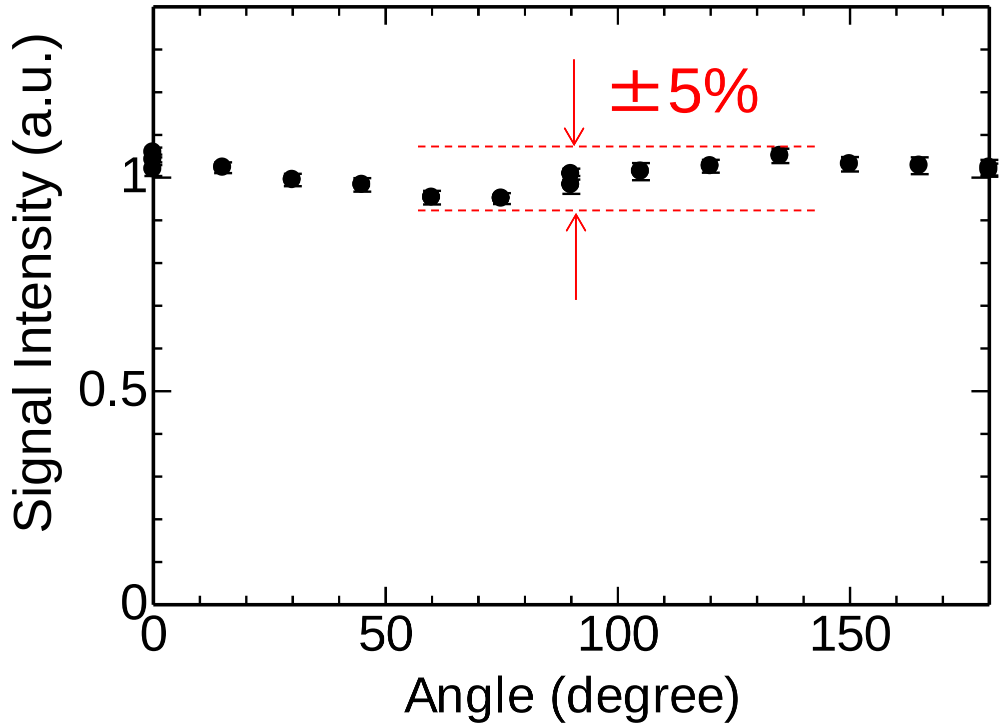

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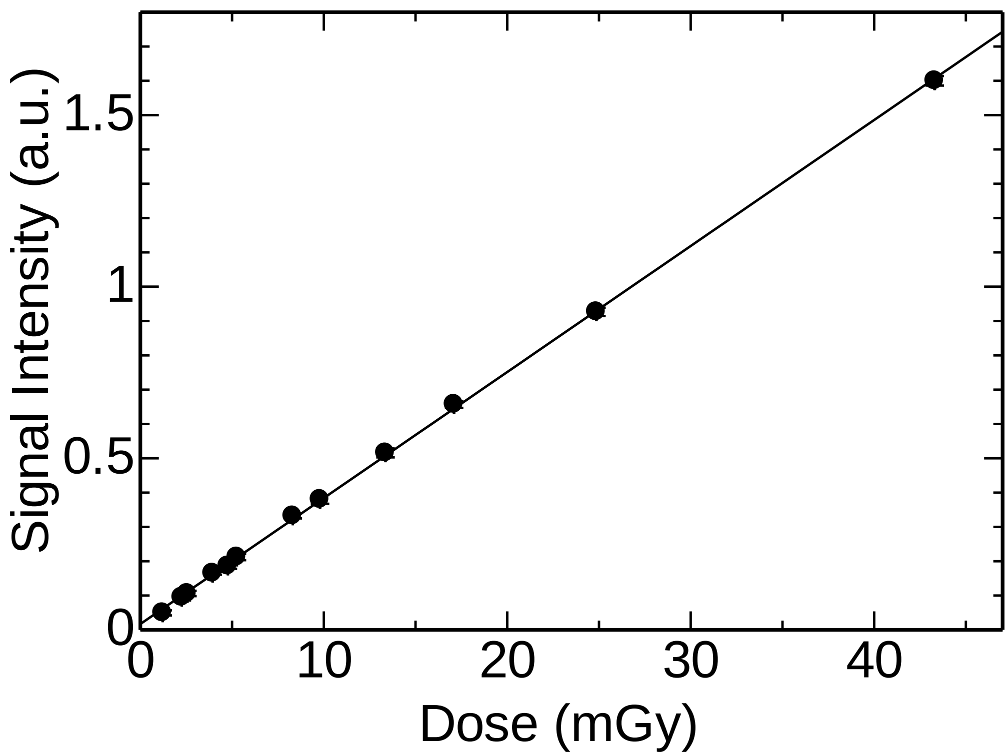

Figure 6 The relationship between the dose and the PSL intensity for the micro-size dosimeter probe with the detectable volume of 100 μm thick and 400 μm diameter and the outer diameter of 900 μm. The error bar is almost within the marker size in this figure. The lower limit is evaluated to be 0.9 mGy by comparison with the noise level.