Abstract—In this study, we have fabricated a scintillating fiber-optic dosimeter (SFOD) for a Co-60 radiotherapy dosimetry. The γ-rays generated from a Co-60 machine are measured using a SFOD and percent depth dose (PDD) curves are obtained according to the different depths of water phantom. Also, Cerenkov lights generated by the interactions of primary or secondary electrons and a plastic optical fiber are measured at different depths of a water phantom using a background optical fiber.

I. INTRODUCTION

A successful radiotherapy treatment depends on the accuracy of radiation delivery to the target volume and it requires a high-precision dosimeter. Generally, for high spatial resolution in dose measurements, a dosimeter should be small enough and for accurate dose measurements without complex calibration processes, its sensing part has tissue-equivalent characteristics [1].

For accurate dosimetry, many kinds of dosimeters such as ion chambers, silicon-diode detectors, diamond detectors, liquid ion chambers and radiographic films have been developed and used. However, they have some disadvantages for measuring doses in radiotherapy dosimetry [2] such as 1) a large sensitive volume, 2) dose rate dependence, 3) complicated construction, and 4) not real-time measurements.

To overcome these problems, a scintillating fiber-optic dosimeter (SFOD) using an organic scintillator has been developed and successfully applied to measure photon and electron beams in radiotherapy dosimetry. There are many advantages of a SFOD over conventional dosimeters used in radiotherapy. A SFOD has a small sensitive volume to get high resolution measurements, and has water or tissue-equivalent characteristics to avoid complex conversions arising from material differences. In addition, it emits visible light proportional to the absorbed dose and dose rate of the photons, electrons or gamma rays [3,4].

However, SFOD has a disadvantage namely unwanted light signals, which are generated by the direct action of the radiation on the organic scintillators or optical fibers themselves with charged particles. This induced light is known as a Cerenkov light, and is produced by a charged particle that passes through a medium with a velocity greater than that of light [5]. Thus, it is always generated as a noise with a scintillating light and interferes with measuring real signals using a SFOD in radiotherapy dosimetry [6]. In case of γ-rays, the Cerenkov light is generated by the interactions of primary or secondly electrons. It was reported that several studies were carried out to minimize or remove Cerenkov light in a SFOD a few methods such as a simple subtraction and optical discrimination methods using a dummy fiber and optical filters, respectively [6,7].

In this study, we have fabricated a SFOD by using of an organic scintillator and a plastic optical fiber (POF) for Co-60
therapies as a light pipe, POFS are used to transmit a scintillating light generated in the organic scintillators; also, photomultiplier tube (PMT) which is commonly used to measure scintillating light due to its high internal gain and reasonable quantum efficiency is used to convert light signals to electrical signals. Using this dosimeter, γ-rays generated by a Co-60 are measured as a function of field sizes, and the percent depth dose (PDD) curves are obtained according to the different depths of a water phantom. The Cerenkov lights are also measured according to field sizes and different depths of a water phantom using a background optical fiber. To remove Cerenkov lights, simple subtraction method is investigated.

II. MATERIALS AND METHODS

As a sensing probe of a SFOD, organic scintillators made out of a polyvinyl-toluene (PVT) with a small amount of wavelength-shifting fluors are used. It was found that typical organic scintillators, which are based on PVT, show no signifying aging for absorbed doses up to 1.0 kGy [8]. This experiment used four kinds of commercially available organic scintillators (BCF-10, BCF-12, BCF-20 and BCF-60, Saint-Gobain Co.) in order to select an organic scintillator which has more light output and is suitable for a light-measuring device such as a PMT. The physical properties of organic scintillators used in this study are listed in Table I.

<table>
<thead>
<tr>
<th>Organic Scintillator</th>
<th>Emission Color</th>
<th>Emission Peak (nm)</th>
<th>Decay Time (ns)</th>
<th>1/e Length (m)</th>
<th># of Photons per MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>BCF-10</td>
<td>Blue</td>
<td>432</td>
<td>2.7</td>
<td>2.2</td>
<td>~8000</td>
</tr>
<tr>
<td>BCF-12</td>
<td>Blue</td>
<td>435</td>
<td>3.2</td>
<td>2.7</td>
<td>~8000</td>
</tr>
<tr>
<td>BCF-20</td>
<td>Green</td>
<td>492</td>
<td>2.7</td>
<td>&gt;3.5</td>
<td>~8000</td>
</tr>
<tr>
<td>BCF-60</td>
<td>Green</td>
<td>530</td>
<td>3.5</td>
<td>3.5</td>
<td>~7100</td>
</tr>
</tbody>
</table>

Commercial grade plastic multimode optical fibers (SH4001, Mitsubishi Ltd.) are used to guide scintillating lights from the sensing probes to the light measuring device. The outer diameter of this fiber is 1.0 mm, and the cladding thickness is 0.01 mm. The refractive indices of the core and the cladding are 1.492 and 1.402, respectively, and the numerical aperture (NA) is 0.510. The NA stands for the light-gathering power and more lights can be guided by a POF with a high NA.

A PMT (H7732-10, Hamamatsu Photonics Inc.) is used as a light-measuring device. The measurable wavelength range of the PMT is from 185 nm to 900 nm, and the peak wavelength is about 400 nm. Typical and maximum dark currents of this PMT are about 3.0 and 50.0 nA, respectively with 1.0 V of control voltage.

Fig. 1 shows the experimental setup for measuring scintillating light using the SFOD irradiated by the γ-rays from Co-60 machine. The sensing probe of a SFOD consists of an organic scintillator and a POF. Before bonding them with an optical epoxy, the cross-sectional surfaces of both the scintillators and the POFS were polished with various kinds of polishing pads in a regular sequence. The outer surface of the SFOD probe was surrounded by reflective-paint-based titanium dioxide (TiO2) to increase the scintillating light-collection efficiency and to intercept light noise from outside with a polyethylene-based black jacket. Co-60 source with a half-life of 5.271 years is used for γ-rays irradiations source. Energies of γ-rays generated from the Co-60 are 1.173, 1.332 MeV and field sizes are 5 × 5, 10 × 10, 20 × 20 and 30 × 30 cm². The source to surface distance (SSD) which means the distance between the Co-60 source and the surface of a water phantom, used in this study is 80 cm. When γ-rays are irradiated on a sensor probe, the scintillating lights generated from organic scintillator are transmitted to the PMT by 15 m length of POF. Eventually, amplified electric signals of a PMT are measured using a LabVIEW.
the same size of its diameter. The R-square value in this result is 0.9999, which means that the accuracy of liner fitting is over 99.99%.

Figure 2. Measurements of scintillating light according to the different kinds and lengths of organic scintillators (a: different kinds of organic scintillators, b: different lengths of the BCF-12).

Figure 3. Measurements of scintillating and Cerenkov lights generated from a scintillating fiber-optic dosimeter according to irradiation field sizes.

Measured scintillating and Cerenkov lights outputs according to the γ-ray field sizes are as shown in Fig. 3. The amount of scintillating light generated from a SFOD without Cerenkov light increases as increasing γ-ray field size due to the increase of γ-ray scatterings with a collimator and a phantom [9]. In the case of Cerenkov light, it also increases as increasing the field size due to the same reason for that of scintillator. Additionally, the Cerenkov light has a dependence on irradiated lengths of an optical fiber because it is generated from the optical fiber [3,6]. Therefore, the amount of Cerenkov light also increases as the field size increases because the γ-ray covers longer part of an optical fiber. As shown in Fig. 3, the gradient of Cerenkov light with the field sizes is higher than that of scintillating light as expected.

Figure 4. Measurements of PDD and Cerenkov light according to the depths of a water phantom (a: Measurements of PDD and Cerenkov light, b: Ratio of Cerenkov/scintillating light).

Fig. 4a shows measured PDD curve and Cerenkov light according to depths of water phantom with a 10 cm × 10 cm field of γ-rays from the Co-60. In clinical practice, the central axis dose distribution is characterized by PDD which can be defined as the quotient of the absorbed dose at any depth $r$ ($d_r$) to the absorbed dose at a fixed reference depth $r_0$ ($d_{r_0}$). PDD can be expressed as [9]:

$$PDD \ (\%) = \frac{d_r}{d_{r_0}} \times 100 \quad (1)$$

For radiation beams used in radiotherapy, the reference depth is usually taken at the position of the peak absorbed
dose. The peak absorbed dose on the central axis is called the \(d_{\text{max}}\) which is defined by the following expression:

\[
d_{\text{max}} = \frac{d}{PDD} \times 100
\]  

(2)

It has been observed that the typical depth of maximum dosage along the central axis for 10 cm \(\times\) 10 cm field in a water phantom is 5.0 mm for \(\gamma\)-rays from the Co-60. In Fig. 4a, the \(d_{\text{max}}\) is 5.0 mm for \(\gamma\)-rays from the Co-60 and this finding is consistent with well known results obtained using an ion chamber (reference data) [10]. In this result, the mean difference between a SFOD and an ion chamber is about 1.4675%. The mean standard deviation of measured data using a SFOD is 1.2118%.

Both trends of PDD and Cerenkov lights are similar but not the same. As shown in Fig. 4b, the ratios of measured amount of Cerenkov to scintillating lights increase slightly as increasing the depths of a water phantom because the field sizes increase slightly as increasing the depths of a water phantom.

IV. CONCLUSIONS

In this study, we have fabricated a SFOD by using of an organic scintillator and a plastic optical fiber (POF) for Co-60 therapy dosimetry. Using this dosimeter, \(\gamma\)-rays generated from a Co-60 source are measured as a function of field sizes and PDD curves are obtained according to the different depths of a water phantom. Cerenkov lights generated by the interactions of primary or secondary electrons are also measured according to the field sizes and the different depths of a water phantom using a background optical fiber. To remove Cerenkov lights, simple subtraction method is investigated.

The SFOD has many advantages over conventional dosimeters in radiotherapy. First, water-equivalent organic scintillators and POFs make it possible to measure dose distribution with a minimal perturbation. Second, small sensitive volume of a SFOD can contribute to measuring dose-distributions and PDDs with a high resolution. Third, there should be no corrections such as temperature, pressure or humidity for accurate dose measurements because a plastic optical fiber is used in the SFOD. Further studies will be carried out to fabricate a SFOD array with smaller diameter of optical fibers and organic scintillators for high spatial resolution and real-time dose distribution measurement in radiotherapy dosimetry. It is expected that the SFOD with an array can be an effective, accurate and convenient tool for measuring dose distributions and PDDs in radiotherapy dosimetry.

REFERENCES


