Characteristics of fiber-optic radiation sensor for passive scattering proton beams

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ABSTRACT: The aims of this study were to investigate the characteristics of a fiber-optic radiation sensor (FORS) that detects the fluorescence light produced by proton beam and to verify its effectiveness in proton therapy quality assurance (QA). Various characteristics of the FORS were investigated, such as the linearity of its relationships to the sensitive length of fiber for the proton beams of intermediate ranges (165.46 and 178.37 MeV) and to the measured dose, as well as its dose rate dependence. In addition, patient specific prescription dose QA was conducted for five patients actually undergoing proton therapy and the results were compared with the doses measured using an ion chamber. The results show that the signal of the FORS is linearly related to the sensitive length of fiber and to the irradiated dose in the range from 1 to 500 cGy. The QA results obtained using the FORS system showed good agreement with the corresponding ion chamber results, with an average difference of 0.40% and a standard deviation of 0.35%. The FORS was dose-rate independent for proton currents up to 5 Gy/min. The profiles of various proton beams obtained using an array of FORS, which were measured as an application of the developed dosimetric system, closely agreed with the profiles acquired using EBT3 film. In summary, the experimental results of FORS demonstrated its effectiveness for use in various proton therapy QA tests.

KEYWORDS: Dosimetry concepts and apparatus; Photon detectors for UV, visible and IR photons (vacuum) (photomultipliers, HPDs, others)

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1 Introduction

Proton therapy beam has unique dosimetric characteristics compared to those of conventional X-ray radiation because of the advantageous percent depth doses, including a near-zero exit dose just beyond the target volume, resulting in reduced doses to normal tissue, with better conformation of the dose to the target volume [1–3]. In proton therapy, the energy deposition increases with the penetration distance, with a maximum near the end of the proton range, called the Bragg peak, followed by a rapid decline within a few millimetres [4, 5]. Even though some advanced photon radiotherapy techniques such as intensity-modulated radiotherapy can produce high dose uniformity across a tumor, proton therapy is a powerful means of decreasing the low-dose volume in the surrounding normal tissue. However, in proton therapy, it is essential to verify the accuracy of the delivered dose and its location because the Bragg peak can produce significant harm if delivered improperly. Therefore, the quality assurance (QA) of clinical proton beams is of great importance in proton therapy.

In general, ion chambers can be considered to be the preferred proton dosimetry detectors due to their accuracy [6, 7]. Although thermo-luminescent dosimeters (TLDs), glass dosimeters, plastic scintillation detectors, and diodes have been used as dosimetric tools for radiotherapy QA, each has drawbacks in proton therapy. For example, although absolute dose measurements determined by TLD-100s and ion chambers differ by less than 3%, TLDs require careful handling, complicated analysis, and TLD reading equipment [8]. While glass dosimeter also shows relatively good agreement with ion chamber results, there can be a mismatch between glass dosimeter and ion chamber measurement. For example, the full widths at half-maximum of Bragg peaks in
proton therapy measured by glass dosimeters are wider than those measured by ion chambers [9].

The plastic scintillation detector is also commonly used in radiotherapy [10–12]. Although the results using plastic scintillation detector showed good agreement for photon and electron beam, the absorbed dose measured by plastic scintillation is non-linear due to quenching effects in proton therapy [13–16]. Generally, the maximum dose measured by scintillator is about 20% lower than that measured with the ion chamber at the Bragg peak point. It has been reported that the non-proportionality, under-response of the light yield to the proton beam dose, is mainly due to the ionization quenching effect in plastic scintillator which is caused by non-radiative de-exitations occurring at high density of deposited energy [13, 17, 18]. Although diode detectors can be used for precise measurements, it is vulnerable to radiation damage [19].

Despite the advantages of ion-chamber-based systems in proton dosimetry, they still have limitations. For example, it is too bulky to use them as a real-time dosimeter during treatment since it is not easy to install this system between snout and patient. The resolution and cost effectiveness of ion chamber array system is another problem in application of this system in active pencil beam treatment.

Recently, Darafsheh et al. conducted spectral analysis obtained from optical fiber during proton beam irradiation and concluded that the fluorescence light produced by proton beam in optical fiber is a good candidate for measuring absorbed dose of proton beams [17, 20]. Unlike plastic scintillation detector, the light signal measured by bare optical fiber does not show the quenching effect for proton dosimetry since the density of the excitation centers along the track is very low. In addition to this, optical fiber has many advantages as a dosimeter since it shows good flexibility, water-equivalence, real-time measurement and cost effectiveness.

Recently there was argument about the physical origins of visible light measured by optical fiber during proton irradiation [17]. In general, Cerenkov Radiation (CR) is generated by delta electrons travelling faster than the speed of light and the threshold corresponds to minimum proton kinetic energy of ∼80 MeV. Helo et al. reported that CR in proton beam is mainly due to fast electrons induced by nuclear reactions which are not proportional to the dose [21]. Based on the spectral analysis obtained from bare fiber during proton beam irradiation, Darafsheh et al. also reported that the signal in fiber optic dosimetry of proton beams is not CR but the fluorescence light of the fiber and the amount of the generated Cerenkov light does not follow the radiation absorbed dose in a medium. Based on the literature review, it is plausible that the visible light signal, which is proportional to the absorbed dose of proton beam, is not coming from the Cerenkov radiation.

Although there was an argument about the physical origin of visible light measured by optical fiber during proton irradiation, the potential of fiber-optic radiation sensors (FORS) as a dosimetric tool was confirmed for the verification of “proton range” by the recent study of Son et al. [22]. To use FORS as a dosimetric tool for patient specific QA, however, the tumor dose (i.e., prescription dose) and 2D dose of each field should also be verified using FORS system. In the present study, we investigated the characteristics of FORS for the measurement of the prescription dose and line profile of each field and evaluated the effectiveness of the FORS system for patient specific proton QA.
2 Methods and materials

2.1 System configuration

We performed proton beam irradiation using the Ion Beam Applications (IBA, Louvain-La-Neuve, Belgium) cyclotron PROTEUS 235 at the Korea National Cancer Center. Proton beams with energies of up to 230 MeV can be produced at this facility, and the maximum proton current of the cyclotron is 300 nA at a radiofrequency of 106 MHz [23]. In the fabricated FORS, the fluorescence light by the proton beam was produced in clear plastic optical fibers (POFs). The FORS system consisted of POFs, a photomultiplier, and a data acquisition system, as shown in figure 1. The employed photomultiplier tube (PMT: H12428-203, Hamamatsu Photonics, Japan) operates at wavelengths of 300–880 nm, and its dark current per channel is typically about 0.2 nA at 800 V (C9525, Hamamatsu Photonics). Each POF was optically coupled to a multi-anode photomultiplier (MAPMT) tube and the MAPMT signals were processed using the National Instruments data acquisition system (NI-DAQ, National Instruments, Texas, U.S.A.). Amplifier circuit with attached to the PMT was used at the output of the PMT. The currents from PMT pass through a current-to-voltage converter circuit and NI-9205 modules acquire voltage signals from the current-to-voltage converters. Real-time monitoring was possible by using a graphic user interface programmed in Labview (version 2013, National Instruments, U.S.A.).

![Figure 1. Systematic figure of FORS system.](image)

Figure 1. Systematic figure of FORS system. Experimental setups for (a) analysis of linearity of relationship between light yield and sensitive length of fiber, (b) chosen sensitive length (i.e., 2 cm) for the experiment such as dose, dose rate and proton beam rage, and (c) FORS array used to measure proton beam field size.
2.2 Sensitive length of fiber vs. light yield

The POFs (PGR-CD2001-30-E, Toray, Japan), whose cross-sections were circles, were multi-mode, step-index optical fibers. Their cores, claddings, and jackets were made of PMMA, fluorinated polymer, and polyethylene, respectively. Each POF with cylindrical shape had an outer diameter of 2.0 mm, a 1.98-mm-thick core, and a 0.02-mm-thick cladding structure. We utilized the subtraction method to determine the active volume, by employing a reference POF and a longer POF. To investigate the relationship between the fluorescence light yield and the sensitive length of fiber for the proton beam, various sensitive lengths of the POFs were irradiated: 2 cm, 4 cm, 6 cm, 8 cm, and 10 cm, as shown in figure 1(a). In experiments for dose, dose rate and proton beam range, we used “2 cm difference in optical fiber lengths” because it was simply the minimum length difference (i.e., the smallest sensitive volume) which shows the stable and reproducible signal compared to ion chamber measurement. If the length in difference is less than 2 cm, then the difference in light yield produced by the two fibers was not large enough and the signal was unstable. In the experiment of sensitive length of fiber vs. light yield, the proton beam energy was 178.37 MeV (arbitrarily chosen in intermediate beam ranges) with a square 10×10 cm² field, and the fluorescence light was measured by scanning at 0.5 mm intervals in a water-phantom. To check the dose rate dependence of the developed system, the fluorescence light yield of the FORS was measured using various dose rates, i.e., various proton currents in water-phantom. In this experiment, the sensor consisted of two POFs with circle shape, each of which had an outer diameter of 1.0 mm (PGS-CD1001-13-E, Toray, Japan), a 0.98-mm-thick core, and a 0.02-mm-thick cladding structure. A 2 cm difference in optical fiber lengths was used to apply the subtraction method, as shown in figure 1(b). The POFs were irradiated by a 165.46 MeV (arbitrarily chosen in intermediate range) proton beam with a square 10×10 cm² field at various dose rates from 1 to 5 Gy/min and were scanned in a water phantom.

2.3 Prescription dose QA

To relate the fluorescence light measurements to the prescribed doses, calibration was first performed with a calibrated ion chamber whose absolute doses were determined according to the IAEA TRS-398 protocol [24]. The difference in length between the two optical fibers (PGS-CD1001-13-E, Toray, Japan) with circle shape was again 2 cm (figure 1(b)). To obtain the calibration curve, the signal with FORS was measured at 6.9 cm depth (mid-SOBP depth) to the proton beam (Range 8.34, SOBP 2.92, 158.73 MeV) at the following dose levels: (1, 2, 5, 10, 15, 20, 50, 100, 150, 200, 300, and 500) cGy. To evaluate the clinical effectiveness of the FORS, we randomly selected five actual patient treatment plans (various treatment site: breast, liver, medial ear and CSI) performed at our National Cancer Center Korea and compared the resulting FORS measurements with the corresponding ion chamber measurements.

2.4 Field size measurement

We developed a finely segmented detector array to measure the field sizes of proton beams. The FORS-based array system, which was arranged in the direction transverse to the proton beam direction, was developed to achieve real-time beam size measurement. To measure the field size of each proton beam, fibers were laid side by side in a supporting frame, as shown in figure 1(c). Optical
fibers with square $1 \times 1 \text{mm}^2$ cross-sections were used to achieve material uniformity. A square POFs (BCF-98, SAINT-GOBAIN CRYSTALS, France) were made of a polystyrene-based core with a refractive index of $n_{\text{core}} = 1.60$ and an acrylic cladding with a refractive index of $n_{\text{clad}} = 1.49$. Figure 2 shows a picture of the system configuration for FORS array. In this system, 64 channels of fiber are optically coupled to MAPMT. MAPMT signals are processed using NI-DAQ system with acquisition frequency of 1 kHz. 100 kΩ resistor is added between theground and each anode output as a passive current to voltage converter for NI 9205. The MAPMT was gain-calibrated because its gain differed in each channel. Uniform gain calibration was achieved using a representative proton beam, which has 178.37 MeV of energy (range of 9.02 cm) with spread-out Bragg peak (SOBP) of 4.25 cm in the water. The experiment was conducted for square proton beam fields of $8 \times 8 \text{cm}^2$ and $3.5 \times 3.5 \text{cm}^2$. The experimental results obtained by the FORS array were compared to the results of measurements performed using Gafchromic EBT3 films (International Specialty Products, New Jersey, U.S.A.) for proton beams.

Figure 2. (a) System configuration including optical fiber, PMT, NI-DAQ, power supply and software, (b) a picture for FORS array.

2.5 Statistical analysis

All the measurements were carried out three times unless otherwise specified and the uncertainty was specified by one sigma (i.e., standard deviation).

3 Results and discussion

3.1 Sensitive length of fiber vs. light yield

Figures 3(a) and 3(b) present the average measured relative depth dose curves of a proton beam and the normalized light yields for various sensitive lengths of fiber, respectively. As shown in figure 3(b), the light yield increases linearly with increasing sensitive length of fiber. The resulting fitting equation has a gradient of 4.25 and an R-square value of 0.9999, indicating the accuracy of the correlation between the measured data and the fitting line. The experimental results illustrate
that the fluorescence light yield of the FORS is proportional to sensitive length of fiber for the proton beam. To check the effectiveness of FORS with 2cm sensitive length, the comparison of representative depth dose distributions measured by FORS with the results with ion chamber was carried out and the results were shown in figure 4.

As seen in figure 4, the percent depth dose (PDDs) measured by FORS for two intermediate proton beam ranges are well matched with the results measured by ion chamber. The average dose ratios measured by FORS and Ion chamber was almost 1 at various depths as seen in figure 4(b).
Figure 5 shows that the relationship between the fluorescence light yield of the FORS and the dose rate. The results show that the ratios for various dose rates are close to 1, which confirms the dose rate independency of developed system.

![Image showing depth dose and fluorescence light yield](image)

**Figure 5.** (a) Measured relative depth doses of proton beams for various proton beam currents. (b) Fluorescence light yield of FORS versus dose rates of proton beams. The uncertainties were added as error bars in (a) and (b).

3.2 Prescription dose QA of clinical cases

Figure 6 presents the light yield versus dose curve, which was calibrated using a calibrated ion chamber. As shown, the fluorescence light yield of the FORS increases linearly with as the dose measured using the ion chamber increases from 1 to 500 cGy. For example, a light yield of 100 (arbitrary units) measured by our FORS system corresponds to a 100 cGy absorbed dose. The results show that the fitting curve is linear, with an R-square value of 0.9999, suggesting that dose measurement based on the light yield of the FORS is possible. The effectiveness of the FORS for patient specific prescription dose measurements was assessed by comparing the results obtained using this system with those obtained using the ion chamber. The results of proton prescription dose QA performed using the FORS for five patients, who were treated by double scattering mode, are listed in table 1. As shown, the prescription doses measured by the FORS differ by less than ~1% from those measured by the ion chamber.

A unique advantage of our developed system is its ability to measure beam signals in real time. Recently, pencil beam scanning (PBS) mode, which enables the best conformity to tumors, was developed and has been used in proton therapy. Unlike in the passive scattering mode, tumors are scanned with very precise beams of protons at high speed in PBS mode. Because the proton beams used for scanning are small and the scanning is performed rapidly, it is very difficult to measure the sizes and locations of pencil beams in real time using conventional dosimetric system. However, it is essential for the dosimeters used in PBS to be able to measure the beam signals in real time. One of the unique features of the developed FORS system is its ability to measure proton beams in real time, suggesting the applicability of our proposed FORS system in QA for proton beam with PBS mode.
Figure 6. Relationship between FORS light yield and output dose measured by ion chamber. Solid line is linear best fit and the uncertainties were added as error bars.

Table 1. Comparison of patient specific prescription dose QA results measured by standard ion chamber and FORS for the treatment fields of five patients. The prescription doses were measured once and three times by ion chamber and FORS, respectively.

| Patient Number | Fields | Field size (cm) | Measurement depth (cm) | Prescription dose (Ion chamber, cGy) | Measured dose (FORS, cGy) | %Diff. |
|----------------|--------|----------------|------------------------|---------------------------------------|--------------------------|--------|
| Patient1       | Field1 | 5.5×13.5       | 6.90                   | 139.9                                 | 138.4±0.76               | 1.10±0.54 |
| (Breast)       | Field2 | 4.0×12.5       | 6.38                   | 139.5                                 | 138.6±0.68               | 0.67±0.49 |
| Patient2       | Field1 | 11.0×7.0       | 5.71                   | 201.4                                 | 201.9±0.95               | 0.23±0.47 |
| (Liver)        | Field2 | 11.5×8.5       | 5.98                   | 149.2                                 | 149.6±2.00               | 0.27±1.34 |
|                | Field3 | 11.0×9.5       | 7.41                   | 149.5                                 | 149.9±0.94               | 0.25±0.63 |
| Patient3       | Field1 | 4.0×7.0        | 9.03                   | 227.0                                 | 227.2±1.19               | 0.10±0.52 |
| (Liver)        | Field2 | 3.1×7.0        | 8.39                   | 226.8                                 | 226.2±0.59               | 0.26±0.26 |
|                | Field3 | 3.0×7.0        | 9.08                   | 227.4                                 | 227.7±1.85               | 0.13±0.82 |
| Patient4       | Field1 | 9.0×7.0        | 4.23                   | 81.0                                  | 81.0±1.05                | 0.04±1.30 |
| (Medial ear)   | Field2 | 9.5×7.0        | 4.79                   | 80.6                                  | 81.5±1.17                | 1.16±1.46 |
|                | Field3 | 8.0×7.0        | 6.55                   | 80.5                                  | 80.8±1.08                | 0.37±1.34 |
| Patient5       | Field1 | 10.0×14.0      | 8.05                   | 175.7                                 | 176.1±0.33               | 0.25±0.19 |
| (CSI)          | Field2 | 11.0×16.5      | 8.79                   | 174.7                                 | 175.4±1.46               | 0.42±0.84 |
3.3 Field size measurement of clinical cases

Figure 7 shows the profiles of two proton fields measured using the FORS array. In this experiment, the field sizes were 8×8 cm² and 3.5×3.5 cm², and the data measured using Gafchormic EBT3 films are also shown in the graph for comparison. The data for measured field sizes and penumbra widths with corresponding uncertainties are listed in table 2, which illustrates that the field sizes measured by the EBT3 film and the FORS agree closely. However, slightly wider penumbra widths were obtained by the FORS, with a maximum difference of 0.57 cm from those measured by using the EBT3 films. The penumbra width differences are attributable to the relatively poor spatial resolution of the FORS system compared to that of the EBT3 film method.

![Figure 7](image_url)

**Figure 7.** Average beam profiles measured by EBT3 film (solid lines) and FORS (full dots) for proton beams with field sizes of 8×8 cm² and 3.5×3.5 cm².

**Table 2.** Measured data by the EBT3 film and an array of FORS for two proton beams. Field sizes of two proton beams were 8×8 cm², 3.5×3.5 cm².

| Dosimetric tool | Reference field size (cm) | Field size (50–50%) | Penumbra (Left) 20–80% | Penumbra (Right) 20–80% |
|-----------------|---------------------------|---------------------|------------------------|------------------------|
| EBT3 Film       | 8.0 × 8.0                 | 7.86±0.02 cm        | 0.30±0.00 cm           | 0.30±0.00 cm           |
| FORS            |                           | 7.84±0.07 cm        | 0.54±0.03 cm           | 0.57±0.03 cm           |
| EBT3 Film       | 3.5 × 3.5                 | 3.54±0.01 cm        | 0.29±0.01 cm           | 0.28±0.02 cm           |
| FORS            |                           | 3.56±0.07 cm        | 0.44±0.01 cm           | 0.44±0.01 cm           |
4 Conclusion

In summary, we studied the characteristics of clinical proton beam measurements performed using a FORS and evaluated the effectiveness of the FORS as a proton therapy QA tool. Our results indicate that the FORS is suitable for use in real-time QA of clinical proton beams.

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