Embedded structure fiber-optic radiation dosimeter for radiotherapy applications

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Abstract: An investigation into a novel in-vivo PMMA (polymethyl methacrylate) plastic fiber-optic dosimeter for monitoring low doses of ionizing radiotherapy radiation in real time and for integrating measurements is presented. The fabricated optical fiber tip possessed an embedded structure. A scintillation material, terbium-doped gadolinium oxysulfide (Gd2O2S:Tb), capable of emitting visible light at around 545 nm which is ideal for transmission through the PMMA when exposed to ionizing radiation was embedded in the PMMA plastic fiber. The dose rate of incident ionizing radiation is measured by analyzing the signal intensity emitted from the scintillation material which propagates through the fiber to a distal MPPC (multi-pixel photon counter). The dosimeter exhibits good repeatability with an excellent linear relationship between the fiber-optic dosimeter output and the absorbed radiation dose with an outstanding isotropic response in its radial angular dependence.

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References and links
cannot provide real-time information concerning the dose delivered. Silicon diodes offer TLDs allows it to provide dose data with relatively high spatial resolution, but the TLD dependent upon the radiation absorbed by the TLD during radiotherapy. The small size of the visible light is emitted from the crystal present in the device. The amount of light emitted is the crystalline material in the device. Following treatment, the TLD detector is heated and treatment. The TLD device absorbs the ionizing radiation during radiotherapy which ionizes effects transistors (MOSFETs) and ionization chambers (ICs) [2–7]. TLD detector devices are thermoluminescent dosimeters (TLDs), silicon diodes, radiochromic films, metal-oxide field-effect transistors and ionization chambers (ICs) [2–7]. TLD detector devices are available as chip, rod or cube types. They can be placed externally on a patient’s body during treatment. The TLD device absorbs the ionizing radiation during radiotherapy which ionizes the crystalline material in the device. Following treatment, the TLD detector is heated and visible light is emitted from the crystal present in the device. The amount of light emitted is dependent upon the radiation absorbed by the TLD during radiotherapy. The small size of the TLDs allows it to provide dose data with relatively high spatial resolution, but the TLD cannot provide real-time information concerning the dose delivered. Silicon diodes offer limited applicability due to their physical size. Radiochromic films are dosimeters that are capable of providing an extensive bidimensional measurement 24–48 hours after irradiation. The radiochromic film response depends on environmental conditions, such as temperature, humidity, which can adversely affect the dose rate. Therefore, radiochromic film response results must be corrected using appropriate factors. Electronic semiconductor diodes can be used directly for relative dose measurement, without requiring corrections for depth dependence. One disadvantage of these sensors is radiation damage, which generates dose intensity dependence. MOSFET detectors measure the difference in voltage shift in the diode before and after exposure to ionizing radiation. However, they accumulate a dose effect, which limits their lifetime. And studies have shown that these dosimeters do not have sufficient sensitivity at distances beyond 10 cm from the source of radiation. The IC is an

1. Introduction

Radiotherapy constitutes the use of ionizing radiation for the treatment of cancer and in China radiation therapy is used to treat more than 70% of all cancer patients. The quality of the delivery of radiation therapy treatment depends on the ability to accurately predict and measure the dose received by the whole volume being irradiated. Consequently, there is a demand for new, in-vivo radiation monitors that can accurately measure body dosage in real time allowing the linear accelerator to deliver accurate radiation dose to the tumor cells without damaging normal cells [1].

Many types of dosimeters have been investigated to measure absorbed dose. These include thermoluminescent dosimeters (TLDs), silicon diodes, radiochromic films, metal-oxide field-effect transistors (MOSFETs) and ionization chambers (ICs) [2–7]. TLD detector devices are available as chip, rod or cube types. They can be placed externally on a patient’s body during treatment. The TLD device absorbs the ionizing radiation during radiotherapy which ionizes the crystalline material in the device. Following treatment, the TLD detector is heated and visible light is emitted from the crystal present in the device. The amount of light emitted is dependent upon the radiation absorbed by the TLD during radiotherapy. The small size of the TLDs allows it to provide dose data with relatively high spatial resolution, but the TLD cannot provide real-time information concerning the dose delivered. Silicon diodes offer limited applicability due to their physical size. Radiochromic films are dosimeters that are capable of providing an extensive bidimensional measurement 24–48 hours after irradiation. The radiochromic film response depends on environmental conditions, such as temperature, humidity, which can adversely affect the dose rate. Therefore, radiochromic film response results must be corrected using appropriate factors. Electronic semiconductor diodes can be used directly for relative dose measurement, without requiring corrections for depth dependence. One disadvantage of these sensors is radiation damage, which generates dose intensity dependence. MOSFET detectors measure the difference in voltage shift in the diode before and after exposure to ionizing radiation. However, they accumulate a dose effect, which limits their lifetime. And studies have shown that these dosimeters do not have sufficient sensitivity at distances beyond 10 cm from the source of radiation. The IC is an

absolute radiation dose measurement instrument, but it measures radiation point by point based on charge generation in an electric field. Although considered to be the ‘gold standard’ instruments for Quality Assurance (QA) purposes, a major disadvantage of these devices is that they use relatively high voltages (Typically many tens or greater) voltages to generate the Electric Field for detection. This makes them completely incompatible for use in-vivo. Additionally ICs require the deployment of dose ionization conversion factors, which strongly depend on the electron beam dose rate.

To improve the shortcomings of these conventional dosimeters, a novel method using scintillators based on a normal plastic optical fiber (POF) made of PMMA (polymethyl methacrylate) is presented. This type of radiation monitor offers major advantages including, small dimensions, low mass, a long operating length, reproducibility, continuous sensitivity, dose linearity, real-time operation, insensitivity to external electromagnetic fields and comprises a simple, robust and clinically compatible measuring system. Scintillating materials can be divided into organic (plastic scintillator) and inorganic scintillators. Radiation dosimeters based on scintillators often employ a tip composed of a scintillating material, which is optically coupled to the end of an optical fiber. As the scintillator is exposed to ionizing radiation, an optical signal is generated and this is guided by the optical fiber toward a detecting device placed remotely from the irradiation zone. Plastic scintillator dosimeters have many desirable dosimetric characteristics when compared with traditional detector systems, as extensively described in the literature [8–17]. However, plastic scintillator dosimeters have at least one shortcoming; a low signal-to-noise ratio (SNR) resulting from the Cerenkov radiation emission. To overcome this problem, the device must use a parallel-paired fiber light guide and two identical photomultiplier tubes (PMTs) to subtract the background signal resulting from the Cerenkov radiation emission. However, this solution significantly enlarges the dosimeter’s volume. By contrast scintillation dosimeters that employ inorganic scintillation materials exhibit a high SNR. As an example, conducted by McCarthy [18], fabricated a dosimeter by injecting a mixture of a scintillating phosphor material (e.g. Gd2O2S:Tb), an epoxy resin and hardener onto an exposed PMMA optical fiber. The device was clinically useful, but the design resulted in a homogeneous problem with relatively low light efficiency, which resulted in the dosimeter failing to exhibit a satisfactory response.

In this reported study, a new kind of inorganic dosimeter based on a novel structure was fabricated to measure a low-energy absorbed radiation dose from a clinical linear accelerator (CLINAC). In addition to the advantages offered by inorganic dosimeters, this reported dosimeter overcomes the homogeneous and low coupling efficiency problems, because of its embedded structure. In this study, the device was assessed for its repeatability and linear response according to the dose rates of CLINAC. A depth-dose curve of this novel inorganic dosimeter for a 6 MV photon beam was also obtained.

2. Dosimeter design and fabrication

The equipment used in this investigation is shown schematically in Fig. 1(a) which details a plastic fiber-optic dosimeter submerged in a water-equivalent tank in a radiotherapy bunker room. The photograph of the immersed dosimeter and IC is shown in Fig. 1(b). The dosimeter was placed in the center of the rectangular radiation beam field formed by a Varian Linear Accelerator. The principle of the plastic fiber-optic dosimeter described in this study relies on the conversion of the incident radiation dose to a measurable visible optical signal by fluorescence [18]. According to this phenomenon, when exposed in the radiation beam, the dosimeter emits a visible light signal which travels through a 25 meter length PMMA fiber to a sensitive photodetector (MPPC) located in the control room, which monitors the intensity of the fluorescence. The intensity of the fluorescence is directly related to the radiation dose rate that is incident on the dosimeter.
The initial dosimeter design used for this investigation is shown in Fig. 2 and consisted of an injection method to coat a mixture of an epoxy resin, hardener and a specific radiation sensitive scintillating material (the ratio of these materials is 8:1:1) onto the optical fiber core. This approach is based on that reported in McCarthy [18]. The initial design was superceded by an improved version which comprises a novel embedded structure, which improves the properties of the dosimeter and is shown in Fig. 3. The fiber-optic dosimeter described in this study comprises a PMMA fiber whose core was micromachined to create a small diameter (0.25 to 0.5 mm) hole at the tip of the fiber. An inorganic scintillating material, terbium-doped gadolinium oxysulfide (Gd₂O₂S: Tb), which fluoresces upon exposure to the incident ionizing radiation (X-Ray or electron beam), was filled and packaged inside the small hole using an epoxy resin adhesive. Because of the unique design the amount of the active material is only about 1 mg. The visible optical signal generated by the inorganic scintillator propagates 25 meters to detection end of the optical fiber from the point of measurement. This distal end of
the fiber was carefully polished and then connected to a Hamamatsu MPPC C11208-350 Avalanche Photodetector Array using an SMA connector where the intensity of the visible fluorescent light signal was measured.

The emission spectrum of the Gd_2O_2S:Tb scintillator exposed to the X-ray ionizing radiation is shown in Fig. 4. There are three discernable peaks present at 490 nm, 545 nm and 590 nm, of which the peaks at 490 nm and 545 nm are both highly suitable for propagating through the PMMA optical fiber.

Fig. 4. The emission spectrum of the Gd_2O_2S:Tb.

3. Results and discussion

The dosimeter was tested at the External Radiation Beam Therapy clinic of the First Affiliated Hospital of the Harbin Medical University, Harbin, China using a Varian Linac. As shown in Fig. 1(b), the fiber-optic dosimeter was fully submersed in a water-equivalent tank mounted on the treatment bed at the center of the ionizing beam with a field size of 10 × 10 cm² at a SSD (Source to Surface Distance) of 100 cm. The IC (TW30012-1) was located in a special dry compartment placed in the water tank immersed in water to obtain the measurement data. Both the IC and fiber-optic dosimeter were exposed to near identical irradiation conditions for reference measurement and data validation.

3.1 Comparing the embedded fiber-optic dosimeter structure with the previous type of dosimeter design

The performance of both the novel fiber-optic dosimeter with the embedded structure and the previously developed type of dosimeter by McCarthy [18], were directly compared by placing them at a submerged depth of 1.5 cm where the dose is the maximum for a beam of 6 MV (photon Energy) as shown in Fig. 1(b). The response of these two dosimeters was determined for 200 MU (Monitor Units) at a dose rate of 600 MU/min with a field size of 10 × 10 cm² and a normal X-ray photon energy of 6 MV. In this investigation, 1 MU is equal to an absorbed dose of 1 cGy under standard reference conditions. A Hamamatsu MPPC C11208-01 Avalanche Photodetector Array with a modified shortened gate time of 0.1 ms was used only for this part of these tests (section 3.1). All the other tests were conducted with the MPPC detector array with a longer gate time which was 0.1 s (section 3.2-3.5). Figure 5 shows the optical intensity of the visible light signal. Dark current noise signals were initially received at the beginning of the test and at the end when the radiation was off in order to establish the ‘dark’ background noise level. At the start of the radiation exposure, the optical signals changed continuously with the pulses of the Linac. Because of the setup conditions where a very short gate is sensitive to the Linac pulses. The pink trace corresponds to the 1000 point moving average value and it is clear that the amplitude of this signal remained stable and continuous throughout the 20 second duration of the Linac irradiation as shown by
the pink line in Fig. 5. The small fluctuation of the signal is due to the Linac delivering unstable X-Ray irradiation as was observed from the instrument panel of the Linac.

![Graph](image1.png)

**Fig. 5.** The results: (a) The signal time resolved response measured by the previous dosimeter design. (b) response measured by the novel dosimeter.

The intensity of the pulse’s peak obtained with the injection method dosimeter was approximately 800 counts and the S/N ratio, calculated for a $10 \times 10$ cm$^2$ field size at $D_{\text{max}}$, was equal to 8.6 dB. But the signal intensity resulting from the novel dosimeter with an embedded structure was about 2500 counts, more than three times greater than the old dosimeter with a S/N ratio of 18.6 dB. It was concluded that the variation in SNR between the two sets of results was due to two factors:

1. A photon produced by the inorganic scintillator was transmitted along the optical fiber only if its angle with respect to the optical fiber axis was less than a given acceptance angle $\alpha$:

$$\alpha = \sin^{-1}\left(\frac{n_{c0}^2 - n_{cl}^2}{n_{c0}^2}\right)$$

where $n_{c0}$ and $n_{cl}$ are the refractive indices of PMMA fiber core and the new cladding, respectively. The new cladding is the mixture of epoxy resin, hardener and scintillation material. As can be seen, in the case of the previous design, at least half of the photons that conform to this condition spread out from the axis of the fiber as shown in Fig. 6(a). By contrast, all of the photons which adhere to the numerical aperture angle are transmitted along the PMMA fiber, because of the embedded structure the scintillation material in the core of the fiber as shown in Fig. 6(b).
ii) The mixture which consists of an epoxy resin and hardener makes the proportion of the inorganic scintillation relatively smaller, so fewer photons are produced. Another drawback of the previous design is that the scintillation material contains a rare earth element, which is potentially toxic. So the injection method dosimeter has to be subsequently further coated which increases the volume of the active material. This greatly decreases its use potential as an in-vivo device. Compared with the previous design, the novel dosimeter overcomes these drawbacks and exhibits excellent performance, because of its small and self-contained embedded structure.

![Fig. 6. The advantage of the new dosimeter: (a) Old type of dosimeter; (b) Novel structure dosimeter.](image)

![Fig. 7. Optical intensity for repeated exposures of 200MU at a dose rate of 600 MU/min at 6MV.](image)

3.2 Dosimeter repeatability

The novel dosimeter was tested in air to determine its reproducibility over five exposures with each exposure being 200 MU dose at a dose rate of 600 MU/min for 20 seconds with a photon beam energy of 6 MV. The data were recorded using a MPPC C11208-350 with 100 ms gate time. Figure 7 exhibits the excellent repeatability of the measurements over the five exposures. The peak intensity monitored during each on phase output remains consistent and stable. But the posterior exposure is a little smaller than the anteriority, because the Linac has a defined instability following extended use, and this would result in the Linac’s output being less than 600 MU/min. To maintain a constant delivered dose, the exposure time is therefore accordingly automatically extended by the Linac control system. The area under each curve during the beam’s active phase corresponds to the dose received during exposure, so integration of the optical intensity during the beam’s active phase yields the respective dose. The integrated intensity in this investigation is presented in Fig. 8 and the data for the five exposure test is shown in Table 1. Since the dose during each exposure is the same, the
integrated intensity of each exposure should also be the same. This can be seen in Table 1 with a maximum percentage error of 0.16%, which demonstrates excellent repeatability of the dosimeter in monitoring the radiation dose and is a value deemed amply suitable for in-vivo clinical use.

![Accumulated Dose](image)

Fig. 8. The accumulated dose measurement using the dosimeter derived from Fig. 7.

<table>
<thead>
<tr>
<th>Exp. No</th>
<th>Integrated Intensity</th>
<th>Percentage Error (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>11702736</td>
<td>0.14</td>
</tr>
<tr>
<td>2</td>
<td>11704001</td>
<td>0.16</td>
</tr>
<tr>
<td>3</td>
<td>11672463</td>
<td>-0.11</td>
</tr>
<tr>
<td>4</td>
<td>11678156</td>
<td>-0.064</td>
</tr>
<tr>
<td>5</td>
<td>11670939</td>
<td>-0.13</td>
</tr>
</tbody>
</table>

### 3.3 Dose linearity

Using the same experimental conditions described in section 3.1 but with the water depth changed to 2cm, the fiber and IC were irradiated at various doses using different dose rates: 100 MU/min, 200 MU/min, 300 MU/min, 400 MU/min, 500MU/min and 600 MU/min. Each exposure lasts 20 seconds. The optical intensity signal with the background (dark signal) subtracted for the varying dose rates is shown in Fig. 9. Figure 9 also shows Cerenkov signal measured by a blank fiber at 600 MU/min, and the result shows that the signal measured by the fiber-optic dosimeter is 735 times as large as the Cerenkov signal. The signal obtained in this way comprises the sum of the Cerenkov as well as pure fluorescence generated in the fibre material (PMMA). To date it has not been possible to ascertain which element is the major and minor contributor in this case as the signal has been too weak to detect using a Spectrometer or filtered detector. Therefore, it is possible to say that in the worst case scenario of Cerenkov comprising 100 percent of this signal that the light signal representing Cerenkov radiation is not a significant interference in our measurement.
From Fig. 9, it can be seen that increasing the dose rate caused an increase in the radiation incident of the dosimeter per unit time, resulting in a corresponding increase in the optical intensity. The measured optical intensity corresponding to a dose rate of 600 MU/min shows almost exactly double the amount of scintillation signal observed at 300 MU/min. The integral sum of the optical intensity for each iteration is directly related to the dose and since the dose was constant, the integral sum should also be a constant. As seen in Fig. 10, the integrated intensity increased stably at a dose rate of 600 MU/min when the beam was activated on and remained invariant when the beam was deactivated. The relationship between the integrated intensity and the dose measured using the IC is shown in Fig. 11. The integrated intensity increased linearly as the dose increased and a linear regression analysis yields the value of $R^2$ (shown in Fig. 11) of 0.9999, which is termed the coefficient of determination representing the accuracy of the match between the measured data and a linear fit. The value of R-square can be obtained from:

$$R^2 = 1 - \frac{\sum (y_i - f_i)^2}{\sum (y_i - \bar{y})^2}$$

where $y_i$ is the value of data set, $f_i$ is the modeled value, and $\bar{y}$ is the mean of the measured data.

It is apparent, therefore, that there is a strong linear relationship between the data obtained using the novel dosimeter and the real dose measured using the IC and the line crossed by the y axis at a count value of 52250. The latter expresses a fraction of the maximum count value of 52250/26791973 = 0.0019(0.19%), i.e. very close to zero.
3.4 Angular dependence

The variation in response of a dosimeter with the incident angle of radiation is known as the angular dependence of the dosimeter. Commercially available dosimeters usually exhibit angular dependence, due to the details of their construction, physical size and the energy of the incident radiation. Angular dependence is important in certain applications, for example in in-vivo dosimetry [19]. The angular dependence of the novel fiber-optic dosimeter was determined at a dose rate of 600 MU/min for 20 seconds with a 10 × 10 cm² field size of 6 MV. The experiment measured the dosimeter axial and radial directional dependence by holding the SSD of 100 cm and rotating the gantry from 0° to 360°. Figure 12 shows the radial angular dependence of the novel fiber dosimeter. This figure also shows the azimuthal angle between the dosimeter and the incident beam at intervals of 30°. The dosimeter dose at any azimuthal angle is normalized to that calculated at 90°. As can be seen, the response at any angle is nearly identical, i.e., the relative response varies at most by 1.15%. So the dosimeter’s response can be considered to be isotropic.
Fig. 12. The radar diagram for measurement radial angular dependence of dosimeter is shown, and how the azimuthal angle is defined too. The thick arrows with angular degrees indicate the direction of the incident radiation beams. For example, at 90°: (a) Radial angular dependence (b) Schematic diagram.

Fig. 13. The radar diagram for measurement axial angular dependence of dosimeter is shown, and how the azimuthal angle is defined too. The thick arrows with angular degrees indicate the direction of the incident radiation beams: (a) Axial angular dependence (b) Schematic diagram.

In the axial angular dependence test, it is known that the IC exhibits an isotropic response, and it is required that for measurement of ambient dose equivalents, the IC must be positioned to within 190° to 210° and 330° to 350° to the reference direction for calibration. The response of the IC will change severely at other angles, particularly at 90° where no response will be recorded [20]. In contrast to the curve for the IC, the novel dosimeter has a strong response at any sensing angle. The axial angular dependence of the dosimeter is shown in Fig. 13. The azimuthal angle between the dosimeter and the incident beam is also shown in this figure at sampling intervals of 45°. The response exhibits axial symmetry at about 90° and 270°, because of the symmetry of the dosimeter’s structure. The result for the axial angular dependence of the novel fiber dosimeter is shown in Table 2. The response of the device at
angles 45° to 135° is stronger than at other angles, because of the construction of the dosimeter. The inorganic scintillator forms a long cylinder. This structure may produce three possibilities for the axial angular dependence of the response: 1) the scintillator is 5 mm long, the depth of each point of the scintillator changes in this cylinder as the gantry rotation, so the angle between the dosimeter and the incident beam must produce a depth-dose response. 2) the visible light generated from the tail end of the scintillator cylinder may be scattered by the scintillator when it is transmitted along the cylinder, but the light produced at the head of the cylinder will be incident to the light guide and with little scattering. 3) the coupling efficiency of the light emerging from the end of the cylinder is greater than that at lateral surface, as validated by D. McCarthy in 2013 [21]. All of these justifications suggest that minimizing the thickness of the scintillator at the head of the dosimeter may reduce the axial angular dependence of the dosimeter’s response.

Table 2. The angular dependence of an oblique angle to dosimeter axis.

<table>
<thead>
<tr>
<th>Angle</th>
<th>0</th>
<th>45</th>
<th>90</th>
<th>135</th>
<th>180</th>
<th>225</th>
<th>270</th>
<th>315</th>
</tr>
</thead>
<tbody>
<tr>
<td>Normalized response</td>
<td>0.95</td>
<td>1.06</td>
<td>1.00</td>
<td>1.05</td>
<td>0.93</td>
<td>0.88</td>
<td>0.83</td>
<td>0.90</td>
</tr>
</tbody>
</table>

3.5 Depth-dose experiment

The dosimeter depth-dose measurement was performed over six exposures with various dose rates that were increased in steps of 100 MU/min from 100 MU/min to 600 MU/min at a photon energy of 6 MV in depth from 1 cm to 10 cm in steps of 1 cm. The results from these measurements were compared with simultaneous measurements recorded using the IC. Figure 14 shows the results of depth-dose profile for the 6 MV beam and the difference between the responses of the dosimeter and IC. It is widely acknowledged that, the $D_{\text{max}}$ is the depth at which the radiation dose reaches a maximum and is determined by the energy of the beam and the composition of the absorbing material. In the case of the experiment of this investigation, for a 6 MV beam, $D_{\text{max}}$ in water is about 1.5 cm. The absorbed dose measured by IC is identified as the standard value in radiotherapy applications, so using the IC the maximum point occurred at a depth of 1.5 cm, while the dosimeter reached a maximum at around 3 cm. Two reasons may explain the anomalous results measured by the novel dosimeter: The first is that the wavelength of the output X-rays will gradually increase with the increase of the distance of the transmitted beam as a result of the Compton Effect. The absorption efficiency of Gd$_2$O$_2$S:Tb which impacts the intensity of the visible light signal will also change with the wavelength of the X-rays input, which is manifest in the absorption spectrum of the Gd$_2$O$_2$S:Tb [22]. Consequently, the depth of the maximum point measured by the novel dosimeter should be different from that of the IC. The second explanation for these results centers on the fact that the terbium-doped gadolinium oxysulfide phosphor is not a water equivalent material. The density and atomic composition of the scintillator is quite different from the water. As a result, with low energy X-ray beams, the predominant mode of interaction in low-Z materials such as water is the Compton Effect, but for Gd$_2$O$_2$S:Tb, a high-Z material, the probability happened photoelectric effect will become more significant. Therefore, it can be expected that the dosimeter will act a quite differently than the IC in
Fig. 14. The normalized dose versus depth and dose rate: (a) measured on the dosimeter (b) measured on IC.

Fig. 15. The ratio between the dose deposited in the scintillator ($D_{\text{dosi}}$) to the dose deposited in water ($D_{\text{IC}}$) change with the depth.
depth-dose experiment. Despite this, the dosimeter exhibits a linear relationship between the tested value of fiber-optic dosimeter and the absorbed dose at each depth and each R-square in the experiment was greater than 0.9999. As the ratio between the radiation dose received by the scintillator ($D_{\text{dosi}}$) to the dose absorbed by water ($D_{\text{IC}}$) confirm, the depth-dose response measured by the dosimeter can be calibrated by multiplying by a simple correction coefficient (the correction coefficient is the reciprocal of the ratio defined above). Figure 15 shows the curve where this ratio is shown to increase monotonically with the depth.

4. Conclusion

Use of a novel ionizing radiation sensitive optical fiber dosimeter for real-time radiotherapy monitoring was investigated and the results have been presented. The detection principle is based on the use of a novel dosimeter which comprises an optical fiber with an embedded microhole positioned at its detection tip that was filled with a radiation sensitive scintillating material terbium-doped gadolinium oxysulfide (Gd$_2$O$_2$S:Tb). Experimental results have shown that this novel device exhibited a strong signal response and a high S/N ratio. The novel dosimeter was also tested for its reproducibility, linear response to radiation dose as well as angular dependence to the same. The results of these experiments showed that the novel dosimeter exhibited excellent repeatability, with a maximum percentage error of 0.16% and linearity (R$^2$ of 0.9999). The dosimeter can be considered to be isotropic with respect to the radial angular dependence of its signal. In a series of depth-dose experiments, it was found that the dosimeter responded in a different manner to a standard IC dosimeter. But the novel dosimeter can be easily calibrated by multiplying the output response by a correction coefficient. The experimental results strongly suggested that the novel fiber-optic dosimeter has great potential to be used as an in-vivo dosimeter for applications such as brachytherapy or intraoperative radiation therapy, where the accumulation of body fluids may cause a significant discrepancy between the prescribed radiation dose and the actual dose delivered to the tumor.

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