Optical fiber dosimeter for real-time in-vivo dose monitoring during LDR brachytherapy

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Abstract: An optical fiber sensor for monitoring low dose radiation is presented. The sensor, based on radiation sensitive scintillation material, terbium doped gadolinium oxysulphide (Gd2O2S:Tb), is embedded in a cavity of 700µm diameter within a 1mm plastic optical fiber. The sensor is compared with the treatment planning system for repeatability, angular dependency, distance and accumulated radiation activity. The sensor demonstrates a high sensitivity of 152 photon counts/Gy with a temporal resolution of 0.1 seconds, with the largest repeatability error of 4.1%, to 0.361mCi of Iodine-125 the radioactive source most commonly used in LDR brachytherapy for treating prostate cancer.

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1. Introduction

Prostate brachytherapy (BT) refers to the implantation of radioactive sources (or “seeds”) into the prostate under image guidance. Low Dose Rate (LDR) brachytherapy is more commonly used in the treatment of prostate cancer and therefore further discussion of brachytherapy within this article is limited to this modality. Prostate brachytherapy is typically performed in the course of 2 hours under spinal or general anaesthesia in an outpatient setting. Many patients tend to be in favour of this option as it typically takes only a few hours with little post-operative recovery [1].

Prior to the procedure, the patient undergoes a transrectal ultrasound or Computed Tomography (CT)/ Magnetic Resonance Imaging (MRI) based volume study. This information then allows the radiation oncologist to pre-plan the 3D seed distribution required to deliver the prescribed dose to the prostate. Seventy five percent of the Iodine-125 seeds are inserted into the peripheral region to avoid overdosing the urethra. In the operating room, patients are placed with their legs in an elevated position, a Foley catheter is inserted, and a template is positioned against the perineum. Ultrasound is the imaging modality utilised for the procedure to allow for the accurate insertion of the brachytherapy needles into the Planned Target Volume (PTV) with the aid of a template. Real-time treatment planning guides the radiation oncologist on where is best to locate the seeds in terms of proximity to organs at risk (OARs) [1]. These seeds are usually Iodine-125, with apparent activities of the individual seeds typically ranging from 0.357mCi to 0.42mCi, and have a half-life of 59.43 days. They measure approximately 4.5 × 0.8 mm, and generally 60 to 100 seeds may be placed, depending on the size of the prostate.

The International Commission on Radiological Protection (ICRP) Report Nos. 86 [2] and 97 [3] as well as International Atomic Energy Agency (IAEA) safety report series 17 [4] describe errors that have occurred in brachytherapy as well as databases, such as ROSIS [5], SAFRON [6] and ACCIRAD [7]. As demonstrated through phantom and computer-simulated error scenarios [8–10], several of the reported BT error types could have been detected with real-time in-vivo dosimetry. Human errors contribute to the majority of mistakes and accidents reported from brachytherapy, although a few are related to technical issues [8]. This can be typically incorrect patient demographics, seed radiation activity, prescribed total dose or simple software issues [8].
Incorrect insertion of the seeds may also occur as it is a manual process and there is also the probability of seed migration post procedure [8].

Uncertainties in the detector position may generate substantial dose uncertainties. It has been reported that positional uncertainties <1.5 mm for the dosimeter are required in order to be able to detect source displacement errors of 5 mm [9,10]. Poor image quality from ultrasound, alongside patient organ movement throughout the procedure, can result in it being very difficult to maintain a positional accuracy of <1.5 mm [10]. An important factor in identifying potential errors with in-vivo dosimetry (IVD) is the ability to place the sensor in a good location alongside the capability to capture this position [8]. Present Treatment Planning Systems (TPS) incorporate the American Association of Physicists in Medicine Task Group (TG)-43 dose calculation protocol [11]. This does not take into account tissue heterogeneities by assuming the patient is made of water. TPS rely on the correct inputted information detail, such as seed radiation activity and the appropriate dose calculation protocol to be used. Variation of the delineation of PTV and OARs are also created by different users of the system leading to further uncertainties [12]. Shortcomings of the TG-43 dose calculation protocol [11] are targeted by developments of model-based dose calculation methods, which in contrast to the TG-43, account for tissue heterogeneities and individualized patient anatomy [13].

The inherent properties of optical fibers, e.g. lightweight, small dimensions, in addition to the water equivalence of PMMA, make them suitable for in vivo radiotherapy applications [14]. This paper presents a radiotherapy dosimeter based on optical fiber sensor technology, offering the unique ability to directly measure the radiation dose administered to the tumour and surrounding tissues. The measured radiation dose can be used to verify the calculated dose distribution that describes the treatment received by the patient. Live measurements during the brachytherapy procedure can allow optimisation during the procedure and result in high quality treatments. The quality of a brachytherapy treatment is directly linked to patient survival and quality of life [15]. Previous work by the authors has demonstrated the applicability of radioluminesence for external beam radiotherapy monitoring, whereby plastic optical fibers are coated with the scintillator [16]. This paper presents recent advancements in the development of an optical fiber radioluminescence sensor developed by Woulfe et al. [17,18], whereby the radiation sensitive scintillator, terbium-doped gadolinium oxysulphide (Gd2O2S:Tb), is embedded within the core of a 1 mm PMMA (Polymethyl methacrylate) plastic optical fiber. The sensor is further characterised for a range of parameters, including angular dependence and distance from the source. The results are further evaluated against theoretical dose calculations performed using TG-43 formulism to independently verify the treatment planning system dose projections. This work has expanded on the verification methods used to validate the sensor and its potential clinical application in brachytherapy.

2. Sensor fabrication

The optical fiber sensor, shown in Fig. 1, is constructed by micromachining a cavity in the 1 mm core of a PMMA (polymethyl methacrylate) plastic optical fiber [17]. The cavity, 700 μm in diameter and 7 mm deep, is filled with a scintillating material, terbium doped gadolinium oxysulphide (Gd2O2S:Tb, GOS) by inserting phosphor into an assembly, as shown in Fig. 2, and sealed with an epoxy. This scintillation material was chosen based on the previously reported work [19] that demonstrates the high light yield of Gd2O2S:Tb to the brachytherapy seeds. The Gd2O2S:Tb type UKL65/F-R1, supplied by Phosphor Technology Ltd, UK has a green emission colour with a median particle size of 3.5 μm. On exposure to ionising radiation, it fluoresces at the primary peak wavelength of 545 nm, with two smaller peaks observed at 490 nm and 590 nm. The resultant emitted fluorescent light penetrates the PMMA optical fiber core and propagates along the fiber to a Hamamatsu Multi-Pixel Photon Counting Module (MPPC) C13366 [20] for monitoring of the optical signal.
The small dimensions of the sensor, with an overall outer diameter of less than 1 mm, mean that they can be guided within existing brachytherapy equipment, for example within the brachytherapy needle, as shown in Fig. 3. This will allow the sensor to be located directly within the prostate, using techniques the radiation oncologist is already familiar with. The optical fiber sensor can also be inserted through the transrectal biopsy guide of the ultrasound probe enabling dosimetry feedback of dose to the surface of the rectal wall. This will provide real-time information of the radiation dose received by the tumour and to the surrounding organs at risk (OARs).
3. Results and discussion

The optical fiber sensor was fixed within a novel prostate phantom, developed in-house using water equivalent RW3 material (polystyrene with titanium oxide admixture), to allow for comparison of results with the TG-43 Dose Calculation [21]. The TG-43 protocol has resulted in significant improvements in the standardisation of both dose-calculation methodologies as well as dose rate distributions used for clinical implementation of brachytherapy. The general, two-dimensional (2D) dose-rate equation (Eq. (1)) from the 1995 TG-43 [21] protocol is retained,

\[ D(r) = S_k \cdot \Lambda \cdot \frac{G_L(r, \theta_0)}{G_L(r_0, \theta_0)} \cdot g_L(r) \cdot \phi_m(r) \]

where \( r \) denotes the distance (cm) from the centre of the active source to the point of interest, \( r_0 \) denotes the reference distance which is specified to be 1 cm in this protocol, and \( \theta \) denotes the polar angle specifying the point of interest, \( P(r, \theta) \), relative to the source longitudinal axis. \( S_k \) defines Air Kerma Strength, \( \Lambda \) is the dose rate constant in water, \( G \) is Geometry function, \( g_L(r) \) is the radial dose function while \( \phi_m(r) \) is 1D Anisotropy function. The reference angle, \( \theta_0 \), defines the source transverse plane [22], specified to be 90° as per Fig. 4 [23]. \( L \) is defined as the active length of the source and \( t \) is the thickness of outer capsule surrounding source.

![Fig. 4. Coordinate system used for brachytherapy dosimetry calculations [23].](image1)

The experimental set-up can be seen in Fig. 5. The brachytherapy seeds were inserted into the prostate phantom for a fixed period of time and the response of the sensor monitored. The sensor was initially tested for its response to 0.361 mCi of Iodine-125, for a comparison of optical signal.

![Fig. 5. Optical fiber sensor within brachytherapy needle and located within the novel prostate phantom.](image2)
3.1. Repeatability

A single Iodine-125 seed, with a total activity of 0.361mCi, is inserted into the prostate phantom and removed over three consecutive cycles to determine the stability and repeatability of the sensor’s response. Due to health and safety in handling the radioactive seeds, the test was limited to three iterations. Figure 6 shows the fluorescent signal for each of the three iterations. As the brachytherapy seeds are inserted into the prostate phantom, there is a clear increase in photon counts. A seed is inserted for a set period of time and then removed, as highlighted over three iterations. The first measurement results show an average photon count of 4920, the second 5299 and finally the third 5107. The average photon count for all 3 repeated measurements was 5130. The position of the seed at coordinates \( x: 0.56\ \text{cm}, \ y: 12.14\ \text{cm} \) and \( z: 9.6\ \text{cm} \) which is 5 mm to the right of sensor equate to a dose of 34.67Gy, according to Eq. (1). For the remainder of the paper, this point was used as the reference point for all measurements. The resultant percentage variation from the mean was -4.1%, 3.3% and -0.45% for the three measurement cycles respectively with a signal variation ranging from 0.56% to 2.4% of the signal. The maximum percentage variation of 4.1% is within current acceptable standards for brachytherapy dosimetry, although it is acknowledged that a more in depth analysis is required with additional measurement sets.

![Fig. 6. Repeatability of 0.361mCi Seed with Gd_{2}O_{2}S:Tb Dosimeter.](image)

3.2. Angular dependency

The angular dependency of the dosimeter was investigated by monitoring the variation in the optical signal at different angles to the dosimeter. Four angles were evaluated: 0, 90, 180 and 270 degrees, each located 0.5 cm from the centre position where the sensor was positioned. These are depicted as 0.5 cm Top (1T), 0.5 cm Right (1R), 0.5 cm Bottom (1B) and 0.5 cm Left (1L). Figure 7 shows the noticeable difference in the response of the GOS at different angles. In terms of photon counts, there is slight decrease of -1.6% for 1R and -5.5% for 1L, and an increase of 5.3% for 1T and 6% for 1B. From above, we can also note the repeatability variation of 4.1%, which potentially equates to a maximum error of 1.9% for angular dependency. This error may
be due to the filling process of the phosphor and the determination of the correct volume of phosphor uniformly distributed within the cavity. High intensity imaging may help to verify this process.

3.3. Distance

The response of the dosimeter with respect to the distance from the radiation seed was also investigated. This was monitored in 0.5 cm steps up to 3 cm distance from the seed, whereby the sensor was placed in the centre of the phantom and the radiation seed was moved in 0.5 cm steps linearly away from the centre. Figure 8 shows that the sensor is capable of detecting activity up to a distance of 3 cm for an activity of 0.361 mCi. 3 cm is the distance required to achieve adequate dosimetry coverage of an average sized prostate gland by having the sensor positioned within the centre of the prostate gland. As can be seen, the response at 0.5 cm is 5498 photon counts decreasing down to 154 photon counts for 3 cm, which is just above background. Any further distance will not be detected adequately. Figure 8(b) shows the sensor’s response aligns well with the theory in demonstrating the radiation’s dependence on the inverse-square law. Further investigation will be performed in an anthropomorphic prostate simulation phantom.

3.4. Accumulated radiation activity

To monitor the sensor’s response to accumulated radiation activity, the sensor is placed in the centre grid position; eight seeds are then inserted in various positions around the sensor. The radio-active sources remain in the prostate phantom at a particular grid position at a depth of 4 cm, as each additional seed is added to the next grid position. The associated measured photon count for a single seed equates to 152 for 1Gy of I-125. The expected point dose as each seed is added to the respective grid positions is calculated from TG43. This is also used within the Treatment Planning System Variseed (Fig. 9). Figure 10 shows the increase in the sensor’s optical signal as each seed is introduced to the phantom.

The apparent activity of brachytherapy seeds, directly relate to the radiation dose the cancer patient will receive during treatment. As the radiation activity increases, there is a discrete and measurable difference in the response of the sensor to the different activity levels of the iodine
Fig. 8. (a) Sensor response at different distances from the radiation source (b) Photon counts as a function of the distance from the source, demonstrating the inverse-square relationship.
Fig. 9. Treatment Planning System (TPS) indicating Point Dose Expected TPS Values (Gy).

Fig. 10. Response of the Optical Fiber Sensor to I-125 seeds ranging from 0.361mCi up to 2.888mCi.

seeds with an increase in photon counts as shown in Fig. 10. The results demonstrate the sensor’s capability of monitoring the placement of individual radiation seeds during LDR brachytherapy, in addition to determining the overall radiation dose. A comparison between the measured dose from the optical fiber sensor and the calculated TPS values is shown in Fig. 11. The dosimeter is in excellent agreement with the expected dose with a maximum error of 4.13% at a seed activity
of 0.361mCi. There is also a good linear response from the dosimeter to the expected dose, with a linear regression of 0.9994.

Fig. 11. Comparison of Dosimeter to TG 43.

4. Discussion and conclusion

An optical fiber sensor, based on radioluminescence, is presented for monitoring the radiation dose a cancer patient receives during seed implantation in brachytherapy. A cavity, 700µm in diameter and 7 mm deep, is micro-machined in the 1 mm core of a PMMA plastic optical fiber and is filled with a radiation sensitive scintillation material. Future work will be focusing on optimisation of the output signal based on the cavity’s dimension and shape. The sensor demonstrates a clear and measurable increase in photon counts when exposed to a single brachytherapy seed and exhibits a linear relationship with increasing seeds. The sensor also demonstrates good repeatability, with a maximum percentage error of 4.13% for 8 seeds when compared to the expected result from the treatment planning system calculation algorithm, TG43. The sensor design is minimally invasive, allowing the sensors to be placed within existing brachytherapy equipment. With the assistance of live feedback of the radiation dose to the planned target volume (PTV) and the organs at risk (OARs) from the sensors, the radiation oncologist is able to ensure accurate placement of the radio-active sources. This will lead to the ultimate goal of improving patient outcomes.

Funding

Royal Historical Society (UF150618); Science Foundation Ireland (18/TIDA/6119).

Acknowledgements

The authors would also like to thank all the staff at the Radiation Physics and Radiation Oncology Departments at the Galway Clinic for their assistance with this work. Dr. O’Keeffe would like to acknowledge the support of the Royal Society and Science Foundation Ireland (SFI), through the
Disclosures
The authors declare no conflicts of interest

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