BJR

Received:
21 October 2014

Revised:
19 February 2015

Accepted:
10 March 2015

doi: 10.1259/bjr.20140702

Cite this article as:
O'Keeffe S, McCarthy D, Woulfe P, Grattan MWD, Hounsell AR, Sporea D, et al. A review of recent advances in optical fibre sensors for *in vivo* dosimetry during radiotherapy. Br J Radiol 2015;88:20140702.

REVIEW ARTICLE

A review of recent advances in optical fibre sensors for *in vivo* dosimetry during radiotherapy

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ABSTRACT

This article presents an overview of the recent developments and requirements in radiotherapy dosimetry, with particular emphasis on the development of optical fibre dosemeters for radiotherapy applications, focusing particularly on *in vivo* applications. Optical fibres offer considerable advantages over conventional techniques for radiotherapy dosimetry, owing to their small size, immunity to electromagnetic interferences, and suitability for remote monitoring and multiplexing. The small dimensions of optical fibre-based dosemeters, together with being lightweight and flexible, mean that they are minimally invasive and thus particularly suited to *in vivo* dosimetry. This means that the sensor can be placed directly inside a patient, for example, for brachytherapy treatments, the optical fibres could be placed in the tumour itself or into nearby critical tissues requiring monitoring, via the same applicators or needles used for the treatment delivery thereby providing real-time dosimetric information. The article outlines the principal sensor design systems along with some of the main strengths and weaknesses associated with the development of these techniques. The successful demonstration of these sensors in a range of different clinical environments is also presented.

Radiation dosimetry deals with methods for quantitative determination of energy deposited in a given medium directly or indirectly by ionizing radiations. A dosemeter can be defined generally as any device that is capable of providing a reading that is a measure of the average absorbed dose deposited in its (dosemeter) sensitive volume by ionizing radiation. There are commonly agreed codes of practice in the UK that define how dosimetric calibration of treatment beams should be performed,[1,2] in addition to defining the types of ionization chambers that may, or may not, be used for these measurements.

Radiotherapy is in a period of rapid scientific and clinical development. The introduction of novel treatment techniques, for example, stereotactic ablative radiotherapy and volumetric modulated arc therapy (VMAT), delivered through the use of technologies such as flattening filter free (FFF) beams and dynamic multi-leaf collimation, is driving the requirement for increasing levels of accuracy and precision in dosimetry. These treatment delivery options are causing existing, well-established, dosimetric equipment to be extended to the limits of its capability. Other recent developments in treatment options include the use of protons and heavy ions, and the availability of small animal irradiation platforms provides additional scope for novel dosimetric systems. Furthermore, the increased use of image-guided radiation therapy, including the use of kilovoltage cone beam CT, MR and positron emission tomography, provides a different set of problems to existing technologies employed within traditional radiotherapy dosimetry.

There are also recommendations for comprehensive quality assurance (QA) programmes[3,4] to assess the performance of all types of radiotherapy treatment equipment, including the treatment planning system (TPS), against known tolerances and for comparison with baseline measurements. The increasing complexity of modern treatment modalities has also introduced a more comprehensive patient-specific QA programme to verify, pre-treatment, an individual patient delivery. Radiotherapy also includes multiple layers of checking from the simple cross-checking of work, through independent monitor unit calculations, and on to

independent audits of treatment centres' planning and dosimetry performance.

In addition to ensuring the correct calibration of treatment beams, and verification of the delivery pre-treatment, it is important to monitor dose delivery during treatment (*in vivo*), rather than verification of the treatment to a phantom. In an ideal scenario, the dose delivered directly within the tumour volume, and/or dose to specific organs at risk (OARs), would be measured while the patient is receiving their treatment. However, this is currently generally carried out by measuring the dose at a "surrogate" position, usually by placing a radiation detector directly on, or near to, the patient's skin surface to provide either an entrance or exit dose value, rather than directly within the tumour itself. There is a growing interest in the need to perform such *in vivo* measurements in part owing to increasing awareness of the potential risks associated with incorrect delivery or planning of radiation treatments, and because of the use of increasing complex delivery techniques such as intensity-modulated radiation therapy (IMRT) and VMAT, and the move towards more hypofractionated treatments delivered with large doses per fraction.

The importance of *in vivo* dosimetry has been further highlighted in recent years as a result of a number of major radiotherapy incidents,[5–7] and whilst the vast majority of radiotherapy sessions are performed without incident, an international review of radiotherapy incidents identified >7000 incidents over three decades (1976–2007). The incidents range from underdosing, leading to a recurrence risk, to overdosing, causing toxicity and even death.[5] The investigations following major incidents have generally recommended that some form of *in vivo* dosimetry measurement would be beneficial,[8] and professional bodies such as the American Association of Physicists in Medicine have recommended that clinics "should have access to TLD or other *in vivo* systems".[9]

There are a number of different options available for use as an *in vivo* dosemeter, with the most commonly used being thermoluminescent detectors (TLDs), diodes, metal-oxide semiconductor field effect transistors (MOSFET), film and electronic portal imaging devices. These options each have relative strengths and weaknesses, and a number of review articles[10–12] have highlighted the merits of each. For a detailed summary of *in vivo* dosemeters, not restricted to optical fibre sensors, see table 1 from Mijnheer et al[10] for dosemeters in external beam radiotherapy and table 3 from Tanderup et al[12] for dosemeters in brachytherapy. Methods to infer the full three-dimensional dose distribution are also being developed primarily by the use of back-projected electronic portal imaging images to reconstruct the dose within the CT volume used to plan the patient's treatment[13–15] or through the analysis of the treatment log files to recreate the multileaf collimator (MLC) positions used during the treatment.[16,17]

In recent years, there has been some interest[10] in investigating alternatives to the established *in vivo* detectors, such as plastic scintillation detectors (PSDs), optically stimulated luminescent detectors, radiophotoluminescent dosemeters and implantable MOSFETs. Some of the main reasons for the development of these alternatives to the existing options include the increasing interest in combining a MRI scanner with a radiotherapy linear accelerator (linac), the development of heavy ion and particle beams in radiotherapy and the introduction of new small animal irradiation platforms for radiobiological investigations. These new technologies present different problems from the effect of magnetic fields on dosemeters,[18] the response of dosemeters in different types of treatment beams,[19,20] the miniaturization of treatment fields[21,22] and the associated complexity of radiation dosimetry at very small field sizes.[23,24]

Optical fibres offer a solution for *in vivo* radiotherapy dosimetry with many advantages over currently employed clinical dosimetry systems. An optical fibre radiation dosemeter is a photonic system based on optical fibre technology, whereby radiation introduces a modification or modulation in some of the characteristics of the optical signal. The optical fibres can be directly affected by the radiation, in which case it is called an intrinsic sensor, or it can be used for the sole purpose of transmitting the optical signal, and is known as an extrinsic sensor. There are a number of different dosimetry techniques that can incorporate optical fibres to further improve the overall system, and these techniques are discussed in turn, together with examples of such optical fibre-based systems.

PROPERTIES OF RADIOTHERAPY DOSEMETERS

Stability and repeatability of the dosemeter, as with any sensor, are important. In order to ensure that this can be achieved, the dosemeter must be immune to a number of environmental conditions and external interferences. The effect of humidity and temperature on the dosemeter must be known and at the least must be compensated for through the use of correction factors.

Ideally, the dosemeter should exhibit a linear response with dose over the range of interest in radiotherapy (typically on the order of milligray or centigray up to 20 Gy). It should also demonstrate independence to dose rate, which is increasingly relevant given the much higher dose rates available clinically since the introduction of FFF beams. The measured response for the overall dose should remain constant regardless of the dose rate applied.

The energy dependence of a dosemeter is an important factor in developing dosimetry systems for radiotherapy. Ideally, the energy response should be independent of the radiation treatment energy used. To achieve this, the dosemeter must be of a material close to that of the medium in which the absorbed dose is measured. Consequently, the dosimetric material should have an atomic number (Z_{eff}) close to that of water ($Z = 7.4$). Materials with a high effective Z number will demonstrate high-energy dependence, particularly at lower energies, which must be considered carefully in measurements.

The dosemeter should also exhibit small directional, or angular, dependence. As the angle of incidence of the radiation can vary considerably from one measurement to the next, assessing any potential angular dependence is particularly important for *in vivo* dosimetry.

The dosemeter should preferably be small in size, allowing for high spatial resolution, and be minimally invasive. They should be robust, easy to handle and not affected by the radiation itself, reducing the need for more frequent recalibration of the detector response. For clinical applications, it is important that the dosemeters are easy to use. For this to be achieved, the system must be easily installed in the area of application without the risk of affecting the measurements. Maintenance should also be at a minimum. In certain applications, it is also important that the readout from the dosemeter is available immediately, while in others, a signal may be readout at a later stage—in which case, the response should be stable and not subject to signal decay or modification post irradiation. A further useful feature can be the option to measure detector response on a beam-by-beam basis, rather than a single cumulative signal arising from the summation of all components of the treatment delivery.

OPTICAL FIBRE SENSOR TECHNOLOGY

Optical fibre sensors offer numerous advantages over conventional dosemeters, such as TLDs and diodes. The most significant feature of an optical fibre dosemeter is that the dose information is transmitted using optical signals as opposed to electrical signals. Consequently, optical fibres are immune to electrical and electromagnetic interferences, which can be a problem for many electronic dosemeters. Electric dosimetry systems, such as ionization chambers, often require a high voltage power supply and must have good electrical insulation.

The small dimensions of optical fibres (typically from 200-μm to 1-mm core diameter) generally allow for excellent spatial resolution of the optical fibre dosemeter. The small size, in addition to being lightweight and flexible, makes such dosemeters suitable for minimally invasive *in vivo* applications. This would allow the radiation dosemeter the potential to be placed either directly into or in close proximity to the tumour, or in the case of a brachytherapy implant alongside the seeds, or radioactive sources, to provide real-time dosimetric information, *e.g.* in close proximity to the implants in the tumour itself or critical tissues requiring monitoring.

The ability to remotely monitor radiation is also an advantage of optical fibres. This has allowed the development of sensors for use in harsh environments, where conditions may make it inappropriate, unsafe or impractical for an individual to be stationed to monitor a parameter of interest.[25–27] There has been interest in the development of sensors for remote monitoring of high levels of radiation and in the deployment of sensors for environmental monitoring of structures, including radiation-hardened sensors for application in the nuclear industry.[28] The ability to locate a sensor several metres from the control electronics means that they can equally be employed for dosimetry applications for the monitoring of radiotherapy treatments, where the console areas outside the treatment rooms can be up to 20 m away from the position where the treatment is being delivered.

Optical fibre sensors can also be multiplexed so that a single controller can monitor a number of sensors. The development of radiation-resistant fibres has also meant that fibre optics can be used within systems whereby they are used solely to transmit the signal through areas of high levels of radiation. This is particularly useful for *in vivo* probes for radiotherapy, which allow for minimally invasive, real-time monitoring of the radiation dose.

A further potential advantage of these sensors is that optical fibres generally comprise only silica (glass) or plastic as their constituent material and therefore are uniquely, and ideally, suited for use in the MRI environment, as they are non-magnetic and do not cause interference on the image and are themselves immune to the intense magnetic field and radio frequency pulses present in the MRI environment. This presents a significant advantage over more traditional dosimetry equipment (*e.g.* ionization chambers, diodes) in allowing accurate dosimetric verification of new technologies that are investigating the potential of combining a MR scanner with a radiotherapy linac.[29,30]

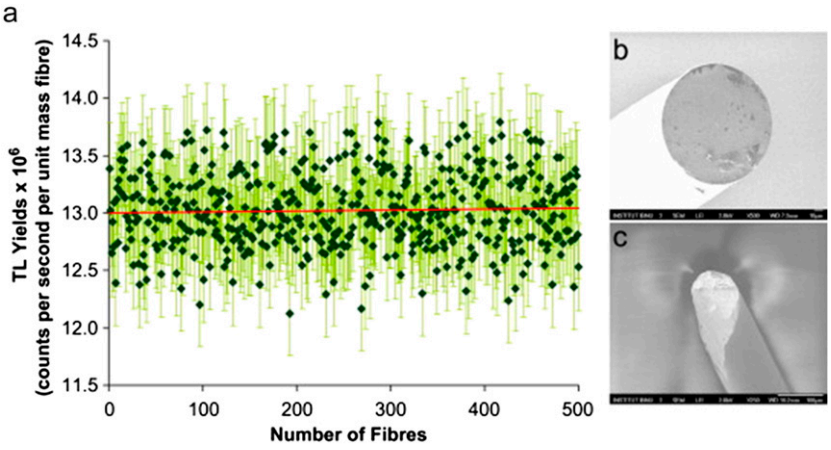
Luminescence dosemeters

Luminescence occurs when a material, having been subjected to radiation, absorbs some of the radiation and as a result emits light with a different wavelength. It is therefore possible, with an understanding of the limitations of performance alongside careful application of appropriate correction factors, to relate the amount of light emitted to the initial dose of radiation received by the material. Different types and forms of radiation can be used to excite a material, and it is these types of radiation that give rise to different types or methods of luminescence. The types predominantly used in radiotherapy dosimetry are thermoluminescence (TL; excitation owing to heat), photoluminescence (PL; excitation is owing to optical or ultraviolet light) and radioluminescence (RL; excitation is owing to alpha, beta or gamma rays, or X-rays).

Thermoluminescent dosemeters

Commercially available germanium (Ge)-doped silica (SiO_2)-type optical fibres have been demonstrated as potential novel forms of TLDs for radiation therapy dosimetry.[31,32] Ge-doped silica optical fibres have been shown to have a higher TL yield than do pure silica fibres or silica fibres doped with other elements, *e.g.* aluminium (Al).[33] Abdul Rahman et al[34] investigated the ability of high spatial resolution (approximately 120 μm) Ge-doped SiO_2 TLDs to measure radiation (Figure 1). At fixed dose rates, the optical fibres gave a flat thermoluminescent yield within 4% and 3% for electron and photon beams, respectively, while demonstrating good reproducibility (±1.5%). They also demonstrated negligible post-irradiation fading and good reusability (0.5%) following thermal annealing.[35] The use of Ge-doped silica fibres has also been proposed by Issa et al[36] for use in *in vivo* brachytherapy applications, while a review article by Bradley et al[37] had summarized the potential radiotherapy dosimetric applications for doped silica fibres as TLDs. Palmer et al[38] compared a Ge-doped SiO_2 TLD fibre, EBT3 Gafchromic® film (International Specialty Products, Wayne, NJ) and Presage® radiochromic material (Heuris Pharma LLC, Skillman, NJ) against a Monte Carlo model for measuring the radial dose distribution of a Cobalt-60 high-dose-rate (HDR) brachytherapy source. They also compared their response against a TPS

Figure 1. (a) Thermoluminescence (TL) yield distribution from selected germanium (Ge)-doped silica (SiO_2) optical fibres irradiated at the nominal photon energy of 6 MV for a fixed dose of 3 Gy (delivered at the rate of 400 cGy min^{-1}). The line indicates the mean value. (b) Scanning electron microscopy image of Ge-doped SiO_2 optical fibres cleaved using an optical fibre cleaver, compared with that cut using a sharp blade (c). Reproduced from Abdul Rahman et al[34] with permission from Elsevier.



calculation of the dose distribution for a typical cervix treatment. They reported that while the fibre presented excellent spatial resolution for measurement in a single direction, its physical size limited the accuracy for complex dose distribution assessment. Although these are highly promising technological developments in the area of optical fibre sensors, they are all based on TL technology. Therefore, a significant drawback to this technique is that owing to the requirement to stimulate the material to produce a signal sometime after irradiation, it cannot deliver real-time dosimetry.

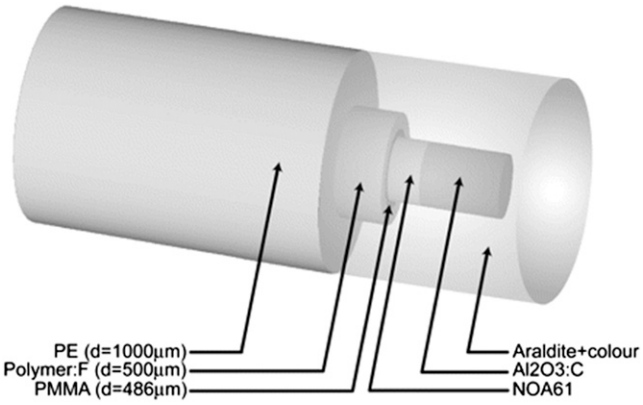
Optically stimulated luminescence

In a similar process to thermoluminescent techniques, PL, or optically stimulated luminescence (OSL), emits the energy stored owing to irradiation, upon exposure to light. Carbon-doped aluminium oxide ($Al_2O_3:C$) is the most prevalent of OSL crystals researched within the medical and personal dosimetry fields. It demonstrates a linear response to doses up to 50 Gy, although there are reports of supralinearity >2 Gy for doses from a 6-MV photon beam. It is stable at room temperature, with negligible fading over a period of 85 days, and its response when varying beam parameters such as dose rate, field size and irradiation direction was found to be excellent.[39] However, it does exhibit a change in sensitivity with irradiation history introducing some uncertainty for radiotherapy dosimetry. Yukihara and McKeever[40] provide a detailed overview of the fundamental theory and practical aspects of OSL systems for medical applications.

Optical fibres pose a clear advantage in the development of OSL dosemeters as they can be used to transmit both the stimulating optical source, typically a laser source, to the OSL material and the luminenscence signal back to the detector, allowing for remote analysis of the OSL response. Optical fibre-based OSL dosemeters are commonly used in conjunction with RL to provide a second dose measurement and to overcome errors caused by undesirable external luminescence, *e.g.* Cherenkov radiation. Marckmann et al[41] couple an $Al_2O_3:C$ crystal to the end of a 500-μm polymethylmethacrylate (PMMA) plastic optical fibre, shown in Figure 2, to provide simultaneous RL/OSL measurements giving real-time dose and dose rate information using RL and further verification of dose post-irradiation using OSL. A green laser [frequency-doubled yttrium aluminium garnet (YAG) laser, 532 nm, 20 mW], used to stimulate the OSL, is transmitted by the optical fibre to the $Al_2O_3:C$ crystal and the resulting light is transmitted back along the fibre to a photomultiplier tube (PMT). The use of an Al_2O_3-OSL fibre sensor has been examined for *in vivo* dosimetry.[42] A further phantom-based study has indicated the suitability of fibre-coupled $Al_2O_3:C$ for brachytherapy with an iridium-192 radiation source.[43]

Gaza and McKeever[44] investigated the use of single-crystal europium-doped potassium bromide (KBr:Eu) for OSL dosimetry using optical fibres. The fibre used in this work exhibited temperature dependence, and the OSL signal was also affected by fading. However, the authors demonstrate the use of optical fibres to allow for quasi real-time analysis of the OSL

Figure 2. Schematic diagram of a polymethylmethacrylate (PMMA) optical fibre optically stimulated luminescence dosemeter based on carbon-doped aluminium oxide ($Al_2O_3:C$). PE indicates the polyethylene sheath and NOA61 the Norland Optical Adhesive 61. Reproduced from Marckmann et al[41] with permission from Oxford University Press.



signal, which can overcome the aforementioned issues. The stimulating laser light is turned on periodically, for 20 ms duration, allowing the OSL signal to be monitored. It is then followed by a 20-ms laser-off period. KBr : Eu has a fast OSL decay time making it suitable for these high-speed OSL measurements.

In a further development to the work on TL of SiO_2 optical fibre,[31–37] their use as an OSL material has also been investigated.[45,46] With these intrinsic sensors, the optical fibres would act as both the sensing and the light-guiding components. While commercially available SiO_2 and Ge-doped SiO_2 optical fibres exhibit a measurable OSL, all work to date has focused on high radiation doses (7–147 Gy) and so their suitability for radiotherapy applications has yet to be examined. Fluoride phosphate optical fibres have been shown to be effective OSL-based optical fibre intrinsic dosemeters for low-level radiation applications. A linear response between 0.16 and 2.00 Gy was observed, with a saturation of the material at 2 Gy. Detection of higher radiation doses is possible if a number of fibres are bundled (*e.g.* six fibres for 8 Gy).[47] However, this would make the sensor relatively bulky for *in vivo* radiotherapy dosimetry.

Radioluminescence

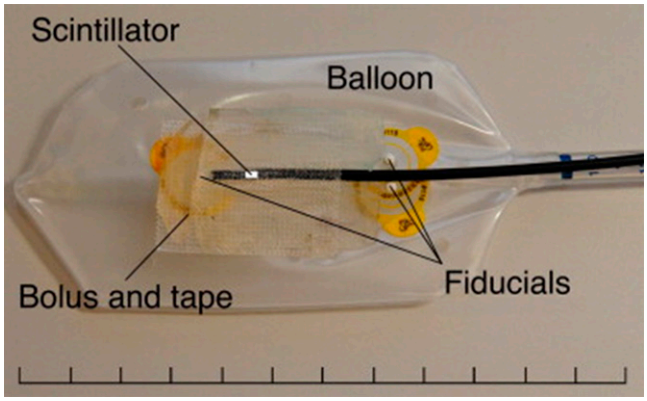
RL, or scintillation, detectors are based on the phenomenon that the material used is capable of converting ionizing radiation into detectable light. A photodiode or PMT subsequently converts it into an electrical signal. Scintillators can be subdivided into two broad groups, namely inorganic and organic. Inorganic scintillators are typically in crystal form and made of alkali halides [*e.g.* caesium iodide (CsI), sodium iodide (NaI)] or oxides (*e.g.* bismuth germanate).[48,49] Scintillation is owing to the crystalline structures that create the energy bands between which transitions of electrons take place. While some crystals scintillate intrinsically, others need activators to allow for scintillation in the visible region. Thallium (Tl) is an example of an activator and is used in one of the most common inorganic scintillators, NaI(Tl).[48,49] Organic scintillators are composed of aromatic hydrocarbons and can be divided further into plastic and liquid scintillators. Organic scintillators scintillate at a molecular level, which means that each scintillator molecule can act as a scintillator centre.[48,49]

Plastic scintillation-based optical fibre dosemeters One of the most widely investigated types of scintillation dosemeters involves the coupling of a plastic scintillator to the tip of an optical fibre probe.[50–52] PSDs offer significant advantages in radiotherapy dosimetry owing to their water equivalence, linearity with dose and energy independence in the megavoltage energy range.[12] When coupled with plastic optical fibres, the dosemeters are completely water equivalent and so they will not interfere with the energy deposition process. However, recent reports on temperature dependence of PSDs will require further evaluation for its impact on radiotherapy applications. One study investigated the effect of temperature on a fibre-coupled organic plastic scintillator that was designed for use in the dosimetry of brachytherapy sources and megavoltage photon beams used in external beam radiotherapy. The study showed that the light yield in the peak regions of the scintillators decreases linearly with increasing temperature. For the blue BCF-12 and the green BCF-60 (Saint-Gobain Crystals, Nemours Cedex, France) temperature coefficients of -0.15 ± 0.01 and $-0.55 \pm 0.04\%\,K^{-1}$, respectively, were observed.[53] These values show that careful consideration should be taken for clinical measurements. A group at the University of Texas MD Anderson Cancer Center (Houston, TX) has also reported significant differences between measurements from detectors inserted inside the patients and those in anthropomorphic phantoms, owing to the temperature dependence of the PSDs.[54]

The plastic scintillating fibres produced by Saint-Gobain Crystals[49] are widely reported for use in radiotherapy applications. These plastic scintillating optical fibres, with fibre diameters ranging from 0.25 to 5.00 mm (typically 1-mm core fibres are used), have a polystyrene core with fluorescent dopants and a PMMA cladding. The trapping efficiency of the fibres is such that between 3% and 7% of the photons are collected for transmission down the fibre. Klein et al[55] presented work using one of these plastic scintillating fibres (BCF-60) mounted onto an endorectal balloon (Figure 3) and positioned within a phantom to verify the dose delivered from sample IMRT and VMAT treatment plans for prostate cancer. The system measured total doses that correlated well with ionization chamber measurements, and it was found that the values were within 1% of the TPS calculations. The results demonstrate the potential that this system may present for real-time dosimetry measurements of patients with prostate cancer undergoing external beam radiotherapy.

The BrachyFOD™ developed by the University of Sydney NSW, Australia, uses the commercially available BC-400 scintillator, from Saint-Gobain Crystals, butt-coupled to a PMMA optical fibre, used to carry the optical signal to a PMT. It has been proposed as a suitable dosemeter for *in vivo* brachytherapy applications, owing to its small size and sufficient sensitivity.[56,57] The difficulty in efficiently coupling the plastic scintillator to the optical fibre is one of the main disadvantages of these types of dosemeters. The probe was further tested for use in monitoring urethral doses during HDR brachytherapy for prostate cancer treatment.[58] 24 patients were enrolled for the clinical trial. After 14 patients, the dosemeter design was improved for more

Figure 3. Partially inflated endorectal balloon with an optical fibre plastic scintillation detector attached. Reproduced from Klein et al[55] with permission from Elsevier.



accurate readings. The results demonstrated a maximum measured dose departure of 9% from the calculated dose for the remaining 10 patients, further indicating the need for *in vivo* dosemeters in radiotherapy.

Therriault-Proulx et al[56] investigated the use of a BCF-60 plastic scintillator for verification of an iridium-192 HDR treatment within a water phantom. They compared the dose and dose rate values from the scintillator to the TPS values and sample doses to the OARs—in this case, the rectal wall and urethra. The group also looked at a mechanism for removal of the stem effect, the ability of a scintillator to detect dwell position errors and its temporal resolution.

In a further study, Therriault-Proulx et al[57] developed a single fibre, multipoint, plastic scintillator for iridium-192 HDR brachytherapy treatment verification in a water phantom. This scintillator contained a three-point detector system comprising BCF-10, BCF-12 and BCF-60 scintillating elements. They compared the accuracy of the measured dose at different source-to-detector distances and looked at the potential of using this system for measuring source position uncertainty. The results indicated a high level of agreement across all the parameters of interest.

Other groups have also investigated the potential for combining large numbers of plastic scintillators together in array patterns. Guillot et al[59] created a two-dimensional (2D) plastic scintillator array, comprising 781 detectors in a grid of size 26 × 26 cm. The performance of this array was then compared with a more traditional 2D array (MatriXX; IBA Dosimetry, Louvain-La-Neuve, Belgium), radiochromic film and TPS for some typical step-and-shoot IMRT treatment MLC sequences. The results indicated a good level of agreement in the gamma pass rate of >97.5% with a tolerance of 3%/3 mm when compared with the TPS. Liu et al[60] created a PMT array with 16 fibre optic dosimeters [Bicron BC400 scintillator (Saint-Gobainn Crystals) coupled to PMMA fibre]. They then characterized the response of this array, and assessed its ability to measure and quantify information relating to how a brachytherapy source is retracted, and beam pulses on a linac. In the work of Gagnon et al,[61] the performance of a plastic scintillator (BCF-60) was compared with a range of existing traditional small field detectors for stereotactic QA measurements. The group also looked at the potential of an array of 49 plastic scintillators. The results compared output factors and dose profiles, and reported a good level of agreement with stereotactic diodes and EBT2 Gafchromic film. In the case of the array, a comparison was made for verification of a stereotactic radiosurgery (SRS) treatment using four non-coplanar arcs, and again, good agreement was reported between the array and the TPS and chambers.

The Exradin W1 scintillator manufactured by Standard Imaging, Middleton, WI,[62] is currently the only commercially available optical fibre dosemeter for radiotherapy applications. The scintillating fibre is a 1-mm core polystyrene-based fibre and is coupled with a PMMA optical fibre for transmission of the optical signal to the detector.[62] However, with more PSDs set to enter the market shortly, focusing on *in vivo* applications (*e.g.* RadiaDyne OARtrac[63]; RadiaDyne, Houston, TX), the potential for widespread clinical adoption of optical fibre-based dosemeters is increasing.

Inorganic scintillation-based optical fibre dosemeters Commonly exploited in radiography applications, inorganic scintillators/phosphors, such as terbium-doped gadolinium oxysulfide ($Gd_2O_2S:Tb$) and thallium-doped caesium iodide ($CsI:Tl$) have also shown some advantages as radioluminescent dosemeters in radiotherapy.[64]

A radiation dosemeter, which comprised a scintillator material, an optical fibre bundle and a light measuring device, was developed to detect tritium in real time. Each scintillator interacts with beta radiation and generates scintillation photons between 455 and 550 nm wavelength of visible light.[65] Three kinds of inorganic scintillator were tested at different distances between the fibre optic sensor and source. These were $Gd_2O_2S:Tb$, cerium-doped YAG ($Y_3Al_5O_{12}:Ce$) and $CsI:Tl$. The results show that the scintillation efficiencies (relative to NaI) of $CsI:Tl$, $Y_3Al_5O_{12}:Ce$ and $Gd_2O_2S:Tb$ are 8%, 5% and 15%, respectively. The $Gd_2O_2S:Tb$-type scintillator was found to give the greatest scintillation response in terms of generated photons.[64] Although this was a useful study for providing a direct comparison between the various materials, it does not point to their prospective performance in a typical high-energy beam used in radiotherapy treatments, as experiments were conducted using a static tritium source.

An optical fibre dosemeter has been developed by McCarthy et al[66,67] and is illustrated in Figures 4 and 5. This fibre dosemeter is constructed by coating the end of an exposed PMMA optical fibre, after the cladding has been removed, with $Gd_2O_2S:Tb$. The phosphor, supplied by Phosphor Technologies Ltd, UK,[68] is

Figure 4. Schematic of an optical fibre dosemeter based on inorganic scintillation.

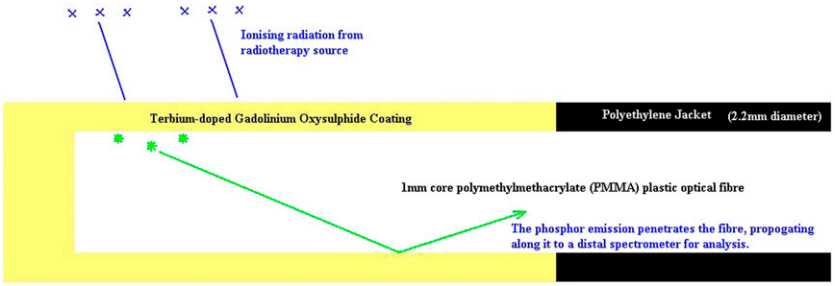
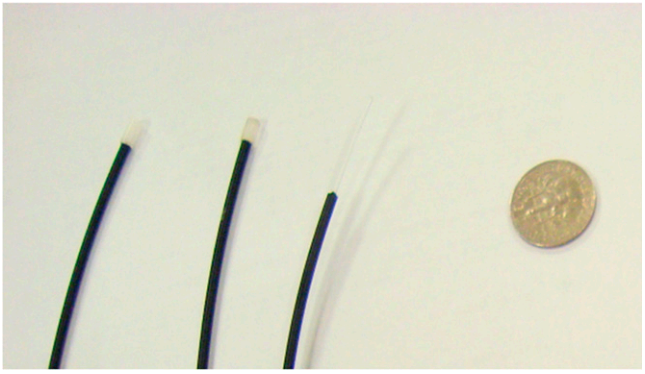


Figure 5. Photograph of two optical fibre inorganic scintillation dosemeters and a bare polymethylmethacrylate optical fibre prior to coating. Reproduced from O'Keeffe et al[70] with permission from SPIE.



mixed with an epoxy mix and injected into a cylindrical mould containing the exposed PMMA fibre optic core and allowed to cure. The radiation-sensitive scintillating material tip of the sensor fluoresces on immediate exposure to ionizing radiation. The resultant emitted fluorescent light penetrates the PMMA optical fibre and propagates along the fibre to a distal scientific-grade spectrometer from Ocean Optics, Dunedin, FL, where the intensity of the peak wavelength of the fluorescent light is measured.

Initial characterization measurements of this sensor have been carried out, with its response being evaluated in water-equivalent phantoms in order to assess whether it would be suitable for potential *in vivo* applications in either brachytherapy or external beam radiotherapy dosimetry. The results demonstrate that the fibre has a high sensitivity and good repeatability across a range of beam energies and types, and demonstrate a linear response from low doses of the order of centigray up to at least 16 Gy in a single delivery. The results also illustrate that the integrated intensity over the radiation beam on time, which corresponds to the dose received, remains relatively constant for dose rates between 100 and 1400 MU min^{-1}. However, the results indicate an energy dependence owing to the presence of high Z materials in the sensor. This also produces an over-response of the sensor measurements with both varying field size[69,70] and depth–dose measurements. The depth–dose data for electron beams show better agreement to that measured using ionization chambers, except for the region of the bremsstrahlung tail owing to the presence of X-ray contamination in the electron beam.

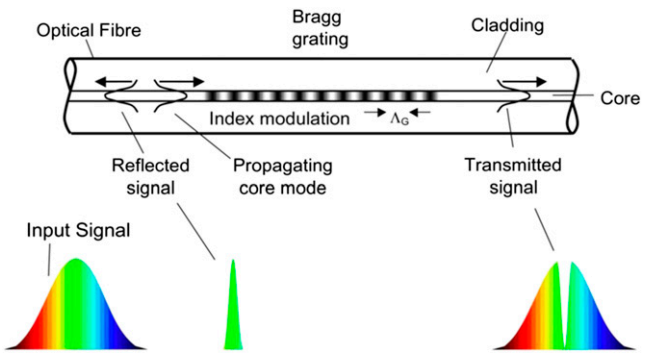
Jackson et al[71] showed the function of a RadLine® system (National Nuclear Laboratory, Warrington, UK), a radiation detector that consists of an inorganic scintillation crystal coupled to a fibre optic cable. The RadLine dosemeter is a small, novel, remotely operated radiation detector. It uses a zinc tungstate ($ZnW0_4$) scintillating crystal that produces scintillation light in response to beta and gamma radiation. The crystal was connected to a fibre optic cable, which transmits the light to a charge-coupled device camera. The RadLine dosemeter was tested in a beta and gamma narrow radiation field of 2.4 GBq, from a caesium-137 (662 KeV) source, at doses rates between 0.125 and 10 mSv h^{-1}. The results show that the lower limit of the device corresponds to a dose rate of 0.2 mSv h^{-1}.[71] Although this appears to be a sensitive system, there is no evidence of how it might perform for measurements of a HDR brachytherapy source.

Another potential dosemeter involved the use of a ruby element ($Al_aO_3:Cr$) of 1-mm diameter, connected to a 30-m fused silica light guide with optically transparent epoxy.[72] The RL light was transmitted through an optical bandpass filter with 689 nm centre wavelength and 29.5 nm bandwidth. The remaining light signal was measured using a Hamamatsu H7421-50 detector (Hamamatsu Protonics KK, Shizuka, Japan). The results showed a rise of the radioluminescent signal with dose, following a linear trend up to 2 Gy. The sensor is stable for larger doses up to at least 500 Gy.[72] Although the system seems very sensitive, it is not clear how it might perform in the typical clinical radiotherapy dosimetry environment.

Overcoming the stem effect

A term known as the stem effect is used to describe Cerenkov radiation and other radiation-induced light that can be produced in optical fibres and thus affect optical fibre-based dosemeter measurements, particularly in luminescent based sensors.[52] Cerenkov radiation is generated when a charged particle passes through a dielectric medium with a velocity greater than that of light in that medium. Cerenkov emissions only occur when the refractive index of the material is >1 and are only produced above a certain energy threshold. For plastics most commonly used in optical fibre scintillation dosimetry, the refractive index is 1.48–1.60, which means that Cherenkov is seen at energies $>$145–180 keV.[73] As the radiation parameters (*e.g.* energy, dose rate) vary considerably in radiotherapy, it is important that Cherenkov radiation is compensated for. A number of techniques are proposed for removing the stem effect in optical fibre-based dosemeters, such as using a dual reference fibre,[74–77] optical filtering[72] and a temporal approach, where the measurement reading is timed such that the stem effect has decayed out.[78] One of the most favoured methods of stem removal is the spectral, or chromatic, method[73,79,80] developed by Fontbonne et al.[81] The technique exploits the difference in optical spectra of the luminescence signal and Cherenkov radiation. It has been shown that by monitoring the light under two different wavelength filters, it is possible to isolate the

Figure 6. Fibre Bragg grating structure with spectral response.[83]



luminescence signal from the Cherenkov signal.[79–82] As a consequence of the large amount of research into this area, it is now widely considered that the issue of the stem effect has been addressed and no longer poses a limitation in the sensors' potential for radiotherapy dosimetry.

Fibre Bragg gratings

Fibre Bragg grating (FBG)-based sensors work by monitoring the wavelength shift of the returned Bragg signal, which changes as a function of the measurand. The Bragg wavelength is related to the refractive index of the material and the grating pitch. The light incident on the grating reflects a narrow spectral component at the Bragg wavelength, and hence in the transmission spectrum, this component is missing. This is further explained by Figure 6.[83] Much work has been carried out in the past number of years investigating the effects of radiation on FBG. This work has primarily concentrated on developing radiation-resistant FBGs for use in temperature and strain measurement applications in nuclear environments.[84–89] Krebber et al[90] demonstrated the use of FBGs for high-dose applications (>2 kGy), where FBGs were written in hydrogen-doped Ge-doped fibres. It was noted that the sensor required a highly stable set-up and a constant temperature. This would indicate that this type of sensor would be unsuitable for radiotherapy applications. However, Bragg grating cavity reflectors have recently been shown to be effective in monitoring low radiation doses (2 Gy), owing to their sensitivity to radiation-induced effects in the silica material of the fibre.[91] However, the high-sensitivity interrogation system used does not allow for real-time measurements of the radiation dose, and to be truly effective for radiotherapy dosimetry would need to be able to measure much lower doses.

CONCLUSIONS AND FUTURE DIRECTION

Optical fibres have been demonstrated to be able to perform accurate radiotherapy dosimetric measurements comparable to existing dosemeters, such as TLDs and diodes, and have the potential for advantages over these conventional systems. They allow for remote monitoring and, in most cases, real-time measurements of the radiation dose. Their small size, lightweight and flexibility have allowed their dosimetric performance to be demonstrated successfully in a range of measurement scenarios. These include both external beam and HDR brachytherapy radiotherapy delivery techniques, in addition to verifying complex treatment deliveries such as IMRT, VMAT and SRS. Their immunity to the intense magnetic field and radio frequency pulses present in the MRI environment may give optical fibre-based dosemeters a significant advantage over conventional dosemeters for real-time monitoring in MRI-guided linacs. The performance of a number of different fibres have been assessed against a wide range of typical radiotherapy dosimetry equipment, including TLDs, diodes, film, 2D arrays and ionization chambers. There is increasing interest in the development of such fibres to allow *in vivo* dosimetric verification of treatments such as HDR and prostate seed brachytherapy.

The majority of optical fibre dosimetry systems are extrinsic, whereby the optical fibre is used to transmit the optical signal relating to the dose information and is not directly involved in the sensing itself, although some intrinsic sensors based on commercial silica fibres have been reported. Luminescence is the predominant technique used, with RL showing the most potential within the area of *in vivo* dosimetry. PSDs have demonstrated significant advantages in the area of radiotherapy dosimetry, owing to their near-water equivalence, and this is reflected in the increasing number of publications in this area. Commercial interest has also focused on PSDs with the first optical fibre dosemeter for radiotherapy applications to reach the market, the Exradin W1, based on this technology, and further PSDs are also set to enter the market shortly. However, the recent reports on temperature dependence of this type of dosemeter must be investigated, and, if necessary, accounted for, to ensure reliability of the measurements. Inorganic scintillation-based optical fibre sensors also demonstrate potential for *in vivo* applications in radiotherapy monitoring. However, the energy dependence of such sensors must be examined and overcome. The issue of stem effect, which can affect optical fibre luminescence measurements, has been addressed, and a number of techniques have been widely demonstrated to successfully overcome this issue and, as such, it is no longer seen as a limitation of optical fibre-based dosemeters.

It is the opinion of the authors that optical fibre sensors could have the potential to provide new opportunities in the way *in vivo* monitoring is currently performed, by providing a method for *in vivo* measurement directly within the tumour volume and/or the dose to specific OARs.

FUNDING

The work was facilitated by COST Action TD1001: Novel and Reliable Optical Fibre Sensor Systems for Future Security and Safety Applications (OFSeSa). The European Commission's seventh Framework "Marie Curie Re-integration" action of the "Peoples" Programme (PERG04-2008-239207).

REFERENCES

1. Institute of Physical Sciences in Medicine (IPSM). Code of practice for high-energy photon therapy dosimetry based on the NPL absorbed dose calibration service. *Phys Med Biol* 1990; **35**: 1355–60.
2. Thwaites DI, DuSautoy AR, Jordan T, McEwen MR, Nisbet A, Nahum AE, et al; IPEM Working Party. The IPEM code of practice for electron dosimetry for radiotherapy beams of initial energy from 4 to 25 MeV based on an absorbed dose to water calibration. *Phys Med Biol* 2003; **48**: 2929–70.
3. IPEM (Institute of Physics and Engineering in Medicine). *Physics aspects of quality control in radiotherapy. IPEM report 81*. York, UK: IPEM; 1999.
4. Klein EE, Hanley J, Bayouth J, Yin FF, Simon W, Dresser S, et al; Task Group 142, American Association of Physicists in Medicine. Task Group 142 report: quality assurance of medical accelerators. *Med Phys* 2009; **36**: 4197–212.

5. Shafiq J, Barton M, Noble D, Lemer C, Donaldson LJ. An international review of patient safety measures in radiotherapy practice. *Radiother Oncol* 2009; **92**: 15–21. doi: 10.1016/j.radonc.2009.03.007
6. Boadu M, Rehani MM. Unintended exposure in radiotherapy: identification of prominent causes. *Radiother Oncol* 2009; **93**: 609–17. doi: 10.1016/j.radonc.2009.08.044
7. International Atomic Energy Agency (IAEA). *Lessons learned from accidental exposures in radiotherapy. Safety reports series 17*. Vienna, Austria: IAEA; 2000.
8. The Royal College of Radiologists; Society and College of Radiographers; Institute of Physics and Engineering in Medicine; National Patient Safety Agency; British Institute of Radiology. *Towards safer radiotherapy*. London, UK: The Royal College of Radiologists; 2008.
9. Kutcher GJ, Coia L, Gillin M, Hanson WF, Leibel S, Morton RJ, et al. Comprehensive QA for radiation oncology: report of AAPM Radiation Therapy Committee Task Group 40. *Med Phys* 1994; **21**: 581–618.
10. Mijnheer B, Beddar S, Izewska J, Reft C. *In vivo* dosimetry in external beam radiotherapy. *Med Phys* 2013; **40**: 070903. doi: 10.1118/1.4811216
11. Van Dam J, Marinello G. *Methods for in vivo dosimetry in external radiotherapy*. 2nd edn. Brussels, Belguium: European Society for Radiotherapy and Oncology, 2006.
12. Tanderup K, Beddar S, Andersen CE, Kertzscher G, Cygler JE. *In vivo* dosimetry in brachytherapy. *Med Phys* 2013; **40**: 070902. doi: 10.1118/1.4810943
13. van Elmpt W, McDermott L, Nijsten S, Wendling M, Lambin P, Mihnheer B. A literature review of electronic portal imaging for radiotherapy dosimetry. *Radiother Oncol* 2008; **88**: 289–309. doi: 10.1016/j.radonc.2008.07.008
14. Wendling M, McDermott LN, Mans A, Sonke JJ, van Herk M, Mijnheer BJ. A simple backprojection algorithm for 3D *in vivo* EPID dosimetry of IMRT treatments. *Med Phys* 2009; **36**: 3310–21.
15. Wendling M, McDermott LN, Mans A, Olaciregui-Ruiz Í, Pecharromán-Gallego R, Sonke JJ, et al. In aqua vivo EPID dosimetry. *Med Phys* 2012; **39**: 367–77. doi: 10.1118/1.3665709
16. Agnew CE, King RB, Hounsell AR, McGarry CK. Implementation of phantom-less IMRT delivery verification using Varian DynaLog files and R/V output. *Phys Med Biol* 2012; **57**: 6761–77. doi: 10.1088/0031-9155/57/21/6761
17. Agnew CE, Irvine DM, Hounsell AR, McGarry CK. Improvement in clinical step and shoot intensity modulated radiation therapy delivery accuracy on an integrated linear accelerator control system. *Pract Radiat Oncol* 2014; **4**: 43–9. doi: 10.1016/j.prro.2013.07.003
18. Reynolds M, Fallone BG, Rathee S. Dose response of selected ion chambers in applied homogeneous transverse and longitudinal magnetic fields. *Med Phys* 2013; **40**: 042102. doi: 10.1118/1.4794496
19. Karger CP, Jäkel O, Palmans H, Kanai T. Dosimetry for ion beam radiotherapy. *Phys Med Biol* 2010; **55**: R193–234. doi: 10.1088/0031-9155/55/21/R01
20. Kron T, Duggan L, Smith T, Rosenfeld A, Butson M, Kaplan G, et al. Dose response of various detectors to synchrotron radiation. *Phys Med Biol* 1998; **43**: 3235–59.
21. Verhaegen F, Granton P, Tryggestad E. Small animal radiotherapy research platforms. *Phys Med Biol* 2011; **56**: R55–83. doi: 10.1088/0031-9155/56/12/R01
22. McKerracher C, Thwaites DI. Assessment of new small-field detectors against standard-field detectors for practical stereotactic beam data acquisition. *Phys Med Biol* 1999; **44**: 2143–60.
23. Aspradakis MM, Byrne JP, Pamans H, Conway J, Rosser K, Warrington JAP, et al. *Small field MV photon dosimetry. IPEM report 103*. York, UK: IPEM (Institute of Physics and Engineering in Medicine); 2010.
24. Newton J, Oldham M, Thomas A, Li Y, Adamovics J, Kirsch DG, et al. Commissioning a small-field biological irradiator using point, 2D, and 3D dosimetry techniques. *Med Phys* 2011; **38**: 6754–62. doi: 10.1118/1.3663675
25. Mieloszyk M, Krawczuk M, Zak A, Ostachowicz W. An adaptive wing for a small-aircraft application with a configuration of fibre Bragg grating sensors. *Smart Mater Struct* 2010; **19**: 085009.
26. Fernandez Fernandez A, Brichard B, O'Keeffe S, Fitzpatrick C, Lewis E, Vaille J-R, et al. Real-time fibre optic radiation dosimeters for nuclear environment monitoring around thermonuclear reactors. *Fusion Eng Des* 2008; **83**: 50–9.
27. Villnow M, Bosselman T, Willsch M, Kaiser J. FBG system for temperature monitoring under electromagnetic immersed and harsh oil and gas reservoir environment. *Proc SPIE* 2014; **9157**: 915797.
28. Delepine-Lesoille S, Phéron X, Bertrand J, Pilorget G, Hermand G, Farhoud R, et al. Industrial qualification process for optical fibers distributed strain and temperature sensing in nuclear waste repositories. *J Sens* 2012; **2012**: 9. doi:10.1155/2012/369375
29. Fallone BG, Murray B, Rathee S, Stanescu T, Steciw S, Vidakovic S, et al. First MR images obtained during megavoltage photon irradiation from a prototype integrated linac-MR system. *Med Phys* 2009; **36**: 2084–8.
30. Raaymakers BW, Lagendijk JJ, Overweg J, Kok JG, Raaijmakers AJ, Kerkhof EM, et al. Integrating a 1.5 T MRI scanner with a 6 MV accelerator: proof of concept. *Phys Med Biol* 2009; **54**: N229–37. doi: 10.1088/0031-9155/54/12/N01
31. Ramli AT, Bradley DA, Hashim S, Wagiran H. The thermoluminescence response of doped SiO2 optical fibres subjected to alpha-particle irradiation. *Appl Radiat Isot* 2009; **67**: 428–32. doi: 10.1016/j.apradiso.2008.06.034
32. Bradley DA, Hugtenburg RP, Hashim S, Okoya OO, Yusoff AL, Hassan AAM, et al. The development of doped radiosensitive glass. In: AIP Conference Proceedings. The 2nd International Conference on Solid State Science and Technology; September 2006; Kuala Terengganu, Malaysia, Sept. 2006.
33. Hashim S, Al-Ahbabi S, Bradley DA, Webb M, Jeynes C, Ramli AT, et al. The thermoluminescence response of doped SiO2 optical fibres subjected to photon and electron irradiations. *Appl Radiat Isot* 2009; **67**: 423–7. doi: 10.1016/j.apradiso.2008.06.030
34. Abdul Rahman AT, Hugtenburg RP, Abdul Sani SF, Alalawi AI, Issa F, Thomas R, et al. An investigation of the thermoluminescence of Ge-doped SiO2 optical fibres for application in interface radiation dosimetry. *Appl Radiat Isot* 2012; **70**: 1436–41. doi: 10.1016/j.apradiso.2011.11.030
35. Abdul Rahman AT, Nisbet A, Bradley DA. Dose-rate and the reciprocity law: TL response of Ge-doped SiO2 optical fibers at therapeutic radiation doses. *Nucl Instr Meth Phys Res A* 2011; **652**: 891–5.
36. Issa F, Abdul Rahman AT, Hugtenburg RP, Bradley DA, Nisbet A. Establishment of Ge-doped optical fibres as thermoluminescence dosimeters for brachytherapy. *Appl Radiat Isot* 2012; **70**: 1158–61. doi: 10.1016/j.apradiso.2011.12.027
37. Bradley DA, Hugtenburg RP, Nisbet A, Abdul Rahman AT, Issa F, Mohd Noor N, et al. Review of doped silica glass optical fibre: their TL properties and potential applications in radiation therapy dosimetry. *Appl Radiat Isot* 2012; **71**: 2–11. doi: 10.1016/j.apradiso.2012.02.001
38. Palmer AL, Di Pietro P, Alobaidli S, Issa F, Doran S, Bradley D, et al. Comparison of methods for the measurement of radiation dose distributions in high dose rate (HDR) brachytherapy: Ge-doped optical fiber, EBT3 Gafchromic film, and PRESAGE® radiochromic plastic. *Med Phys* 2013; **40**: 061707. doi: 10.1118/1.4805100
39. Jursinic PA. Characterization of optically stimulated luminescent dosimeters, OSLDs,

for clinical dosimetric measurements. *Med Phys* 2007; **34**: 4594–604.
40. Yukihara EG, McKeever SW. Optically stimulated luminescence (OSL) dosimetry in medicine. *Phys Med Biol* 2008; **53**: R351–79. doi: 10.1088/0031-9155/53/20/R01
41. Marckmann CJ, Andersen CE, Aznar MC, Bøtter-Jensen L. Optical fibre dosemeter systems for clinical applications based on radioluminescence and optically stimulated luminescence from Al2O3:C. *Radiat Prot Dosimetry* 2006; **120**: 28–32.
42. Magne S, de Carlan LC, Bordy JM, Isambert AL, Bridier A, Ferdinand P. Multichannel dosimeter and α-Al2O3: C optically stimulated luminescence (OSL) fiber sensors for use in radiation therapy—evaluation with photon beams. *IEEE Trans Nucl Sci* 2011; **58**: 386–94.
43. Andersen CE, Nielsen SK, Greilich S, Helt-Hansen J, Lindegaard JC, Tanderup K. Characterization of a fiber-coupled Al2O3: C luminescence dosimetry system for online *in vivo* dose verification during 192Ir brachytherapy. *Med Phys* 2009; **36**: 708–18.
44. Gaza R, McKeever SW. A real-time, high-resolution optical fibre dosemeter based on optically stimulated luminescence (OSL) of KBr: Eu, for potential use during the radiotherapy of cancer. *Radiat Prot Dosimetry* 2006; **120**: 14–19.
45. Bradley DA, Aziz AZ, Knapton AJ, Abdul-Rashid HA, Mizanur Rahman AKM, Amin YM, et al. Optically stimulated luminescence in beta irradiated Ge-doped optical fibre. In: Proceedings 2013 IEEE 4th International Conference on Photonics (ICP); October 2013; Melaka, Malaysia: IEEE, 38–40.
46. Espinosa G, Bogard JS. Optically stimulated luminescence response of commercial SiO_2 optical fiber. *J Radioanal Nucl Chem* 2008; **277**: 125–9.
47. Kalnins CAG, Spooner NA, Ebendorff-Heidepriem H, Monro TM. Luminescent properties of fluoride-phosphate glass for radiation dosimetry. *Opt Mater Express* 2013; **3**: 960–7.
48. Saint Gobain Crystals. Product Comparisons: Inorganic and Organic Crystals, 2003. [Cited 18 September 2014.] Available from: www.crystals.saint-gobain.com
49. Saint Gobain Crystals. [Cited 20 August 2014.] Available from: http://www.crystals.saint-gobain.com/
50. Suchowerska N, Lambert J, Nakano T, Law S, Elsey J, McKenzie DR. A fibre optic dosimeter customised for brachytherapy. *Radiat Meas* 2007; **42**: 929–32.
51. Jang KW, Yoo WJ, Moon J, Han KT, Park BG, Shin D, et al. Multi-dimensional fiber-optic radiation sensor for ocular proton therapy dosimetry, *Nucl Instr Meth Phys Res A* 2012; **695**: 322–5.
52. Beddar AS. Plastic scintillation dosimetry and its application to radiotherapy. *Radiat Meas* 2006; **41**: S124–33.
53. Buranurak S, Andersen CE, Beierholm AR, Lindvold LR. Temperature variations as a source of uncertainty in medical fiber-coupled organic plastic scintillator dosimetry. *Radiat Meas* 2013; **56**: 307–11.
54. Beddar S. On possible temperature dependence of plastic scintillator response. *Med Phys* 2012; **39**: 6522. doi: 10.1118/1.4748508
55. Klein D, Briere TM, Kudchadker R, Archambault L, Beaulieu L, Lee A, et al. In-phantom dose verification of prostate IMRT and VMAT deliveries using plastic scintillation detectors. *Radiat Meas* 2012; **47**: 921–9.
56. Therriault-Proulx F, Brier TM, Mourtada F, Aubin S, Beddar S, Beaulieu L. A phantom study of an *in vivo* dosimetry system using plastic scintillation detectors for real-time verification of 192Ir HDR brachytherapy. *Med Phys* 2011; **38**: 2542–51.
57. Therriault-Proulx F, Beddar S, Beaulieu L. On the use of a single-fiber multipoint plastic scintillation detector for 192Ir high-dose-rate brachytherapy. *Med Phys* 2013; **40**: 062101. doi: 10.1118/1.4803510
58. Suchowerska N, Jackson M, Lambert J, Yin YB, Hruby G, McKenzie DR. Clinical trials of a urethral dose measurement system in brachytherapy using scintillation detectors. *Int J Radiat Oncol Biol Phys* 2011; **79**: 609–15. doi: 10.1016/j.ijrobp.2010.03.030
59. Guillot M, Beaulieu L, Archambault L, Beddar S, Gingras L. A new water-equivalent 2D plastic scintillation detectors array for the dosimetry of megavoltage energy photon beams in radiation therapy. *Med Phys* 2011; **38**: 6763–74. doi: 10.1118/1.3664007
60. Liu PZ, Suchowerska N, Abolfathi P, McKenzie DR. Real-time scintillation array dosimetry for radiotherapy: the advantages of photomultiplier detectors. *Med Phys* 2012; **39**: 1688–95. doi: 10.1118/1.3690465
61. Gagnon JC, Thériault D, Guillot M, Archambault L, Beddar S, Gingras L, et al. Dosimetric performance and array assessment of plastic scintillation detectors for stereotactic radiosurgery quality assurance. *Med Phys* 2012; **39**: 429–36. doi: 10.1118/1.3666765
62. Standard Imaging Inc.. [Cited 20 August 2014.] Available from: http://www.standardimaging.com/exradin/exradin-scintillator/
63. Klawikowski SJ, Zeringue C, Wootton LS, Ibbott GS, Beddar S. Preliminary evaluation of the dosimetric accuracy of the *in vivo* plastic scintillation detector OARtrac system for prostate cancer treatments. *Phys Med Biol* 2014; **59**: N27–36. doi: 10.1088/0031-9155/59/9/N27
64. van Eijk CW. Inorganic scintillators in medical imaging. *Phys Med Biol* 2002; **47**: R85–106.
65. Jang KW, Cho DH, Yoo WJ, Seo JK, Heo JY, Park J-Y, et al. Fiber-optic radiation sensor for detection of tritium. *Nucl Instr Meth Phys Res A* 2011; **652**: 928–31.
66. McCarthy D, O'Keeffe S, Lewis E, Sporea D, Sporea A, Tiseanu I. Optical fibre X-ray radiation dosimeter sensor for low dose applications. Proceedings of IEEE Sensors Conference; October 2011; Limerick, Ireland, 121–4. doi: 10.1109/ICSENS.2011.6127086
67. McCarthy D, O'Keeffe S, Lewis E, Sporea DG, Sporea A, Tiseanu I, et al. Radiation dosimeter using an extrinsic fiber optic sensor. *IEEE Sens J* 2014; **14**: 673–85.
68. Phosphor Technologies Ltd. [Cited 18 September 2014.] Available from: http://www.phosphor-technology.com/
69. Woulfe P, O'Keeffe S, McCarthy D, Grattan MWD, Hounsell AR, Cronin J, et al. Characterisation of radioluminesence based optical fibre dosimeter in radiotherapy beam applications. In: Proceedings of SPIE, International Society for Optics and Photonics: 23rd International Conference on Optical Fiber Sensors; June 2014; Santander, Spain, 915799.
70. O'Keeffe S, Grattan MWD, Hounsell AR, McCarthy D, Woulfe P, Cronin J, et al. Radiotherapy dosimetry based on plastic optical fibre sensors. Proceedings of SPIE Fifth European Workshop on Optical Fibre Sensors, 879418, Vol 8794; 20 May 2013. Available from: http://dx.doi.org/10.1117/12.2025864
71. Jackson SF, Monk SD, Lennox K. Testing of a scintillator and fibre optic based radiation sensor. *Radiat Meas* 2013; **59**: 50–8.
72. Teichmann T, Sommer M, Henniger J. Dose rate measurements with a ruby-based fiber optic radioluminescent probe. *Radiat Meas* 2013; **56**: 347–50.
73. Beaulieu L, Goulet M, Archambault L, Beddar S. Current status of scintillation dosimetry for megavoltage beams. *J Phys Conf Ser* 2013; **444**: 012013.
74. Beddar AS, Mackie TR, Attix FH. Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: I. Physical characteristics and theoretical considerations. *Phys Med Biol* 1992; **37**: 1883–900.
75. Beddar AS, Mackie TR, Attix FH. Cerenkov light generated in optical fibres and other light pipes irradiated by electron beams. *Phys Med Biol* 1992; **37**: 925.

76. Lambert J, Yin Y, McKenzie DR, Law SH, Ralston A, Suchowerska N. A prototype scintillation dosimeter customized for small and dynamic megavoltage radiation fields. *Phys Med Biol* 2010; **55**: 1115–26. doi: 10.1088/0031-9155/55/4/014
77. Ishikawa M, Bengua G, Sutherland KL, Hiratsuka J, Katoh N, Shimizu S, et al. A feasibility study of novel plastic scintillation dosimetry with pulse-counting mode. *Phys Med Biol* 2009; **54**: 2079–92. doi: 10.1088/0031-9155/54/7/015
78. Clift MA, Johnston PN, Webb DV. A temporal method of avoiding the Cerenkov radiation generated in organic scintillator dosimeters by pulsed mega-voltage electron and photon beams. *Phys Med Biol* 2002; **47**: 1421–33.
79. Archambault L, Beddar AS, Gingras L, Roy R, Beaulieu L. Measurement accuracy and Cerenkov removal for high performance, high spatial resolution scintillation dosimetry. *Med Phys* 2006; **33**: 128–35.
80. Guillot M, Gingras L, Archambault L, Beddar S. Beaulieu L. Spectral method for the correction of the Cerenkov light effect in plastic scintillation detectors: a comparison study of calibration procedures and validation in Cerenkov light-dominated situations. *Med Phys* 2011; **38**: 2140–50.
81. Fontbonne JM, Iltis G, Ban G, Battala A, Vernhes JC, Tillier J, et al. Scintillating fiber dosimeter for radiation therapy accelerator. *IEEE Trans Nucl Sci* 2002; **49**: 2223–7.
82. Therriault-Proulx F, Beaulieu L, Archambault L, Beddar S. On the nature of the light produced within PMMA optical light guides in scintillation fiber-optic dosimetry. *Phys Med Biol* 2013; **58**: 2073–84. doi: 10.1088/0031-9155/58/7/2073
83. Mihailov SJ. Fiber Bragg grating sensors for harsh environments. *Sensors (Basel)* 2012; **12**: 1898–918. doi: 10.3390/s120201898
84. Fernandez Fernandez A, Brichard B, Berghmans F, Decreton M. Dose-rate dependencies in gamma-irradiated fiber Bragg grating filters. *IEEE Trans Nucl Sci* 2002; **49**: 2874–8.
85. Fernandez Fernandez A, Rego G, Gusarov A, Brichard B, Santos JL, Berghmans F. Evaluation of long-period fiber grating temperature sensors in nuclear environments. In: Proceedings of SPIE International Society for Optics and Photonics, Second European Workshop on Optical Fibre Sensors; June 2004; Santander, Spain, 88–91. doi: 10.1117/12.566600
86. Gusarov A, Berghmans F, Deparis O, Fernandez Fernandez A, Defosse Y, Megret P, et al. High total dose radiation effects on temperature sensing fiber Bragg gratings. *IEEE Photonic Technology Letters* 1999; **11**: 1159–61.
87. Gusarov A, Fernandez Fernandez A, Vasiliev S, Medvedkov O, Blondel M, Berghmans F. Effect of gamma-neutron nuclear reactor radiation on the properties of Bragg gratings written in photosensitive Ge-doped optical fiber. *Nucl Instr Meth Phys Res B* 2002; **187**: 79–86.
88. Kakuta T, Yamagishi H, Iwamura T, Urakami M. Development of new nuclear instrumentation based on optical sensing—irradiation effects on fiber Bragg grating sensors. *Proc SPIE* 2000; **4185**: 816–19.
89. Maier RR, MacPherson WN, Barton JS, Jones J, Scott M, Fernandez Fernandez A, et al. Fibre Bragg gratings of type I in SMF-28 and B/Ge fibre and type IIA B/Ge fibre under gamma radiation up to 0.54 MGy. In: Proceedings of SPIE: International Society for Optics and Photonics. 17th International Conference on Optical Fiber Sensors; May 2005; Bruges, Belgium, 511–14.
90. Krebber K, Henschel H, Weinand U. Fibre Bragg gratings as high dose radiation sensors? *Meas Sci Technol* 2006; **17**: 1095–102.
91. Avino S, D'Avino V, Giorgini A, Pacelli R, Liuzzi R, Cella L, et al. Radiation dosimetry with fiber Bragg gratings. Proceedings of SPIE: International Society for Optics and Photonics: 23rd International Conference on Optical Fiber Sensors; June 2014; Santander, Spain, 91574.