Miniaturized scintillator dosimeter for small field radiation therapy

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Abstract

The concept of a miniaturized inorganic scintillator detector is demonstrated in the analysis of the small static photon fields used in external radiation therapy. Such a detector is constituted by a 0.25 mm diameter and 0.48 mm long inorganic scintillating cell (1.6 × 10⁻⁵ cm⁻³ detection volume) efficiently coupled to a narrow 125 μm diameter silica optical fiber using a tiny photonic interface (an optical antenna). The response of our miniaturized scintillator detector (MSD) under 6 MV bremsstrahlung beam of various sizes (from 1 × 1 cm² to 4 × 4 cm²) is compared to that of two high resolution reference probes, namely, a micro-diamond detector and a dedicated silicon diode. The spurious Čerenkov signal transmitted through our bare detector is rejected with a basic spectral filtering. The MSD shows a linear response regarding the dose, a repeatability within 0.1% and a radial directional dependence of 0.36% (standard deviations). Beam profiling at 5 cm depth with the MSD and the micro-diamond detector shows a mismatch in the measurement of the full widths at 80% and 50% of the maximum which does not exceed 0.25 mm. The same difference range is found between the micro-diamond detector and a silicon diode. The deviation of the percentage depth dose between the MSD and micro-diamond detector remains below 2.3% within the first fifteen centimeters of the decay region for field sizes of 1 × 1 cm², 2 × 2 cm² and 3 × 3 cm² (0.76% between the silicon diode and the micro-diamond in the same field range). The 2D dose mapping of a 0.6 × 0.6 cm² photon field evidences the strong 3D character of the radiation-matter interaction in small photon field regime. From a beam-probe convolution theory, we predict that our probe overestimates the beam width by 0.06%, making our detector a right compromise between high resolution, compactness, flexibility and ease of use. The MSD overcomes problem of volume averaging, stem effects, and despite its water nonequivalence it is expected to minimize electron fluence perturbation due to its extreme compactness. Such a detector thus has the potential to become a valuable dose verification tool in small field radiation therapy, and by extension in Brachytherapy, FLASH-radiotherapy and microbeam radiation therapy.

1. Introduction

Treatment of cancer using radiotherapy is always a balance between successfully killing cancerous tissue with lethal doses of ionizing radiation and minimizing the risks of harmful effects on surrounding healthy tissue.

Stereotactic radiotherapy (stereotactic radiosurgery and stereotactic body radiotherapy) has become a standard procedure in the treatment of small tumors (Aitken and Hawkins 2015, Videtic et al 2017, Graber et al 2019, Spencer et al 2019). The dose is here delivered using highly-controlled narrow radiation beams and small number of fractions, yielding a concentrated and accurate treatment.

Unlike large treatment fields, the dosimetry of small fields has to face several challenges including lack of electronic equilibrium, steep dose gradients and partial occlusion of radiation source (Bouchard et al 2019, Graber et al 2019, Spencer et al 2019).
Therefore, to accurately measure the dose deposition in small fields and thus perform robust quality assurance programs, a dosimeter has to show high spatial resolution, high compactness and/or water-equivalence (to avoid risks of electron transport alteration), while being convenient to use. Unfortunately, the dosimetric signals delivered by standard detectors, which almost perfectly merge for large fields, noticeably diverge in the small field regime (Westermark et al 2000, Laub and Wong 2003, Tanny et al 2015, Godson et al 2016).

Among available dosimetry strategies, plastic scintillator dosimeters (PSD) have retained much attention (Beddar et al 1992, 1992, 2001, Beaulieu and Beddar 2016, Pasquino et al 2017, Archer et al 2017, Galavis et al 2019). Being water-equivalent, PSDs avoid perturbations of the electron fluence while limiting over-responses and correction factors (Beaulieu and Beddar 2016). Water-equivalence is however obtained at the expense of a probe lateral size larger than 0.5 mm (Galavis et al 2019), due to limits in signal-to-noise ratio induced by spurious Cerenkov light (Beddar et al 1992, Archambault et al 2007, De Boer et al 1993). Despite a modest volume averaging of the order of 1% within small photon fields, such probes may represent limits in the design of detection architectures and need careful calibration regarding Cerenkov signal (Morin et al 2013).

As an alternative to the PSD approach, we recently introduced the first miniaturized inorganic scintillator dosimeter from a narrow 125 μm outer diameter silica optical fiber (Suarez et al 2019). The miniaturization of inorganic dosimeters (Ding et al 2020) provides a unique opportunity to combine unprecedented spatial resolution and low electron fluence perturbation (even with water non-equivalent fibers and scintillators). Miniaturization is here rendered possible by the high radiation-to-light conversion efficiency of inorganic scintillators (Shionoya et al 1998, Van Eijk 2002, Weber 2002) and the superior guiding performances of silica fibers regarding plastic fibers. Fiber probes based on inorganic scintillators deliver a signal level a few orders of magnitude higher than that from fiber detectors based on plastic scintillators, leading to a smaller Cerenkov contribution in dosimetry (Kertzscher and Beddar 2017, 2019, Linares Rosales et al 2020).

In this paper, we demonstrate a miniaturized scintillator detector (MSD) in the dosimetry of small photon fields. After a preliminary specification of the MSD in terms of linearity, repeatability and directional dependence, the response of the detector to small photon fields is compared to that of a micro-detector and a silicon diode (i.e. two reference high resolution detectors in small photon fields). Then, a highly-resolved 2D map of a 0.6 × 0.6 cm² field with the MSD reveals a strongly inhomogeneous field distribution in a few cubic-centimeter volume. We predict from convolution-based theory that volume averaging effects are negligible. Given its high resolution, extreme compactness and negligible stem effect, the inorganic MSD has the potential to become a valuable and practical dose verification tool in small field radiation therapy, and by extension in Brachytherapy, FLASH-radiotherapy (Bourhis et al 2019a, 2019b) and microbeam radiation therapy (Soliman et al 2020, Bartzsch et al 2020).

2. Material and methods

2.1. Probe fabrication

The MSD is engineered from a narrow biocompatible 125 μm multimode fiber from SEDI-ATI. The fiber tip is first processed in a buffered fluorhydric acid. Next, an inorganic scintillator (Gd₂O₂S:Tb) in the form of a powder is mixed with an acrylic glue. Then, a small amount of the resulting scintillating mixture is attached onto the fiber tip and polymer is hardened. Terbium-doped gadolinium oxysulfide (Gd₂O₂S:Tb) is known to emit visible light upon exposure (figure 1(b)) with good efficiency, stability and linearity at room temperature (Qin et al 2016, Alharbi et al 2018, Hu et al 2018). Note that under fixed irradiation conditions, the luminescence intensity of terbium-doped gadolinium oxysulfide remains almost constant over a broad temperature range (O’Reilly et al 2020).

Figure 1(a) displays a magnified optical image of a resulting MSD. The length and diameter of the scintillating cell are 0.48 mm and 0.25 mm, respectively. The two-meter long fiber is surrounded by an opaque 0.9 mm black hytrel cladding to be light protected. Both the scintillating cell and the last 1.8 cm of the fiber extend beyond the extremity of the opaque cladding to be directly in contact to the phantom (figure 1(d)). Despite its small size, the resulting bare fiber probe shows a scintillating cell that is clearly visible with naked eyes, thereby enabling an accurate probe alignment to the source crosshair. The probes are robust to contact, temperatures up to 150 °C and moisture.

2.2. Experimental set-up and protocol

Our MSD is tested in the 6 MV bremsstrahlung of a TrueBeam linear accelerator (Varian), in a water tank (PTW MP3; see figure 1(c)). The source-surface distance is fixed to 100 cm. The water tank is equipped with a 3D motorized stage driven with the MEPHYSTO software. The MSD is fixed to this three-axis translation stage with...
a 3D printed holder (in blue in figure 1(d)). In a coordinate frame (0, x, y, z) where the vertical (Oz)-axis defines the beam propagation direction, the MSD is oriented along (0x)-axis (see figure 1(d)).

The output optical signal from the MSD is recorded with a standard camera placed in the control room (figure 1(e)). The probe and the camera are interconnected with a ten-meter fiber extension. Fiber connection is realized with FC/PC connectors. To record the output optical signal from the MSD, the camera is equipped with a 35 mm objective (Fujinon HF35SA) and operates at an exposure time of 100 ms and a 4 × 4 pixel binning. It is driven under Labview to be used as a photodetector. Each image of the fiber output is automatically integrated over a specific region of interest, leading to an analog 10 Hz electric signal directly proportional to the MSD optical signal. Prior to light acquisition, the background level of the camera is measured from images acquired with the source off.

A narrow chromatic filter (544/24 nm band pass filter from Semrock) is positioned in front of the camera to selectively transmit the narrow-band luminescence from the Gd2O2S:Tb (figure 1(b)) and reject most of the broadband Cerenkov signal (96.18% rejection is measured in a 10 × 10 cm² field at 5 cm depth). Figure 1(e) shows the intensity of the Cerenkov signal detected by our fiber probe in a 4 × 4 cm² field with and without chromatic filter. The Cerenkov signal is measured with a scintillator-free etched fiber (see inset of figure 1(e)) positioned at 5 cm depth and at the beam center in the water phantom. In the presence of the chromatic filter, the intensity of the detected signal is within the noise level of the camera and is considered to be zero. We also tested a tapered fiber with only glue at the tip. In that case, no signal difference has been detected, thus confirming negligible Cerenkov and fluorescence signals from the glue. We also measured that the glue does not modify the emission spectrum of the scintillator.

The MSD response in terms of dose linearity, repeatability is assessed with the detector positioned at the center of a 4 × 4 cm² photon field. Measurements are realized at 5 cm depth in water, with the MSD axis along (0x). The dose linearity of the MSD is tested with a constant dose rate of 600 MU min⁻¹ and at 2, 5, 10, 20, 50, 100, 200, 300, 500, 1000 and 5000 MU (1 MU corresponds to an absorbed dose of 1 cGy under TRS 398 recommendations (IAEA, 2000)). The repeatability of the MSD is tested over ten exposures at 300 MU.

Percentage depth doses (PDD) and dose profile (DP) are measured in 1 × 1 cm² – 4 × 4 cm² fields, in the water phantom. The probe scan is defined in a step-by-step mode with the MEPHYSTO software. During result post-processing, the output signal from the camera is synchronized with the various probe positions to form the beam profiles and PDDs. The response of the MSD is compared to that of detectors commonly used in small field dosimetry: a micro-diamond detector (TM60019 from PTW company) and a silicon diode (TM60018 from PTW company). Data from these two probes are recorded with the MEPHYSTO software. The micro-diamond detector and the silicon diode are aligned along the (0z)-axis for PDD and DP measurements. For all detectors, the PDD is measured at the beam center, from water surface down to 15 cm depth (along (0z)), and it is normalized to the maximum dose. The DP is measured perpendicularly to the beam axis (along (0y)) at a depth of 5 cm in water. It is normalized to the dose value at the beam’s central axis.
The 2D dose signal of a 0.6 × 0.6 cm² photon field is accumulated in the (yOz)-plane from successive PDD acquisitions along (0z) spaced from 0.25 to 0.75 mm apart along (0y) (see figure 1(d)). All PDD curves are acquired in a continuous scan mode at a constant velocity.

The directional dependence of the probe is assessed in air to ensure environment homogeneity. To this end, the water tank is removed and the probe is attached on a narrow plastic rod used as a holder that induces negligible perturbation on acquisitions. The source-detector distance is fixed to 100 cm, i.e. the probe is centered at the isocenter of the linear accelerator. Measurements are realized by rotating the gantry around the probe. The radial angular diagram is plotted in the (yOz)-plane perpendicular to the detector axis (Ox). The azimuthal angular diagram is obtained in the orthogonal (xOz)-plane. Signals are recorded in a 4 × 4 cm² field, at 300 MU with a dose rate of 600 MU min⁻¹. For each diagram, the optical signal is normalized to the signal at 0°.

2.3. Prediction of the volume averaging

The effect of the finite size of the MSD in the dose profiling can be mathematically anticipated by the convolution of the detector point spread function (the detector response to a point-like excitation) and real dose distribution (Goodman 2005). The measured DP Fm along the (0y)-axis reads:

\[ F_m(y) = \int_{-\infty}^{+\infty} D(y_0)F(y - y_0)dy_0, \]  

where \( D \) is the point spread function of the detector and \( F \) is the real DP. In the Fourier space, the convolution product is turned into the regular product. The point spread function of the detector is usually analytically described either with a parabolic function (Sibata et al 1991, Higgins et al 1995) or a Gaussian function (Garcia-Vicente et al 1998, Ulmer and Kaissl 2003). Here, a rectangular function is preferred as it ensures the upper limit of the volume averaging for a given detector size. The width of the rectangle is fixed to 0.25 mm to match the smaller diameter of the ellipsoidal detection cell of the MSD, which defines the probe resolution in our experimental approach. The real DP is approximated with a Gaussian function whose 1/e diameter fits the 0.6 cm nominal beam diameter. Numerical calculations are realized with the Matlab software.

2.4. Results

The response of the MSD to irradiation at 600 MU min⁻¹ is first quantified at 5 cm depth in the water tank by measuring the optical power at the probe output with a photon counter (from Aurea technology). We find for beam widths of 1, 2, 3 and 4 cm an output power of 3.41 × 10⁶, 4.08 × 10⁷, 4.51 × 10⁷ and 4.94 × 10⁷ photons s⁻¹, respectively, with signals far exceeding fluctuation-noise level (minimum signal-to-noise ratio of 180 in 1 × 1 cm² field size).

The dose linearity of the MSD is reported in figure 2(a). The repeatability of the probe is addressed in figure 2(b). For the ten exposures at 300 MU, the light intensities accumulated by the detector show standard deviation from signal average of 0.1%. The directional response of the MSD is shown in figure 3. The signal deviation over the measured angular range is within 0.36% of the signal average for the radial angular directions (figure 3(a)) and 3.5% in the orthogonal azimuthal angular plane (figure 3(b)).

Figures 4(a)–(d) show beam DPs of the MSD at 5 cm depth in 1 × 1 cm², 2 × 2 cm², 3 × 3 cm² and 4 × 4 cm² field sizes, respectively, compared to that of a micro-diamond detector and a silicon diode. For each field size, the profiles acquired with the three probes (downer panel) are represented together with the micro-diamond-to-MSD and micro-diamond-to-diode profile differences (upper panel). As the field broadens, the MSD shows an over-response at the tails of the profile (dose <20%) as compared to the micro-diamond detector.
Figure 3. Directional response of the MSD in (a) the radial ($y_0z$) and (b) the azimuthal ($x_0z$) angular planes. The response is normalized to 1 at 0°, i.e. when the beam direction points toward the floor. For a better view, schematics of the probe cross-sections along these two orthogonal planes (in light gray) are superimposed to the angular diagrams.

Figure 4. 6 MV field profile comparison between the MSD, a micro-diamond detector and a silicon diode. Profiles are measured at 5 cm depth in (a) $1 \times 1$ cm$^2$, (b) $2 \times 2$ cm$^2$, (c) $3 \times 3$ cm$^2$ and (d) $4 \times 4$ cm$^2$ fields (downer panels). Profile difference between detectors is reported for each field size (upper panels).
and silicon diode. Profile discrepancy between the MSD and the two other probes also starts to be observed right at the inner edge of the beam center (dose > 80%) for the larger field sizes (3 × 3 cm² and 4 × 4 cm²). Profile difference (in magnitude) across the scan is of 0.65% ± 0.80% between the MSD and the micro-diamond detector and 0.74% ± 0.77% between the silicon diode and the micro-diamond detector. When the field area is enlarged to 3 × 3 cm², the discrepancies of the MSD and the silicon diode regarding the micro-diamond detector are similar in the beam center (> 80%) and penumbra (80%–20%), but deviate in beam tails (< 20%). At beam center, differences are within 0.60% ± 0.43% and 0.54% ± 0.40%, respectively. In left penumbra, they become 1.82% ± 0.71% and 1.47% ± 0.78%, respectively (1.51% ± 0.78% and 1.38% ± 0.86%, respectively, in right penumbra). In the left beam tail, differences are within 5.00% ± 0.52% and 0.32% ± 0.30%, respectively (5.16% ± 0.39% and 0.44% ± 0.56%, respectively, in the right tail).

Figure 5 shows an analysis of the beam width measured with the MSD, the micro-diamond detector and the silicon diode. The beam width is assessed for all probes at 80%, 50% and 20% of the field maximum. Only the full width at 20% of maximum is impacted by the MSD over-response at the beam tail. For the 3 × 3 cm² beam, the full widths at 80% and 50% of maximum measured with the MSD and the silicon diode deviate from the micro-diamond response by 0.13 mm and 0.25 mm, respectively. Such differences between detectors in the measurement of the full width at 20% of maximum, which is within 0.3 mm with the silicon diode, increases to 0.9 mm with the MSD.

The percentage depth dose (PDD) curves from the three detectors are reported in figure 6. Figures 6(a)–(d) show PDDs measured from 0 to 15 cm depths in 1 × 1 cm², 2 × 2 cm², 3 × 3 cm² and 4 × 4 cm² field sizes, respectively. For each field size, the PDD acquired with the three probes (downer panel) are represented together with the micro-diamond-to-MSD and micro-diamond-to-diode PDD differences (upper panel). In the buildup region, PDD mismatch between the MSD and micro-diamond detector are within 4%, and 6% deviation is observed between the silicon diode and the micro-diamond detector. In the decay region, the maximum discrepancy between the micro-diamond detector and the MSD remains below 2.3% for the 1 × 1 cm², 2 × 2 cm² and 3 × 3 cm² field sizes. The corresponding mismatch between the micro-diamond detector and the diode is within 0.76%.

To evidence the full potential of the MSD, we plot on figure 7(a) a 2D field map of the smaller photon beam available with jaws on the TrueBeam irradiator (0.6 × 0.6 cm² field size). From a basic convolution-based imaging theory, we estimate the volume averaging of the MSD to artifactually broaden the detected profile by 0.0034 mm, i.e. 0.06% of the beam width (figure 7(b)).
Figure 6. Percentage depth dose curve measured with the MSD, the micro-diamond detector and the silicon diode in (a) $1 \times 1 \text{ cm}^2$, (b) $2 \times 2 \text{ cm}^2$, (c) $3 \times 3 \text{ cm}^2$ and (d) $4 \times 4 \text{ cm}^2$ fields (downer panels). Upper panels: PDD differences between the detectors.

Figure 7. (a) 2D field plot of a $0.6 \times 0.6 \text{ cm}^2$ beam in the vertical $(y0z)$-plane perpendicular to the MSD axis $(0x)$, see figure 1(d). (b) Theoretical $0.6 \text{ cm}$ Gaussian profile ($1/e$ width) and its convolution-based image with a $0.25 \text{ mm}$ large scanning probe.
The reproducibility of our approach is addressed in figure 8. We report the beam DPs of a $1 \times 1 \text{cm}^2$ field measured at 5 cm depth with three MSDs fabricated under the same conditions and with the same fabrication parameters. The three detectors show scintillation cells of the same length of $480 \mu\text{m}$ and whose widths vary from $250$ to $268 \mu\text{m}$ (see figure inset). The measured profile difference between the three detectors is below 1%.

3. Discussion

The MSD has been evaluated in small photon fields and compared to the micro-diamond detector and silicon diode. The dose linearity and the repeatability of the MSD are excellent and agree well with other scintillating probes such as the commercial W1 and W2 Extradin detectors (Pasquino et al 2017, Galavis et al 2019).

The 0.36% radial directional dependence (standard deviation) of the MSD witnesses a good probe axis-symmetry, which is a critical morphological parameter for beam profiling in a plane perpendicular to the detector axis. The higher variability (3.5%) of the detection regarding azimuthal beam angles may be due to the slight elongated shape of the scintillating cell along the MSD axis. It might also be due to magnitude variation of the Čerenkov signal relative to the scintillation signal, because more or less fiber is in the beam according to the azimuthal beam angle. However, the chromatic filter appears to be efficient enough to cancel this effect. Further investigation of this directional dependence is warranted. Note that the probe directional response is shown regarding the primary photon beam (in air) to benefit from environment isotropy as the gantry rotates around the detector. Given the energy response of the Gd$_2$O$_2$S:Tb (Alharbi et al 2019, 2020), we anticipate similar directional response in a tissue-equivalent environment.

Profile mismatch between the three detectors are within 2%–3% for the $1 \times 1 \text{cm}^2$ and $2 \times 2 \text{cm}^2$ field sizes, which proves that, even non corrected, the MSD accuracy is in the range of standard reference probes. The little profile mismatch between the MSD and the two other probes observed as the beam broadens (over-response at beam tail and slight under-response right at the beam edge) is likely to be due to the higher density of the scintillator. The larger atomic number of the scintillator constituents results in a higher sensitivity to low-energy photons as compared to water (Qin et al 2019) due to a higher photoelectric cross section (Rikner and Grusell 1985). Profile discrepancy could however be overcome by applying correction factors, as usually done for detectors. The measurement of the beam width at 80% and 50% of the maximum is however not affected by the water non-equivalence of the detector in the small field regime. These results tend to show that, despite correction factors to apply to the detector, the MSD is particularly well adapted to perform dose profiling in small photon fields.

Depth dose measurements realized with the MSD, micro-diamond detector, and silicon diode are quite consistent for beam sizes below $3 \times 3 \text{cm}^2$. In the buildup region, the differences between the three detector are on the same range regardless of the field size. In the decay region, the energy dependence of our water non-equivalent Gd$_2$O$_2$S:Tb based detector (Alharbi et al 2019, 2020) explains the appearance of an over-response for field sizes larger than $3 \times 3 \text{cm}^2$ field (Qin et al 2019). Such an over-response has been already found in the depth dose characterization with inorganic scintillator detectors (Molina et al 2013, Alharbi et al 2018). Here again,
better agreement between the non corrected MSD and the two reference standard probes is found for the smaller field sizes, with a PDD difference within 2.3% below \(3 \times 3\) cm\(^2\) field size. Even non corrected, the MSD accuracy is of the level of cutting-edge dosimeters. These results make the MSD very promising for small field dosimetry with high resolution and low detection footprint. Note that the discrepancy between the signals delivered by the commercial probes and the MSD just underneath the water surface may be due to the energy dependence of the Gd\(_2\)O\(_2\)S:Tb.

The MSD enables high resolution mapping of sub-centimeter photon fields. A longitudinal 2D map of such narrow beam provides a straightforward view of the strong dose inhomogeneities within the tiny irradiated volume. A complete set of dose data (profiles, PDD, etc) becomes readily accessible. Despite very small field size, volume averaging is predicted to be negligible. Our basic convolution-based theoretical model implies a probe inducing minimum electron transport alteration, a condition which is supposed to be fulfilled with our probe of extreme compactness. Over the area of the scintillation cell (about 0.1 mm\(^2\)), the water-equivalent thickness of our MSD is smaller than 1.7 mm. Across the optical fiber of diameter 0.125 mm, the water-equivalent depth is lower than 0.275 mm.

Despite modest technical facilities, our fabrication process of the MSDs lead to very similar detectors that generate field profiles whose differences are smaller than 1%. Automatizing the production of our probes would improve the reproducibility of the technique, especially in terms of the width of the scintillation cell which varies by 3.5% in our test of fabrication reproducibility.

4. Conclusion

In this work, we have evaluated the recently introduced concept of a MSD (Suarez et al 2019) in the dosimetry of small photon fields as used in stereotactic radiotherapy and radiosurgery. Miniaturization means here a detection volume typically eight times as small as that of the tighter PSD (Galavis et al 2019) and an optical fiber 4–5 times narrower than conventional plastic fibers. Owing to miniaturization, such a detector is expected to combine high spatial resolution and low electron fluence perturbation, even with inorganic water non-equivalent materials, due to its extreme compactness. This tiny inorganic scintillator detector has been shown to develop accuracy in DP and PDD acquisition at the level of two reference dosimeters (a micro-diamond detector and a silicon diode) though no correction factor is applied to the new dosimeter. For field size larger than \(3 \times 3\) cm\(^2\), some mismatches originating from secondary radiations of increased magnitude need to be corrected. This confirms that our inorganic detector is better suited to characterized small photon fields than the large fields used in conventional radiotherapy. Moreover, the MSD shows excellent detection linearity and repeatability as well as low directional dependence (which agree well with the organic W1 and W2 Extradin detectors Pasquino et al 2017, Galavis et al 2019). The production technique of our detector is also highly reproducible. As a potential application, 2D field map is plotted in a 0.6 \(\times\) 0.6 cm\(^2\) photon beam with a volume averaging predicted to be negligible. The MSD overcomes stem effect, and despite its water non-equivalence it is expected to minimize electron fluence perturbation due to its extreme compactness. Such a detector thus has the potential to become a valuable dose verification tool in small field radiation therapy. As a perspective, its field of application can also be extended to all radiation therapies based on steep dose gradients such as brachytherapy and in a longer term FLASH-radiotherapy and microbeam radiation therapy.

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