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Six-probe scintillator dosimeter for treatment verification in HDR-brachytherapy

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Abstract

Background: *In vivo* dosimetry (IVD) is gaining interest for treatment delivery verification in HDR-brachytherapy. Time resolved methods, including source tracking, have the ability both to detect treatment errors in real time and to minimize experimental uncertainties. Multiprobe IVD architectures holds promise for simultaneous dose determinations at the targeted tumor and surrounding healthy tissues while enhancing measurement accuracy. However, most of the multiprobe dosimeters developed so far either suffer from compactness issues or rely on complex data post-treatment.

Purpose: We introduce a novel concept of a compact multiprobe scintillator detector and demonstrate its applicability in HDR-brachytherapy. Our fabricated seven-fiber probing system is sufficiently narrow to be inserted in a brachytherapy needle or in a catheter.

Methods: Our multiprobe detection system results from the parallel implementation of six miniaturized inorganic $Gd_2O_2S:Tb$ scintillator detectors at the end of a bundle of seven fibers, one fiber is kept bare to assess the stem effect. The resulting system, which is narrower than 320 microns, is tested with a MicroSelectron 9.14 Ci Ir-192 HDR afterloader, in a water phantom. The detection signals from all six probes are simultaneously read with a sCMOS camera (at a rate of 0.06 s). The camera is coupled to a chromatic filter to cancel Cerenkov signal induced within the fibers upon exposure. By implementing an aperiodic array of six scintillating cells along the bundle axis, we first determine the range of inter-probe spacings leading to optimal source tracking accuracy (first tracking method). Then, three different source tracking algorithms involving all the scintillating probes are tested and compared. In each of these four methods, dwell positions are assessed from dose measurements and compared to the

39 treatment plan. Dwell time is also determined and compared to the treatment plan.

40

41 **Results:** The optimum inter-probe spacing for an accurate source tracking ranges from
42 15 mm to 35 mm. The optimum detection algorithm consists of adding the readout
43 signals from all detector probes. In that case, the error to the planned dwell positions
44 is of 0.01 ± 0.14 mm and 0.02 ± 0.29 mm at spacings between the source and detector
45 axes of 5.5 and 40 mm, respectively. Using this approach, the average deviations to the
46 expected dwell time are of -0.006 ± 0.009 s and -0.008 ± 0.058 s, at spacings between
47 source and probe axes of 5.5 mm and 20 mm, respectively.

48

49 **Conclusions:** Our six-probe $Gd_2O_2S:Tb$ dosimeter coupled to a sCMOS camera can
50 perform time-resolved treatment verification in HDR brachytherapy. This detection
51 system of high spatial and temporal resolutions (0.25 mm and 0.06s, respectively)
52 provides a precise information on the treatment delivery via a dwell time and position
53 verification of unmatched accuracy.

54

55

56 *This is a sample note.*

57

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1. Introduction

High dose rate brachytherapy (HDR-BT) is a standard modality in cancer treatment which offers advantages of highly localized dose distributions and minimum number of treatment fractions.^{1,2,3,4} To ensure that the planned dose is properly delivered, time-resolved in vivo dosimetry (IVD) has been proposed for monitoring treatments and detecting errors^{5,6,7,8,9}. Among time-resolved IVD approaches, optical fibers coupled to scintillators have shown promise in time-resolved verification of the dose rate^{10,11,12,13,14,15,16,17}, as well as dwell position and dwell time monitoring of a stepping radioactive source^{7,15,18,19,20}.

IVD in multiprobe architectures has recently attracted attention for its ability to increase the spatial extent of treatment monitoring to volumes including the targeted tumor and surrounding healthy tissues. By use of individual detectors in various parallel catheters, Wang et al.²¹ and Guiral et al.²² performed extended source tracking along the source catheter. Cartwright et al.²³ implemented a source tracking with an array of 16 plastic scintillator dosimeters embedded in a 20 mm-diameter rectal applicator. However, the acquisition rate of the detector was limited to 1 s, short dwell times of the source could not be assessed. Moreover, the diameter of the resulting multiprobe dosimeter in the centimeter range limits its field of application in BT. Therriault-Proulx et al.²⁴ developed a three-probe plastic scintillator detector sufficiently narrow to be inserted within a BT needle or catheter, thereby making multiprobe tracking applicable in a broader range of BT. Because they involve one-millimeter outer diameter fibers, the three scintillator probes were engineered on the same fiber to be insertable into a BT needle or catheter. In this specific setup, where the luminescence from all three scintillators travels through the same physical path within the fiber, the optical signals are entangled at the fiber output. By analyzing the optical signal with a spectrometer, dosimetry has been successfully demonstrated with a detection rate as large as 3 seconds. A recent study has introduced a hyper-spectral optical detection system enabling time-resolved dose rate monitoring as well as dwell position and dwell time verification with such a fiber detector^{19,25}. To determine the dose received by each individual scintillator from the detected optical signals, a calibration process based on the AAPM TG-43 dose parameters²⁶ has been implemented²⁵. The detected optical signals were combined in a 4×4 linear equation system to obtain a detector overall response that is free from the stem effect^{25,27}. With this system, triangulation approaches for source tracking

116 were demonstrated.

117 In this paper, we use the miniaturized scintillator detector (MSD) approach^{28,29,30} to
118 demonstrate a seven-channel multiprobe detector that is narrow enough to perform time-
119 resolved treatment monitoring within a single BT needle or catheter. Our 320-micron outer
120 diameter fiber detector consists of 6 scintillating probes and a bare test-fiber, engineered
121 at the end of a narrow seven-fiber bundle. The parallel measurement of the seven optical
122 signals at the bundle output with a sCMOS camera avoids inter-probe cross-talk, i.e., signal
123 entanglement at the bundle output. As a result, the calibration of our multiprobe detector
124 does not require the use of the AAPM TG-43 algorithm. Moreover, Cerenkov signal is simply
125 removed by positioning a bandpass filter in front of the camera. Each MSD of the detection
126 system ensures minimum volume averaging within the steep dose gradients of BT sources,
127 leading to unmatched source tracking performances in space and time.

128 II. Material and Methods

129 II.A. Multiprobe system

130 The multiprobe detector (MPD) shown in Fig. 1(b) involves a 10-meter-long bundle of seven
131 biocompatible fibers arranged in a hexagonal lattice (cf. Fig. 1(a); fabricated by SEDI-ATI).
132 Each fiber is of 80-micron outer diameter (50-micron core diameter) and is covered with a 5-
133 micron-thick polyimide protective coating. The total width of the bundle is of 270 microns.
134 Each fiber tip is tapered in the form of a leaky-wave nano-optical antenna^{28,31} aimed at
135 improving the transfer of the X-ray excited luminescence from the scintillators to the fiber.
136 Scintillating powder ($\text{Gd}_2\text{O}_2\text{S:Tb}$) is locally attached to the tapered tip of six of the seven
137 fibers to form the probes P1-P4, P6 and P7 (see Fig. 1(b)). $\text{Gd}_2\text{O}_2\text{S:Tb}$ is chosen as the
138 scintillating material owing to its high scintillation efficiency, stability, linearity and fast
139 temporal response^{32,33,34} and with very low sensitivity to temperature (in the range of 15°-
140 40°)^{35,36}. This inorganic scintillator shows an energy dependence³⁰ that need to be corrected
141 if a direct dose rate measurement, rather than a source tracking, is targeted. The last bare
142 fiber, labelled as P5, is used to evaluate the level of spurious Cerenkov signal generated
143 within fibers upon irradiation. The overall fabrication process is detailed in Refs.^{29,30}. The
144 MPD is locally enlarged to diameters around 320 microns by the presence of the scintillating

145 cells (see Fig. 1(c)). The scintillation cells forming the six parallel probes are shown in
146 Fig. 1(d). The scintillation volume varies from 0.008 mm^3 (P1) to 0.009 mm^3 (P3), due
147 to slight imperfections in our fabrication process, in terms of reproducibility. Four different
148 inter-probe spacings along the bundle axis are defined by adapting the length of each optical
149 fiber of the bundle. P1 and P2, P2 and P3, and P4 and P6 are spaced by Δ , 2Δ , and 4Δ ,
150 respectively, where $\Delta = 5 \text{ mm}$, whereas P3 and P4 as well as P6 and P7 are both spaced
151 by 1.7Δ . Interprobe spacings are defined with an accuracy of 0.3 mm . Measurements were
152 realized by positioning the MPD onto a calibrated motorized stage coupled to a binocular
153 equipped with a crosshair. The fiber bundle is positioned within a black 0.9-mm hytrel
154 cladding to minimize collection of the background light from the test room. This opaque
155 shield stops about 10 cm before the first probe P1 so that all six probes plus the bare fiber
156 are directly in contact to the water phantom. To avoid contribution from external spurious
157 light, the experiments are carried out in the dark. In addition, an opaque cover is used to
158 protect the water phantom from residual room light.

159 II.B. Optical readout

160 The optical signals at the end of the MPD are simultaneously recorded with a sCMOS camera
161 (Andor Technology, Zyla 4.2 model) whose maximum detection yield spectrally matches
162 the emission of the $\text{Gd}_2\text{O}_2\text{S:Tb}$ material. A 35 mm camera objective (Fujinon HF35SA)
163 is positioned in front of the camera to image the bare output face of the fiber bundle at
164 a rate of 0.06 s. Prior to acquisitions, we define seven regions of interest (ROI) tightly
165 enclosing the seven light spots that are observable in the image (one spot per probe, see
166 the green circles in Fig. 1(e) delimiting the ROIs). Our code, developed under Labview
167 environment, automatically defines the ROIs as 16-pixel diameter disks centered with respect
168 to the maximum signal. The image pixels located within each ROI are integrated to obtain
169 seven detection signals sampled at 0.06 s. A chromatic filter (544/24 nm band pass filter
170 from Semrock) is positioned in front of the camera to filter out the spurious Cerenkov signal
171 (stem effect) generated in the fibers upon exposure.²⁹

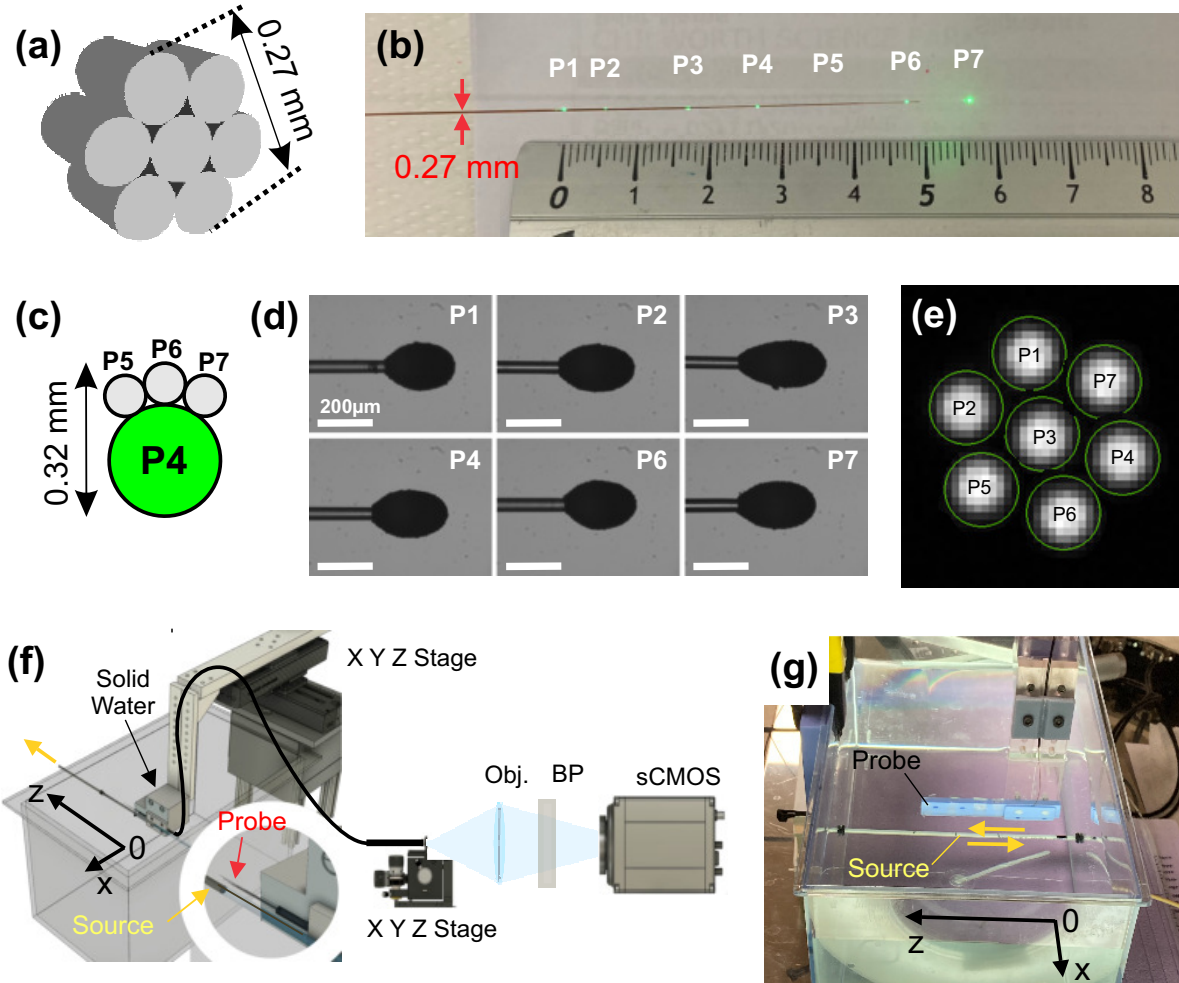


Figure 1: (a) Schematic of the bundle of 7 fibers arranged in an hexagonal lattice. (b) Photograph of the multiprobe detector. Green laser light is coupled to the free bare facet of the fiber bundle to identify in the image the six scintillation cells (six green spots are observed due to light scattering of the fiber modes by the scintillators). (c) Schematic of a cross-section of the MPD at the location of probe P4 (cf. (b)). The green and gray disks represent the cross-sections of the scintillating cell and fibers, respectively. (d) Magnified optical images of the six scintillation cells at the end of the fiber bundle. (e) Image of the bare face of the fiber bundle by the sCMOS camera when white light is projected onto the seven probes. (f) Schematics of the experimental setup involving a water tank, the multiprobe detector positioned onto a 2D motorized stage and a photometer based on a sCMOS camera coupled to an objective (Obj.) and a band-pass filter (BP). (g) Photograph of the experimental setup.

172 II.C. Brachytherapy system

173 A MicroSelectron afterloader with a 9.14 Ci Ir-192 HDR source (Air kerma strength of 37309
174 U) is used for irradiation.

175 II.D. Phantom

176 The probe characterization is conducted in a 40x30x30 cm³ water tank. The source catheter
177 crosses the tank widthwise (Figs. 1(f) and (g)). During experiments, temperature in the
178 water phantom varies from 17° to 19° (temperature assessment with a thermometer before
179 and after the experiments). The MPD is fixed to a solid-water holder that is attached to a
180 2D translation stage via a plastic adaptor (Figs.1(f) and (g)). The fiber probe is set parallel
181 to the source catheter. A coordinate frame of the set-up is defined so that the origin of the
182 frame coincides with the scintillators of the proximal probe *P1*. The source-probe spacing
183 along the (0x) and (0z) axes is determined with the motorized stage and the afterloader,
184 respectively.

185 II.E. Detector calibration

186 The MPD is calibrated along seven lines parallel to the (0z)-axis. These lines are spaced
187 by 5.5, 8, 10, 15, 20, 30 and 40 mm (along (0x)) from the axis of the source catheter.
188 First, the MPD is positioned at the desired source-probe inter-catheter spacing x with the
189 motorized stage. Then, the source is displaced along the fixed source catheter by 2.5-mm
190 steps. The calibration curves are obtained by integrating 165 images per source position.
191 An interpolation (performed with the "interp" function of Matlab software) is applied to
192 the measured profiles to obtain a 0.1-mm sampling rate. During calibration, we verify with
193 the scintillator-free fiber probe P5 that no optical signal (stem effect) is detected with the
194 chromatic filter positioned in front of the camera.

195 The signal-to-noise ratio (SNR) of the six scintillating probes forming the detector is
196 assessed at the dwell positions corresponding to the maxima of all the above-mentioned
197 calibration curves, at the seven source-probe spacings x ranging from 5.5 mm to 40 mm.
198 The SNR is the average amplitude of the signal divided by its standard deviation.

199 II.F. Dwell position and dwell time verification

200 II.F.1. Measurements

201 To test the MPD, an irradiation protocol consisting of 40 dwell positions is applied for
 202 each value of source-probe spacing along ($0x$). The dwell positions are spaced by 2.5 mm
 203 and the dwell time is fixed to 10 s. Note that 2.5-mm is the minimum displacement step
 204 allowed by our afterloader. With a source tracking accuracy of 0.023 ± 0.077 mm³⁰, our sub-
 205 millimeter scintillating probes are suitable to accurately track shorter sources steps (e.g., 1
 206 mm steps). The dwell time value has been fixed to 10s for a direct comparison of our source-
 207 tracking results with those of Linares et al.¹⁹. Note that our scintillating probes have already
 208 demonstrated their effectiveness in monitoring dwell times ranging from 0.2s to 11s.³⁰.

209 II.F.2. Source position monitoring

210 The instant position of the source at each acquisition time is retrospectively determined from
 211 the output signals of the MPD and the source activity using the calibration curves presented
 212 in section II.E.. To ensure that the transitory phases between two successive dwell positions
 213 are not taken into account in the source tracking, the first and the last signal points for each
 214 dwell position are ignored. These transitory phases, which last a few tens of milliseconds³⁷,
 215 correspond to the displacement of the radioactive source between two dwell positions.

216 During a treatment delivery, each probe j ($1 < j < 7$) of the MPD delivers a temporal
 217 signal $S_j(t)$. At each instant $t = k\tau$, where $k \in \mathbb{N}$ and τ is the acquisition time of the
 218 camera, the instant source position is deduced from the readout signal S_j as follows.

219 First, function f_j^k is defined for each probe j as:

$$220 \quad f_j^k(z) = |C_j(z) - S_j^k|, \quad (1)$$

221 where $C_j(z)$ and S_j^k are the calibration curve and the readout signal of the j^{th} probe at
 222 the k^{th} time step, respectively. z corresponds to the spatial coordinate along the axis of the
 223 source catheter. Calibration curves being symmetric regarding the z -coordinate, each probe
 224 provides two likely instant source positions located on both sides of function f_j^k . Therefore,
 225 at minimum two probes are necessary to unambiguously determine the position of a stepping

226 BT-source.

227 The instant source position Z^k is determined using four different methods involving
 228 various manipulations of functions f_j^k . To find the inter-probe spacing which optimizes source
 229 tracking accuracy (for detector design purpose), source position verification is realized from
 230 a "two-probe" dosimetry using Eq. 2 ($m = 1$). $j_1 \in [1, 6] \setminus 5$ and $j_2 \in [2, 7] \setminus 5$ are the indices
 231 of the probes forming the 15 probe pairs (j_1, j_2) allowed by our detector. The inter-probe
 232 spacing ranges from 5 mm to 52 mm. Probe P5, which is bare to assess in-fiber Cerenkov
 233 effect, is not involved in the source position monitoring. Source tracking is systematically
 234 analyzed from each of the 15 probe pairs of the detector.

$$235 \quad Z_m^k(z) = \min \left[f_{j_1}^k(z) + f_{j_2}^k(z) \right], \quad (2)$$

236 We also tested and compared three different source tracking algorithms. In each case,
 237 all the six scintillating probes are involved in the source position monitoring during the
 238 treatment delivery. The two first values of the instant source position (Z_m^k , $m = 2$ and 3)
 239 are calculated from the readout signals of probe pairs dynamically chosen among the six
 240 available probes of the MSD. In both of these two-probe measurements, the source position
 241 is defined from Eq. 2 with indices j_1 and j_2 which vary with the source position along (0z).
 242 Probe pairs are dynamically chosen to provide the higher readout signals ($m=2$) or on the
 243 basis of a maximum gradient of their calibration curves at the source position ($m=3$). The
 244 last z -coordinate of the source Z_4^k is determined by adding functions f_j^k of all seven probes.
 245 We have:

$$246 \quad Z_4^k(z) = \min \left[\sum_{i=1}^7 f_j^k(z) \right], \quad (3)$$

247 II.F.3. Dwell time verification

248 The monitoring of an HDR-BT treatment is known to produce a staircase temporal sig-
 249 nal^{14,15,29}. The dwell times of the stepping source, which correspond to the duration of the
 250 plateaus in between two successive signal edges, can be simply determined from an edge de-
 251 tection within all readout signals of our MPD system. Our edge detection approach involves
 252 function F^k defined as:

$$F^k = \sum_{j=1}^6 (S_j^{k+1} - S_j^k), \quad (4)$$

By adding the signal derivatives from all probes, our algorithm is expected to reduce undesired fluctuations in the edge detection function (due to readout noise), as compared to that of a single probe detector.

III. Results

III.A. Detector specification and calibration

The specification of the scintillating probes forming the MPD can be found in Ref. ³⁰. The linearity coefficient of the probes exceeds 0.999 regardless of the source-probe spacings x . The deviation to repeatability remains below 2% for all values of x . The energy dependence of the fiber probes forming the MPD, referred to as the single-probe MSD, has already been characterized in a past study³⁰. Figure 2(b) shows the calibration curves of the MPD acquired at source-probe spacings x of 10, 20 and 30 mm. Six gaussian-like profiles are shown per source-probe inter-catheter spacing (one profile per probe), whose maxima coincide with the probe positions along (0z), as depicted in Fig. 2(a). The SNR of the probes forming the detector varies from 110-140 down to 20-25 at source-probe spacings x of 5.5 mm and 40 mm, respectively.

III.B. Source position monitoring

III.B.1. Optimal probe-to-probe spacing

The aperiodic scintillator array of our MPD enables 15 probe pairs whose inter-probe spacings vary from 5 mm to 52 mm. To identify optimal inter-probe spacings for the future MPD designs, we analyzed the accuracy of the source position verification over these 15 probe pairs, versus the inter-probe distance δz along the detector axis (0z) (see Fig. 3). The instant source position is determined from Z_1^k function, cf. Eq. 2 ($m = 1$). The displacement range of the BT source along (0z) varies with the inter-probe distance δz as $\delta z + 2 * 0.75$ cm (see inset of Fig. 3(b)).

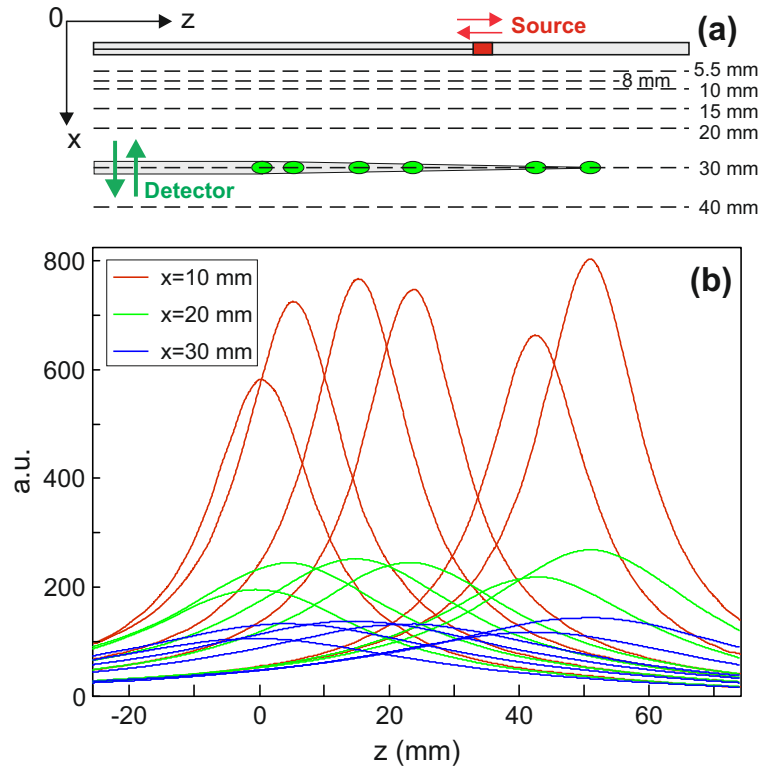


Figure 2: (a) Schematic of the experimental set-up in the water tank. The coordinate frame defining the axis convention is shown in the top-left corner. The source and MPD are represented in red and green, respectively. All seven positions of the MPD defining source-probe spacings x ranging from 5.5 to 40 mm are represented with dashed lines. The source and MPD move along the $(0z)$ and $(0x)$ axes, respectively. (b) Calibration curves of the MPD used for source tracking at x equal to 10, 20 and 30 mm.

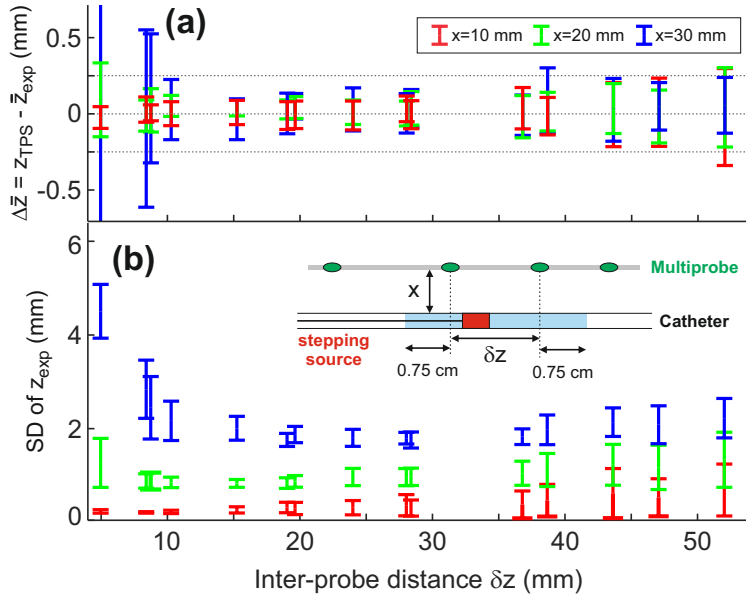


Figure 3: Optimal inter-probe spacing for source tracking. (a) Offset between the measured and planned dwell positions along ($0z$) (\bar{z}_{exp} and z_{TPS} , respectively) as a function of the inter-probe distance δz . The measured dwell position \bar{z}_{exp} corresponds to the average of the instant source position Z_1^k (see Eq. 2) over a dwell time. Each error bar shows for one source-probe inter-catheter spacing x (see inset of (a)) the offset analysis to the planned dwell positions (mean and standard deviation). This analysis is performed over z -coordinates spanning over $\delta z + 2 * 0.75$ cm (see inset of (b)). (b) SD of the experimentally determined source position Z_1^k versus the inter-probe distance δz . Each error bar shows the distribution (mean and standard deviation) of the SD for dwell positions within $\delta z + 2 * 0.75$ cm (see inset of (b)), at a given source-probe inter-catheter spacing x (see inset of (a)).

278 We see from Fig. 3(a) that the minimum deviation to the planned dwell positions
 279 occurs at δz values in-between 15 mm and 35 mm, regardless of the source-probe inter-
 280 catheter spacing x . In that δz range, the offset distribution to the planned dwell positions
 281 does not exceed 0.05 ± 0.15 mm at $x=30$ mm (0.15 ± 0.41 mm at $x=40$ mm). The SD of the
 282 measured instant source position Z_1^k reaches minimum values at δz in between $0.87x$ and x
 283 (see Fig. 3(b)). This property, which is observed for all values of x ranging from 5.5 mm to
 284 40 mm, is imputed to the broadening of the calibration curves along ($0z$) as the source-probe
 285 inter-catheter spacing x increases (cf. Fig 2). The tighter distributions of SD are of $0.2 \pm$
 286 0.022 mm, 0.82 ± 0.12 mm and 1.78 ± 0.12 mm at $x = 10, 20$ and 30 mm, respectively.

III.B.2. Source tracking: two-probe versus all-probe detection strategies

A six-probe detector enables numerous detection strategies for source tracking. In Fig. 4, we compare three algorithms which involve dosimetry either from probe pairs dynamically chosen during the treatment delivery (cf. Z_2^k and Z_3^k of Eq. 2) or from all the probes of the detector (cf. Z_4^k of Eq 3).

At source-probe spacings x below 15 mm, all three source-tracking algorithms show similar accuracy. The mean and standard deviation of the offset to the planned dwell positions do not vary by more than 0.008 and 0.014 mm, respectively, from one method to another (Fig. 4(a)). With the two higher signal method (cf. Z_2^k of Eq. 2), the deviation to the planned dwell positions, which is of 0.007 ± 0.138 mm at $x=20$ mm, increases up to 0.20 ± 1.12 mm at $x=40$ mm. As a comparison, the dwell position verification from the two signals of steeper gradients (calculation of Z_3^k) leads to a mismatch to the treatment plan of 0.027 ± 0.115 mm and -0.11 ± 0.68 mm at x equal to 20 and 40 mm, respectively. Source tracking from all detected signals (calculation of Z_4^k) is much less impacted by the enhancement of the source-probe inter-catheter spacing x . The offset to the planned dwell positions is of 0.029 ± 0.078 mm at $x=20$ mm and 0.02 ± 0.19 mm at $x=40$ mm, respectively.

The SD of the instant source position determined from the three above-mentioned methods is reported in Fig. 4(b). For source-probe distances below 15 mm, all three methods determine the instant source positions with almost the same accuracy. When x exceeds 20 mm, the dwell position verification from the two higher detected intensities (calculation of Z_2^k) is the less accurate. A detection from all probes (cf. Z_4^k) minimizes signal fluctuations.

Fig. 5 displays a detailed representation of the source position determined from all scintillating probes, which correspond to the analysis shown in Fig. 4 (cf. black error bars). Fig. 5(a) reports the error to the planned dwell positions, which corresponds to the difference between the planned and measured dwell positions (z_{TPS} and \bar{z}_{exp} , respectively). Fig. 5(b) reports the SD of the instant source position Z_4^k , i.e., the SD of the distribution of source positions measured at a rate of 0.06s during a dwell time. Noticeable enhancement of the SD is observed when the source is positioned out of the region where the probes are located, i.e., in between the detector and probe P_1 ($\bar{z}_{exp} < 0$) or beyond probe P_7 ($\bar{z}_{exp} > 5.2$ cm; cf. Fig. 1(b) and Fig. 2). This local fluctuation enhancement is maximum when $x=5.5$

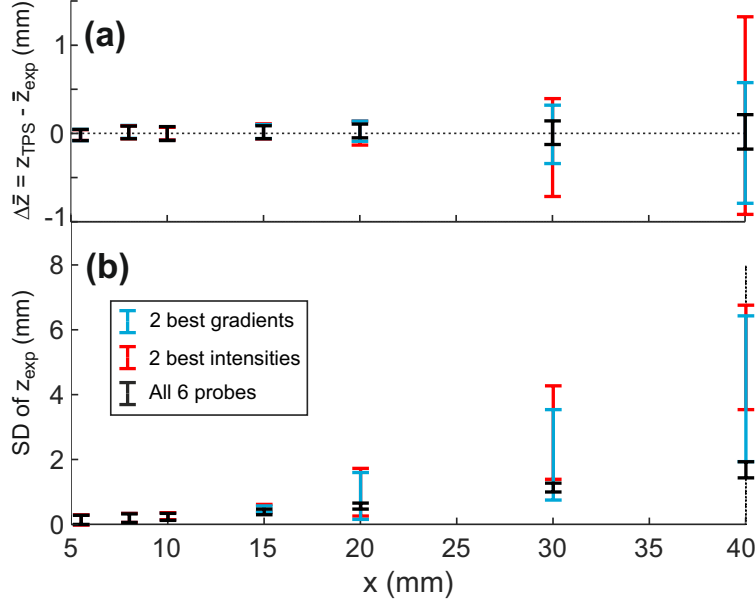


Figure 4: Source tracking: two-probe versus all-probe algorithms. (a) Offset between the measured and planned dwell positions along ($0z$) (\bar{z}_{exp} and z_{TPS} , respectively) as a function of the source-probe inter-catheter spacing x . The measured dwell position \bar{z}_{exp} corresponds to the average of the instant source position over a dwell time. The instant source position corresponds to Z_2^k , Z_3^k or Z_4^k (cf. Eqs. 3 and 2; see inset of (b)). Each error bar shows for one algorithm (see inset of (b)) the analysis of the offset to the planned dwell positions (mean and SD). This analysis is performed over a range of source positions along ($0z$) of 10 cm. (b) SD of the experimentally determined instant source positions Z_2^k , Z_3^k and Z_4^k . Each error bar shows for one algorithm (see inset of (b)) the analysis (mean and SD) of the distribution of SD for dwell positions spanning over 10 cm (along ($0z$)).

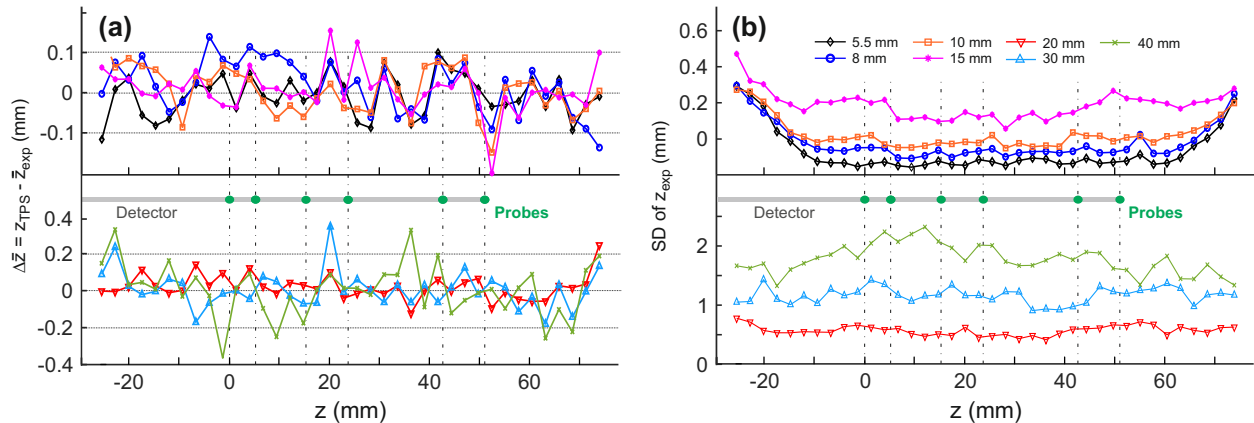


Figure 5: Analysis of an "all-probe" source tracking. (a) Offset between the planned and experimentally determined dwell positions along ($0z$) (z_{TPS} and \bar{z}_{exp} , respectively). \bar{z}_{exp} is the average of the measured instant source position Z_4^k over a dwell time. The mismatch to the planned dwell positions is shown for seven values of the source-probe inter-catheter spacing x (see legend of (b)). (b) SD of the measured instant source position Z_4^k along ($0z$) as a function of the z -coordinate. Here again, seven source-probe spacings x are considered (see legend). Insets of (a) and (b), schematic of the multiprobe detector which identifies the z -coordinates of the scintillators on the graphs (with dashed lines).

317 mm and lessens as the source-probe spacing x increases, to finally vanish at $x=20$ mm. On
 318 the contrary, the offset to the planned dwell positions remains at the same level over the
 319 entire displacement range of the stepping source (i.e., 10 cm), regardless of the source-probe
 320 spacing x (cf. Fig 5(a)).

321 III.C. Dwell time verification

322 In Fig. 6, we report the analysis of the offset ΔT between the measured and planned dwell
 323 times (T_{exp} and T_{TPS} , respectively) as a function of the source probe inter-catheter spacing
 324 x . The error bars show the mean and SD of the offset to the planned dwell times (calculated
 325 over 40 dwell positions). The average of ΔT is estimated to be -0.006 ± 0.009 s.

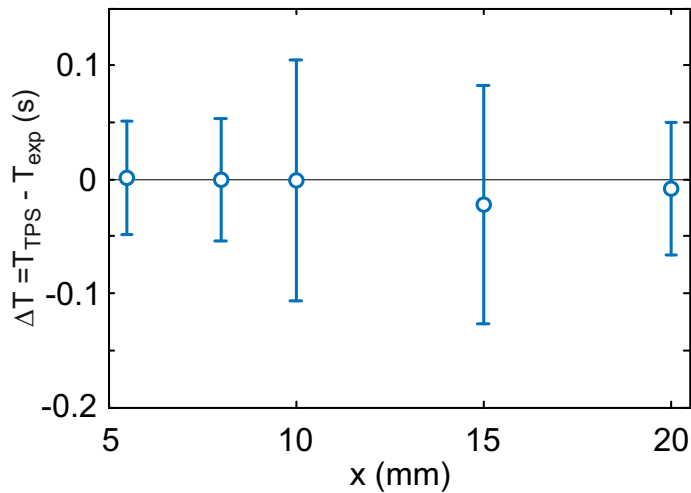


Figure 6: Mismatch ΔT between the measured and planned dwell times (T_{exp} and T_{TPS} , respectively) as a function of the source-probe spacing x .

326

IV. Discussion

327

IV.A. Detector specification and calibration

328

Our MPD shows an acceptable SNR for an accurate dwell time and position verification

329

(larger than 20). At a source probe spacing x of 20 mm, Linares et al. reported a SNR

330

ranging from 13 to 23 with their multipoint detector system involving three photomultiplier

331

tubes (PMT) as photometers²⁵. At the same source-probe spacing, we measure a SNR

332

spanning from 33 to 39 with a sCMOS camera, a detection rate of 0.06 s and sensitive

333

volumes that are 250 to 650 times smaller than those of Linares's multipoint detector²⁵.

334

SNR could even be enhanced either by slightly broadening the scintillation cell and the fiber

335

core (cf. Ref.³⁰) or by replacing the sCMOS camera by six PMTs which provide higher light

336

detection sensitivity and an acceptable detection speed. However, six (or seven) PMTs may

337

be more expensive than a sCMOS camera. Moreover, the performance of these ultrasensitive

338

photometers can be seriously reduced after misuse (such as, e.g., exposing the PMT to room

339

light). Note that the range of SNR observed across the probe array for each source-probe

340

spacing x is due to small discrepancies in scintillation volume and in-fiber luminescence

341

coupling efficiency of the probes. We did not observe noticeable signal variation of the

342

MPD with the slight temperature drift in the water phantom during our experiments, thus

343

confirming the very low sensitivity of the $Gd_2O_2S:Tb$ to changes in temperature^{35,36}.

344

Beyond the opportunity of inserting a seven-probe detector within a brachytherapy

345 needle, the advantage of the miniaturization of a water nonequivalent fiber dosimeter resides
 346 in a smaller volume averaging effect in the strong dose gradients of the radioactive source,
 347 as well as a minimum electron fluence perturbation.

348 Miniaturization does however not resolve the energy dependence of inorganic scintil-
 349 lator detectors. In contrast to the multipoint detector approach based on organic materi-
 350 als^{19,24,25,38}, our water non-equivalent multiprobe system requires a space-dependent correc-
 351 tion factor for a direct accurate determination of the dose rate at a measured dwell position³⁰.
 352 To overcome this issue, our strategy is to define a dose rate value from the dwell position and
 353 source activity by use of the AAPM TG-43 dose parameters²⁶. This indirect dosimetry can
 354 be seen to bring the required correction factor³⁰. Note that the space-dependent calibration
 355 of the organic multipoint detector also relies on the AAPM TG-43 formalism²⁵.

356 IV.B. Dwell position verification

357 IV.B.1. Optimal probe-to-probe spacing

358 At an interprobe spacing δz of 36.8 mm and at a source-probe spacing $x=5.5$ mm, our MPD
 359 used as a two-probe detector ensures a deviation to the planned dwell positions of $0.03 \pm$
 360 0.15 mm. We followed the measurement process depicted in Fig. 3(b), with source positions
 361 expanding over 51.8 mm along ($0z$) (i.e., $36.8 + 2 * 7.5$ mm). As a comparison, the deviation
 362 to the planned dwell positions is estimated to be 0.45 ± 0.3 mm (1SD) with the three-
 363 probe fiber detector of Linares et al¹⁹, at a source-probe spacing $x = 5$ mm and a source
 364 activity of 10.73 Ci. Linares's detection system consists of three 10-mm spaced scintillation
 365 cells integrated at the end of an individual fiber detection line. Two reasons may explain
 366 the higher source-tracking accuracy of our two probe device, despite detection efficiencies
 367 of equivalent SNR. First, the sensitive volume of our probes is two orders of magnitude
 368 smaller, thereby limiting the averaging effect in the steep dose gradients near an HDR-BT
 369 source. Second, the source tracking accuracy of Linares's system may be constrained due to
 370 the utilization of non-exclusively experimental calibration, which incorporates the AAPM
 371 TG-43 formalism²⁵.

372 With a 21-mm spaced four-probe detection system, Guiral et al demonstrated a mis-
 373 match to the planned dwell positions of 0.11 ± 0.7 mm at a 0.1 s detection rate and along a

374 60 mm portion of source catheter. At a similar source probe spacing estimated to be of 20
 375 mm, we find a deviation to the planned dwell positions of 0.03 ± 0.14 mm over dwell positions
 376 expanding over 50 mm. Our sensitive volume being 45 times smaller than Guiral et al.'s²²,
 377 volume averaging effect is reduced, thereby improving dwell position verification.

378 IV.B.2. Source tracking: two-probe versus all-probe detection strategies

379 Cascading scintillating probes along the source catheter allows for extending optimum source
 380 tracking capabilities over longer source paths. The source tracking accuracy reported above
 381 can be improved when the probe pair is dynamically chosen among the six available probes of
 382 the detector to follow the source during the treatment delivery (all six probes are sequentially
 383 involved in the treatment monitoring). The detection accuracy for the "two best gradients"
 384 surpass that of the "two best intensities" algorithm. At a 10 s dwell time, source tracking
 385 accuracy from these two algorithms is only slightly better than for the fixed two-probe
 386 system but the gap should be noticeably enhanced for shorter dwell times, especially the
 387 sub-second regime. The two investigated algorithms could also show improved performances
 388 if a constant interprobe spacing was considered.

389 Source tracking accuracy is noticeably enhanced when the source position monitoring
 390 is done via the accumulation of all the six readout signals of the detection system. In
 391 that case, the deviation to the planned dwell positions is reduced to -0.017 ± 0.063 mm and
 392 0.029 ± 0.078 mm at the source-probe spacings x of 5.5 and 20 mm, respectively. At $x = 5.5$
 393 mm, we observe a maximum offset to the treatment plan of 0.12 mm. As a comparison,
 394 Linares et al reported a maximum deviation of 1.8 mm at $x = 5$ mm and at similar source
 395 displacement range (10 cm) and dwell time (10s)¹⁹. At $x = 20$ mm, Guiral et al found a
 396 discrepancy to the treatment plan of 0.11 ± 0.7 mm with their four-probe detection system
 397 (at a dwell time of 5s)²². The accuracy of the MPD in dwell position verification is fully
 398 compatible with the requirements of HDR brachytherapy in terms of medical treatment and
 399 quality assurance^{1,7,39,40}.

400 Over the entire displacement range of the stepping source (i.e., 10 cm) and at a dwell
 401 time of 10 s, the "all-probe" detection approach provides a dwell position verification that is
 402 not affected by the fluctuations of the instant source position \bar{z}_{exp} , whatever the source-probe
 403 spacing x . Although the SD of \bar{z}_{exp} varies by one order of magnitude during the treatment at

404 $x = 5.5$ mm (Fig. 5(b)), the dwell position measurement remains at the same accuracy level
 405 (Fig. 5(a)). The offset to the planned dwell position shows a narrow distribution of 0.002
 406 ± 0.049 mm and a maximum value which does not exceed 0.12 mm. Such performances
 407 exceed those of competing multipoint scintillator detectors¹⁹. Note that all these results are
 408 obtained at a noticeably long dwell time. Measurements within a range of shorter dwell times
 409 will be studied in the future. A measurement accuracy of 0.023 ± 0.077 mm has already
 410 been shown with a single MSD for dwell times ranging from 0.1 to 11 s³⁰.

411 IV.C. Dwell time verification

412 Across the source-probe spacings x ranging from 5.5 to 20 mm, the total average of the
 413 differences ΔT between the measured and planned dwell times is of -0.006 ± 0.009 s, at a
 414 detection rate of 0.06 s. At $x = 20$ mm, ΔT is found to be -0.008 ± 0.058 s. With their
 415 four-probe detector, Guiral et al measured a difference ΔT of 0.05 ± 0.9 s at approximately
 416 similar source-probe spacing, a detection rate of 0.1 s and for 5-s dwell times²². In their
 417 study, the source positions expanded over 6 cm within the source catheter, rather than 10 cm
 418 as in our case. At $x = 5.5$ mm, we find an offset ΔT of 0.0009 ± 0.0497 s. As a comparison,
 419 Linares et al reported an offset to the planned dwell time of 0.33 ± 0.37 s at $x = 5$ mm and
 420 a dwell times of 1s.

421 It is noteworthy that the measurement errors of the dwell time discussed here are all
 422 obtained from an edge detection in a staircase detection signal. In all the proposed methods,
 423 dwell times are defined as the elapsed time in between two successive signal edges, which
 424 are identified from a signal derivative calculation. Since all the considered dwell times in the
 425 above-cited studies are at least 50 fold longer than the integration time of the photometers
 426 used, one can assert that the measurement accuracy will not significantly change as the dwell
 427 time increases. We recently verified this property in a single probe detection³⁰. Therefore,
 428 the results from Guiral's and Linares's multiprobe detectors obtained with 5 s and 1 s dwell
 429 times, respectively, can be directly compared to our results measured at a dwell time of 10
 430 s. As an example, the lower measurement accuracy of Linares's approach may be partly
 431 explained by their shorter 1 mm inter-dwell spacing, leading to noticeably smaller temporal
 432 edges (given the strong dose gradients involved), rather than the use of a shorter dwell time
 433 of 1s.

434 Finally, note that the individual component of the MPD, referred to as the single-probe
 435 MSD, has already demonstrated its capability in monitoring prostate treatment sequences
 436 encompassing dwell times ranging from 0.2 to 11 seconds.³⁰ With our probe approach for a
 437 MPD, 94% of the 966 dwell positions were successfully identified, with an average deviation
 438 to the planned dwell times of 0.005 ± 0.060 and a 100 % detection rate for dwell times
 439 exceeding 0.5 s (17%, 86%, 91% and 95% of the dwell times of 0.2 s, 0.3 s, 0.4 s and 0.5 s
 440 were successfully identified, respectively).

441 IV.D. **Clinical use**

442 Our multiprobe detector is compatible with clinical applications. It indeed consists of bio-
 443 compatible elements and it is sufficiently narrow to be inserted in a BT needle or in a
 444 catheter. As a preliminary step, we successfully positioned our detector in a one-millimeter
 445 wide sealed encapsulation pipe made of PEEK material.

446 *In vivo* applications forbid the use of our motorized stage for probe positioning. In
 447 gynecologic BT, the probe and the source would be inserted in two parallel catheters of
 448 an applicator. The inter-catheter spacing would be precisely known, as in the case of the
 449 present study. In prostate BT, the probe and the source are inserted in two independent
 450 needles which are implanted manually in a patient. Therefore, x and z coordinates (cf.
 451 Fig. 2(a)) are usually coupled since the needles are rarely implanted perfectly parallel from
 452 each other, due to operational uncertainties. x and z coordinates of the source relatively to
 453 the probe could however be simultaneously determined by various triangulation approaches
 454 that would be rendered possible by our six-probe detection system (see for instance Ref.¹⁹).
 455 Source tracking via a triangulation process requires a refined 2D calibration plot of the
 456 system, which would not represent a challenge here³⁰. Moreover, parallel IVD from a pair of
 457 multi-probes detectors connected to the same camera and inserted in two different needles
 458 would allow a 3D positioning of the source by triangulation³⁸ with minimum equipment.

459 Note that further experiments are required to assess the compatibility of our probe with
 460 various clinical scenarios. Firstly, it is necessary to repeat measurements using sources of
 461 lower activities, specifically those nearing the end of their clinical lifespan (around 3 or 4
 462 months). This will help evaluate the probe's performance under non-optimum conditions
 463 leading to lower SNR. Secondly, it is important to test our fiber detector while it is positioned

464 within a plastic catheter or either a plastic or metallic needle, depending on the specific
465 treatment being targeted. This evaluation will provide insights into the probe's functionality
466 and reliability when used in conjunction with different types of delivery devices potentially
467 attenuating or spectrally modifying radiations at probe locations. Lastly, it is recommended
468 to consider treatment sequences that involve dwell times of fractions of seconds, various
469 source steps (reduced to 1 mm), and source-probe spacings larger than 4 cm. By examining
470 these parameters, we could gain a comprehensive understanding of the probe's performance
471 across a wider range of clinical scenarios.

472 As mentioned earlier, the individual component of the MPD, specifically the single probe
473 fiber detector, has already been successfully demonstrated in accurately monitoring dwell
474 times down to 0.2 seconds, with source-probe distances of up to 4.7 cm³⁰. In cases where
475 the source-probe distance exceeds 4 cm, an alternative multiprobe architecture is envisioned.
476 This architecture would involve multiple MPDs, each incorporating 2 to 4 probes equally
477 spaced apart. These MPDs, inserted within different catheters or needles and engineered
478 from the same fiber bundle, would be positioned parallel to each other, and separated by
479 approximately 3 cm within the treatment volume. This proposed configuration would provide
480 the capability to accurately monitor the source in three dimensions (3D) throughout the
481 larger treatment volumes typically encountered in brachytherapy.

482 V. Conclusion

483 We have demonstrated in a water phantom a monitoring device for HDR-BT based on a
484 six-probe scintillator dosimeter coupled to a sCMOS camera. Being engineered at the end of
485 a narrow 270- μ m diameter fiber bundle, our miniaturized probes combine high spatial reso-
486 lution and high detection speed while ensuring a minimum perturbation of the therapeutic
487 process, even if water nonequivalent (inorganic) materials are used. Moreover, the overall
488 dosimeter is totally free from inter-probe cross-talk. The use of a sCMOS camera, rather
489 than seven photomultiplier tubes of higher sensitivity offers the possibility of a simultaneous
490 parallel readout of the six scintillating probe signals, plus the residual Cerenkov signal after
491 chromatic filtering (from the bare fiber), in a simple and low cost architecture that is suitable
492 for clinical use. The lower SNR of the camera is compensated by a higher probe detection
493 efficiency enabled by our concept of an IVD micro-pixel based on a nano-optical interface in

494 between scintillators and a fiber²⁸.

495 First, we found a range of probe-to-probe spacings which minimizes source tracking
496 uncertainties. This will be an important information for future MPD designs. We then
497 studied and compared three different source tracking algorithms from the large array of
498 options available within our six-probe system. The best detection approach was found by
499 adding the parallel readout signals from all the probes of the detector. Realizing a source
500 tracking based on this overall accumulated readout signal led to an offset to the planned
501 dwell position as small as 0.01 ± 0.14 mm and 0.02 ± 0.29 mm over a 10-cm long source
502 displacement in the source catheter and at spacings between source and probe catheters of
503 5.5 and 40 mm, respectively. Using this method, we also measured deviations to the planned
504 dwell time of -0.006 ± 0.009 s and -0.008 ± 0.058 s, at source-probe spacings x of 5.5 mm and
505 20 mm, respectively (detection rate of 0.6 s). All the studied configurations were found to
506 surpass current fiber-integrated multiprobes and multipoints detection systems. The next
507 steps will be to test our detection system with various treatment plans used for instance in
508 prostate brachytherapy. The detection performances demonstrated here need to be assessed
509 at shorter dwell times down to a fraction of a second. Triangulation approach will also be
510 realized to simultaneously define the dwell time and the 2D coordinates x and z of a stepping
511 HDR-BT source.

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