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Experimental determination of dosimetric characteristics of Best® 125 I brachytherapy source

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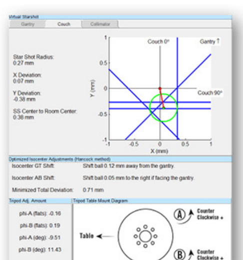

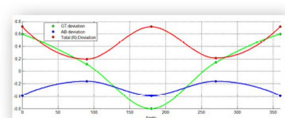

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Experimental determination of dosimetric characteristics of Best® ^{125}I brachytherapy source

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^{125}I brachytherapy sources are being used for interstitial implants in tumor sites such as the prostate. Recently, the Best® ^{125}I source became commercially available for interstitial brachytherapy treatment. Dosimetric characteristics (dose rate constant, radial dose function, and anisotropy function) of this source were experimentally determined, following the AAPM Task Group 43 recommendations, and were related to the NIST 1999 calibration assigned to this source. Measurements were performed in Solid Water™ phantom using LiF thermoluminescent dosimeters. The results indicated a dose rate constant, Λ, of $1.01 \pm 0.08\,\mathrm{cGy\,h^{-1}\,U^{-1}}$ for the new source. The radial dose function, $g(r)$, of the new source was measured at distances ranging from 0.5 to 10.0 cm. The anisotropy function, $F(r,\theta)$, of the new source was measured at distances of 2, 5, and 7 cm from the source center. These data compare favorably with those from the Nycomed/Amersham Models 6711 and 6702 sources. The anisotropy constant, ϕ_{an}, of the Best® ^{125}I source was found to be 0.982. Complete dosimetric parameters of the new source are presented in this paper. © *2000 American Association of Physicists in Medicine.* [S0094-2405(00)01709-0]

Key words: ^{125}I, dosimetry, thermoluminescent dosimeter, brachytherapy

I. INTRODUCTION

^{125}I and ^{103}Pd brachytherapy sources are being used for interstitial implants in various tumor sites, such as the prostate. ^{125}I sources are used in part because their low-energy photon emissions allow a rapid decrease in dose with increasing distance, and hence minimize the dose to normal tissue. The practice of using ultrasound-guided brachytherapy seed implants for prostate cancer has increased greatly in the last few years. The primary reasons for this increase can be attributed to the ease of the procedure for the patient, reduced side effects compared to radical prostectomy, and great cost effectiveness.[1–4] Because of an increase in the frequency of this procedure, a shortage of available seeds became apparent. To meet the rising demand, several manufacturers, including Best Medical International (7643 Fullerton Road, Springfield, VA) developed ^{125}I and ^{103}Pd sources. Nath and Mellilo studied one such ^{125}I source, the Model 2300.[5]

Recently, the National Institute of Standards and Technology calibrated a newly introduced ^{125}I source, (Best® ^{125}I source, Model 2301) to the 1999 standard. Our goals of this project were to experimentally determine the dose distribution around the Best® ^{125}I brachytherapy source, and to intercompare with earlier versions of this model and other commercially available sources. These determinations were performed according to the methodology outlined in AAPM Task Group No. 43 (TG-43)[6,7] and in accordance with the American Association of Physicists in Medicine's recommendations for source calibration.[8,9]

II. MATERIALS AND METHODS

A. ^{125}I source

Best Medical International designed the Best® ^{125}I source with a physical length of 5 mm and an outer diameter of 0.8 mm. This source was manufactured using a 3.8 mm long by 0.25 mm diameter tungsten rod, coated with a 0.1 mm thick organic matrix containing ^{125}I, encapsulated in double titanium capsules. Laser welding seals the two capsules. The double encapsulation is designed to provide thinner walls at the ends of the source, relative to other sources, such as the Model 6711. This design will improve the anisotropy function near the ends of the source. The improvement is due to less self-absorption at the end of the source due to the lesser thickness. This source is manufactured and distributed by Best Medical International with activities ranging from 0.2 up to 10 mCi. A total of nine Best® ^{125}I sources, received in three separate shipments, were evaluated during the course of this project. The air kerma strength of these seeds were measured based on the Wide Angle Free AIR Chamber (WAFAC) calibration of the seeds at NIST in 1999.

B. Thermoluminescent dosimeters

Dose distributions around the Best® ^{125}I source were measured in a Solid Water™ phantom (Radiation Measurements Inc., RMI, Middletown, WI) using LiF thermoluminescence dosimeters (TLD-100, Harshaw/Bicron 6801 Cochran Rd., Solon, OH 44139). For these measurements, slabs of Solid Water™ phantom material were machined to accommodate the source and LiF TLD chips of dimensions $(3.1\times 3.1\times 0.8\,\mathrm{mm^3})$ and $(1.0\times 1.0\times 1.0\,\mathrm{mm^3})$. Figure 1 shows the schematic diagram of the experimental setup for measurements of the dose rate constant and radial dose function. The position of the TLDs within the phantom was selected in order to minimize interference of any one TLD to the absorbed dose of any other TLD chip. Figure 2 shows the phantom that was designed and constructed for the measurement of the anisotropy function of a brachytherapy source.

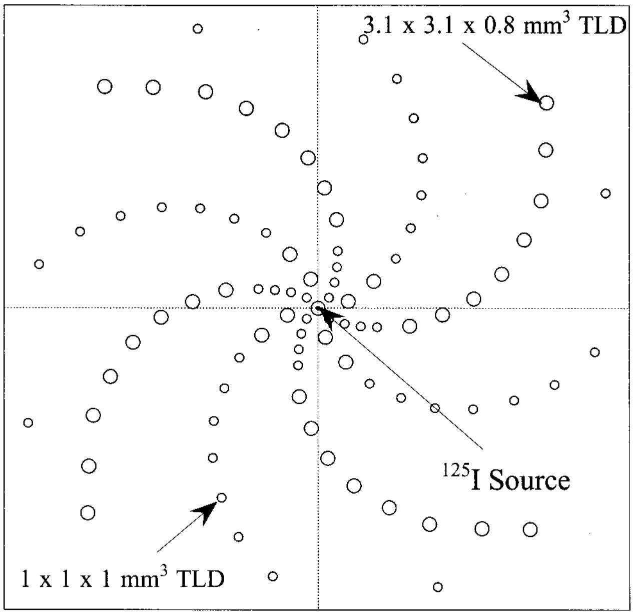

FIG. 1. Schematic diagram of the Solid Water™ phantom designs for TLD measurements of radial dose function, $g(r)$, and dose rate constant, Λ. In this arrangement, the position of the source was adjusted such that the source center was at the level of the TLD chips.

This phantom accommodates LiF TLD chips in concentric circles relative to the source center. The TLD positions ranged from 0° through 360° in 10° degree intervals relative to the source axis. The source was held in position in a specially designed removable Solid Water™ phantom plug. This design allowed for measurement of any other sources by replacing the removable plug. Small Solid Water™ plugs were made to fill any TLD position not used during measurements. The source and the TLD chips were surrounded with at least 10 cm of Solid Water™ phantom material in order to provide full scattering conditions. Each measurement consists of data from at least eight TLD chips. The TLD chips were read using a Harshaw TLD reader (Model 3500) and they were annealed using a standard technique.[10] The TLD responses from this irradiation were used to determine the dosimetric data that are described below;[10]

$$\frac{\dot{D}(r,\theta)}{S_K} = \frac{R}{S_K \varepsilon T E(r) F_{\rm lin}}, \tag{1}$$

where R is the TLD response, corrected for the physical differences of the TLD chips using the predetermined chip factors, T is the experimental time (hours), S_K is the measured source strength at the time of measurement, and ϵ is the calibration factor for the TLD response (nC/cGy) measured with a 6 MV x-ray beam from a linear accelerator (Clinac 2100C/D, Varian Oncology Systems, 3045 Hanover Street, Palo Alto, CA). $E(r)$ in Eq. (1) is the correction factor for the energy dependence of the TLD between the calibration beam and the ^{125}I photons. A value of 1.4 was used for $E(r)$ in this project.[11] $F_{\rm lin}$ is the nonlinearity correction of the TLD response for the given dose. In these measurements, the experimental times were selected such that the absorbed doses ranged from 10–100 cGy, the range over which the TLD response is linear.[10]

C. Dosimetry technique

Characteristics of the Best® ^{125}I source were determined experimentally according to TG-43.[6] Following this protocol, the dose distribution around a sealed brachytherapy source can be determined using the following formalism:

$$\frac{\dot{D}(r,\theta)}{S_K} = \frac{\Lambda G(r,\theta)}{G(r_0,\pi/2)} g(r) F(r,\theta), \tag{2}$$

where: S_K is the air kerma strength of the source, Λ is the dose rate constant, $G(r,\theta)$ is the geometry factor, $g(r)$ is the radial dose function, $F(r,\theta)$ is the anisotropy function, and $r_0 = 1$ cm.

The above quantities are defined and discussed in detail in AAPM TG-43.[6] However, they are briefly reviewed here together with our technique of measurement.

D. Dose rate constant, Λ

The dose rate constant, Λ, is defined as the dose rate per unit air-kerma strength at a reference point along the transverse axis of the source. This quantity is expressed in units of $\mathrm{cGy\,h^{-1}\,U^{-1}}$, where U is the unit of the air-kerma strength of the source, which is defined as $1\,\mathrm{U} = 1\,\mathrm{Gy\,m^2\,h^{-1}} = 1\,\mathrm{cGy\,cm^2\,h^{-1}}$. The dose rate constant of the Best® ^{125}I source was measured using LiF TLD chips. The dose rate constant of the Best® ^{125}I in Solid Water™ phantom material was obtained from the rearrangement of Eq. (2) as shown below:

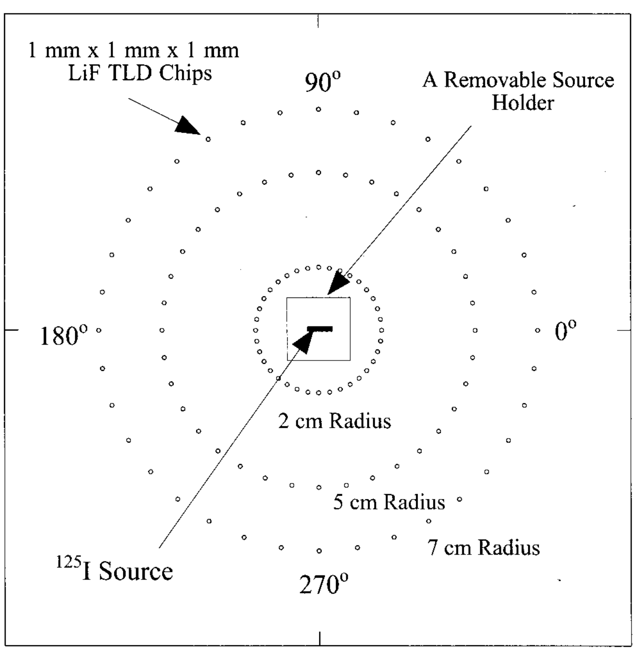

FIG. 2. Schematic diagram of the Solid Water™ phantom designs for TLD measurements of anisotropy function, $F(r,\theta)$.

$$\Lambda = \frac{\dot{D}(r,\theta)}{S_K}. \tag{3}$$

From this value, a dose rate constant in water was determined by multiplying the dose rate constant in Solid Water™ by 1.05.[12]

E. Radial dose function, $g(r)$

Radial dose function, $g(r)$, represents the tissue attenuation of the photons that are emitted by a brachytherapy source. The radial dose function is defined as

$$g(r) = \frac{\dot{D}(r,\pi/2)G(r_0,\pi/2)}{\dot{D}(r_0,\pi/2)G(r,\pi/2)}, \tag{4}$$

where $\dot{D}=(r,\pi/2)$ and $\dot{D}=(r_0,\pi/2)$ are the dose rates measured at distances of r and r_0, along the transverse bisector of the source. r_0 is a reference distance and usually is defined to be 1 cm, as is the case in this project. $G(r,\theta)$ is known as the geometric factor, which takes into account the effect of the distribution of radioactive material inside the source on the dose distribution at a given point. The geometric factor is defined by the AAPM[6] as

$$G(r,\theta) = \begin{cases} \dfrac{1}{r^2} \quad \text{point source approximation,} \\[2ex] \dfrac{\tan^{-1}\left(\dfrac{x+L/2}{y}\right) - \tan^{-1}\left(\dfrac{x-L/2}{y}\right)}{Ly} \\[2ex] \quad \text{linear source approximation.} \end{cases} \tag{5}$$

As seen in Eq. (4), the value of the radial dose function is equal to 1 at the reference point along that source's transverse bisector. An active length of 4.0 mm was used in the calculation of the geometry factor.

The TLD measurements were performed using the Solid Water™ phantom material shown in Fig. 1. The pattern of the TLD locations was selected to minimize the interference of any one TLD to the absorbed dose of other TLD chips. The measurements were performed from 0.5 to 10 cm at 0.5 and 1 cm increments. The larger TLD chips were used for distances greater than 2 cm and the smaller chips were used at distances of 2 cm or less. This range was divided into three regions in order to minimize attenuation along inner radii and to limit TLD doses to a zone of linear response. TLD responses were converted to absorbed dose using Eq. (1) and the radial dose function was extracted using Eq. (4). The TLD data at each distance was obtained from the average of at least eight TLD chips with a combined uncertainty (1σ) of about $\pm 5\%$.

F. Anisotropy function, $F(r,\theta)$

The anisotropy function, $F(r,\theta)$, represents the variation of dose distribution around a brachytherapy source due to the distribution of radioactivity within the source, self-absorption, and oblique filtration of the radiation in the capsule material. From AAPM TG-43, the anisotropy function is defined as

$$F(r,\theta) = \frac{\dot{D}(r,\theta)G(r_0,\pi/2)}{\dot{D}(r,\pi/2)G(r,\theta)}. \tag{6}$$

The measurements were performed at distances of 2, 5, and 7 cm from the source center using a Solid Water™ phantom that was accurately machined to accommodate the TLD chips. Anisotropy was measured for individual radii at angles ranging from 0° through 360° at 10° degree intervals relative to the source axis (Fig. 2). The results of individual dose rates measured at various angles were folded into the first quadrant. At least eight TLD chips were exposed for each data point. Exposures were scheduled for individual radii to minimize interference of TLD chips at any one radius to the response of TLD chips at any other radius. TLD responses were converted to absorbed dose using Eq. (1) and then the anisotropy functions were extracted using Eq. (6). The anisotropy factor, $\phi_{an}(r)$, at each radial distance was calculated following the TG-43 recommendation as

$$\phi_{an}(r) = \frac{\int \dot{D}(r,\theta)\,d\Omega}{4\pi \dot{D}(r,\pi/2)}. \tag{7}$$

The anisotropy constant, $\overline{\phi_{an}}$, of the new source was determined by averaging of the individual anisotropy factors for a given medium.

III. RESULTS

The dose rate constant, Λ, of the Best® ^{125}I source was measured in Solid Water™ as $0.961 \pm 0.07\ \mathrm{cGy\,h^{-1}\,U^{-1}}$. The error propagation for this experiment is shown in Table I. In this table, ''Type A'' errors are the statistical fluctuations in data obtained by repeating measurements. ''Type B'' errors are based on the uncertainty of the information that is used in the data analysis for which the investigator has no control of the magnitude of these errors. Total error is a quadrature combination of these two types of errors. A correction factor, accounting for attenuation differences between water and Solid Water™, of 1.05[12] was used to obtain a dose rate constant in water of 1.01 ± 0.08. This value is for clinical application, as recommended by TG-43. A comparison of these values with other commercially available sources can be found in Table II.

The radial dose function, $g(r)$, of the Best® ^{125}I source was measured at distances ranging from 0.5 to 10 cm using two different sizes of LiF TLD chips. Figure 3 shows a comparison of the radial dose function of the Best® ^{125}I source in Solid Water™ with other commercially available sources. The final TLD data in Solid Water™ are presented in Table III. For clinical application of these data the measured data in the range of 0.5–10 cm was fitted to a fourth-order polynomial function as follow:

$$g(r) = a_0 + a_1 r + a_2 r^2 + a_3 r^3 + a_4 r^4,$$

TABLE I. Uncertainty determinations in experimental measurements of the dose rate constant.

Component of uncertainty	Type A (%) (statistical uncertainty)	Type B (%) (systematic uncertainty)
Repetitive TLD measurements	4.0	
TLD dose calibration (including uncertainty in linac calibration)		2.0
Correction for energy dependence of LiF		5.0
Seed and TLD positioning		1.0
Quadrature combination	4.0	5.5
Total uncertainty (1 sigma)	6.8	
NIST uncertainty in S_K	0.5	
Total combined uncertainty	6.8	

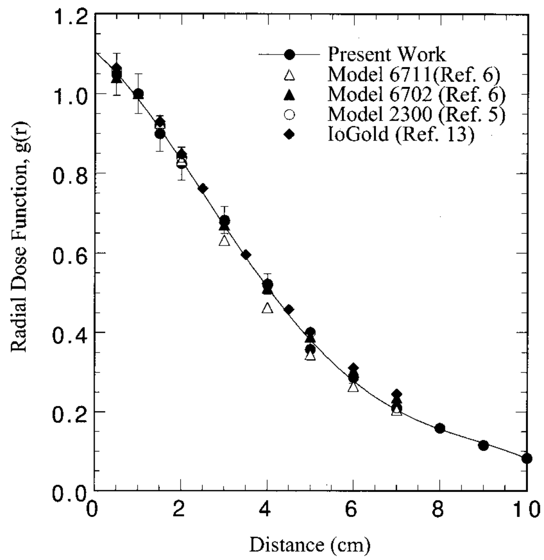

FIG. 3. A comparison of the measured radial dose functions of the Best® ^{125}I brachytherapy source in Solid Water™ versus other commercially available sources. The solid line represents a fourth-order polynomial fit to the measured data. The error bars represent $\pm 5\%$.

where $a_0 = 1.10$, $a_1 = -9.01\text{E-}2$, $a_2 = -3.33\text{E-}2$, $a_3 = 5.77\text{E-}3$, and $a_4 = -2.56\text{E-}4$, with $R = 0.9995$.

The anisotropy function, $F(r,\theta)$, of the Best® ^{125}I was measured with LiF TLD chips at distances of 2, 5, and 7 cm from the source center in Solid Water™. Figure 4 shows the result of these measured anisotropy functions. The uncertainty of the measured data is $\pm 5\%$. Figure 5 shows the comparison of the anisotropy function of the Best® ^{125}I source with several other commercially available sources. No significant differences were observed between Best® ^{125}I and other commercially available sources. From the anisotropy functions the anisotropy factors, $\phi_{\text{an}}(r)$, and anisotropy constants, $\overline{\phi_{\text{an}}}$, of the Best® ^{125}I were extracted [Eq. (7)]. Table IV shows the tabulated anisotropy functions, anisotropy factors, and anisotropy constants for the Best® ^{125}I in Solid Water™. The anisotropy constant of the Best® ^{125}I in Solid Water™ was found to be 0.982.

IV. DISCUSSIONS AND CONCLUSIONS

A double encapsulated ^{125}I source has been designed and fabricated by Best Medical International for interstitial brachytherapy seed implant. Dosimetric characteristics [i.e., dose rate constant, Λ, radial dose function, $g(r)$, and anisotropy function, $F(r,\theta)$] of the Best® ^{125}I have been measured in Solid Water® using LiF TLD chips. These measurements were performed following the AAPM TG-43 task group recommendation.

The dose rate constant, Λ, of the Best® ^{125}I source in Solid Water™ was measured to be $0.961 \pm 0.07\,\text{cGy}\,\text{h}^{-1}\,\text{U}^{-1}$. Table II shows a comparison of the

TABLE II. A comparison of dose rate constant, Λ, of the Best® ^{125}I brachytherapy source with the dose rate constants of Model 6711, Model 6702, Model 2300, and IoGold (Model MED3631-A/M) sources. The unit of dose rate constant is $\text{cGy}\,\text{h}^{-1}\,\text{U}^{-1}$, where $1\,\text{U} = 1\,\text{cGy}\,\text{cm}^2\,\text{h}^{-1}$.

Source model	Method	Medium	Dose rate constant[a] (cGy/h/U)
Best® ^{125}I	Measurements	Solid Water™	0.961 ± 6.8%
Present work	calculated[b]	Water	1.010 ± 6.8%
Model 6711	Monte Carlo simulation	Water	0.973
Williamson *et al.* (Ref. 12)	Monte Carlo simulation	Solid Water™	0.934
Model 6702	Monte Carlo simulation	Water	1.03
Williamson *et al.* (Ref. 12)	Monte Carlo simulation	Solid Water™	0.998
Model 2300	Measurements	Solid Water™	0.955 ± 3%
Nath and Mellilo (Ref. 5)			
IoGold	Measurements[c]	Water	1.06 ± 5%
MED363 A/M			
Wallace *et al.* (Ref. 13)			

[a]Based on 1999 NIST calibration standards.
[b]Calculation based on the measured value in Solid Water™ multiplied by a correction factor of 1.05 (Ref. 12).
[c]Measured in RW1 water equivalent phantom and multiplied by 1.033 to arrive at the dose rate constant in liquid water (Ref. 13).

TABLE III. Experimentally determined radial dose function, $g(r)$, or Best® ^{125}I brachytherapy source in Solid Water™.

Distance from source center (cm)	Radial dose function, $g(r)$
0.50	1.048
1.00	1.000
1.50	0.899
2.00	0.824
3.00	0.683
4.00	0.522
5.00	0.358
6.00	0.287
7.00	0.208
8.00	0.159
9.00	0.115
10.00	0.083

dose rate constant of the Best® ^{125}I source with the dose rate constant of the Model 6711 and Model 6702 sources by Williamson,[12] and the IoGold source by Wallace and Fan.[13] The tabulated values of the dose rate constants are as reported in water, based on NIST 1999 standards for air-kerma strength. The dose rate constant in water for the Best® ^{125}I source is calculated by the multiplication of measurements made in Solid Water™ by 1.05.[12] This conversion factor is derived through Monte Carlo calculations and accounts for differences in absorption and attenuation between water and Solid Water™. The calculated dose rate constant shows excellent agreement, within experimental uncertainty, with any and all of the other commercially available sources against which comparisons have been made. The clinical dose rate constant recommended for use is $1.01\pm.08\,\mathrm{cGy\,h^{-1}\,U^{-1}}$. This value represents the calculated dose rate constant in water.

The radial dose function, $g(r)$, of the Best® ^{125}I was measured in 0.5 cm increments in the range of 0.5–2.0 cm and 1 cm increments for distances between 2.0–10 cm using two

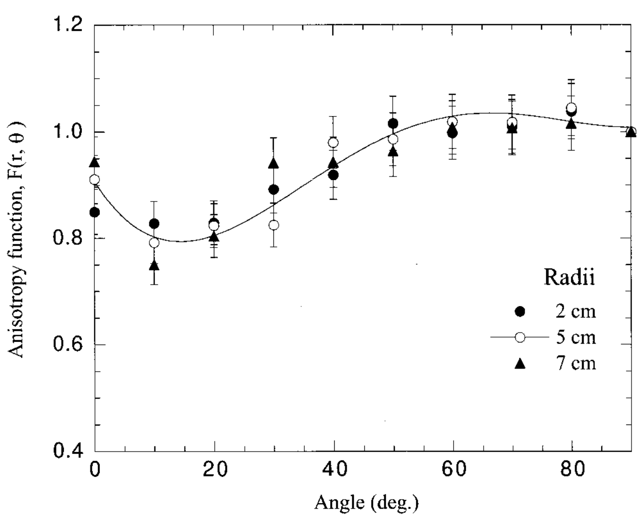

FIG. 4. A comparison of the anisotropy functions, $F(r,\theta)$, at 2, 5, and 7 cm measured in Solid Water™ with TLD dosimeters of the Best® ^{125}I brachytherapy source. The solid line represents a fourth-order polynomial fit to the 5 cm measured data points. The error bars represent ±5%.

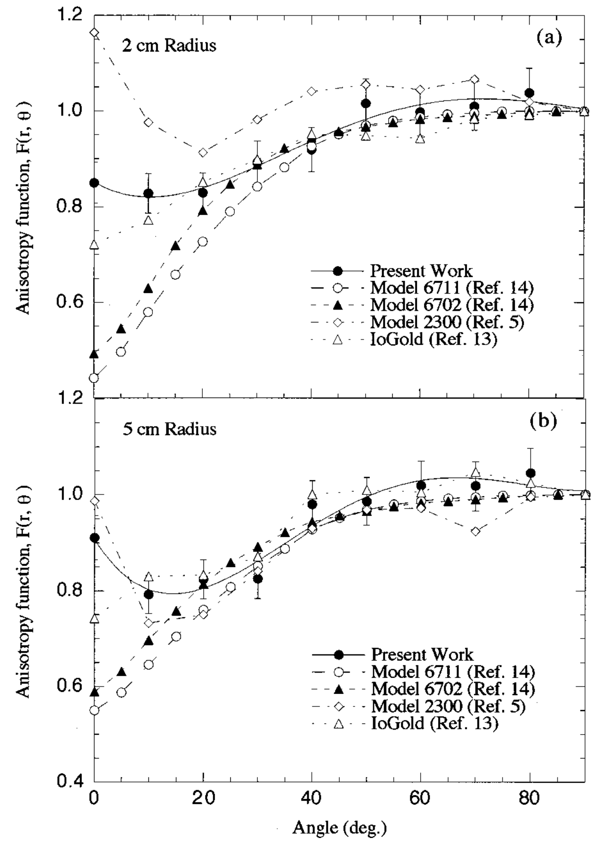

FIG. 5. A comparison of the measured anisotropy functions, $F(r,\theta)$, of the Best® ^{125}I brachytherapy source in Solid Water™ at distances of 2 cm [(a) upper panel] and 5 cm [(b) lower panel] from the source center with other commercially available sources. The solid line represents a fourth-order polynomial fit to the measured data for the Best® ^{125}I source. The dashed lines are connected data points for other data. The error bars represent ±5%.

different sizes of LiF TLD chips. The results indicate that the measured radial dose function for the Best® ^{125}I source is in an excellent agreement (within experimental uncertainty) with the radial dose function of Models 6711 and Model 6702 as reported in TG-43,[6] Model 2300 as measured by

TABLE IV. Experimentally determined anisotropy function, $F(r,\theta)$, of Best® ^{125}I brachytherapy source in Solid Water™.

Anisotropy function			
	Distance from the source center (cm)		
Angle (degrees)	2 cm	5 cm	7 cm
0	0.850	0.911	0.944
10	0.828	0.792	0.750
20	0.829	0.821	0.804
30	0.892	0.930	0.942
40	0.919	0.980	0.943
50	1.016	0.986	0.964
60	0.998	1.019	1.008
70	1.010	1.018	1.007
80	1.038	1.045	1.016
90	1.000	1.000	1.000
$\phi_{an}(r)$	0.990	0.988	0.970
$\overline{\phi}_{an}$		0.982	

Nath and Mellilo,[5] and IoGold as measured by Wallace and Fan.[13]

The anisotropy function, $F(r,\theta)$, of Best® ^{125}I was measured at angles ranging from 0°–360° at 10° intervals relative to the source axis (Fig. 2) at distances of 2, 5, and 7 cm using LiF TLD. Through the application of Eq. (7), the anisotropy factor, $\phi_{an}(r)$, and anisotropy constant, $\overline{\phi_{an}}$, of the Best® ^{125}I was extracted from measurements in Solid Water™.[14] Table IV shows the tabulated anisotropy functions, anisotropy factors, and anisotropy constants for the Best® ^{125}I in Solid Water™. The anisotropy constant of the Best® ^{125}I in Solid Water™ was found to be 0.982. This value is an improvement when compared to the published value of 0.95 for the IoGold source by Wallace and Fan,[13] 0.93 for Model 6711,[6] and 0.95 for Model 6702.[6] However, contrary to the data published by Nath and Mellilo[5] for the model 2300 ^{125}I source, no significant variation of the anisotropy function with depth were observed for the Best® ^{125}I source. In conclusion, dosimetric characteristics of the Best® ^{125}I source were experimentally determined based on TG-43 recommendations. The parameters for the Best® ^{125}I source recommended for use in clinical applications are 1.01 $\mathrm{cGy\,h^{-1}\,U^{-1}}$ for the dose rate constant, 0.982 for the anisotropy constant, and the tabulated radial dose function in Table III. The coefficient for a fourth-order polynomial fit to the radial dose function is also given within the text. These values are comparable to the reported values for other commercially available sources.

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1. K. J. Russell and J. C. Blasko, ''Recent advances in interstitial brachytherapy for localized prostate cancer,'' Probl. Urolo. **7**, 260–278 (1993).
2. D. Nori, ''Conformal brachytherapy of prostate cancer: An effective outpatient treatment,'' Cancer Watch **1**, 124–126 (1992).
3. M. S. Porrazzo, B. S. Hilaris, C. R. Moorthy, A. E. Tchelebi, C. A. Mastoras, L. L. Shih, L. Stabile, and N. Salvaras, ''Permanent interstitial implantation using palladium-103: The New York Medical College preliminary experience,'' Int. J. Radiat. Oncol., Biol., Phys. **23**, 1033–1036 (1992).
4. J. C. Blasko, K. Wallner, P. D. Grimm, and H. Ragde, ''Prostate specific antigen based control following ultrasound guided ^{125}I implantation for Stage T1/T2 Prostate carcinoma,'' J. Urol. (Baltimore) **154**, 1096–1099 (1995).
5. R. Nath and A. Mellilo, ''Dosimetric characteristics of a double wall ^{125}I source for interstitial brachytherapy,'' Med. Phys. **20**, 1475–1483 (1993).
6. R. Nath, L. L. Anderson, G. Luxton, K. A. Weaver, J. F. Williamson, and A. S. Meigooni, ''Dosimetry of interstitial brachytherapy sources: Recommendations of the AAPM Radiation Therapy Committee Task Group No. 43,'' Med. Phys. **22**, 109–234 (1995).
7. A. S. Meigooni, J. F. Williamson, and R. Nath, ''Single source dosimetry for interstitial brachytherapy,'' *Brachytherapy Physics*, edited by J. F. Williamson, B. R. Thomadson, and R. Nath (Medical Physics Publishing, Madison, WI, 1995), Chap. 10, pp. 210– 233.
8. J. F. Williamson, B. M. Coursey, L. A. DeWerd, W. F. Hanson, R. Nath, and G. S. Ibbott, ''Guidance to users of Nycomed Amersham and North American Scientific, Inc. ^{125}I interstitial sources: Dosimetry and calibration changes,'' Med. Phys. **26**, 570–573 (1999).
9. J. F. Williamson, B. M. Coursey, L. A. DeWerd, W. F. Hanson, R. Nath, and G. S. Ibbott, ''Dosimetric prerequisites for routine clinical use of new low energy photon interstitial brachytherapy sources,'' Med. Phys. **25**, 2269–2278 (1998).
10. A. S. Meigooni, V. Mishra, H. Panth, and J. F. Williamson, ''Instrumentation and dosimeter-size artifacts in quantitative thermoluminescence dosimetry of low-dose fields,'' Med. Phys. **22**, 555–561 (1995).
11. P. J. Muench, A. S. Meigooni, R. Nath, and W. L. McLaughlin, ''Photon energy dependence of the sensitivity of radiochromic film and comparison with silver halide film and LiF TLDs used for brachytherapy dosimetry,'' Med. Phys. **18**, 769–775 (1991).
12. J. F. Williamson, ''Comparison of measured and calculated dose rates in water near ^{125}I and ^{192}Ir Seeds,'' Med. Phys. **18**(4), 776–785 (1991).
13. R. E. Wallace and J. J. Fan, ''Report on the dosimetry of a new design ^{125}I brachytherapy source,'' Med. Phys. **26**, 1925–1931 (1999).
14. K. Weaver, ''Anisotropy functions for ^{125}I and ^{103}Pd sources,'' Med. Phys. **25**, 2271–2278 (1998).