Dosimetric characteristics of a double wall ^{125}I source for interstitial brachytherapy[a)]

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Recently, a newly designed encapsulated source of ^{125}I has become commercially available for use in permanent and temporary interstitial brachytherapy. The ^{125}I sources in current use come in two different configurations: the Model 6711 source (Medi Physics/Amersham) for permanent implants has radioactive iodine adsorbed on the surface of a silver wire, and the Model 6702 (Medi Physics/Amersham) source for temporary implants has radioactive iodine absorbed in three spherical resin balls. Both of these iodine sources are encapsulated in a thin-walled shell (0.05-mm thick) made of titanium. The newly designed ^{125}I source (Best Industries Model 2300 series) contains radioactive iodine adsorbed on a tungsten wire that is encapsulated by two walls of titanium. This double-walled ^{125}I source offers the following potential advantages: (i) Because it contains radioactive iodine on the ends as well as the circular surface of the tungsten wire, it can produce a more isotropic dose distribution than the sources in current use; (ii) because it is available in a wider range of source strengths, it is suitable for both temporary and permanent implantation; (iii) because it has a tungsten radiographic marker, source localization is considerably easier than the ^{125}I Model 6702 source that has no radiographic marker; and (iv) because it uses a double-walled encapsulation the risk of radioactive contamination due to source rupture is considerably reduced. In this work, dose distributions produced by the new design ^{125}I source (Model 2300) for interstitial brachytherapy have been measured using LiF TLD's in a Solid Water phantom. Dosimetric characteristics of the new ^{125}I sources are compared with those of the currently available ^{125}I sources. Radial dose function for the Model 2300 source is found to be similar to that for the 6702 source, as expected by the lack of silver characteristic x rays in the photon spectrum from the 2300 source. Using the calibration of source strength based upon the 6702 standard, the dose-rate constant for the 2300 source was determined to be 0.86 cGy h^{-1} U^{-1} [equal to 1.10 cGy h^{-1} mCi^{-1} (app)]. From the measured two-dimensional dose distributions around the source, the anisotropy function for the new source was determined as a function of radial distance and angle. The dose distribution produced by the Model 2300 source was considerably more isotropic than those produced by the 6711 and 6702 sources.

Key words: iodine-125, interstitial brachytherapy, dosimetry, TLD

I. INTRODUCTION

Interstitial brachytherapy can be either temporary or permanent. Most temporary interstitial implants employ dose rates in the range of 30–90 cGy/h. Commonly used radioisotopes for temporary implants in the United States are ^{192}Ir and ^{125}I, and to a lesser extent ^{137}Cs. Photons from ^{125}I have an average energy of about 28 keV, which is considerably smaller than the energy of photons from ^{192}Ir (on average, 360 keV). For distances close to the source, the depth dose is primarily governed by the inverse square law and tissue attenuation has only minor impact. Therefore, the dose distributions in the tumor volume produced by an implant are largely determined by inverse square law and ^{125}I implant dose distributions in tumors are similar to those from ^{192}Ir implants. But at large distances from the source, the effects of tissue attenuation would dominate. Therefore, the exposure levels from ^{125}I implants at patient organs away from the implant site, and of personnel and family members, are considerably lower than those for ^{192}Ir implants. In addition, the ^{125}I photons are much more easily shielded by thin layers of high atomic number materials than the high energy photons from ^{192}Ir. For example, a light weight apron of 0.1-mm Pb thickness reduces the dose to personnel by a factor of about 100 for ^{125}I sources.

Typical dose rates at the onset of permanent implants using ^{125}I are about 5–7 cGy/h, resulting in a total dose to full decay of about 16 000 cGy.[1] Thus, the permanent implants with ^{125}I irradiate tumor at a low dose rate protracted over several months. In contrast, temporary implants with ^{125}I or ^{192}Ir produce dose rates in the range of 30–90 cGy/h, delivering doses in the range of 2 000–6 000 cGy in 2–3 days.

Primarily because of the radiation safety advantages offered by ^{125}I sources over ^{192}Ir, the use of ^{125}I sources in clinical implants has been steadily growing in the past 15–20 years.[1] The ^{125}I sources were clinically introduced by Lawrence Soft X-Ray Corporation in the 1970's and the first clinical applications were pioneered at the Memorial Sloan-Kettering Institute in New York. The source design has undergone several modifications in the past. The two

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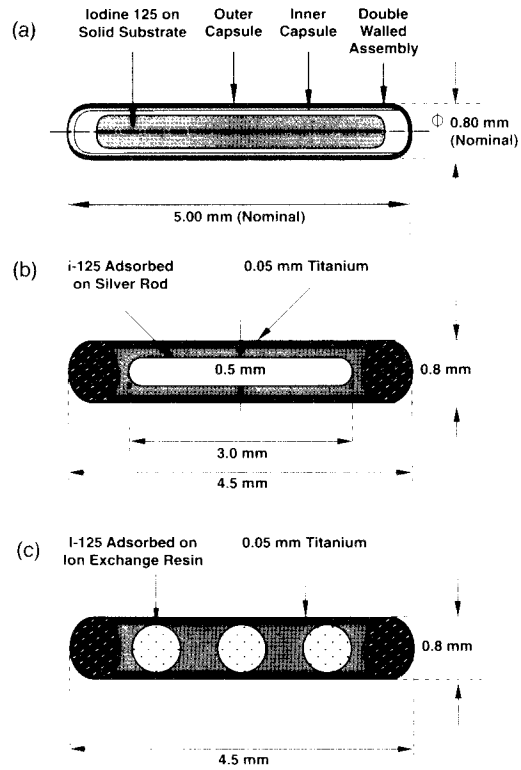

FIG. 1. Schematic drawings of ^{125}I sources (a) Best Industries Model 2300, (b) MediPhysics/Amersham Model 6711, and (c) MediPhysics/Amersham Model 6702.

commercially available models of ^{125}I sources are in current use. Both of these designs, Model 6702 and 6711 sources, are manufactured and distributed by Medi Physics/Amersham. The Model 6702 source contains three resin balls with up to 40 mCi of ^{125}I in a 0.05-mm thick titanium shell welded on both ends [Fig. 1(c)]. Because of this fabrication technique, using plasma arc welding, an indefinitely shaped glob of molten titanium is produced on each end, and precise control over the length and diameter of the source is not possible. The diameter has a range from 0.8 to 0.95 mm, and a length from 4.2 to 4.9 mm. This variation in size sometimes leads to jamming of applicator guns and rupture of radioactive sources. The Model 6702 source does not contain a radiographic marker and is designed for temporary interstitial brachytherapy guided by CT.[1] These sources are barely visible on ordinary radiographs. On the other hand, Model 6711 contains up to 5.5 mCi of ^{125}I adsorbed on a silver wire radiographic marker [Fig. 1(b)] and is designed for permanent implantation.[1] The 6711 sources show up clearly as lines in a radiograph, indicating the silver wire in the source. The designs of currently used ^{125}I sources have not been substantially modified since 1983 and warrant reexamination for potential improvements.

Recently, a newly designed ^{125}I source, which consists of a tungsten wire with adsorbed ^{125}I encapsulated in double-walled titanium, has been introduced by Best Industries (Model 2300).[2] The Nuclear Regulatory Commission (NRC) and the Food and Drug Administration (FDA) have already approved the use of new ^{125}I sources from Best Industries in humans. The assembled source [Fig. 1(a)] is sealed on one end using precision laser welding. Compared to the plasma arc welding technique used in fabrication of sources in current use, the laser welding should provide superior precision resulting in more consistent length and diameter of the fabricated source. Because the source is double-walled, the risk of leakage from a ruptured source is greatly reduced. Also, the source contains radioactivity even on the two ends of the tungsten wire, which increases the dose rate along the source axis. As shown in the present work, this configuration reduces the anisotropy of dose distribution compared to the ^{125}I sources in current use.[2] Finally, the manufacturer is able to deliver, in the same, single model radioactive source, a wide range of activities of up to 40 mCi, providing a ^{125}I source with a radiographic marker for both temporary and permanent implants. The overall objective of this work was to determine basic dosimetry parameters for the new ^{125}I source using the methodology developed by the Interstitial Collaborative Working Group (ICWG: multi-institutional contracts No. 1-CM-57776 to 8, funded by the NIH from 1985–1988 at Yale University, the Memorial Sloan-Kettering Cancer Center, and the University of California at San Francisco). The ICWG has developed consensus guidelines on the physical, biological, and clinical aspects of interstitial brachytherapy using the ^{125}I sources in current use.[1] Dosimetry of the new design ^{125}I source using the same methodology offers the advantages of consistency of dosimetry data for different models of ^{125}I sources.

II. METHODS AND MATERIALS

A. Radiation characteristics of ^{125}I

^{125}I is produced by neutron capture in ^{124}Xe. It decays by electron capture to the first excited state of ^{125}Te, which undergoes internal conversion 93% of the time and otherwise emits a 35.5-keV gamma ray. The average photon energy is 28.5 keV. Encapsulated ^{125}I sources also emit 4-keV Ti characteristic x rays, which are important only for measurements performed in air.[3] Because the Model 6711 source contains a silver wire, it also emits silver characteristic x rays, which lowers the average photon energy to about 27 keV for Model 6711.[4] The spectrum for the new ^{125}I source has been analyzed quantitatively by Rustgi.[2] It indicates a spectrum similar to that for Model 6702,[5] therefore radial dose function of the Model 2300 source is expected to be similar to that of Model 6702.

B. Dose measurements using LiF TLD

Measurements of dose rate around brachytherapy sources are difficult because of steep dose gradients in the vicinity of these sources. In addition, low dose rates make it impractical to employ a small volume ionization chamber for these measurements because the ionization current is very low compared with leakage current and background noise. Larger volume ionization chambers have an adequate signal-to-background ratio but poor spatial resolution near the sources. For these reasons, doses around

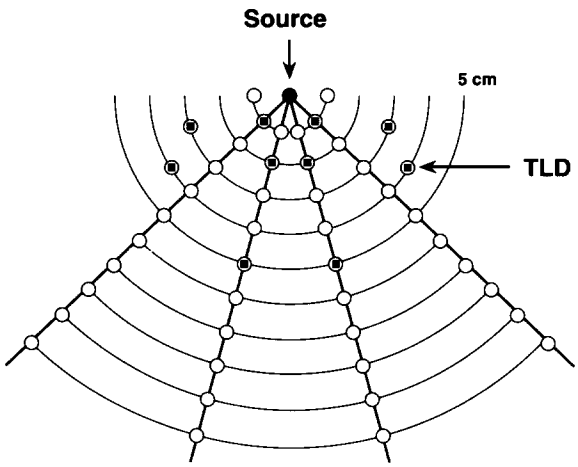

FIG. 2. Schematic drawing of the Solid Water slab (1 cm thick) containing the source and TLD's. The source axis is perpendicular to the slab.

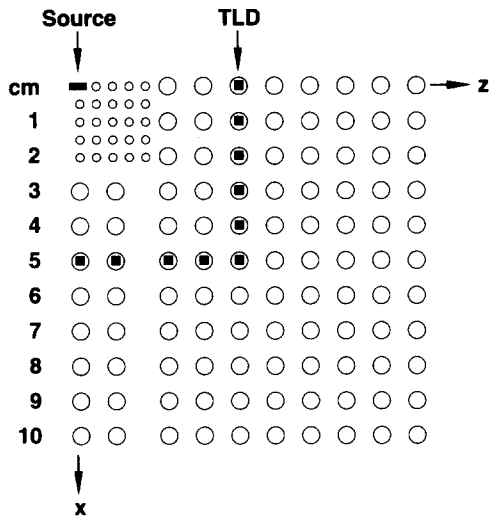

FIG. 3. Schematic drawing of the Solid Water slab (1 cm thick) containing the source and TLD's. The source lies in a hole with its axis along the z direction. Smaller TLD's ($1\times1\times1$ mm) are used in the smaller holes near the sources, whereas the larger TLD chips are used in holes further away.

brachytherapy sources are usually measured with small solid-state dosimeters such as diodes and thermoluminescent dosimeters (TLD's). Following the ICWG, we have chosen to use LiF TLD for the proposed dose measurements. TLD's have to be calibrated against a NIST-calibrated ionization chamber if they are to be used for absolute dose measurement. For a high dose-rate source the response of solid-state dosimeters can be directly compared with the dose rate measured with an ionization chamber. However, for conventional low dose-rate brachytherapy sources, calibration is usually performed indirectly. In this method sensitivity (response per unit dose) of TLD's is measured with a calibrated high dose-rate photon beam (such as ^{60}Co) and then corrected for the difference in response to the energy of the brachytherapy source versus that of the calibration beam.

We have measured the energy dependence of the sensitivity of LiF TLD's with photon energy using a calibrated ionization chamber irradiated by a set of orthovoltage and megavoltage beams. We find that the sensitivity of LiF (LiF-100) to 28-keV (average) photons from ^{125}I is about 40% higher than for megavoltage photons;[6] this is in agreement with previous observations.[7,8] Since the photon spectrum of ^{125}I does not change much with depth in phantom[9] a single energy correction of the sensitivity of LiF can be used at all depths. This information was used to account for energy dependence of LiF TLD.

C. LiF TLD

Doses around brachytherapy sources were measured with LiF TLD's placed in shallow holes precisely machined in a slab of Solid Water phantom for transverse axis measurements as shown in Fig. 2, and for two-dimensional dosimetry as shown in Fig. 3. The configuration in Fig. 2 has the advantage of providing four simultaneous measurements at a given radial distance along the transverse axis, thus reducing uncertainties due to statistical fluctuations and small deviations in the positioning of TLD's and source. It was chosen to measure the transverse axis data, which included the dose-rate constant and the radial dose function. The configuration of Fig. 3 was used for relative 2-D dosimetry measurements.

In our current brachytherapy dosimetry system, an elaborate protocol for TLD measurements is followed and is described here briefly. Four batches of 100 commercially available LiF TLD chips each with dimensions of $3.1\times3.1\times0.89$ mm (TLD-100, Harshaw Co.) were used. For distance less than 2 cm, small TLD's with dimensions of $1\times1\times1$ mm were used. Before each experiment, TLD's were annealed in an aged aluminum tray at 400 °C for 1 h and then kept at room temperature for 45 min followed by 80 °C heating for 24 h. After irradiation and a waiting period of at least 24 h, the responses of the TLD chips were measured with a Harshaw TLD reader (Models 2000A and 2000B). Responses of several chips irradiated simultaneously were averaged to improve the statistical quality of the experimental data. The response of individual chips was corrected for differences in their physical properties such as mass, size, etc., using a chip factor described in Ref. 9.

D. Interchip effects

Because of the predominance of the photoelectric effect for the low energy photons of ^{125}I, it is possible that the dose at a point in a phantom could be affected by replacing part of the phantom with a LiF chip. In particular, there can be an interchip shielding effect when one chip is in the shadow of another and there can be an interchip scattering effect when chips are very close to each other. The interchip effect, taken as the ratio of the response of the isolated chip to that of the four chips at the same distance from a ^{125}I source, has been measured. The largest interchip effect of 6% was found in Solid Water at 1 cm depth, the depth at which the four chips used for the depth-dose measure-

ments are closest to each other.[9] These interchip effects were reduced by using only two TLD's at short distances from the sources (less than 2 cm).

E. Choice of phantom material for ^{125}I dosimetry

Due to high dose gradients in the vicinity of small radioactive sources, the dosimetry of radioactive sources used in clinical brachytherapy is critically dependent on the accuracy of source-to-detector distance measurements. In addition to being more convenient than water, solid materials can be precisely machined to accommodate sources and detectors, and distances can be accurately determined. However, the radiation characteristics of solid phantoms may not be equivalent to those of water, especially for the low energy photons emitted by ^{125}I. Our measurements and Monte Carlo calculations show that Solid Water is within 3%–5% of water for the dosimetry of ^{125}I sources; however, polystyrene and PMMA are not equivalent to water. However, Williamson[11] has shown recently that a correction of about 4.3% is needed if dose-rate constant in water is to be derived from Solid Water phantom measurements. Results of our very recent (unpublished) Monte Carlo calculations in Solid Water, water, PMMA, polystyrene, and various body tissues, indicate that Solid Water is a better substitute for muscle than water for the ^{125}I Model 6702 source.

F. ICWG dose calculation formalism for interstitial brachytherapy

The ICWG has recommended a formalism for dosimetry of interstitial brachytherapy sources.[1] The ICWG recommends that the source strength should be specified in terms of air-kerma strength, S, which is defined as the product of air-kerma rate at a large distance l from the source in air along the perpendicular bisector of the source axis and the distance l squared:[12]

$$S=\dot{K}l^2, \tag{1}$$

where $\dot{K}$ is the air-kerma rate at a distance of l, which should be large enough for point source geometry to be valid. The recommended unit of air-kerma strength is 1 μGy m^2 h^{-1}, which is equivalent to 1 cGy cm^2 h^{-1}. This unit has been noted by the symbol U in some of the recent literature on brachytherapy dosimetry.[3,11,13] A ^{125}I source with air-kerma strength of 1.27 U is equivalent to 1.00 mCi (apparent).[13]

The ICWG protocol recommends that the dose rate in a medium is best calculated from quantities measured solely in the medium. The dose rate at a point (r,θ) near a source can be expressed as

$$\dot{D}(r,\theta)=S\Lambda\frac{G(r,\theta)}{G(1,\pi/2)}F(r,\theta)g(r), \tag{2}$$

where Λ is the dose-rate constant, which is defined as the dose rate to water at 1 cm in the medium along the transverse axis produced by a unit strength source. θ is the angle subtended by the source axis and radial vector from the source center to the point (r,θ). The function, $G(r,\theta)$, is a geometry factor that accounts for distribution of radioactive material within the source, and is given by

$$G(r,\theta)=r^{-2} \quad \text{for a point source,}$$
$$=\frac{\theta_2-\theta_1}{Ly} \quad \text{for a line source.} \tag{3}$$

Distances and angles for the line source are shown in Ref. 13. The angular anisotropy function, $F(r,\theta)$, a function normalized to unity at $\theta=90°$ for each r, accounts for absorption and scatter in the medium and encapsulation. This function can be obtained from relative dose measurements. The radial dose function, $g(r)$, accounts for absorption and scatter along the transverse axis, $(\theta=90°)$, and, by definition, is unity at 1 cm.[14–16] Using Eq. (2), the radial dose function can be obtained from relative doses measured along the source's transverse axis as given by

$$g(r)=\frac{\dot{D}(r,\pi/2)G(1,\pi/2)}{\dot{D}(1,\pi/2)G(r,\pi/2)}. \tag{4}$$

Likewise the dose anisotropy function can be obtained by relative dose measurements in 2-D and is given by

$$F(r,\theta)=\frac{\dot{D}(r,\theta)G(r,\pi/2)}{\dot{D}(r,\pi/2)G(r,\theta)}. \tag{5}$$

For points along the transverse axis, the dose rate at distance r is given by the simplified expression:

$$\dot{D}(r)=\Lambda S\frac{G(r,\pi/2)}{G(1,\pi/2)}g(r) \tag{6}$$

that, for r greater than or equal to 1 cm along the transverse axis, can be approximated by

$$\dot{D}(r)\approx\Lambda S\frac{g(r)}{r^2}. \tag{7}$$

For the point source approximation, dose rate at any (r,θ) can be obtained by using the anisotropy factor of the source as follows:

$$\dot{D}_{\text{point}}(r,\theta)\approx\Lambda S\frac{g(r)}{r^2}\phi(r), \tag{8}$$

where $\phi(r)$ is the anisotropy factor, which is defined as the ratio of 4π-averaged dose rate at a given radial distance divided by the dose rate at the same distance along the transverse axis. Because $\phi(r)$ varies slightly with r, an average value, termed anisotropy constant ϕ_{avg} is usually a good approximation.

It should be noted that this formalism, produced by the ICWG, represents consensus developed by a collaborative group after three years of intensive effort. Also, the AAPM Task Group No. 43 on Dosimetry of Interstitial Brachytherapy Sources has adopted the ICWG formalism. To summarize, this protocol requires the following dosimetry parameters for clinical dosimetry:

- dose-rate constant Λ,
- radial dose function $g(r)$,

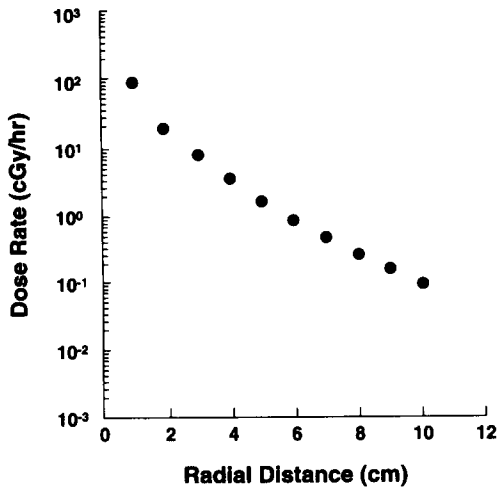

FIG. 4. Measured dose rate along the transverse axis of an ^{125}I Model 2300 source with an air-kerma strength of 100 U.

- anisotropy function $F(r,\theta)$,
- anisotropy factor $\phi(r)$,
- anisotropy constant ϕ_{avg}.

In this work, we present measured data for these parameters for the new ^{125}I source.

G. Source strength specification and calibration

Because the new ^{125}I sources do not yet have an air-kerma standard traceable to the National Institute of Standards and Technology (NIST), we used the source strength provided by the manufacturer, who is using a well-type ionization chamber calibrated for the ^{125}I Model 6702 source.[17] Also, source strength was determined by us using a reentrant ionization chamber calibrated for 6702 sources. From the spectrum data of Rustgi[2] and our radial dose function data (presented in the next section) for the new ^{125}I source, it is reasonable to expect that calibration of the new ^{125}I source would be similar to that of Model 6702. In this work, dose-rate constant and other dosimetry data are expressed in terms of source strength derived by the procedures used by the manufacturer, which are based on the 6702 standard. Dosimetry in patients would be consistent with the data from the present study provided the users employ the same method for specification of source strength and that the manufacturer continues to use the same standard consistently. Ultimately, when the NIST air-kerma standard for the new ^{125}I source becomes available, it will be used and the dose-rate constant will be modified appropriately. This, however, will not affect the patient dosimetry.

III. RESULTS

A. Dose-rate constant Λ

Dose rates along the transverse axis of a ^{125}I Model 2300 source were measured in a Solid Water phantom for distances up to 10 cm using LiF TLD's (Fig. 4). Dose-rate constant in Solid Water was determined to be 0.86 ± 0.03 cGy h^{-1} U^{-1}. In older and obsolete units of source strength (i.e., mCi apparent), it is 1.10 ± 0.03 cGy h^{-1} mCi^{-1} (apparent). Currently accepted values of dose-rate constants for other ^{125}I sources are also shown in Table I. The dose-rate constant for the new ^{125}I source is very similar to that of Model 6711 and is 7% smaller than that for Model 6702.

TABLE I. Dose-rate constants in Solid Water for ^{125}I sources.

Source	Dose-rate constant Λ (cGy h^{-1} U^{-1})	Dose-rate constant Λ (cGy^{-1} h^{-1} mCi^{-1})
^{125}I Model 6702	0.92[a]	1.16[a]
^{125}I Model 6711	0.85[a]	1.07[a]
^{125}I Model 2300	0.86	1.10

[a]From ICWG.

B. Radial dose function $g(r)$

From the measured dose rates, at points along the transverse axis (Fig. 4), the radial dose function was calculated. Results for the new ^{125}I source, together with our previous data of ^{125}I Model 6702 and 6711 sources, are shown in Fig. 5. These data support the theoretical expectation that radial dose function for the new source should be similar to that for the Model 6702 source because the new ^{125}I source produces a photon spectrum more similar to that of the ^{125}I Model 6702 source than Model 6711.

C. Two-dimensional dose distributions

Our two-dimensional dosimetry data around the new ^{125}I source and Models 6702 and 6711 are shown in Figs. 6 and 7. From these data it is apparent that the dose distributions around the new ^{125}I source are more isotropic than those around the currently used ^{125}I source, especially at short distances as illustrated also in Fig. 7.

The anisotropy effect can also be estimated from the ratio of dose at a point on the source axis to that on the

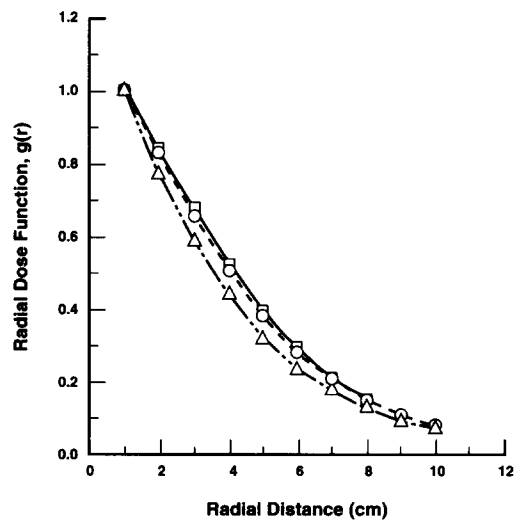

FIG. 5. Radial dose functions $g(r)$ for ^{125}I sources Model 2300 (squares), Model 6702 (circles), and Model 6711 (triangles).

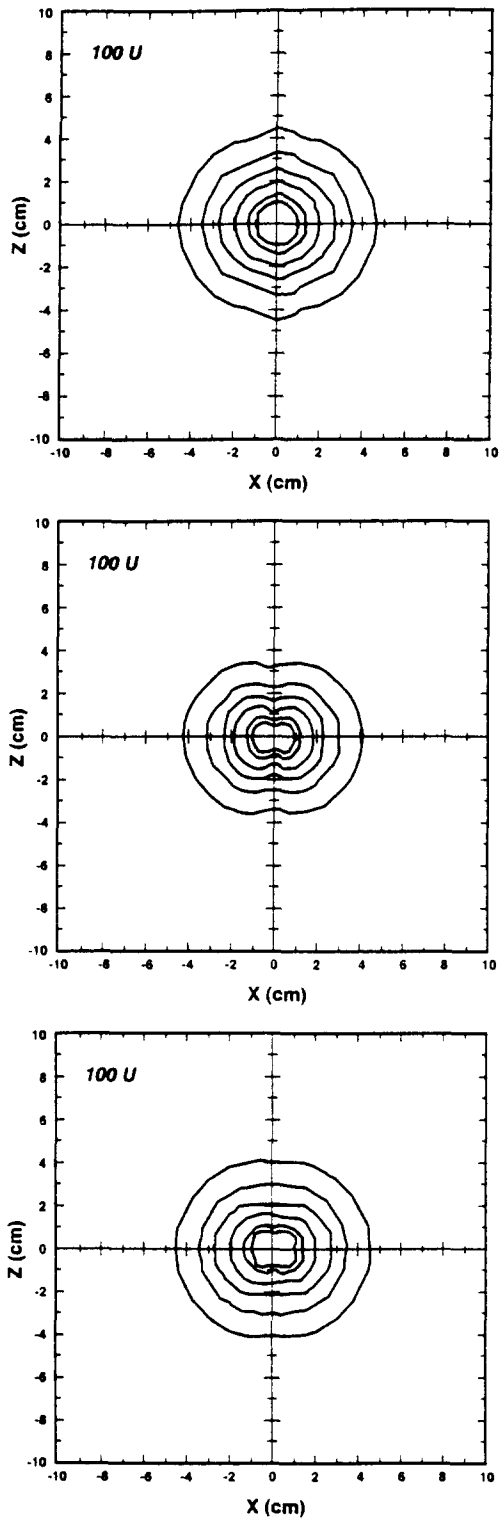

FIG. 6. Measured isodose curves in cGy/h from a 100 U ^{125}I source Model 2300 (top panel), Model 6711 (middle panel), and Model 6702 (bottom panel). The dose-rate levels are 2, 5, 10, 20, 50, and 100 cGy/h.

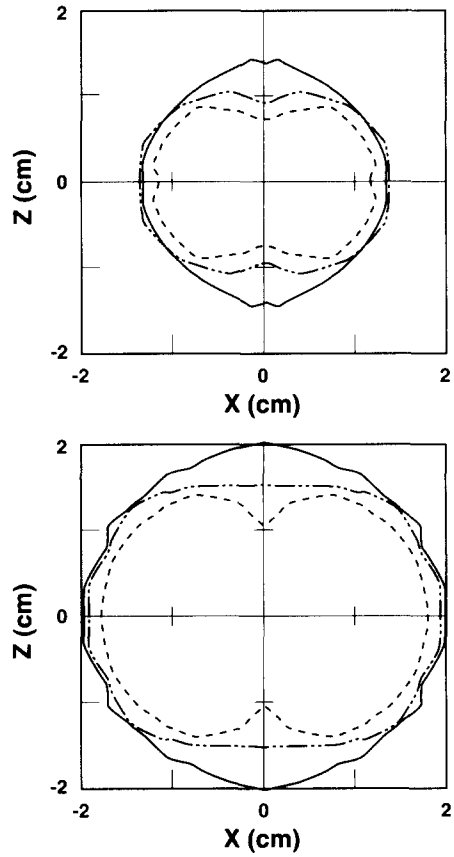

FIG. 7. Isodose curves of 20 cGy/h (bottom panel) and 50 cGy/h (top panel) from ^{125}I sources, Model 2300 (solid line), Model 6711 (broken line), and Model 6702 (dotted line).

transverse axis in Fig. 8. It is observed that this ratio is much closer to unity than that for Models 6702 and 6711. It should also be noted that only a small fraction of the 4π solid angle is subtended by the polar areas. To assess the effect of anisotropy on the overall dose distribution, it is necessary to investigate the anisotropy function $F(r,\theta)$, the anisotropy factor $\phi(r)$, and the average anisotropy constant ϕ_{avg}. These are discussed now.

Using the 2-D dosimetry data and Eq. (5), the anisotropy functions, $F(r,\theta)$ were determined. As shown in Fig.

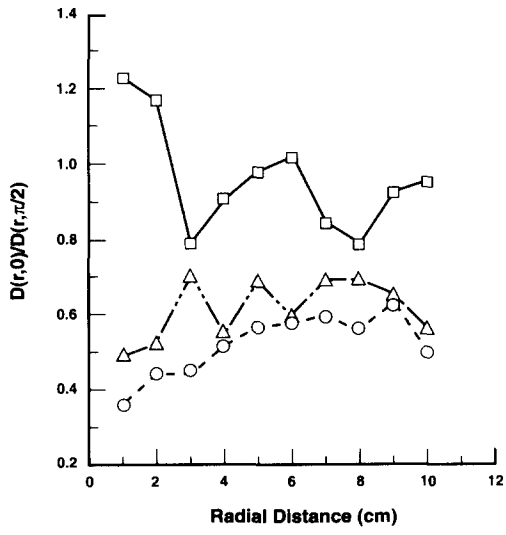

FIG. 8. Dose rate along the source axis divided by dose rate at the same radial distance along the transverse axis for ^{125}I sources, Model 2300 (squares), Model 6702 (triangles), and Model 6711 (circles).

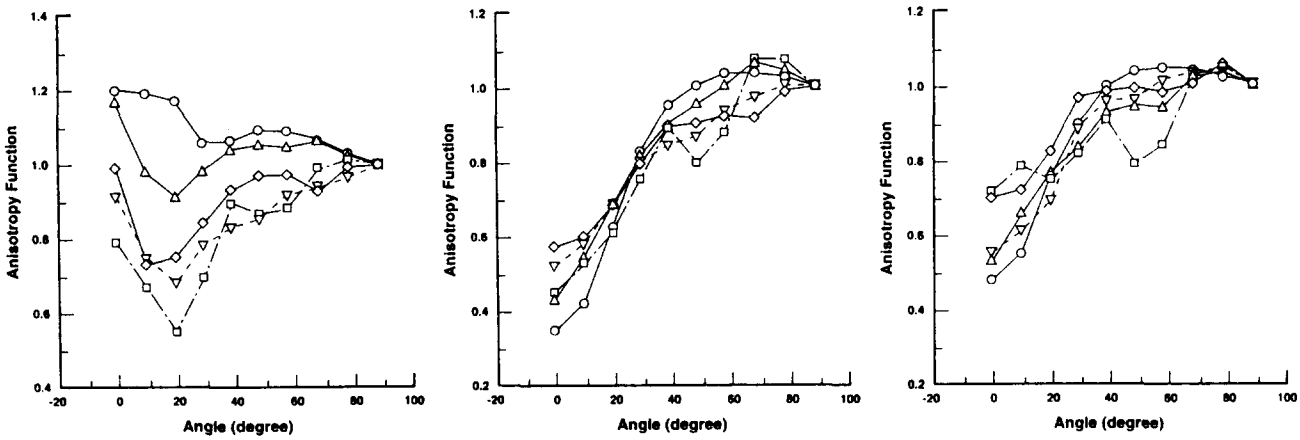

FIG. 9. Anisotropy functions for ^{125}I source, Model 2300 (left panel), Model 6711 (middle panel), and Model 6702 (right panel) at radial distances of 1 cm (circles), 2 cm (triangles), 3 cm (squares), 4 cm (inverted triangles), and 5 cm (diamonds).

9, the anisotropy functions for the Model 2300 source are closer to unity than those for Models 6702 and 6711, especially at short distances (less than and equal to 2 cm). Because the dose rate falls rapidly as the radial distance from the source increases, the dose distributions produced in a multisource implant of new ^{125}I sources would be expected to be much more isotropic than sources in current use because most of the dose at any given point in a multisource implant arises from the sources that are in close vicinity of the point of interest.

The anisotropy factor $\phi(r)$ defined by Eq. (8) was also calculated as a function of r (Fig. 10). The anisotropy factor for Model 2300 source is significantly closer to unity than that for Models 6702 and 6711, especially at distances closer than 2.0 cm. An average of $\phi(r)$, weighted by inverse square of distance r, was calculated using Eq. (9). For the Model 2300 source, the weighted-average value was 0.99, which is closer to unity than the values for Models 6702 and 6711 by 4%–6% (Table II). The data for Models 6702 and 6711 were taken from similar measurements using identical techniques as presented here[18] and are in agreement with values recommended by ICWG.[1]

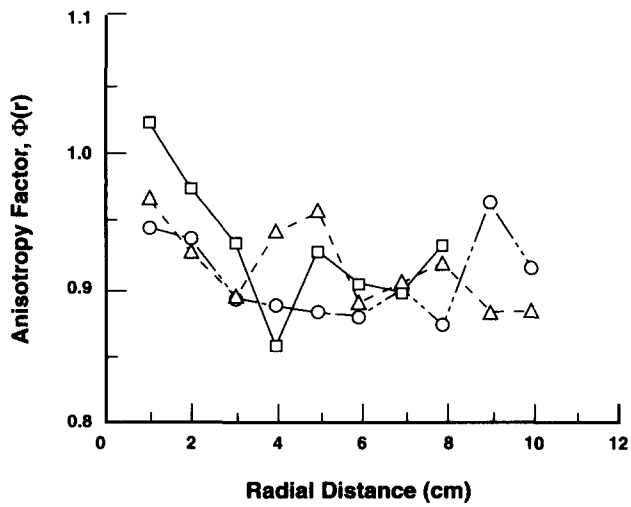

FIG. 10. Anisotropy factor as a function of radial distance from the center of an ^{125}I Model 2300 source (squares) compared to those for Model 6711 (circles), and Model 6702 (triangles) sources.

Finally, dose rates as a function of radial distance using the point source approximation [Eq. (8)] were calculated for ^{125}I Model 2300 sources (Table III). For this calculation, the radial dose function was fitted to the expression,

$$g(r)=\exp(-a_4 r)(a_1+a_2 r+a_3 r^2). \quad (9)$$

The best values of parameters a_1, a_2, a_3, and a_4 were determined to be 1.105 55, 0.323 35, −0.021 44, and 0.341 93, respectively.

IV. DISCUSSION AND CONCLUSIONS

The overall objective of this was to study the physical and dosimetric properties of a newly designed ^{125}I source, which has become available for temporary and permanent implants in interstitial brachytherapy. Compared to the currently used ^{125}I sources, the newly designed source offers several practical advantages, which can significantly improve clinical implementation of interstitial brachytherapy.

Up until recently, one of the problems with the ^{125}I sources has been that the photon emission from the available sources is highly anisotropic. The anisotropy of dose distributions produced by a ^{125}I source leads to uncertainties in clinical dosimetry, because in typical implants with 50–100 sources it is impractical, if not impossible, to determine the orientation of these sources relative to the patient's frame of reference. In practice, it is generally assumed that dose distributions around ^{125}I sources are

TABLE II. Anisotropy constants for ^{125}I sources.

Source	ϕ_{avg}
^{125}I Model 6702	0.96[a]
^{125}I Model 6711	0.94[a]
^{125}I Model 2300	0.99

[a]From ICWG.

TABLE III. Radial dose function and dose rate x distance squared for an ^{125}I source with air-kerma strength of 1 U (using the point source approximation).

Distance along transverse axis (cm)	Radial dose function $g(r)$	Dose rate $\times r^2$ (cGy h^{-1} cm^2)
0.5	1.06	0.91
1.0	1.00	0.85
1.5	0.92	0.79
2.0	0.84	0.72
2.5	0.76	0.65
3.0	0.68	0.58
3.5	0.60	0.51
4.0	0.52	0.45
4.5	0.46	0.39
5.0	0.40	0.34
5.5	0.34	0.29
6.0	0.29	0.25
6.5	0.25	0.21
7.0	0.21	0.18

isotropic and that one can use point-source approximation for clinical dosimetry. This approximation is perhaps reasonable for multisource implants containing a large number of sources with randomly distributed orientation.[5,19] However, it is clearly inadequate when few sources are used in regularly placed catheters, e.g., in the case with stereotactic implantation of brain tumors with ^{125}I sources. This problem has been partially solved because the new ^{125}I source indeed produces a more isotropic dose distribution.

Only recently, direct measurements of dose rates in a medium from interstitial brachytherapy sources of ^{125}I in current use have become available.[1,8,10,20–23] In the past, dose rates in tissue were determined using the exposure rate constant, exposure-to-dose conversion factor (f factor) and tissue attenuation correction factors. However, newer protocols[1] recommend the use of dose-rate constants, radial dose and anisotropy functions for dose calculations. In this study, the physical and dosimetric characteristics of this newly designed source were investigated using the methodology developed by the Interstitial Collaborative Working Group. When the ICWG started its work in 1985, it was soon realized that the single source dosimetry data for ^{125}I sources was based upon relatively old and somewhat unreliable measurements.[1] Therefore, an extensive series of measurements for determining the dosimetry parameters of ^{125}I sources was initiated. After three years of effort, the three independent ICWG research groups have generated a new dosimetry calculation formalism and have recommended new dosimetry parameters for both ^{125}I Model 6702 and 6711 (and ^{192}Ir) sources.[1] The recommended dose-rate constant for ^{125}I Model 6702 is 1.17 cGy h^{-1} mCi^{-1}, which is 11% smaller than the currently used value of 1.30. For Model 6711, the recommended value is 1.08 cGy h^{-1} mCi^{-1}, which is 20% smaller than the value in current use. Thus, the ICWG discovered a 10%–20% error in the dosimetry of ^{125}I sources in current use. A more recent Monte Carlo evaluation of ^{125}I dosimetry confirms most of the ICWG dosimetry. However, it indicates that the dose-rate constants in a water phantom are about 4% higher than those in a Solid Water phantom.[11]

These recommendations would have the effect of revising the dose prescription for the two models of ^{125}I by as much as 16% and would effect the retrospective analysis of clinical data for ^{125}I implantation. It should also be noted that before 1983 the permanent implants were performed using the now obsolete Model 6701 source, which has a dose-rate constant similar to that of Model 6702. Thus, the replacement of Model 6701 by Model 6711 in 1983 has introduced an error in the dosimetry that must be corrected in the future retrospective analyses. The new dosimetry parameters may not affect most of the relative biological effectiveness (RBE) experiments because most of these experiments had their own direct dosimetric measurements; i.e., they did not rely upon dose-rate constants as is the practice in clinical dosimetry. However, if an RBE experiment did use the old dosimetry parameters, then the RBEs reported in that work may be in error by up to 20%. This may explain, in part, the wide variation of RBEs reported.[1,24] Therefore, good dosimetry data on new and old sources is essential for future radiobiological and clinical studies.

To exploit these potential benefits from a newly designed ^{125}I source in clinical implementation of interstitial brachytherapy, we are of the opinion that it is essential to critically examine the physical and dosimetric characteristics of the new ^{125}I sources relative to the ^{125}I sources in current use, and to determine its dosimetry parameters to the same level of accuracy as the ^{125}I sources in current use. Knowledge of accurate dosimetry, the quality of the analyses of clinical outcome data and the RBEs of different isotopes in clinical use can be significantly improved. We hope to have contributed to this effort by the presentation.

At this time, the cost of the newly designed ^{125}I source is expected to be about the same as that of ^{125}I sources currently in use.

In summary, the newly designed ^{125}I source offers the following advantages over the sources in current use for interstitial brachytherapy:

- Dose distributions around the new ^{125}I source are more isotropic, thus improving the dose distributions within a multisource implant.
- The new ^{125}I source with a radiographic marker is available in a wider range of source strengths, thus making it possible to use the same design source for both permanent and temporary implants.
- The new ^{125}I source is encapsulated in double-walled titanium, which considerably reduces the risk of radioactive contamination due to source rupture.

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