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Invited Review

Recent developments in brachytherapy source dosimetry

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ABSTRACT

Application of radioactive isotopes is the treatment of choice around the globe for many cancer sites. In this technique, the accuracy of the radiation delivery is highly dependent on the accuracy of radiation dosimetry around individual brachytherapy sources. Moreover, in order to have compatible clinical results, an identical method of source dosimetry must be employed across the world. This problem has been recently addressed by Task Group 43 (TG43) from the American Association of Medical Physics (AAPM) with a protocol for dosimetric characterization of brachytherapy sources. This new protocol has been further updated using published data from international sources, by a new Task Group from the AAPM. This has resulted in an updated protocol known as TG43U1 that has been published in March 2004 issue of Medical Physics. The goal of this presentation is to review the original TG43 protocol and associated algorithms for brachytherapy source dosimetry. In addition, the shortcomings of the original protocol that has been resolved in the updated recommendation will be highlighted. I am sure that this is not the end of the line and more work is needed to complete this task. I invite the scientists to join this task and complete the project, with the hope of much better clinical results for cancer patients. *Iran. J. Radiat. Res., 2004; 2 (3): 97-105*

Keywords: *Brachytherapy, radiation dosimetry.*

INTRODUCTION

Brachytherapy, the treatment of cancer patients by placing sealed radioactive elements at the short distances relative to the tumor, was born soon after Henri Becquerel's discovery of radioactivity in Paris (1896) and Marie Curie's discovery of radium (1898). However, within the past several years there have been great improvements in the technical aspect of the brachytherapy procedures (Blasko and Grimm 1993, Blasko *et al.* 1995, Grimm *et al.* 1994, 2001), and better understanding in the radiobiological aspect of low dose rate radiation (Brenner and Hall 1991, Orton 1993) have helped us to advance in this treatment modality. Moreover, development of the protocols regarding the dosimetric characterization of the brachytherapy sources (Nath *et al.* 1995, Williamson *et al.* 1998) and development of treatment-planning systems, which could incorporate the CT, MRI and Ultrasound images for the 3D dose calculations (Schoeppel *et al.* 1993, Ling *et al.* 1987), have further improved this field. These advancements have lead to the clinical results, which are superior or at least comparable to the other treatment modalities for cancer patients. In this presentation the advancement of the brachytherapy source dosimetry is being reviewed.

Reviewing the brachytherapy data prior to 1995 indicates a large variation of the dosimetric characteristics of sources, due to the inconsistency in technique of the dosimetry or use of various

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dosimeters and phantom materials. For example, in 1975, Hilaris *et al.* have measured the dose rate constant of ^{125}I, Model 6701, to be 1.68 cGy.cm^2/mCi.hr. However, in the same year, Anderson *et al.* (1975) have found a dose rate constant of 1.3 cGy.cm^2/mCi.hr for ^{125}I, Model 6701, using the point source approximation. Moreover, Holt *et al*. (1975) measured a value of 1.03 cGy.cm^2/mCi.hr using a spherical ionization chamber, and Anderson and Ding (1975), measured a value of 1.18 cGy.cm^2/mCi.hr using TLD measurement in plastic. These results were indicating more than 60% differences in dose rate constant of the same source within one year. These differences were attributed to the differences in the phantom material, dosimeters, as well as the dosimetry techniques, particularly for low energy photon emitter sources.

In 1995, the AAPM (American Association of Physicists in Medicine) Task Group 43 (Nath *et al.* 1995) has introduced a protocol which has minimized the large variation of the dosimetric information determined by various investigators around the world. Using this recommendation, dosimetric characteristics (dose rate constant, radial dose function, anisotropy function, and anisotropy factors) of several new designs of ^{125}I and ^{103}Pd brachytherapy sources have been determined and published by various investigators. These characterizations were performed using experimental and Monte Carlo simulation techniques. This protocol has introduced a universal dosimetry technique for the brachytherapy sources, which is briefly described in the following sections.

TG-43 Recommendation for brachytherapy dosimetry

Characteristics of a brachytherapy source could be determined using both experimental and theoretical methods following the AAPM recommendations published in the TG-43 report (Nath *et al.* 1995). Following this protocol, the dose distribution around a sealed brachytherapy source can be determined using the following formalism:

$$\dot{D}(r,\theta)=\frac{\Lambda S_k G(r,\theta)}{G(r_o,\theta_o)} g(r)F(r,\theta) \qquad (1)$$

Where

Λ is the dose rate constant

G(r, θ) is the geometry function

g(r) is the radial dose function

F(r, θ) is the anisotropy function

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

Dose rate constant

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 value is expressed in units of cGy.h^{-1}U^{-1}, where U is the unit of air-kerma strength of the source and is defined as 1U=1μGym^2/h^{-1}=1 cGycm^2/h^{-1}. The dose rate constant of the source is measured using LiF TLDs in Solid Water^{TM} (water equivalent phantom material) and calculated using Monte Carlo simulation technique in water and Solid Water^{TM} as:

$$\Lambda=\frac{\dot{D}(1cm,\pi/2)}{S_k} \qquad (2)$$

Radial Dose Function, g(r)

Radial dose function, g(r), describes the attenuation in tissue of the photons emitted from a brachytherapy source. Radial dose function is defined as:

$$g(r)=\frac{\dot{D}(r,\pi/2)G(r_o,\pi/2)}{\dot{D}(r_o,\pi/2)G(r,\pi/2)} \qquad (3)$$

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, respectively, along the transverse bisector of the source. r_0 is the reference distance and usually is defined to be 1 cm, as is the case for this project. G(r, θ) is known as the geometry function which takes into account the effect of the physi-

cal shape of the radioactive material inside the source on the dose distribution at a given point. The geometry function is defined by the AAPM (Nath *et al.* 1995) as:

$$G(r,\theta)=\begin{cases}1/r^2 & \text{Point Source approximation}\\ \dfrac{\tan^{-1}(\frac{x+L/2}{y})-\tan^{-1}(\frac{x-L/2}{y})}{Ly} & \text{Linear Source approximation}\end{cases} \quad (4)$$

Where L is the active length of the source as shown in figure 1.

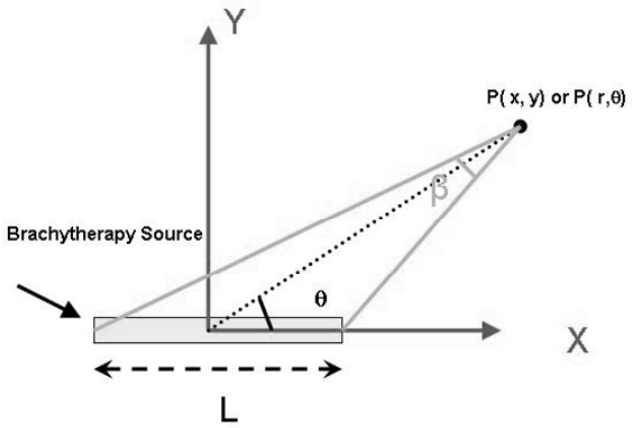

Figure 1. Schematic diagram of the source orientation and point of interest, P(x,y).

A. Anisotropy Function, F(r, θ)

The anisotropy function, F(r, θ), describes the variation in dose distribution around a brachytherapy source due to self-absorption and oblique filtration of radiation in the capsule material. The anisotropy function is defined as:

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

Where $\dot{D}(r,\theta)$ and $\dot{D}(r,\pi/2)$ are the dose rates measured at distances of r and angles of θ and π/2 relative to the longitudinal axis of the source, respectively. The anisotropy factor, ϕ_{an} (r), is defined following the TG-43 recommendations as:

$$\phi_{an}(r)=\frac{\int \dot{D}(r,\theta)\sin(\theta)d\theta}{2\dot{D}(r,\pi/2)} \quad (6)$$

The anisotropy constant, ϕ_{an}, of the new source was determined by averaging the individual anisotropy factors in a given medium.

TLD dosimetry technique

Dose distributions around the brachytherapy sources are normally measured in a Solid Water™ phantom material (Model 457, Radiation Measurements Inc., RMI, Middletown, WI) using TLD-100 LiF thermoluminescent dosimeters (Harshaw/Bicron 6801 Cochran Rd., Solon, OH 44139). For these measurements, slabs of Solid Water™ phantom material are machined to accommodate the source and LiF chips of dimensions (3.1 × 3.1 × 0.8 mm^3) and (1.0 × 1.0 × 1.0 mm^3). Figure 2 shows the schematic diagrams of two samples of experimental setup for measurements of the dose rate constant and radial dose function at University of Kentucky (2A) (Meigooni *et al.* 2002a) and University of Wisconsin (2B) (Peterson and Thomadson 2002). The specially designed patterns of the TLD locations were selected to minimize the interference of any one TLD to the other TLD chips.

Figure 3 shows the phantom design that was used for the measurement of the anisotropy function at University of Kentucky (Meigooni *et al.* 2002a) while the phantom design shown in Figure 2B is used at university of Wisconsin (Peterson and Thomadson 2002).

The TLD measurements are performed by surrounding the source and the TLD chips with at least 10 cm of phantom material to provide full scattering conditions. TLDs are then read using a TLD reader and responses are converted into dose using the following equation (Meigooni *et al.* 1995).

$$\dot{D}(r,\theta)=\frac{R}{T.S_k.\varepsilon.E(r).dT.F_{lin}} \quad (7)$$

Where $\dot{D}(r,\theta)$ is the dose rate to water in the medium of measurement at a given distance and angle relative to the longitudinal axis of the source. R is the TLD response, corrected for

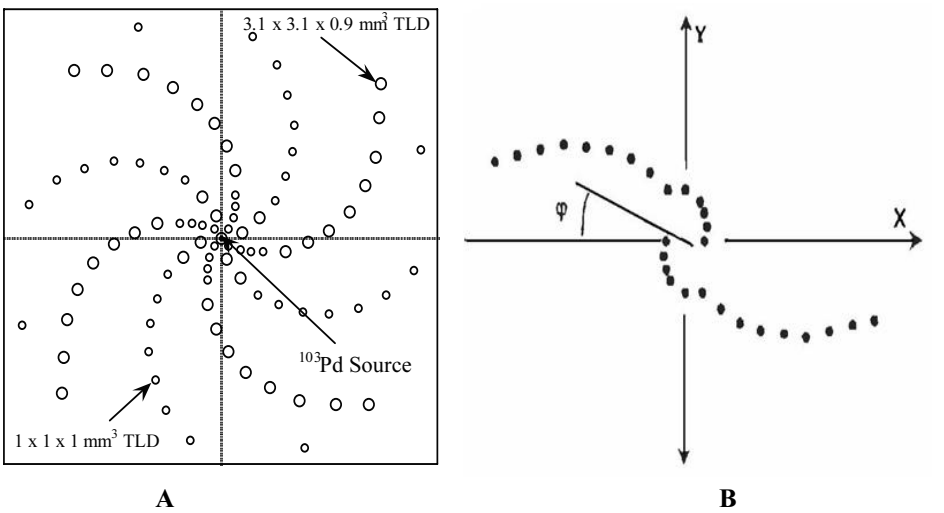

Figure 2. Schematic Diagram of the experimental setup at University of Kentucky (**A**) and Schematic Diagram of the experimental setup at University of Wisconsin (**B**).

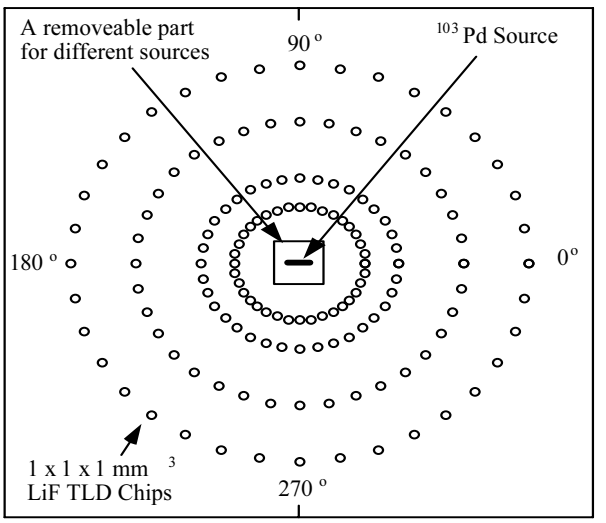

Figure 3. The schematic diagram of the experimental setup for measurements of anisotropy functions at University of Kentucky.

background and the physical differences between individual TLDs of the same batch (Meigooni *et al.* 1995), T is the experimental time (hours), S_k is the source air-kerma strength at the beginning of the measurement. e is the calibration factor for the TLD response (nC/cGy) measured with a 6 MV X-ray beam from a linear accelerator. E (r) is a correction factor for the energy dependence of the TLD response between the calibration beam and the brachytherapy source. dT is a correction factor used to account for the decay of the source during irradiation. F_{lin} is the nonlinearity correction of the TLD response for the given dose.

Monte Carlo Simulations

The Monte Carlo simulation has become an invaluable technique in the characterization of new brachytherapy sources within the last several years. Presently several Monte Carlo codes, such as MCNP, PTRAN and EGS4 are commonly used for dosimetric characterization of the brachytherapy sources (Williamson 1987, 1988). These codes simulate the interactions as emitted from the source until they are exited from the phantom or their energy is fully absorbed. The photon cross section library used in this code was DLC-200) (DLC = Data Library Code) is distributed by the Radiation Shielding Information Computing Center (RSICC) (Roussin *et al.* 1983). Also, this code uses the corresponding mass-energy absorption coefficients by Hubbell and Seltzer (1995), for converting energy fluence into absorbed dose. The Monte Carlo code allows the simulation in various phantom material, particularly the liquid was in which the experimental procedure is either impossible or it is really difficult. The simulation in air will provide the air kerma strength, which is currently the

used in brachytherapy dosimetry in place of source activity.

RESULTS AND DISCUSSION

Reviewing the recent dosimetry of the brachytherapy sources performed by various investigators (Meigooni *et al.* 2000, 2001, 2002b, Wallace and Fan 1999a & b, Weaver 1998, Williamson 2000, Li *et al.* 2000, Rivard 2001, Wallace 2000, Nath and Yue 2000, Karaiskos *et al.* 2001, Kirov and Williamson 2001, Reiners *et al*. 2001) following the TG-43 recommendation for various brachytherapy sources shows the following results.

a. ***Dose rate Constant:*** Table 1 shows a comparison of the measured and Monte Carlo calculated dose rate constants of several different types of ^{125}I, by various investigators, following the TG-43 protocol. These results show that there is a good agreement between different investigators on the same source. Moreover, different investigators have shown the same impact of the source geometry on the dose rate constant of different source models from the same isotope.

b. ***Radial Dose Function:*** Figure 4 shows a comparison between the measured and Monte Carlo simulated radial dose function of a ^{125}I source (Meigooni *et al.* 2002c). This figure indicates a good agreement between the two methods of dosimetry. Therefore, knowing the accurate source geometry, one could reproduce the experimental data using Monte Carlo simulation technique. Figure 5 shows a comparison between the radial dose functions of various models of ^{125}I brachytherapy sources, in water. This figure indicates that the following the TG-43 protocol will lead to a consistent result for the radial dose function of the brachytherapy sources. Similar consistent results were observed for other brachytherapy sources such as ^{103}Pd and ^{192}Ir.

Table 1. Measured or calculated dose rate constants, Λ, of ^{125}I and ^{103}Pd brachytherapy sources in water.

Source and Model, ^{125}I	Dose Rate Constant, Λ (cGy*cm^2/hr*U)	Source of Data	Reference
Model 6702	1.037 1.039	TG-43 Monte Carlo	(Nath *et al.* 1995) (Williamson, 1991)
Model 6711	0.981 0.978	TG-43 Monte Carlo	(Nath *et al.* 1995) (Williamson, 1991)
MED3631A/M	1.060 1.083 1.067	TLD TLD Monte Carlo	(Wallace and Fan ,1999) (Li *et al.* 2000) (Rivard *et al.* 2001)
InterSource^{125}	1.065 1.013 1.050 1.020	TLD Monte Carlo TLD Monte Carlo	(Meigooni *et al.* 2002a) (Meigooni *et al.* 2002a) (Reniers *et al.* 2001) (Reniers *et al.* 2001)
SelectSeed	0.954	Monte Carlo	(Karaiskos *et al.* 2001)
I-Plant, Model 3500	1.010	TLD	(Duggan and Johnson, 2001)

Source and Model, ^{103}Pd	Dose Rate Constant, Λ (cGy*cm^2/hr*U)	Source of Data	Reference
Model 200	0.680 0.650	Monte Carlo TLD	(Williamson, 2000) (Nath and Yue, 2000)
MED3633	0.680 0.677 0.682	TLD Monte Carlo Diode	(Wallace and Fan, 1999) (Li *et al.* 2000) (Li *et al.* 2000)

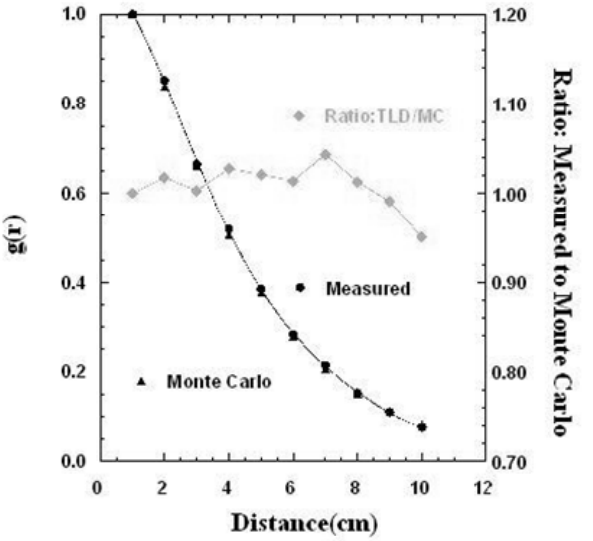

Figure 4. Comparison between the Monte Carlo simulated and measured radial dose function of ^{125}I source (Meigooni *et al.* 2000).

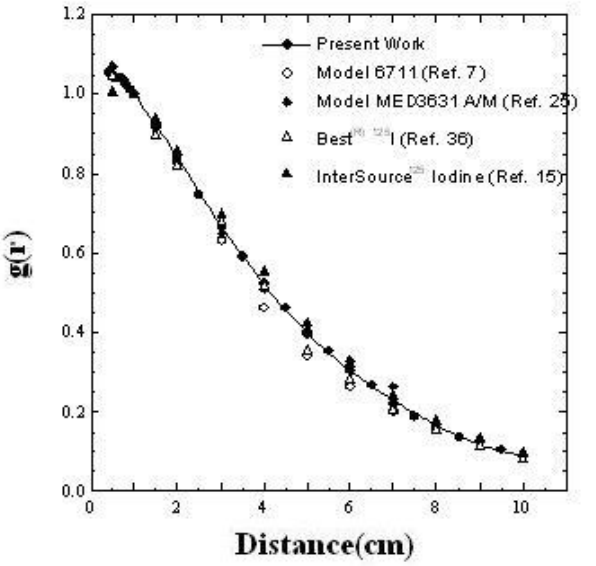

Figure 5. Comparison between the radial dose function of different commercially available ^{125}I sources.

c. ***Radial Dose Function:*** Figure 6 shows a comparison between the measured and Monte Carlo simulated anisotropy functions of a ^{125}I source (Meigooni *et al.* 2002c). This figure indicates a good agreement between the two methods of dosimetry. Figure 7 shows a comparison between the anisotropy functions of various models of ^{125}I brachytherapy sources, in water. This figure indicates that the following the TG-43 protocol will lead to a consistent result for the radial dose function of the brachytherapy sources. The differences between the anisotropy functions at the small angles are attributed to the shape and sizes of the end caps. Table two shows the anisotropy

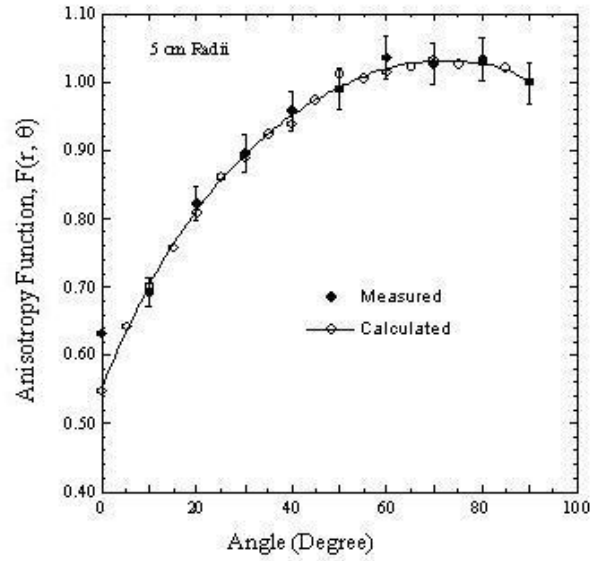

Figure 6. Comparison between the Monte Carlo simulated and measured anisotropy function of ^{125}I source (Meigooni *et al.* 2000).

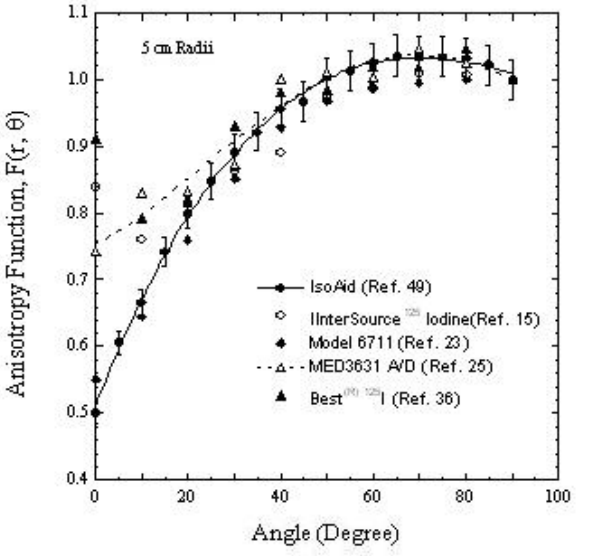

Figure 7. Comparison between the anisotropy functions of different commercially available ^{125}I sources.

factors at different radial distances and anisotropy constants of the commercially available ^{125}I and ^{103}Pd sources.

CONCLUSION

The TG-43 protocol has introduced a recommendation and formalism that leads to a consistent brachytherapy dosimetry. Moreover, these dosimetric parameters are easily measurable today and can be produced using the Monte Carlo simulation technique. The variation of the dosimetric parameters of a given source by different investigators is less than 5% as compared to the 60% variation before TG-43. This consistency of determination of dose distribution around the brachytherapy sources along with improvement in the technological aspect of the brachytherapy procedures leads to a better and more promising clinical results for cancer patients.

It is good to know that an updated TG-43 has recently become available (Rivard *et al.* 2004) to cover the short coming of this recommendation as listed below:

a. During the process of the brachytherapy source dosimetry, it has been found that the Monte Carlo simulated anisotropy function was highly sensitive to the thickness and depth of the active layer within the source. This parameter was not always accurately known by the vendors, therefore, some investigators selected the thickness of the active layer to match experimental data. The effect of the active layer thickness on the TG-43 dosimetric parameters of a sample source will be presented in order to justify that the Monte Carlo simulated data by itself is not sufficient for clinical application.
b. Several investigators have argued that the geometric function of the source, G(r, theta), as defined in the TG-43 report was not applicable for some source designs. For instance, the active length of a source with a long active wire cannot be defined in the same way as a source with active beads at each end. Some investigators proposed a Monte Carlo simulated geometric function; however, no analytical method is available at this time. With these

Table 2. Anisotropy factors, $\phi_{an}(r)$ and anisotropy constant, ϕ_{an} of ^{125}I and ^{103}Pd brachytherapy sources.

Model and Source	0.75 cm	1.0 cm	2.0 cm	3.0 cm	4.0 cm	5.0 cm	6.0 cm	7.0 cm	ϕ_{an}
Model 6702		0.960	0.952	0.951	0.954	0.954			0.960*
Model 6711		0.944	0.941	0.942	0.943	0.944			0.948*
MED3631 A/M	0.965	0.952	0.945	0.946	0.947	0.948			0.948
InterSource^{125}		0.959	0.947	0.983	0.943	0.949	0.949	0.965	0.956
selectSeed		0.933	0.936	0.941	0.943	0.945	0.946		0.936
BT-125-I		0.976	0.973	0.970		0.968		0.964	0.975
Best ^{125}I			0.990			0.988		0.970	0.982
Model 200		0.866	0.862	0.868	0.871	0.872			0.872*
InterSource^{103}		0.901	0.891	0.889		0.892		0.889	0.894
MED3633		0.927	0.919	0.916	0.927	0.917			0.921
Best ^{103}Pd		0.895	0.869	0.880	0.877	0.886	0.907	0.841	0.880

* The anisotropy constants were calculated as the mean value of the anisotropy factors.

and other shortcomings in AAPM brachytherapy dosimetry techniques, a review and revision of the TG-43 protocol is warranted to include newer source designs.

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