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A systematic evaluation of the dose-rate constant determined by photon spectrometry for twenty-one different models of low energy photon-emitting brachytherapy sources

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Abstract

Purpose—To perform a systematic comparison of the dose-rate constant (Λ) determined by the photon spectrometry technique (PST) with the consensus value (Λ_{CON}) recommended by the American Association of Physicist in Medicine (AAPM) for twenty-one low energy photon-emitting interstitial brachytherapy sources.

Method and Materials—A total of 63 interstitial brachytherapy sources (21 different models with 3 sources per model) containing either ^{125}I (14 models), ^{103}Pd (6 models), or ^{131}Cs (one model) were included in this study. A photon spectrometry technique (*Med. Phys.* **34**, 1412-1430, 2007) was used to determine the dose-rate constant (Λ_{PST}) for each source model. Source-dependent variations in Λ_{PST} were analyzed systematically against the spectral characteristics of the emitted photons and the consensus values recommended by the AAPM brachytherapy subcommittee.

Results—The values of Λ_{PST} for the encapsulated sources of ^{103}Pd , ^{125}I , and ^{131}Cs varied from 0.661 $\text{cGyh}^{-1}\text{U}^{-1}$ to 0.678 $\text{cGyh}^{-1}\text{U}^{-1}$, 0.959 $\text{cGyh}^{-1}\text{U}^{-1}$ to 1.024 $\text{cGyh}^{-1}\text{U}^{-1}$, and 1.066 $\text{cGyh}^{-1}\text{U}^{-1}$ to 1.073 $\text{cGyh}^{-1}\text{U}^{-1}$, respectively. The relative variation in Λ_{PST} among the six ^{103}Pd source models, caused by variations in photon attenuation and in spatial distributions of radioactivity among the source models, was less than 3%. Greater variations in Λ_{PST} were observed among the fourteen ^{125}I source models; the maximum relative difference was over 6%. These variations were caused primarily by the presence of silver in some ^{125}I source models and, to a lesser degree, by the variations in photon attenuation and in spatial distribution of radioactivity among the source models. The presence of silver generates additional fluorescent x-rays with lower photon energies which caused the Λ_{PST} value to vary from 0.959 $\text{cGyh}^{-1}\text{U}^{-1}$ to 1.019 $\text{cGyh}^{-1}\text{U}^{-1}$ depending on the amount of silver used by a given source model. For those ^{125}I sources that contain no silver, their Λ_{PST} was less variable and had values within 1% of 1.024 $\text{cGyh}^{-1}\text{U}^{-1}$. For the 16 source models that currently have an AAPM recommended Λ_{CON} value, the agreement between Λ_{PST} and Λ_{CON} was less than 2% for 15 models and was 2.6% for one ^{103}Pd source model.

Conclusions—Excellent agreement between Λ_{PST} and Λ_{CON} was observed for all source models that currently have an AAPM recommended consensus dose-rate constant value. These results demonstrate that the PST is an accurate and robust technique for the determination of dose-rate constant for low-energy brachytherapy sources.

Keywords

Dose-rate constant; palladium-103; iodine-125; cesium-131; energy spectrum; photon spectrometry

I. Introduction

In a commonly used formalism, i.e., the American Association of Physicists in Medicine (AAPM) Task Group 43 (TG-43) formalism, the radiation dose distribution produced by a radioactive source for interstitial brachytherapy is calculated as a product of the source strength (in unit of air-kerma strength), the dose-rate constant, and some 3-dimensional relative dose distribution parameters such as the radial dose function and anisotropy factor (Nath, Anderson et al. 1995). The dose-rate constant (Λ) is defined by the AAPM TG-43 as the absolute dose-rate delivered by the source to a reference point in water (1 cm away from the source on the transverse bisector) per unit air-kerma strength. It provides the necessary link between the relative dosimetry properties of the source and the absolute dose to be delivered to the entire irradiated volume in clinical dosimetry (Nath, Anderson et al. 1995). Any error introduced in the determination of Λ would, therefore, propagate directly to patient dosimetry and, unlike random errors, such an error will affect all patients in the same direction for as long as the inaccurate value of Λ continues to be used.

The need to have an accurate determination of Λ becomes inevitable when more than one source models are made available for clinical use. With only one source model, the impact of a systematic error in Λ for a given clinical application could be minimized, in theory, by conducting clinical trials to establish the necessary prescription dose for a desired clinical outcome. As long as the prescription dose and the associated treatment technique established by the trial are used consistently for patients of similar characteristics, the clinical impact of the systematic error in Λ would be minimal. However, when another new source model (even with the same radionuclide) is introduced, the clinical experience gained from using the existing source model (e.g. the prescription dose) cannot be applied safely to the new source model if large uncertainties exist in the dose-rate constants (and/or other basic relative dosimetry parameters) of the two source models. In the last decade, more than twenty new source models containing ^{125}I , ^{103}Pd , or ^{131}Cs with different internal source designs have been introduced for interstitial brachytherapy. Unless independent clinical trials are conducted to establish the prescription dose for each new source model, which is impractical and costly, one must have an accurate and consistent dosimetry characterization for all source models being introduced into the clinical practice.

An accurate determination of Λ for brachytherapy sources emitting low-energy (<50 keV) photons, such as those encapsulated with ^{125}I , ^{103}Pd , or ^{131}Cs , has been challenging because of the lack of a simple dosimeter for accurate measurement of absolute dose close to the source. The most commonly used method has been LiF thermoluminescent dosimeter (TLD) which is very complex to use. Uncertainties in the Λ determined by TLD dosimetry are approximately 8-10% at one standard deviation and 16% at the 95% confidence level (Rivard, Coursey et al. 2004). Indeed, relatively large variations in the Λ values determined by different techniques and/or research groups for the same source model have been observed in the published reports (Rivard, Coursey et al. 2004). To minimize the impact of uncertainties associated with individual Λ -determinations on clinical dosimetry, AAPM currently recommends the use of a consensus dose-rate constant value, $\text{CON}\Lambda$, for each source model. For a given source model, the AAPM brachytherapy subcommittee examines the peer-reviewed literature and identify candidate dose-rate constant values obtained from experimental measurements ($\text{EXP}\Lambda$) and Monte Carlo simulations ($\text{MC}\Lambda$). The $\text{CON}\Lambda$ is established as the equally weighted average of the separately averaged experimental and Monte Carlo Λ values, $\text{CON}\Lambda = [\langle \text{EXP}\Lambda \rangle + \langle \text{MC}\Lambda \rangle]/2$. (Rivard, Coursey et al. 2004)

In 2007, we reported a photon spectrometry technique (PST) for the determination of dose-rate constant for low energy photon-emitting brachytherapy sources (Chen and Nath 2007). The PST technique utilizes a high-resolution gamma-ray spectrometer to measure the

source-specific photon characteristics emitted by each source and the well-established dose-deposition properties of mono-energetic photons in water to determine the dose-rate constant. This technique eliminates many of the difficulties arising from the detector size, the energy dependence of detector sensitivity, and the use of non-water-equivalent solid phantoms in absolute dose rate measurements using TLD or diodes. It also circumvents the uncertainties that may arise from using assumed nominal source geometry and composition in Monte Carlo simulations, i.e. the actual sources may differ from the nominal geometry and composition due to variations in manufacturing processes. The estimated overall uncertainty of the PST technique was within 4%, which is significantly smaller than the reported 8-10% uncertainty associated with the current TLD technique. The initial applications of the technique to selected source models have helped uncover a 15% discrepancy in the dose-rate constant determined for an earlier version of ^{131}Cs source (Chen, Bongiorni et al. 2005) and delineate the dependence of dose-rate constant on source structure (Nath and Chen 2007; Khan, Chen et al. 2008; Chen and Nath 2010). The aim of this work was to perform a systematic evaluation of the dose-rate constants determined by the PST technique for 21 available low-energy interstitial brachytherapy sources and to assess the accuracy and robustness of the PST technique by comparing the measured $\rho_{\text{PST}}\Lambda$ with the AAPM consensus values.

II. Method and Materials

A total of 63 low-energy interstitial brachytherapy sources (21 different models with 3 sources per model) containing ^{125}I (14 models), ^{103}Pd (6 models), or ^{131}Cs (one model) were obtained from various source manufacturers. The basic source information and structural characteristics of these sources are summarized in Table 1. To assist the discussion of different source designs and their impact on the measured dose-rate constants, graphical illustrations of the source structure obtained from published literature and/or source manufacturer brochures (see footnote of Table 1 for specific references) are also included.

In this work, the dose-rate constant of each source was determined using the photon spectrometry technique described earlier by Chen and Nath (Chen and Nath 2007). Briefly, the PST technique consists of both experimental measurement and theoretical calculation in the determination of dose-rate constant. Because the dosimetric characteristics of an encapsulated source are fundamentally determined by its emitted photons, the experimental part of PST involved direct measurement of the photon energy spectrum emitted by each source. The measured photon energy spectrum captures the effects of source construction and encapsulation on the photons emitted by the bare radionuclide (for examples, the photon absorption and scattering, and the production of characteristic x-rays in certain source construction materials). With the measured photon energy spectrum, the dose-rate constant of each source was then calculated from first-principles based on well-characterized dose deposition properties of mono-energetic photons in water.

The photon energy spectrum emitted along the radial direction in the transverse bisector of each source was measured in air by using a high-resolution and high-purity germanium spectrometer. Photon attenuation through air before reaching the detector was taken into account explicitly in the energy response calibration for the spectrometer. The dose-rate constant ($\rho_{\text{PST}}\Lambda$) was calculated from the measured relative energy spectrum according to (Chen and Nath 2007)

$$\Lambda = \frac{\sum_i n(E_i) E_i (\mu_{en}(E_i) / \rho)_{air} \Lambda(E_i)}{\sum_i n(E_i) E_i (\mu_{en}(E_i) / \rho)_{air}}, \quad (1)$$

where $n(E)$ denotes the number of emitted photons with energy E , $[\mu_{en}(E) / \rho]_{air}$ is the mass energy absorption coefficient of air, and $\Lambda(E)$ is the dose-rate constant of a mono-energetic photon source with energy E given by

$$\Lambda(E) = [\mu_{en}(E) / \rho]_a^w \frac{1}{S_K} \int_{-L/2}^{L/2} dx \rho(x) e^{-\mu(E) \sqrt{x^2+y^2}} (x^2+y^2)^{-1} B_{en}(\mu(E) \sqrt{x^2+y^2})|_{y=1cm}. \quad (2)$$

In Eq.(2), S_K is the air-kerma strength of the source, $\rho(x)$ is the linear density of source strength along the longitudinal axis of the source (with $\int_{-L/2}^{L/2} \rho(x) dx = S_K$), $[\mu_{en}(E) / \rho]_a^w$ is the ratio of mass energy absorption coefficients of water to air, $\mu(E)$ is the mass attenuation coefficient of water, and $B_{en}(\mu(E)r)$ is the energy buildup factor that accounts for the relative dose contribution from the scattered photons in water. (Berger 1968) The x -axis in Eq.(2) coincides with the longitudinal axis of the source, the origin of the x and y axes coincides with the center of the source, and L denotes the active length of the source. In numerical calculations, the mass attenuation and mass energy absorption coefficients were taken from the database maintained by NIST (Hubbell and Seltzer 2004) and the energy buildup factor was taken from the data published by Angelopoulos *et al.*, (Angelopoulos, Perris *et al.* 1991) which included both Rayleigh scattering and Compton scattering with bound atomic electrons as well as the transport of secondary electrons.

Source-model dependent variations in the measured $p_{ST}\Lambda$ were analyzed with respect to the spectral characteristics of the emitted photons and the spatial distributions of radionuclide in each source model. The utility of photon spectrometry technique for dose-rate constant determination was assessed by comparing the measured $p_{ST}\Lambda$ with the AAPM consensus value, $CON\Lambda$, for each source model that currently has such a value.

III. Results

A. ^{103}Pd sources

Table 2 summarizes the relative photon energy spectra and dose-rate constants determined for the six ^{103}Pd source models. The average photon energy emitted by each source was calculated from the energies of the emitted photons weighted by the measured energy spectrum. The average photon energy of the six ^{103}Pd sources varied from 20.55 keV (for the IBt 1032P source) to 20.69 keV (for the NASI MED3633 and Draximage Pd-1 sources). The measured $p_{ST}\Lambda$ varied from 0.661 $\text{cGy} \cdot \text{h}^{-1} \cdot \text{U}^{-1}$ (for the Draximage Pd-1 source) to 0.678 $\text{cGy} \cdot \text{h}^{-1} \cdot \text{U}^{-1}$ (for the Theragenics model 200 source), with a maximum relative difference of 2.5%.

The variation in $p_{ST}\Lambda$ among the six ^{103}Pd source models can be traced to variations in both the emitted photon energy spectra and the spatial distributions of radioactivity among these source models. A first-order estimation of the photon-energy-induced variation in $p_{ST}\Lambda$ can be obtained from the measured average photon energies by treating the sources as point sources. According Chen and Nath (Chen and Nath 2001), the maximum relative variation in $p_{ST}\Lambda$ over the average photon energies measured for the ^{103}Pd sources (in Table 2) would be approximately 1.5%, which is less than the observed 2.5% overall variation in $p_{ST}\Lambda$. The

additional variation in the $p_{ST}\Lambda$ arises from different spatial distributions of radioactivity in these source models. For example, the IBt 1032P source uses a novel biocompatible polymer shell for source encapsulation instead of the traditional titanium encapsulation used in other low-energy sources. The use of polymer encapsulation produced less spectral hardening of the photons emitted by ^{103}Pd as compared to the titanium encapsulation. As a result, the average photon energy emitted by the IBt 1032P source was lower than other sources. Because the dose-rate constant of a point source decreases monotonically with decreasing photon energy in this energy range (Chen and Nath 2001), the $p_{ST}\Lambda$ measured for the IBt 1032P source was indeed lower than all other ^{103}Pd sources, except the Draximage Pd-1 source even though the average photon energy emitted by the Draximage Pd-1 source was at the high end of the six sources. The heavy concentration of ^{103}Pd radioactivity near the two ends of the Draximage Pd-1 source, as compared to the uniform distribution of ^{103}Pd in the central region of the IBt 1032P source, results in a lower $p_{ST}\Lambda$ for the Draximage Pd-1 source. For the five source models that use titanium encapsulation, the attenuation of ^{103}Pd photons by the source encapsulation and internal structures produced different levels of hardening of the ^{103}Pd bare spectrum. Nonetheless, the variation in their $p_{ST}\Lambda$ values did not follow exactly the magnitude of their emitted average photon energies due to the additional variations in the spatial distribution of radioactivity.

B. ^{125}I sources

Table 3 summarizes the relative photon energy spectra and dose-rate constants determined for the fourteen ^{125}I source models. The average photon energy emitted by the sources ranged from 27.23 keV (for the SL-125 source) to 28.53 keV (for the STM1251 source). The $p_{ST}\Lambda$ of these source models varied from 0.959 to 1.024 $\text{cGyh}^{-1}\text{U}^{-1}$ with a maximum relative difference greater than 6%. The variation in $p_{ST}\Lambda$ among the ^{125}I source models was greater than that observed among the ^{103}Pd source models. This larger variation was due primarily to the presence of silver in some of the source models which generates variable amount of low-energy silver fluorescent x-rays in addition to the source-structure-induced variations in the emitted ^{125}I photon energy spectrum.

For the five source models that do not contain any silver, the emitted photon energies were within 0.5% of each other. Their $p_{ST}\Lambda$ values were less variable and were within 1% of a nominal value 1.024 $\text{cGyh}^{-1}\text{U}^{-1}$. The small differences in their $p_{ST}\Lambda$ can be traced to the differences in their source designs. For example, the MED3631-A/M source had the lowest $p_{ST}\Lambda$ value ($1.017 \pm 0.038 \text{ cGyh}^{-1}\text{U}^{-1}$) among the five sources because of its placement of radioactivity carries near the two ends of the source.

For the nine source models that contained silver, their $p_{ST}\Lambda$ values exhibited a strong dependence on the silver content (Nath and Chen 2007). Because the energy of silver fluorescent x-rays is lower than the photons emitted by ^{125}I , the average energy emitted by these sources was lower (by as much as 4.5%) than the sources that contain no silver. As a result, $p_{ST}\Lambda$ values were generally lower than those of silver-free sources. As the silver content decreased, the $p_{ST}\Lambda$ increased from a value of 0.959 $\text{cGyh}^{-1}\text{U}^{-1}$ to 1.019 $\text{cGyh}^{-1}\text{U}^{-1}$, approaching the values of silver-free sources. These results indicate that the numerical value of dose-rate constant can be significantly affected by the amount of fluorescent materials used in a given source model. In addition to ensuring correct geometric dimension and radioactivity distribution, one also needs to precisely control the amount of fluorescent materials used in each source during source production, so that the nominal dose-rate constant value established initially for the source model remains valid in ongoing clinical dosimetry. Alternatively, designing sources with non-fluorescent materials can eliminate the potential uncertainties discussed earlier with fluorescent materials on dose-rate constant determination.

C. ^{131}Cs source

Table 4 presents the relative energy spectra and the dose-rate constants determined for the model CS-1 Rev.1 and CS-1 Rev.2 ^{131}Cs sources. As shown in Table 4, the portion of the energy spectrum arising from the principal emissions of ^{131}Cs was nearly identical along the transverse axis of the source for the two production versions. However, the fluorescent x-ray yield, generated by the photoelectric interactions of ^{131}Cs photons with niobium present in the source construction materials, was significantly lower in the Rev.2 source. The Nb- K_{α} and Nb- K_{β} fluorescent x-ray intensities emitted by the Rev.2 sources were only about 35% of those measured from the CS-1 Rev.1 model (Chen and Nath 2010). Nonetheless, the average photon energy emitted by the two versions of ^{131}Cs sources were approximately the same (~ 30.5 keV). Consequently, the $\text{pST}\Lambda$ values for the Rev.1 and CS-1 Rev.2 sources were similar, $1.066 \pm 0.040 \text{ cGyh}^{-1}\text{U}^{-1}$ and $1.073 \pm 0.041 \text{ cGyh}^{-1}\text{U}^{-1}$, respectively. The reduced emission of lower-energy niobium fluorescent x-rays from the Rev.2 sources resulted in a slight (0.7%) increase in its dose-rate constant. As reported earlier by Chen et al, the $\text{pST}\Lambda$ value determined for the CS-1 Rev.1 source was about 15% greater than a TLD-determined value reported initially for the CS-1 Rev.0 source (Chen, Bongiorno et al. 2005). Subsequent dosimetric characterizations of the Rev.1 and Rev.2 sources by other research groups have also reported dose-rate constant values ranging from $1.032 \text{ cGyh}^{-1}\text{U}^{-1}$ (Monte Carlo using MCNP code) (Wittman and Fisher 2007) to $1.046 \text{ cGyh}^{-1}\text{U}^{-1}$ (Monte Carlo using MCNP code) (Rivard 2007), $1.048 \text{ cGyh}^{-1}\text{U}^{-1}$ (Monte Carlo using MCNP code) (Wang and Zhang 2008), and $1.063 \text{ cGyh}^{-1}\text{U}^{-1}$ (TLD in liquid water) (Taylor, Ibbott et al. 2008) which are closer to the $\text{pST}\Lambda$ value than that reported for the Rev.0 source (Murphy, Piper et al. 2004).

Compared to the data in Tables 2 and 3, the average photon energy emitted by the ^{131}Cs sources is significantly greater than those emitted by the ^{125}I (by $> 7\%$) and the ^{103}Pd (by $> 30\%$) sources. As a result, the dose-rate constant of ^{131}Cs sources is greater than those of ^{125}I and ^{103}Pd sources as expected (Chen and Nath 2001).

D. Comparison with AAPM recommended consensus dose-rate constant values

For the 21 source models included in this study, AAPM consensus dose-rate constant values were available for 13 ^{125}I source models and 3 ^{103}Pd source models (Rivard, Coursey et al. 2004; Rivard, Butler et al. 2007). No AAPM consensus value is currently available for the ^{131}Cs source. As shown in Tables 2 and 3, the $\text{pST}\Lambda$ values were within $\pm 2\%$ of the $\text{CON}\Lambda$ for all source models except the Best 2335 ^{103}Pd source. The relative difference for the Best 2335 ^{103}Pd source was 2.6%, which was the largest relative difference observed among the sixteen source models that currently have an AAPM determined $\text{CON}\Lambda$ value.

IV. Discussions

Permanent interstitial brachytherapy (PIB) involves implantation of low-energy photon sources directly into the tumor volume to create a therapeutically potent dose distribution that conforms to the shape of tumor volume with minimum radiation to the surrounding normal tissues. By reducing radiation exposure to the normal tissues, PIB has the potential to improve the therapeutic ratio of radiation therapy by delivering higher radiation dose to the tumor without additional increase in normal tissue toxicity. Radioactive sources encapsulated with photon-emitting radionuclide, such as ^{125}I , ^{103}Pd , or ^{131}Cs , are currently used in PIB for treating cancers in brain, head and neck, and prostate. In particular, PIB using image-guided transperineal implantation has remained a popular form of radiation therapy for carefully selected prostate cancer patients. The PIB procedure is minimally invasive and is usually performed in a one-day surgery unit on an outpatient basis. The treatment-related morbidity such as urinary incontinence and sexual impotence is fairly low

for PIB(Arterbery, Frazier et al. 1997), and control of the cancer in selected patients is as good as that with other treatment modalities such as surgery(Potters 2000; Ragde, Korb et al. 2000; Stokes 2000). For these reasons the public interest in PIB has grown rapidly in the last decade and more than 20 new source models have been introduced.

However, as discussed in the Introduction, relatively large uncertainties (on the order of 8 to 10% at one standard deviation) have been noted in the experimental determination of dose-rate constant for the low-energy sources. Uncertainties of this magnitude can negatively affect the clinical application of PIB. For example, dose variations on the order of 10% prescription dose have been shown to have significant effect on both biochemical recurrence-free survival rates (PSA-RFS)(Stock, Stone et al. 1998; Potters, Cao et al. 2001) and long-term recurrence-free survival rates (RFS) at 5, 10, and 15 years ($p=0.001$) for prostate PIB(Zelefsky and Whitmore 1997). Using the dose delivered to 90% prostate volume, D_{90} , as a dosimetry index, Stock et al found that the PSA-RFS at 4 years had a dramatic 20% reduction between implants with $D_{90} \geq 90\%$ prescription dose (PSA-RFS of 92%) and those with $D_{90} < 90\%$ prescription dose (PSA-RFS 68%)(Stock, Stone et al. 1998). Similar reduction in PSA-RSF (about 12%) was also seen by Potters et al when D_{90} changed from greater than 90% to less than 90% prescription dose(Potters, Cao et al. 2001). These results strongly suggest that the current level of systematic uncertainty in absolute dosimetry needs improvement. An error in Λ that causes a systematic under-dose of 10% could result in sub-optimal care for all patients using that source model.

The results shown in Tables 2 and 3 demonstrate that the photon spectrometry technique has consistently produced dose-rate constant values that are in good agreement with the AAPM consensus values for those source models that currently have such a value. As mentioned in the Introduction, the PST technique is different from the traditional experimental and Monte Carlo simulation techniques. The hybrid usage of photon characteristics actually emitted by the source and of well-defined dose-deposition properties of mono-energetic photons in water allows PST to avoid the difficulties associated with direct measurement of absolute dose-rate near a source and the uncertainties in modeling source geometry and composition amid manufacturing tolerance. As a result, the overall uncertainty of the PST technique is significantly smaller than the traditional TLD technique. It suggests that the ${}_{PST}\Lambda$ determined by the photon spectrometry technique can be used either as an independent check of the ${}_{EXP}\Lambda$, ${}_{MC}\Lambda$, and the consensus values or as an addition to ${}_{EXP}\Lambda$ and ${}_{MC}\Lambda$ in the derivation of consensus values.

It is worth noting that, once the spectrometry system is set up and calibrated, the PST technique is quick (~ 2 hours per source) and robust for dose-rate constant determination. This feature has already enabled us to determine the dose-rate constants for 21 source models with the *same* photon spectrometry equipment and analysis method, which is essential for investigating the relative variations of dose-rate constants among different source designs. When the systematic uncertainties associated with the dosimetry technique and analysis method are kept the same for all source models, the relative difference of dose-rate constants between different source models (or different production versions of the same source model) can be quantified with greater precision (Chen, Bongiorni et al 2010). As illustrated in the Results section, even small differences between two source models can be delineated with respect to the variations in source design. This suggests that the PST technique can also be used as a QA tool for monitoring the constancy of dose-rate constant for sources of the same model but manufactured at different times.

V. Conclusions

A systematic assessment of the dose-rate constants determined by the photon spectrometry technique was carried out for 21 different models of low-energy interstitial brachytherapy sources. Excellent agreement between the dose-rate constants determined by the photon spectrometry technique and the AAPM recommended consensus values was observed for all source models that currently have a consensus value. It is demonstrated that the photon spectrometry technique can be used as an accurate and robust technique for the determination of dose-rate constant for low energy photon-emitting brachytherapy sources.

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Table 1

Source Information

Manufacturer	Name	Model	Type	Structural Characteristics	Schematics
Mills Pharmaceuticals Inc	ProstaSeed	SL-125	^{125}I	<p>Encapsulation: Titanium tube with 0.05-mm-thick wall and 0.3-mm-thick end-welds.</p> <p>Internal components: Five 0.50-mm-diameter silver spheres with radioactive iodine adsorbed to the surface of the spheres down to a nominal depth of several micrometers.</p>
Imagyn Medical Technologies Inc.	IsoStar	IS-1205I	^{125}I	<p>Encapsulation: Titanium tube with laser welded ends.</p> <p>Internal components: Five 0.56-mm-diameter silver spheres with silver iodide adhered to the spheres.</p>
Amersham Health	OncoSeed	6711	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and welded end caps.</p> <p>Internal components: A 3.0-mm long silver rod onto which ^{125}I is adsorbed.</p>
Amersham Health	EchoSeed	6733	^{125}I	<p>Encapsulation: Titanium capsule with circular grooves on external surface; 0.05-mm-thick wall with welded end caps.</p> <p>Internal components: A 3.0-mm-long and 0.5-mm-diameter silver rod onto which ^{125}I is adsorbed.</p>
IsoAid	Advantage	IAL-125A	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and rounded end caps welded at each end.</p> <p>Internal components: A 3-mm-long and 0.5-mm-diameter cylindrical silver core onto which ^{125}I is uniformly adsorbed.</p>
Nucletron Corp	SelectSeed	130.002	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall.</p> <p>Internal components: A 3.4-mm-long and 0.5-mm-diameter silver rod on which a layer of ~3 μm thick radioactive AgI is coated.</p>
Draximage Inc.	BrachySeed	LS-1	^{125}I	<p>Encapsulation: Titanium capsule, laser welded at the center of the seed with a 0.05-mm-thick wall.</p> <p>Internal components: ^{125}I is uniformly impregnated in two 0.5-mm-diameter ceramic alumina-silicate beads, separated by a 2.97-mm-long Pt/Ir radio-opaque marker.</p>
Implant Sciences Corp	I-Plant	3500	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and 0.25-mm-thick end welds.</p> <p>Internal components: Two conical silver markers flank a 0.64-mm-diameter quartz tube, on which ^{124}Xe ions are implanted in a thin silicon layer. Neutron activation converts the ^{124}Xe into ^{125}Xe which decays to ^{125}I before the source is shipped.</p>

Manufacturer	Name	Model	Type	Structural Characteristics	Schematics
Bebig/Theragenics Corp	IsoSeed/I-Seed	125.S06	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and ~0.44 to 0.48-mm-thick spherical end-welds.</p> <p>Internal components: ^{125}I is uniformly distributed in a 3.5-mm-long cylindrical ceramic (alumina, 2.88 g/cm³) annulus, which covers a 3.5-mm-long and 0.17-mm-diameter gold marker.</p>
Amersham Health	OncoSeed	6702	^{125}I	<p>Encapsulation: Titanium capsule.</p> <p>Internal components: ^{125}I is adsorbed on three anion exchange resin spheres.</p>
North America Scientific Inc.	Prospera I-125	MED3631-A/M	^{125}I	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and 0.05 to 0.15-mm-thick spherical end welds.</p> <p>Internal components: ^{125}I is uniformly distributed in four 0.5-mm-diameter polystyrene resin beads, which are separated into two groups by two gold-copper markers of the same diameter.</p>
IBt	Intersource-125	1251L	^{125}I	<p>Encapsulation: Two concentric titanium tubes, each with a 0.04-mm wall thickness, laser welded at the edges.</p> <p>Internal components: An x-ray marker composed of 0.045-mm-thick Pt/Ir alloy is attached to the inner tube. ^{125}I is deposited on the inner tube in three printed bands. The distance between the outermost edges of the activity bands is 3.7 mm.</p>
SourceTech Medical	^{125}I Implant Seed	STM1251	^{125}I	<p>Encapsulation: Titanium shell with 0.08 mm radial thickness and 0.13-mm-thick end caps.</p> <p>Internal components: A 0.36-mm-diameter gold radiographic marker embedded in a 0.51-mm-diameter and 3.81-mm-long aluminum tube. The inner and outer surfaces of the tube are coated with 2-μm-thick nickel and copper, respectively. A 17-nm thin layer of radioactive and cold iodine is deposited on these surfaces.</p>
Best Medical International Inc.	Best I-125	2301	^{125}I	<p>Encapsulation: Double wall titanium capsule with thinner walls at the ends of the source.</p> <p>Internal components: ^{125}I is distributed within a low atomic number cylindrical annulus that surrounds a tungsten rod.</p>
Theragenics	TheraSeed	200	^{103}Pd	<p>Encapsulation: Titanium tube (0.056 mm thick) with the tube ends closed by inverted "end-cups" (0.040 mm thick).</p> <p>Internal components: Two right circular cylindrical graphite pellets (of 0.56-mm diameter and 0.89-mm length), upon which a mixture of radioactive and nonradioactive palladium is added, separated by a cylindrical lead marker.</p>
North America Scientific Inc.	Prospera Pd-103	MED3633	^{103}Pd	<p>Encapsulation: Titanium capsule with 0.05-mm-thick wall and 0.05-mm to 0.15-mm-thick spherical end welds.</p> <p>Internal components: ^{103}Pd is uniformly distributed in four 0.5-mm-diameter polystyrene resin beads, which are separated into two groups by two gold-copper markers of the same diameter.</p>

Manufacturer	Name	Model	Type	Structural Characteristics	Schematics
Best Medical International Inc	Best Pd-103	2335	^{103}Pd	Encapsulation: Double-wall titanium capsule, with a total wall thickness of 0.08 mm and thinner ends.
				Internal components: Three ^{103}Pd -coated polymer spheres (0.56 mm diameter) flank each side of a 1.2-mm-long tungsten x-ray marker.	
IBt	OptiSeed ¹⁰³	1032P	^{103}Pd	Encapsulation: Biocompatible polymer shell in contrast to titanium capsule used by other seeds.
				Internal components: ^{103}Pd is distributed uniformly in two cylindrical polymer-based activity carriers (0.7-mm long and 0.4-mm diameter) which flank a 1.944 mm long radio-opaque cylindrical gold marker.	
Draximage Inc.	BrachySeed	Pd-1	^{103}Pd	Encapsulation: Titanium capsule with 0.05-mm-thick wall.
				Internal components: Two ^{103}Pd containing ceramic pellets (0.55-mm diameter) flank a 3.1-mm-long and 0.38-mm-diameter Pt/Ir x-ray marker. Surround the x-ray marker is a 1.13mm long Ti annulus flanked by two 1.189 mm long Ti spacers.	
IsoAid	Advantage	IAPD-103A	^{103}Pd	Encapsulation: Titanium tube with 0.05-mm-thick wall.
				Internal components: ^{103}Pd is distributed uniformly on the four carbon matrix spheres, two on each side of a 1.25-mm long and 0.5-mm diameter cylindrical silver marker.	
IsoRay	Proxcelan	CS-1 Rev.1	^{131}Cs	Encapsulation: Titanium tube with welded end-caps.
				Internal components: ^{131}Cs radionuclide is chemically attached to an inorganic substrate that surrounds a 0.25-mm-diameter gold wire.	
IsoRay	Proxcelan	CS-1 Rev.2	^{131}Cs	Encapsulation: Titanium tube with welded end-caps.
				Internal components: ^{131}Cs radionuclide is chemically attached to an inorganic substrate that surrounds a 0.27-mm-diameter gold wire.	

Schematic drawings for the Nucletron 130.002, IBt OptiSeed¹⁰³, and IsoAid IAPD-103A sources are taken from respective manufacturer's brochures. Schematic drawings for the IsoRay CS-1 Rev.1 and Rev.2 sources are taken from Chen, Bongiorno, et al 2005. The schematic drawing for the Draximage Pd-1 source was taken from Meigooni et al. (*Applied Radiation and Isotopes*, **58**, 533-541, 2003). Schematic drawings for all other sources are taken from AAPM TG-43 Update (Rivard et al 2004) and Supplement to 2004 Update (Rivard et al 2007) reports for consistency.

Table 2
Relative photon energy spectrum and dose rate constant measured for the Pd-103 sources

Manufacturer and Source Model	Relative Energy Spectrum			Average Energy (keV)	Λ (cGyhr ⁻¹ U ⁻¹)		
	¹⁰³ Pd - K _α (20.1 keV)	¹⁰³ Pd - K _β (22.9 keV)	¹⁰³ Pd - γ (39.7 keV)		This Study ^y ¹⁰³ Pd- Λ	TG-43 Consensus ^{CON} Λ	Rel. Difference 100*(¹⁰³ Pd- Λ / ^{CON} Λ -1)
Bare ¹⁰³Pd	1	0.185	0.001	20.55	-	-	-
Theragenics 200	1	0.228	0.002	20.65	0.678 ± 0.026	0.686 ± 0.033	-1.2
NASI MED3633	1	0.252	0.002	20.69	0.676 ± 0.026	0.688 ± 0.033	-1.7
Best Medical 2335	1	0.241	0.002	20.67	0.667 ± 0.025	0.685 ± 0.033	-2.6
IBt 1032P	1	0.199	0.001	20.58	0.664 ± 0.025	na	na
Draximage Pd-1	1	0.249	0.002	20.69	0.661 ± 0.025	na	na
IsoAid IAPD-103A	1	0.229	0.002	20.65	0.676 ± 0.026	na	na

Table 3
Relative photon energy spectrum and dose rate constant measured for the I-125 sources

Manufacturer and Source Model	Relative Energy Spectrum				Average Energy (keV)	Λ (cGyhr ⁻¹ U ⁻¹)		
	Ag - K _α (22.1 keV)	Ag - K _β (24.9 keV)	¹²⁵ I - K _α (27.3 keV)	¹²⁵ I - K _β (31.0 keV)	¹²⁵ I - γ (35.5 keV)	This Study _{PST} Λ	TG-43 Consensus _{CON} Λ	Rel. Difference 100*(_{PST} Λ/ _{CON} Λ-1)
Bare Source	—	—	1	0.226	0.058	—	—	—
MBI SL-125	0.274	0.064	1	0.245	0.066	0.959 ± 0.037	0.953 ± 0.046	0.6
Imagyn IS-12051	0.272	0.067	1	0.247	0.067	0.959 ± 0.036	0.940 ± 0.045	2.0
Amersham 6711	0.268	0.067	1	0.249	0.067	0.960 ± 0.037	0.965 ± 0.046	-0.5
Amersham 6733	0.268	0.069	1	0.251	0.069	0.961 ± 0.036	0.980 ± 0.047	-1.9
IsoAid IAI-125A	0.260	0.061	1	0.244	0.066	0.962 ± 0.037	0.981 ± 0.047	-1.9
Nedatron 130.002	0.243	0.059	1	0.246	0.066	0.962 ± 0.037	na	na
Draximage LS-1	0.082	0.020	1	0.257	0.072	0.977 ± 0.037	0.972 ± 0.047	0.5
Implant Sciences 3500	0.061	0.014	1	0.247	0.066	1.004 ± 0.038	1.014 ± 0.049	-1.0
Bebig/Theragenics 125.S06	0.011	0.006	1	0.245	0.065	1.019 ± 0.039	1.012 ± 0.049	0.7
6702 - OncoSeed	—	—	1	0.241	0.064	1.024 ± 0.039	1.036 ± 0.050	-1.2
NASIMED3631-A/M	—	—	1	0.242	0.064	1.017 ± 0.038	1.036 ± 0.050	-1.8
Best 2301	—	—	1	0.250	0.068	1.021 ± 0.039	1.018 ± 0.049	0.3
IBt - 125IL	—	—	1	0.248	0.068	1.024 ± 0.039	1.038 ± 0.050	-1.3
SourceTech Med. STM1251	—	—	1	0.256	0.071	1.024 ± 0.039	1.018 ± 0.049	0.6

Table 4

Relative photon energy spectrum and dose rate constant measured for the Cs-131 source

Manufacturer and Source Model	Relative Energy Spectrum					Average Energy (keV)	Λ (cGyh ⁻¹ U ⁻¹)		
	¹³¹ Cs - L (4.1 keV)	Nb-K _α (16.6 keV)	Nb-K _α (18.7 keV)	¹³¹ Cs-K _α (29.7 keV)	¹³¹ Cs-K _{β1,β3} (33.6 keV)	¹³¹ Cs-K _{β2} (34.4 keV)	This Study ^{PST} Λ	TG-43 Consensus CONΛ	Rel. Difference 100%(^{PST} Λ/ ^{CON} Λ-1)
Bare ¹³¹ Cs	0.143	—	—	1	0.178	0.035	-	-	
Isoray CS-1	0.0	0.007	0.001	1	0.201	0.050	1.066 ± 0.040	na	na
Isoray CS-1 Rev.2	0.0	0.003	0.001	1	0.202	0.050	1.073 ± 0.041	na	na