Original paper

Dosimetric comparison between microSelectron iridium-192 and flexi cobalt-60 sources in high-dose-rate brachytherapy using Geant4 Monte Carlo code

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Abstract

Purpose: Manufacturing of miniaturized high activity iridium-192 (^{192}Ir) sources have been made a market preference in modern brachytherapy. Smaller dimensions of the sources are flexible for smaller diameter of the applicators, and it is also suitable for interstitial implants. Presently, cobalt-60 (^{60}Co) sources have been commercialized as an alternative to ^{192}Ir sources for high-dose-rate (HDR) brachytherapy, since ^{60}Co source have an advantage of longer half-life comparing with ^{192}Ir source. One of them is the HDR ^{60}Co Flexisource manufactured by Elekta. The purpose of this study was to compare the TG-43 dosimetric parameters of HDR flexi ^{60}Co and HDR microSelectron ^{192}Ir sources.

Material and methods: Monte Carlo simulation code of Geant4 (v.11.0) was applied. Following the recommendations of AAPM TG-43 formalism report, Monte Carlo code of HDR flexi ^{60}Co and HDR microSelectron ^{192}Ir was validated by calculating radial dose function, anisotropy function, and dose-rate constants in a water phantom. Finally, results of both radionuclide sources were compared.

Results: The calculated dose-rate constants per unit air-kerma strength in water medium were 1.108 cGy h^{-1}U^{-1} for HDR microSelectron ^{192}Ir, and 1.097 cGy h^{-1}U^{-1} for HDR flexi ^{60}Co source, with the percentage uncertainty of 1.1% and 0.2%, respectively. The values of radial dose function for distances above 22 cm for HDR flexi ^{60}Co source were higher than that of the other source. The anisotropic values sharply increased to the longitudinal sides of HDR flexi ^{60}Co source, and the rise was comparatively sharper to that of the other source.

Conclusions: The primary photons from the lower-energy HDR microSelectron ^{192}Ir source have a limited range and are partially attenuated when considering the results of radial and anisotropic dose distribution functions. This implies that a HDR flexi ^{60}Co radionuclide could be used to treat tumors beyond the source compared with a HDR microSelectron ^{192}Ir source, despite the fact that ^{192}Ir has a lower exit dose than HDR flexi ^{60}Co radionuclide source.

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Key words: Monte Carlo dosimetry, brachytherapy, flexi HDR ^{60}Co, microSelectron HDR ^{192}Ir, Geant4.

Purpose

Certainly, Monte Carlo (MC) simulation is the most accurate approach of dose computation showing many advantages [1]. For instance, brachytherapy source photon beams can be modelled accurately, and characteristics of photon and electron transports in a heterogeneous media can consider [1]. Accurate modelling of brachytherapy source requires full details of the geometry information and components' compositions. This is because, if there are errors in the geometry information and components' compositions, increased uncertainties of dose calculation can occur. Therefore, these details are usually provided by manufacturers.

Recently, many works in brachytherapy have studied dose distributions around radioactive sources. Monte Carlo codes, including MCNPX [2], PENELOPE [3], Geant4

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[4, 5], and EGSnrc [6], are specifically built to simulate charged particles transport and radiation interaction, with matter at energy levels used in medical physics [7].

Geant4 Code [8] originally designed for high energy physics experiments has been used in many applications, such as nuclear physics and space sciences. While allowing simulation of low energy particles, its applications are also extended to medical physics. This study comprised Geant4 code MC-based dosimetry in accordance with AAPM TG-43UI [9] protocol for flexi high-dose-rate (HDR) ^{60}Co source (Elekta AB, Stockholm, Sweden) [10] and microSelectron HDR ^{192}Ir (Nucletron, The Netherlands) [2]. The goal of this study was to compare the various dosimetric parameters of TG-43UI [9] of the two sources in a water phantom. In the literature, some authors have published relevant dosimetry data with different methodology for both the sources. Vijande *et al*. [10] reported a dosimetric characterization of flexi ^{60}Co source using Geant4 [4, 5] and PENELOPE [3]. Recently, Almansa *et al*. [5] published a complete dosimetric characterization of Elekta flexi ^{60}Co HDR source using PENELOPE Monte Carlo code. Moreover, Williamson and Li [11] and Kirov *et al*. [12] published an extensive base of dosimetry data based on MC calculations, thermoluminescent dosimeter (TLD), and diode measurements, including dose-rate constants and anisotropy functions for the microSelectron HDR ^{192}Ir radionuclide source. Similarly, Russel and Ahnesjö [13] performed MC calculations based on EGS4 code of the same quantities in a water phantom, and considering electron binding energy of scattering atom in incoherent scattering process. Karaiskos *et al*. [14] also calculated dose-rate distribution in a water around this source using analytical MC simulation. Furthermore, Zabihzadeh *et al*. [2] calculated TG-43 parameters based on MCNPX Monte Carlo code.

In this study, the Geant4 [15] Monte Carlo code was used to calculate the TG-43 dosimetric parameters, with the same methodology for the two sources. Dose rates were determined considering the photon transport only. Anisotropy functions $F(r, \theta)$ and radial dose $g(r)$ were compared with other published data for the sources and between each other, whereas dose-rate constants were compared with the published data only. To the best of our knowledge, no research has been published comparing TG-43 parameters of both microSelectron ^{192}Ir and flexi ^{60}Co HDR radionuclide sources using Geant4 Monte Carlo simulation.

Material and methods

Source description

MicroSelectron HDR ^{192}Ir source (model No., 105.002) consists of a 3.50 mm long active source, with a diameter of 0.60 mm enclosed in a 0.85 mm diameter 304 stainless steel capsule (density of 7.8 g/cc). The tip of encapsulation is assumed to be a 0.108 mm thick conical section, with a half angle of 23.6° and radius of the face of 0.17 mm. The conical section is attached to a 0.49 mm long solid cylindrical section, followed by a 3.6 mm long hollow section with an inner diameter of 0.335 mm. Following the hollow section of a 0.40 mm long solid conical section with a half-angle of the cone is assumed to be 24° attached to the conical section of a 0.5 mm section of 304 stainless steel cable. The reason to consider only a 5 mm long cable and not a longer one was based on the specifics of clinical setting. In clinical applications, where the portion of a cable near the source is aligned with the axis of a source, doses around the cable are, in general, insignificant. Consequently, the cable length considered in the simulations is not a crucial issue [8, 16]. Figure 1A shows the geometry of microSelectron HDR ^{192}Ir source used in MC simulation.

The source design, dimensions, and materials of flexi ^{60}Co HDR source was provided by the manufacturer, and shown schematically in Figure 1B. It is composed of a central cylindrical active core made of metallic ^{60}Co, with a density of 8.9 g/cc, 3.5 mm in length and 0.5 mm in diameter [10, 14, 17]. The active core of flexisource ^{60}Co radionuclide source is covered by a cylindrical 316L stainless steel of elemental composition by weight

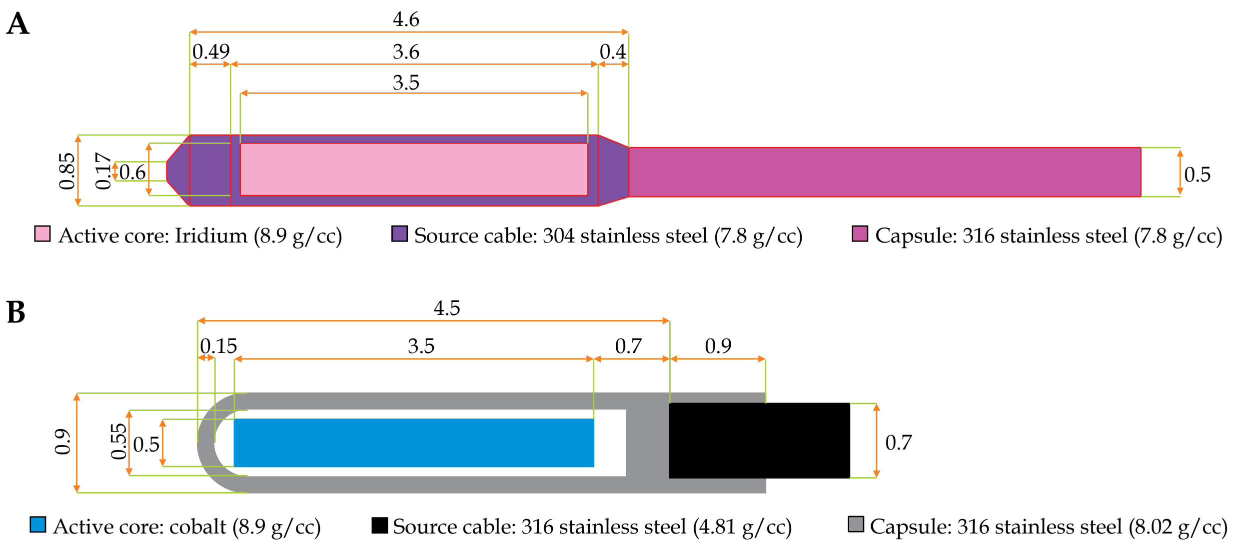

Fig. 1. A schematic drawing illustrating the structure of: **A**) HDR microSelectron ^{192}Ir source [2] and **B**) HDR flexi ^{60}Co source [10]

(67% Fe, 11% Ni, 18% Cr, 2% Si, 2% Mn) [5, 10], with density of 8.02 g/cc. For this study, 2 mm length and a 0.7 mm diameter 316L steel cable was included, with a mass density of 4.81 g/cc. Interstitial areas for ^{60}Co source between the active element and the cover were surrounded by an air shell of 0.05 mm thick, and air shell was encapsulated by a layer of 0.9 mm of external diameter stainless steel [2, 5]. These dimensions were applied to re-create a model of the source in Geant4.

Monte Carlo calculations

Monte Carlo methods are statistical simulation methods that provide approximate solutions to a wide variety of physical and mathematical problems by performing statistical sampling processes, which utilize a sequence of random numbers and probability to obtain an approximation to the problem under simulation. For this work, the geometry and tracking (Geant4 toolkit) [15] version 11.0 was applied for the simulation study. Geant4 is an object-oriented MC toolkit written in C^{++} language for simulating the passage of particles through matter [18]. It covers a broad range of functionalities, including complex geometry definition, physical processes modeling, particle tracking, and hits recording. This toolkit has been applied extensively to MC simulation related to brachytherapy [18]. The accuracy of Geant4 simulation methods for external radiotherapy and internal brachytherapy have been previously established [8, 19].

Geant4 defines particle type and physics processes in physics list class [20]. We used low energy package of Geant4 in the physics list class for modeling of Compton scattering, photo-electric effect, and Rayleigh scattering processes [21]. Class library in Geant4 provides several random number engines. In our application, the RanecuEngine was chosen at the beginning of the main program [18]. It was then possible to reset random number seed between runs in our program. General particle source method was applied to define a primary source particle. In this method, many internal commands for definition of the source position, geometry, and type of emitted particle were defined in the G4General-Particle-Source class. These commands were done through a macro file without any change in the main program.

We considered the ^{60}Co flexisource used in this study composed of two gamma energies: 1.173 MeV and 1.332 MeV as well as the radiation spectrum of microSelectron of ^{192}Ir source obtained from the National Nuclear Data Center database [20]. The β spectrums of both of simulated sources were neglected because it did not contribute to dose-rate distribution due to encapsulation within the stainless steel [8, 21]. Dosimetric characterization of brachytherapy sources has been one of the medical physics applications of the code [5, 16, 22]. TG-43 formalism calculated the dose rate distributions around brachytherapy sources according to the following equation [6, 9, 18]:

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

where S_k is air-kerma strength of brachytherapy source, which can be specified in terms of air-kerma rate, at a point along the transverse axis of the source in free space [6, 9]. It is defined as the product of air-kerma rate (K_{air} (d)) at a calibration distance, d in free space, measured along the transverse bisector of the source and the square of the distance (d) [6, 23].

$$S_k = \dot{K}_{air}(d, \theta_o)d^2\ [\mu Gym^2h^{-1}] \tag{2}$$

where θ_o = 90°; the air-kerma rate ($\dot{K}_{air}$), in Gy/s of source activity (A) in Becquerel and the number of photons per decay (Np) was determined from:

$$(\dot{K}_{air}) = K_{air}\ \mathrm{Np}\ [\mathrm{Gy/s}] \tag{3}$$

The air-kerma strength per unit source activity was then calculated as:

$$\frac{S_k}{A} = \frac{\dot{K}_{air}(d, \theta_o)d^2}{A} = K_{air}\ \mathrm{Np}d^2[Gym^2s^{-1}Bq^{-1}] \tag{4}$$

Or

$$\frac{S_k}{A} = \frac{\dot{K}_{air}(d, \theta_o)d^2}{A} = 3.6 \times 10^9(K_{air}\ \mathrm{Np}d^2)[\mathrm{U}Bq^{-1}] \tag{5}$$

where U = $\mu Gym^2\ h^{-1} = cGycm^2\ h^{-1}$.

The dose-rate constant in water (Λ) is the ratio of the dose rate at the reference distance (r_o = 1 cm, θ_o = 90°) on the transverse axis per a unit air-kerma strength, and its unit is $cGyU^{-1}h^{-1}$.

$$\Lambda = \frac{\dot{D}(r_o = 1\ \mathrm{cm}, \theta_o = 90°)}{S_k} \tag{6}$$

The dose-rate constant depends on both the radionuclide and source model, and is influenced by both the source internal design and the experimental methodology used by the primary standard to realize S_K.

$G(r, \theta)$ is the geometry function, which accounts for the effect of the distribution of radioactive material inside the source on the dose distribution at a given point according to:

$$G(r, \theta) = \begin{cases} \dfrac{\Delta\beta}{L\ rsin\theta} = \dfrac{tan^{-1}\left(\dfrac{rcos\theta + L/2}{rsin\theta}\right) - tan^{-1}\left(\dfrac{rcos\theta + L/2}{rsin\theta}\right)}{L\ rsin\theta} & \text{if } \theta \,\#\, 0 \\ (r^2 - L^2/4)^{-1} & \text{if } \theta = 0 \end{cases} \tag{7}$$

where L is the active length of the source, β is the angle subtended by the active source with respect to the point (r, θ), and $G(r_0, \theta_0)$ is the geometric function at reference point (r_0 = 1 cm and θ_0 = 90°).

The radial dose function $g(r)$ accounts for dose fall-off on the transverse plane due to photon scattering and attenuation:

$$g(r) = \frac{\dot{D}(r, \theta_o = 90°)\ G(r_o = 1\ \mathrm{cm}, \theta_o = 90°)}{\dot{D}(r_o = 1\ \mathrm{cm}, \theta_o = 90°)\ G(r, \theta)} \tag{8}$$

The anisotropy function $F(r, \theta)$ represents the variation of the dose distribution around a brachytherapy source due to the distribution of radioactivity within the source, self-absorption, and oblique filtration of the radiation in the capsule material, which defines as:

$$F(r, \theta) = \frac{\dot{D}(r, \theta)\, G(r_o = 1 \text{ cm}, \theta_o = 90°)}{\dot{D}(r, \theta_o = 90°)\, G(r, \theta)} \quad (9)$$

From different parameters of TG-43 equations S_k, Λ, and $G(r, \theta)$ refer to the source specification and its geometrical shape according to the manufacture design.

To estimate the dose-rate constant, the radial dose function $g(r)$ and the anisotropy function $F(r, \theta)$ of the sources were located in the center of a 2 × 2 × 2 m^3 cube of a phantom. The dose-rate constant in water (Λ) was the ratio of the dose-rate at the reference distance on the transverse axis per unit of air-kerma strength. The air-kerma strength was calculated along the transverse axis of the source in the air-filled cubic phantom using mesh scoring with a voxel size of 2 × 2 × 2 mm^3. The air composition recommended in table XIV of the TG-43 update [9] for air was applied, with a relative humidity of 40%. Deposited doses in air and water were calculated around the source using the PrimitiveScore class and mesh scoring method. PrimitiveScore is an abstract base class representing a detector for dose scoring in Geant4. The dimension of each voxel in mesh scoring was set to be 2 × 2 × 2 mm^3. The deposited doses in air and water in these voxels were then converted to the TG-43 calculated parameters, such as Λ, $g(r)$ and $F(r, \theta)$ using equations (6), (8), and (9), respectively.

Uncertainty analysis

In this study, uncertainties associated with MC process were considered according to the AAPM task group report No. 138 and GEC-ESTRO recommendations [18, 24]. The type A (∇_Λ) uncertainty for MC methods was due to the numbers of events, where Poisson statistics applies; this uncertainty decreases by the inverse square root of the number of particles. The number of photons simulated was set to 10^9 to achieve an uncertainty of less than 0.1%. Type B uncertainties arise from uncertainties in the source dimensions, internal component location, volume averaging, and material composition. This affects the MC simulation, more specifically, the dose-rate constant, radial dose, and anisotropic functions for a specific simulated brachytherapy source. Uncertainties associated with the dose-rate constant ∇_Λ, radial dose $\nabla_{g(r)}$, and anisotropy functions $\nabla_{F(r,\theta)}$ for both ^{60}Co and ^{192}Ir sources were considered according to a comparison of our results from the MC simulation with values of the AAPM task group report No. 138 and GEC-ESTRO reports [24]. Finally, the total dose uncertainty was calculated through the following equation:

$$\nabla_{\text{Total}} = \sqrt{\nabla^2_\Lambda + \nabla^2_\Lambda + \nabla^2_{g(r)} + \nabla^2_{F(r,\theta)}} \quad (10)$$

where $\sqrt{\nabla^2_\Lambda + \nabla^2_{g(r)} + \nabla^2_{F(r,\theta)}}$ is the error propagation for type B uncertainties.

Results

The results of calculations of the-dose rate constant for HDR microSelectron ^{192}Ir and HDR flexi ^{60}Co sources obtained from the simulations by Geant4 code in the 2 × 2 × 2 m^3 cube of water phantom with a voxel size of 2 × 2 × 2 mm^3 are presented in Table 1. The dose-rate constant was used to specify the radioactive sources and depended on both the radionuclide and source model. The percentage dose-rate constant difference for ^{60}Co and ^{192}Ir between the MC result and the published report was 0.2% and 1.1%, respectively.

Figure 2A and B shows a comparison of the radial dose function between our MC results and published data for ^{192}Ir and ^{60}Co, respectively [26, 28], with percentage differences of 1.1% for ^{192}Ir and 1% for ^{60}Co (Table 2). Figure 2C and D also demonstrates a comparison between the anisotropy dose function at 1 cm radial distance between our MC results and published data for ^{192}Ir and ^{60}Co, respectively [26, 28], with percentage differences of 1.5% for ^{192}Ir and 1.3% for ^{60}Co. In addition, the type A and B uncertainties for both HDR microSelectron ^{192}Ir and HDR flexi ^{60}Co sources obtained from a comparison between the MC calculated values are indicated in Table 2. The total uncertainty of MC simulation for the HDR flexi ^{60}Co source was 1.7% and that of HDR microSelectron ^{192}Ir was 2.2%. The error intervals in each case were in accordance with the AAPM TG No. 138 [24].

Figure 3A compares the radial dose functions for the two sources from 0.5 cm to 60 cm radial distances. Radial dose function associated with ^{192}Ir was higher than that of ^{60}Co at radial distances from 3 to 20 cm, with the maximum percentage difference of 8.1% observed at 5 cm distance, whereas it was lower at larger distances from the source. The radial dose function of HDR flexi ^{60}Co source was linearly fall-off from 1 cm to 20 cm of radial distances, whereas for ^{192}Ir it was from 5 cm to 20 cm.

Compared with the values of HDR microSelectron ^{192}Ir source, anisotropic dose distributions of HDR flexi ^{60}Co source were significantly higher at the longitudinal sides of the source. The anisotropy functions for the two sources at radial distances of 1 cm, 2 cm, and 4 cm are compared in Figure 3B-D.

The primary photons from the HDR flexi ^{60}Co source had a greater range, and were attenuated at a lower level when considering the results of radial and anisotropic dose distribution functions, whereas the primary photons from the lower energy HDR microSelectron ^{192}Ir source had a smaller range and were partially attenuated. This means that compared with a ^{192}Ir source, ^{60}Co radionuclide could be utilized to treat malignancies farther away from the source, even though the ^{60}Co radionuclide source has a larger exit dose than iridium.

Table 1. Dose-rate constant values ($cGyh^{-1}U^{-1}$) for HDR microSelectron ^{192}Ir and HDR flexi ^{60}Co source in a water phantom

	Present study	Other studies	Diff. (%)
^{60}Co	1.089	1.087 [25, 26]	0.2
^{192}Ir	1.097	1.108 [27]	1.1

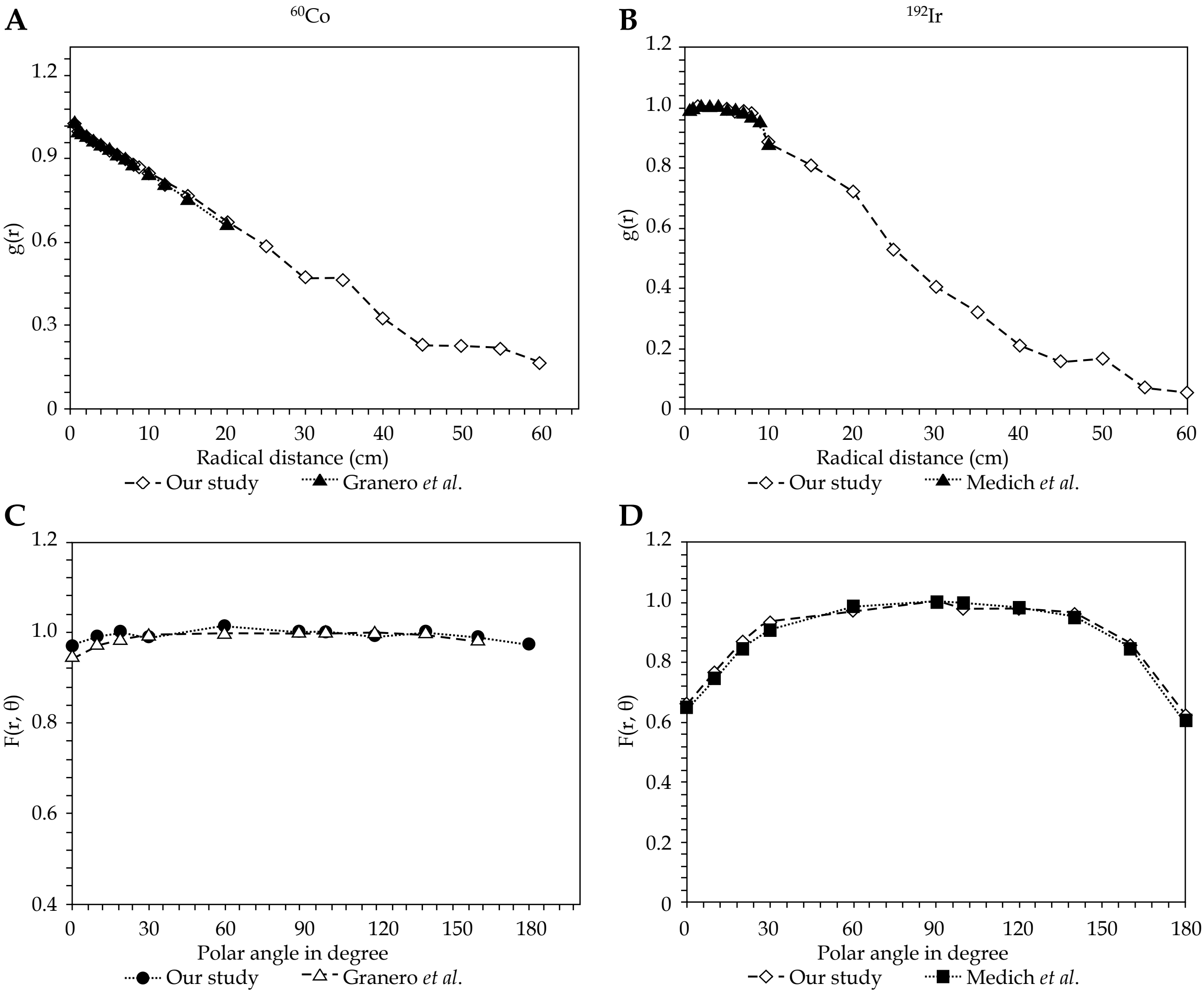

Fig. 2. A comparison between our MC results and published data at 1 cm depth. **A**) Radial dose function of HDR flexi ^{60}Co, **B**) radial dose function of HDR microSelectron ^{192}Ir, **C**) 2D anisotropy function of HDR flexi ^{60}Co, and **D**) 2D anisotropy function of HDR microSelectron ^{192}Ir

Discussion

In this study, the dosimetric comparison between the microSelectron HDR ^{192}Ir and the flexi ^{60}Co sources was evaluated in a liquid water phantom, and the calculations were performed with Geant4 Monte Carlo simulations. As a result, the dose-rate constants were verified based on MC simulation of the BEBIG ^{60}Co HDR source [25, 26] for the flexi ^{60}Co HDR source and flexisource HDR ^{192}Ir source [27] for the microSelectron ^{192}Ir source. The percentage difference between the dose-rate constant value from this study and the published data was 0.2% and 1.1% respectively, as indicated in Table 1. Papagiannis *et al.* [6] deduced the dose-rate constants for point source of ^{60}Co HDR, and reported an uncertainty value of 0.2%.

Table 2. Percentage dose difference obtained from a comparison between MC calculated values and published data [26, 28] as well as uncertainties of ^{192}Ir and ^{60}Co sources determined from MC simulation

Dosimetric quantities		^{192}Ir	^{60}Co
% dose differences	∇_{Λ}	1.1%	0.2%
	$\nabla_{g(r)}$	1.1%	1.0%
	$\nabla_{F(r,\theta)}$	1.5%	1.3%
Uncertainty			
Type A	Due to the number of simulated events	0.1%	0.1%
Type B	Due to source dimension, internal component location, volume averaging, and material composition	2.2%	1.7%
∇_{Total}		2.2%	1.7%

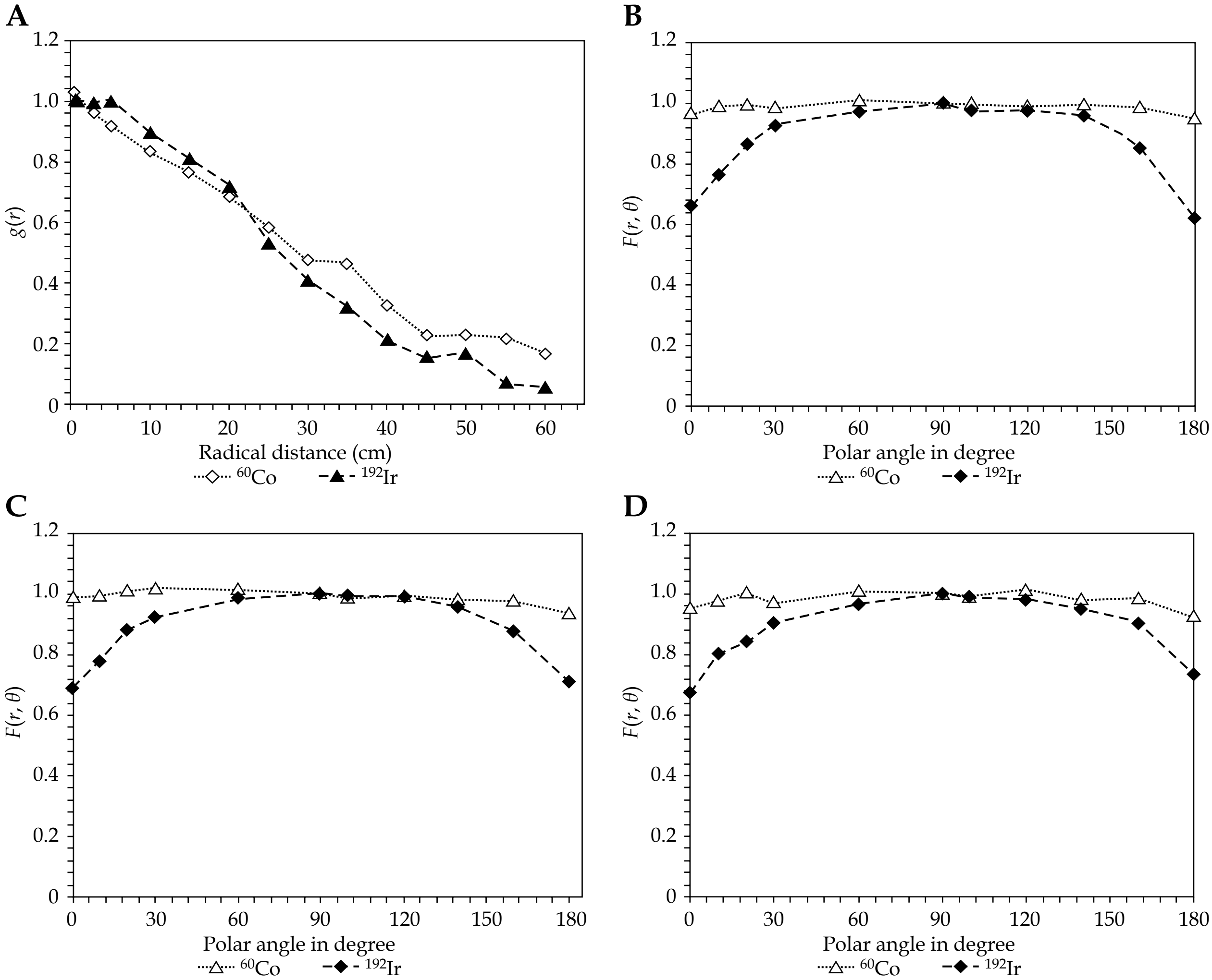

Fig. 3. A comparison between MC results of HDR flexi ^{60}Co and microSelectron ^{192}Ir. **A**) Radial dose function, **B**) 2D anisotropy function at 1 cm, **C**) 2D anisotropy function at 3 cm, and **D**) 2D anisotropy function at 4 cm

In another study, Ranjbar *et al.* [18] have shown also approximated 0.2% percentage difference between the calculated and published date. Zabihzadeh and Arefian [2] in their work on tumor dose enhancement by nano-particles during high-dose-rate ^{192}Ir brachytherapy reported the uncertainty value between the calculated and published values as 1.06%.

The dose fall-off on the transverse plane caused by photon attenuation and scattering in the water medium is accounted for by the radial dose function $g(r)$. Additionally, the anisotropy function $F(r, \theta)$ and the geometry factor $G(r, \theta)$ have an impact. The geometry component is influenced by the source's physical characteristics, such as its length and radius. The sources' construction must be identical to ensure the same geometry factors. In clinical dose distribution, the isodose curve is influenced by the anisotropy factor $F(r, \theta)$. These two functions are necessary for comparing various brachytherapy sources [6]. Results of the radial dose function and the anisotropy dose functions for both sources were verified at 1 cm depth using BEBIG ^{60}Co HDR [26] and HDR ^{192}Ir [28] sources, respectively. As demonstrated in Table 2, the percentage dose difference of each function with the corresponding published data was below 2%, in an acceptable range according to the recommendation of AAPM TG43-U1 [9] and TG 138 [24].

Figure 3A compares the radial dose functions, and at 5 cm depth, the microSelectron ^{192}Ir source recorded 8.1% higher dose than the flexi ^{60}Co source. The values of the radial dose function of ^{60}Co are lower than those of ^{192}Ir within the range of 3 cm to 20 cm radial distances and linearly fall-off from 1 cm to 20 cm distances, but are higher from 0.5 cm to 1 cm and above 20 cm. From 5 cm to 20 cm, the value of the ^{192}Ir radial decline linearly. Employing the EGSnrc Monte Carlo transport algorithm, Islam *et al.* [29] evaluated the radial dose distribution for the BEBIG HDR ^{60}Co and microSelectron HDR ^{192}Ir v. 2 sources. They found that the ^{60}Co source had greater values in the region of 0.18 cm to 1 cm and above 22 cm of radial distance compared with the ^{192}Ir source.

In comparison with microSelectron ^{192}Ir source HDR values, the anisotropic dose distributions of the flexi ^{60}Co HDR source are significantly higher at the longitudinal sides of the source. The primary difference may result from the sources' varied geometric constructions, source types, choice of angle interval, and average energy differences [6, 16, 30].

Conclusions

Both ^{60}Co and ^{192}Ir isotopes have a long history in the field of brachytherapy. Several hundred HDR afterloading units equipped with ^{60}Co or ^{192}Ir sources have been put into use in the past. The average energy of a brachytherapy source determines the penetrability of photon particles emitted from the source. High energy sources allow a higher dose to target at larger distances to the sources. Therefore, the primary photons from the lower-energy HDR microSelectron ^{192}Ir source have a limited range, and are partially attenuated when considering the results of radial and anisotropic dose distribution functions. This implies that a HDR flexi ^{60}Co radionuclide could be used to treat tumors farther from the source than a ^{192}Ir source despite the fact that HDR microSelectron ^{192}Ir has a lower exit dose than HDR flexi ^{60}Co radionuclide source. Hence, the treatment planning system should adjust the isodose distribution using dwell time positioning technology in a clinically relevant manner.

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Conflict of interest

The authors report no conflict of interest.

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