Dosimetric Characterization of an Intensity-modulated X-Ray Brachytherapy System

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Abstract

Purpose: An intensity-modulated X-ray brachytherapy system is being developed for various clinical applications. This new system makes it possible for clinical staff to control energy as well as dose rate for different tumor sites according to their sizes and radiobiological characteristics.

Materials and Methods: This system is mainly composed of an X-ray tube, guide tube collimation, and secondary (pseudo) target. Due to its configuration, convenient modulations of fluorescent X-ray energy and intensity are possible. To observe applicability of this novel system for various primary and secondary target combinations, Monte Carlo simulation using MCNP5 was performed, and air measurements were done. As a primary and pseudo-target combination, silver–molybdenum (Ag-Mo), tungsten–neodymium (W-Nd), and tungsten–erbium (W-Er) were used for the calculation of dose profile. Specifically, a dose distribution was calculated around each of these target combinations. Dose distributions as a function of target angles were also calculated. The Ag-Mo combination was analyzed for Cartesian coordinates of xy, xz, and yz planes of the pseudo-target to observe dose distribution as a function of the angle of secondary target. Results: The results showed that radial dose fall-off of Ag-Mo was greater than commercially available brachytherapy sources (103Pd and 125I) due to its low characteristic X-ray energy. Conclusions: Dose distribution variance should be considered in beam modulation for clinical application. Dynamic movement of the pseudo-target is feasible and remains as a subject for future research.

Keywords: Brachytherapy, dosimetry, intensity modulated, X-ray

INTRODUCTION

Radiation therapy is one of the important treatment methods of killing cancer cells. It can be divided into two main methods: external beam therapy (EBT) and brachytherapy. These two methods rapidly developed after their introduction. The EBT, however, has developed far more than the latter one because of the intensity-modulated radiation therapy technique. Adaptation of intensity modulation into brachytherapy should be a significant improvement of treatment modality. The new brachytherapy system used in this report, a so-called needle-based system, controls energy, intensity, and beam shape by changing target needle and position along guide tube. This system is composed of X-ray tube with collimating optics, a guide tube to an insertable pseudo-target needle. As shown in Figure 1, a conventional X-ray tube generates the primary X-ray beam which is guided to the secondary target. Here, a secondary X-ray beam consisting mainly of the characteristic lines of the secondary (pseudo) target is produced for a treatment. This two-stage X-ray production eliminates any complex designing for insertion of a small X-ray-generating device. This configuration overcome the features of the past devices. One limitation of the past devices was low power that yielded a very low dose rate because X-ray source had to be small enough to be inserted into the body. A second limit was that it was difficult and dangerous to insert a high-vacuum, high-voltage (up to 90 kVp) device (miniature X-ray tube) into the human body. In the new system, since the primary beam is generated out of the patient body, medical staff can increase primary X-ray...
power without any limitation, and energy to the tumor can be modulated by changing pseudo-target. The efficiency of the reemitted X-ray power (X-ray fluorescence) from the secondary target is determined by a quantum yield of the X-ray fluorescence radiation. Specifically, the quantum yield of X-ray fluorescence radiation for Mo (Z = 42) which was used in this study is determined mainly by K-shell excitation (K-fluorescence of Mo) and equal to about 74%.[12] From the various combinations of primary X-ray and pseudo-targets, one can find the appropriate beam for a specific clinical application. In this research, the dose profiles of the recently developed image-guided intensity-modulated X-ray brachytherapy (IMXBT) as a function of primary and pseudo-target combination and wedge angle of the pseudo-target were simulated using Monte Carlo calculation.

**MATERIALS AND METHODS**

**Experimental setup and scintillation (NaI (Tl)) detector measurement in air**

In the experiment, relative fluorescent X-ray beam intensity measurement in air was performed using a scintillation counter that was attached to a goniometer. Rotation of detector in the horizontal plane and gradual rotation of pseudo-target angle perpendicularly to the detector moving plane made it possible to reconstruct three-dimensional dose distributions.

Since the operation voltage of X-ray tube used in the experiment was <90 kVp, plausible combinations of primary and pseudo-target were limited. It is also obvious that Kα X-ray lines of pseudo-target should be less than that of primary X-ray energy to excite electrons in K-shell. Therefore, the three target combinations studied in this study were selected because they produced maximum efficiency (X-ray quantum yield).

In this study, several combinations of primary and pseudo-target were simulated to observe feasibility. First, primary targets were Ag and W. For the MCNP calculation and the measurement, only an Ag primary target was used. Incident fluence on the secondary target material is different from the Ag primary as compared to the W primary target. Delivered power to the primary target of the system was determined by the following estimation:

\[
P_{\text{target}} = P \times \text{Efficiency} \times \frac{\pi r^2}{(4 \pi R^2)}
\]

where, \( P \): X-ray tube power,
\( r \): Pinhole radius of the optical collimator,
\( R \): Distance from the focus of the X-ray tube to the pinhole,

Efficiency: \( 9 \times 10^{-10} Z Y \)[13]

From this, obtained fluence incidents on the secondary target were estimated to be \( 2.27 \times 10^{10} \) photons/s and \( 8.46 \times 10^6 \) photons/s for Ag and W primary target, respectively.

In this experiment, Ag primary target and Mo pseudo-target combination were used for the measurement in air. As shown in Figure 1, a commercial 1.5 kW SEIFERT X-ray tube with Ag anode was used as the primary X-ray source. The tube has a fine linear focus of 8 mm \( \times \) 0.4 mm. By arranging a take-off angle of about 6°, a point source with a projected focal spot of 0.8 \( \times \) 0.4 mm was extracted from the tube.

The collimator was tightly pressed to the output window of the X-ray tube. The optical scheme used allows forming a quasi-parallel X-ray beam with a different divergence angles. In our experiments, an optical scheme was used with a divergence angle about 0.5° along the beam axis. The collimators also have a turning flange connected to the needle via a Morse cone for precise mounting of the needle and rotating it around the beam axis.

The needle X-ray device is connected to the X-ray tube through an optical collimator. The hollow needle made of aluminum alloy (99% Al) has a length of 80 mm and an external diameter of 2.2 mm which was the same diameter used in the MCNP5 calculations. A molybdenum (Mo) target was installed inside the needle a few millimeters from its far end. The target is movable along the needle axis and can be rotated around the needle axis for beam intensity adjustment. The used target angles ranged from 10° to 45° and shaped as wedge to reflect primary X-rays.

The measurement arrangement includes a Huber goniometer with a detector connected with electronic systems and computer placed in the separate room. A computer-controlled two-circle Huber goniometer was positioned next to the needle X-ray device. The center of the installed inside the needle target was set in the center of rotation of the goniometer using two computer-controlled translation tables. The goniometer had a moving arm with an attached single-channel NaI (Tl) scintillation detector (Radican LTD, St. Petersburg, Russia). The arm could move the detector in the horizontal plane around the center of the goniometer. The detector had a pulse-height discriminator controlled by two threshold voltages that allows to measure not only the X-ray intensity, but also to evaluate the energy distribution of the emitted by the target radiation. The digitized count pulses are fed to a controller module connected to a computer.

Figure 1: Pilot device of the intensity-modulated X-ray brachytherapy system
Monte Carlo calculation
Monte Carlo calculation is widely used in medical physics for accelerator calibration and characterization, detector simulation, brachytherapy source simulation, and development of new treatment. Monte Carlo simulation is a practical method for designing and choosing of target material and its geometry for this novel brachytherapy system. In this study, dose distribution by the IMXBT was simulated by Monte Carlo code. The IMXBT needs baseline data of dose distribution around the source in both air- and water-equivalent phantoms. It also needs various ranges of X-ray energies as well as various shapes and materials of the needle’s pseudo-target. To accomplish the calculation of this purpose, as mentioned earlier, Monte Carlo codes such as MCNP5 or EGSnrc codes are widely used. Besides, before its application to real treatment, it is necessary to predict and establish dose distribution around the single brachytherapy source. Eventually, it is also necessary to incorporate calculation values into treatment planning system. To do this, MCNP version 5 Monte Carlo code developed by Los Alamos National Laboratory was used. This Monte Carlo calculation package could realistically simulate particle behavior of electron, photon, and neutron. Furthermore, this is a new MCNP version that includes Doppler energy broadening of low-energy photons that was neglected until in the previous version (4C). This is a precollision motion of the electron in incoherent scattering. In addition, this new version of MCNP5 uses photo-atomic data derived from ENDF/B-VI.8 data library that are derived from EPDL97. This new data was included in MCPLIB04 that was released in 2002. Therefore, this study expected to obtain more accurate result than using previous version 4C. MCNP5 is capable of simulate photon transport accurately where fluorescent X-rays such as K_α′, K_α, K_β, and K_γ are dominant. The energy of the photon is transported down to 1 keV. In this study, detector volume was varied as a function of distance, such as smaller volumes for short distance and larger volume for further distance from center of the source. In spite of the fact that several X-ray lines are generated by a primary target, only K_α′ energy from the primary target was included in the MCNP input. This simplified assumption of that monoenergetic emission of primary target was not significantly affected the result between measurement and calculation which was within statistical error range in the calculation and positioning error in TLD measurement. Calculations were performed from 50 million to 2 billion particle histories depending on the situation. Unlike other radioisotope brachytherapy sources, the IMXBT emits X-ray asymmetrically according to pseudo-target wedge. Therefore, smaller dose collecting volumes should be used. However, the use of these smaller volumes significantly lengthens the calculation time for *F8 tally to obtain reasonable error range. All the secondary particles included electrons ([mode p e] in the MCNP input) and were simulated using energy deposition estimator (*F8 tally) for absolute dose calculation. In the case of calculating radial dose fall-off and for other relative comparison purpose, *F4 tally (track length fluence estimator) was used. This tally method was much faster than *F8 tally for the same number of histories and the MCNP embedded statistics were better. However, using this tally limits transport of secondary electrons produced by photons (only [mode p] in the MCNP input). Geometry used in the calculation is briefly shown in Figure 2. As shown in Figure 2, the overall geometry was focused on the center of the source defined by the center of pseudo-target’s wedge and peripheral areas where detector volume to be placed. Experimental specifications in air measurement are as follows: external diameter of the needle, internal diameter (identical to diameter of pseudo-target), and diameter of the primary beam incident on the pseudo-target were 2.2 mm, 2 mm, and 1.6 mm, respectively. For a comparison purpose, radial dose functions of ^10^3Pd and ^125^I brachytherapy sources were compared. The following are TG 43 in American Association of Physicists in Medicine (AAPM) guidelines for radial dose function:

\[
g(r) = \frac{D(r, \theta_0) G(r_0, \theta_0)}{D(r_0, \theta_0) G(r, \theta_0)}
\]

where,

\[
G(r, \theta) = \frac{\beta}{L_{\text{rsin}} \theta}
\]

For point source approximation

\[
G(r, \theta) = \frac{1}{r^2}
\]

For line source approximation

In this study, point source approximation was used because the length of the source is different from definition of AAPM TG 43 for conventional radioisotope sources.

Pseudo-target angles used in the MCNP calculation were 10°, 30°, 45°, 60°, and 80° for each primary and pseudo-target combination. Since dose distributions from the secondary X-rays are anisotropic in the static status of the needle due to its unique feature and center point definition, it is necessary to observe dose distribution of three Cartesian planes of two-dimensional data. These results will meet clinical application requirements. For Ag primary and Mo pseudo-target (Ag-Mo) combination, calculations along xy, xz, and yz plane of the needle were performed to observe anisotropy characteristics as a function of wedge angle of pseudo-target.

![Image of MCNP calculation set up](https://example.com/image.png)

**Figure 2:** Cross-sectional view of the MCNP calculation set up for the intensity-modulated X-ray brachytherapy system (not to scale)
RESULTS
Dose fall-off for different primary and pseudo-target combinations

The dose rate as a function of the pseudo-target’s angle as calculated by the MCNP for different combination is presented in Figure 3. This calculation performed under an assumption that 0.00008 W of the same X-ray power is incident on pseudo-target.

The Ag-Mo combination, for example, produced 286 cGy/h at 1 cm from center of the source. In the experiment, it was possible to increase power up to 0.00016 W, and thus, twice the dose rate can be achieved.

As shown in Figure 3, lower the degree of the angle, the higher the dose rate observed along +y axis. This trend is also identical for all other primary and pseudo-target combinations such as tungsten-neodymium (W-Nd) and tungsten-erbium (W-Er). The results indicate that the choice of primary and pseudo-target and determination of angle of pseudo-target greatly affect the results on the dose distribution. If clinic staffs want to irradiate small tumor volumes, they may need to choose pseudo-target with smaller angle. On the contrary, they may need pseudo-target with large angle for larger tumor volumes.

Comparison between calculation and measurement in air

MCNP dose calculations in water as well as in air calculations were performed in order to compare calculated and measured values. Since most of the prominent fluorescent X-ray energy from Mo pseudo-target is only about 17.5 keV, different geometry of the detector between experimental and calculation setup contributed to the discrepancies of the two methods. The MCNP calculation and in air measurement with a scintillation counter showed reasonable agreement [Figure 4] for silver primary target and molybdenum secondary target combination. The dose rate was normalized to 1 at 1 cm from the source for comparison purpose. In spite of geometrical difference of the detector between calculation and experiment, the results agreed well within the calculation error range. This result confirms that the device is working properly.

Radial dose function

Since the dominant emitted fluorescent X-ray energy from pseudotargets is about 17.5 keV [Table 1], the radial dose fall-off was comparable to commercially available ^{103}\text{Pd}$ and $^{125}\text{I}$ brachytherapy sources. Comparison of Ag-Mo combination with $^{125}\text{I}$ and $^{103}\text{Pd}$ is shown in Table 2. As shown in the Table 2, dose rate fall-off was much significant than conventional interstitial brachytherapy sources due to its low X-ray energy for Ag-Mo combination. W-Nd and W-Er showed less dose radial function due to higher energy
than those of $^{125}\text{I}$ and $^{103}\text{Pd}$. At 0.5 cm from the center of the needle, the radial dose function was 2.387–2.516 depending on the tube material and thickness, while it was <0.2 at 3.0 cm. Most radioisotope sources are covered with 0.05 mm titanium, while the IMXBT which was used in the experiment is covered with 0.1 mm aluminum tube. This different casing material affected just a little portion of overall dose fall-off as shown in the result. Furthermore, to use the intensity modulation feature, reduction of case thickness would be difficult because other mechanical devices such as motor to control pseudo-target should be accommodated.

**Dose distribution along xy plane: Anisotropy**
Dose rate distribution along xy plane of the needle was calculated specifically for Ag-Mo combination. It was evident that the beam shape was not isotropic due to the angle of the pseudo-target. As shown in Figure 5a-e, as the angle of the pseudo-target increases, the 3D radiation dose decreases. For visual purpose, normalized dose rate was shown in Figure 5. As depicted in Figure 5, the primary X-ray beam is coming in from 180° to 0° to heat secondary target. To quantify this effect with the conventional radioisotope brachytherapy, anisotropy function was introduced which was suggested by AAPM TG 43.[13] Although it is quite different from the geometry used in TG 43 formalism which defines the finite length of the radioisotope source, anisotropy of this IMXBT was calculated according to point source approximation for comparison purpose. Each dose point was normalized to 1 at 90° where perpendicular to beam direction as shown in Figure 5. The anisotropy factors are plotted in Figure 6 for various wedge angles at 1 cm from center of the pseudo-target. As shown in Figure 6, anisotropy was significant up to 1.96 for 80° pseudo-target wedge because dose was normalized at 90° where insignificant dose distribution exists.

**Dose distribution along xz plane: Anisotropy**
The Figure 7 illustrates only half of the coordinate because it is assumed that the shape of dose distribution expected to be symmetric as in Figure 5. Unlike Figure 5, and Figure 7a-e plots absolute dose along the xz plane. In xy plane, increase of secondary target angle significantly affects the anisotropy. The highest anisotropy was observed for 80° up to 8.94 (270° along xz plane was normalized to unity). While the anisotropy was only less than 2.4 for 10° [see Figure 8].

**Dose distribution along yz plane: Anisotropy**
Dose distribution along yz plane of the needle was calculated which is perpendicular to xy plane. This also illustrates only half of the coordinate. As shown in Figure 9a-e, dose distribution was increased as wedge angle increased. This trend is vice versa of previous case for xy and xz plane. However, it was natural result because beam from the center of source emanates obliquely through the aluminum tube. In Figure 9, half side (180°–360°) of yz plane was shown. The dose rate was normalized to unity at 90° and anisotropy was plotted. The anisotropy was significantly increased between 120° and 180°. However, for 80°, anisotropy was more than 0.1 while others were <0.6 for polar angles between 200° and 260°.

**Table 1: Combination of primary and secondary (pseudo) target used in MCNP calculation**

| Primary target | $K_{\alpha}$ (keV) | Pseudo-target | $K_{\alpha}$ (keV) |
|----------------|-------------------|---------------|-------------------|
| Ag             | 22.16             | Mo            | 17.48             |
| W              | 69.53             | Nd            | 37.36             |
| Er             |                   |               | 49.13             |

Ag: Silver, W: Tungsten, Mo: Molybdenum, Nd: Neodymium, Er: Erbium

**Table 2: Comparison of radial dose functions**

| Distance (cm) | $^{103}\text{Pd}$ (Thera 200) | $^{125}\text{I}$ (6711) | $^{125}\text{I}$ (6702) | Ag-Mo (1 mm Al) | Ag-Mo (0.5 mm Al) | Ag-Mo (0.5 mm Ti) | W-Nd (0.5 mm Ti) |
|---------------|-------------------------------|------------------------|------------------------|-----------------|-------------------|-------------------|-----------------|
| 0.5           | 1.30                          | 1.071                  | 1.030                  | 2.487           | 2.516             | 2.387             | 1.005           |
| 1.0           | 1.00                          | 1.00                   | 1.00                   | 1.00            | 1.00              | 1.00              | 1.00            |
| 1.5           | 0.749                         | 0.908                  | 0.935                  | 0.570           | 0.570             | 0.597             | 0.987           |
| 2.0           | 0.555                         | 0.832                  | 0.851                  | 0.284           | 0.372             | 0.391             | 0.974           |
| 3.0           | 0.302                         | 0.632                  | 0.679                  | 0.171           | 0.166             | 0.196             | 0.869           |

Ag-Mo combination in this study was averaged for all secondary wedge angles (normalized to unity at 1 cm, $^{103}\text{Pd}$ and $^{125}\text{I}$ are from reference,[14] and are covered with 0.05 mm thick titanium capsule. This comparison based on the point source approximation for IMXBT). Ag-Mo: Silver–molybdenum, W-Nd: Tungsten–neodymium.
Characteristic X-rays with imaginary HPGe detector

Primary characteristic X-ray emitted from Mo is about 17.5 keV. However, there are other miscellaneous X-rays emitted from Mo. These photons were simulated using F8 spectrum tally embedded in the MCNP code. The needle was put 1 cm above the HPGe detector (3 cm diameter × 1 cm height) and the pseudo-target wedge faced toward the detector face. Wedge angle used in this simulation was 10°.

**Figure 5:** Normalized dose rate distribution of Ag-Mo combination for various pseudo-target angles along xy plane. Half of the polar angle is shown here because dose distribution is expected to be symmetric. (a) 10°, (b) 30°, (c) 45°, (d) 60°, and (e) 80°.

**Figure 6:** Anisotropy along xy plane for various pseudo-target angles (90° of polar angle around pseudo-target needle was normalized to unity)

**Discussion**

As stated in the result section, the dose rate of 2.9 Gy/h was readily achievable through the experimental device setup. This dose rate is quite enough for low dose rate treatment, but still relatively low for high dose rate (HDR) (application which requires more than 12 Gy/h). Since secondary X-ray beam cannot be delivered without primary beam supply, nor be delivered through winded pathway, only HDR applications are a feasible option at this point. However, HDR can be obtained by increasing power of the X-ray system by using a high-capacity X-ray tube or even a linac. If a linac was the primary source of X-ray target, cost problem may occur. For clinical application, primary beam generation by X-ray tube would be a viable option. The better conformity of tumor is possible by this needle-based system according to pseudo-target geometry. For example, such as cervix cancer application in HDR using CT, when needle location is very close to organ at risk (OAR, e.g., rectum), beam can irradiate only toward the tumor site not affecting the OAR by simply changing the pseudo-target into with larger angle such as 80° wedge. In conventional brachytherapy, it was a very challenging task because the source irradiate isotropically.
Figure 8: Anisotropy along xz plane for various pseudo-target angles (270° of polar angle around pseudo-target needle was normalized to unity)

Therefore, this system can be used for better conformity in the cases where neighboring OARs be concerned.

Conclusion

A novel intensity-modulated X-ray brachytherapy system was devised and tested successfully with conventional low energy X-ray tube. Specifically, feasibility of this system was tested with Monte Carlo calculation along with experiment. The results showed good agreement for both calculation and measurement in air in this study and in water (TLD measurement) in the previous study. This system overcomes some drawbacks that previous X-ray miniature brachytherapy system had. This new brachytherapy system could be applied to treat localized cancers such as breast, prostate, and brain and other tumors by simply changing input power, combination of X-ray targets and changing of angle of the pseudo-target. The treatment can be performed with single or multiple target needles with with lesser number of needles as compared to conventional brachytherapy using radioisotopes. Furthermore, quality of brachytherapy would be enhanced and expense of treatment could be saved because the treatment was carried out by just one system for various tumor treatments without replacing radioisotope sources. With this obtained result and feasibility of this device, more complex application such as beam modulation by changing the needle position using motor-driven device would be possible and remains for future research.

Acknowledgement

Authors appreciate late Dr. George Gutman who initiated this research idea.
Financial support and sponsorship
Nil.

Conflicts of interest
There are no conflicts of interest.

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