

Preliminary Monte Carlo Investigation of Using Ir-192 as the Source for Real Time Imaging Purpose

Chengyu Shi¹, Brian Wang²

¹Department of Medical Physics, Memorial Sloan Kettering Cancer Center, New York, NY, USA

²Department of Radiation Oncology, James Brown Cancer Center, The University of Louisville, Louisville, KY, USA

Email: shic@mskcc.org, brian.wang@louisville.edu

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Abstract

The purpose of this study is to investigate the potential use of Ir-192 as the source for real time imaging during HDR (High Dose Rate) brachytherapy treatment. Phantom measurement was performed to determine outside of the body dose. Monte Carlo code, EGSnrcMP egs_inprz, was used for the simulation to calculate the outside of the body x-ray signal for CT reconstruction. Matlab code was developed to reconstruct the Ir-192 source and for 3D visualization in order to assess reconstructed CT resolution, signal-to-noise ratio, and imaging dose information. The measured dose was 0.67 ± 0.04 cGy, which was comparable to the Monte Carlo simulation result 0.71 ± 0.20 cGy. The reconstructed source diameter dimension was 1.3 mm compared with 1.1 mm for the real source dimension. The signal-to-noise ratio was 19.91 db following de-noising. Source position was within a 1 mm difference between programmed and simulated results. Although the Ir-192 signal is weak for CT imaging, it is possible to use it as a CT imaging x-ray source for HDR treatment localization, verification and dosimetry purposes. Further study is needed for the detailed design of an outside of the body CT-like device for use in brachytherapy imaging.

Keywords

Monte Carlo, Ir-192, HDR, Imaging

1. Introduction

With the goal of sparing organs at risk (OARs), new radiation therapy techniques have been developed in the past two decades to treat the target while avoiding healthy tissues. Intensity Modulated Radiation Therapy (IMRT) tech-

nique has been widely applied to external beam radiation therapy. In order to ensure precision and accuracy of treatment, IMRT requires combined use with Imaging Guided Radiation Therapy (IGRT). However, in brachytherapy, the implementation of IMRT is delayed due to treatment space and physical limitations. Several studies have tried to apply IMRT concept to brachytherapy. In 2002, Ebert investigated the use of Intensity-Modulated Brachytherapy (IMBT) with radially asymmetric internally applied radiation sources [1]. In 2008, Hiatt *et al.* introduced the concept of Intensity Modulated Electronic Brachytherapy (IMEB) [2], and then in 2009 further explored depth dose modulation electronic brachytherapy to spare skin in intracavitary breast treatment [3]. In 2010, Shi *et al.* used dosimetry algorithm and inverse treatment planning to calculate three-dimensional (3D) IMBT dose distributions [4]. All these studies focused on the IMBT concept and dosimetry. Recently, Liu *et al.* developed a rotating-shield device in 2013 and later modified it for application in different brachytherapy treatments [5]-[10], further expanding IMBT's clinical capabilities. Similar to IMRT's dependence on IGRT for external beam treatment, IMBT also relies on imaging prior to treatment, during treatment, and even following treatment with verification. Brachytherapy treatment also requires imaging guidance or real-time imaging for localization and verification purposes. Image guidance is actually more important in brachytherapy than external treatment due to the large dose variation as a result of applicator position uncertainty. Several studies showed there is approximately a 5% change in dosimetry for the target and nearby critical structures per mm of applicator shift [11] [12]. In a recent review of HDR (High Dose Rate) safety, authors reported about 8 events per 33,000 treatments each year [13]. The causes of these events were mostly due to wrong length, distance, or applicator. Therefore, imaging verification during treatment is important for preventing adverse events.

Real-time monitoring of applicator position has been explored with use of infrared markers [14], but this technique is burdened by bulky equipment and potential marker shifts relative to the applicators. Other existing imaging modalities, such as ultrasound, MRI (Magnetic Resonance Imaging), and C-arm x-ray, CT (Computed Tomography) either need an extra imaging source, which is not convenient in brachytherapy treatment, or lack real-time imaging potential. There are many potential benefits if the Ir-192 source itself for HDR treatment can be used as the imaging source and if a special CT type detector for imaging can be designed. This could reduce unnecessary x-ray sources and exposure, be performed in real-time, and provide inherent source localization information. Safavi-Naeini *et al.* developed a BrachyView in-body imaging system for HDR prostate brachytherapy in 2013 and 2015 [15] [16] [17], which was capable of providing imaging, localization and dose validation. However, due to physical space limitations, the device may not be suitable for all types of HDR treatment.

In this study, we investigated the potential of using Ir-192 as the CT source for brachytherapy imaging purposes. Monte Carlo simulation was carried out to study the accuracy of localization with the Ir-192 source and to assess imaging

resolution, signal-to-noise ratio, imaging dose information.

2. Materials and Methods

2.1. Phantom and Treatment Planning

An ACR (American College of Radiology) CT accreditation adult abdomen phantom (diameter = 16 cm) was used for this study and is shown in **Figure 1(a)** and **Figure 1(b)**. The phantom was scanned with a GE (General Electric Company, Milwaukee, Wisconsin, USA) 4-slice LightSpeed CT scanner using a brachytherapy protocol (1.25 mm CT slice thickness, 50 cm field-of-view (FOV)). Images were then transferred to an Oncentra treatment planning system (Elekta Medical System Ltd., Stockholm, Sweden, version 4.3). A treatment plan was developed with a 3.0 cm diameter cylinder simulated. The prescription was 500 cGy per fraction at 5 mm above the cylinder surface. Thus, 2.0 cm away from the center in the radius direction would receive 500 cGy per fraction. The treatment plan was then transferred to the microSelectron HDR unit. A Fluke (Fluke Electronics Corporation, Everett, WA) Model 451 P pressurized ion chamber survey meter was placed with the detector sensitive region touching the phantom surface for dose rate measurement. The University of Wisconsin Accredited Dosimetry Calibration Laboratory calibrated the survey meter with $\pm 6.5\%$ uncertainty. **Figure 1** shows the CT scanning and treatment setup for the adult abdomen phantom. The purpose of this phantom study is to get the range of dose rate on the phantom surface and to calculate the calibration factor for the following Monte Carlo simulation to convert fluence to absolute dose per simulated particle.

2.2. Monte Carlo (MC) Simulation

The EGSnrcMP `egs_inprz` (version 1.0) program (National Research and Council Canada) was used for the Monte Carlo simulation. EGSnrcMP `egs_inprz` is a graphic user interface (GUI) for the NRC RZ user-codes

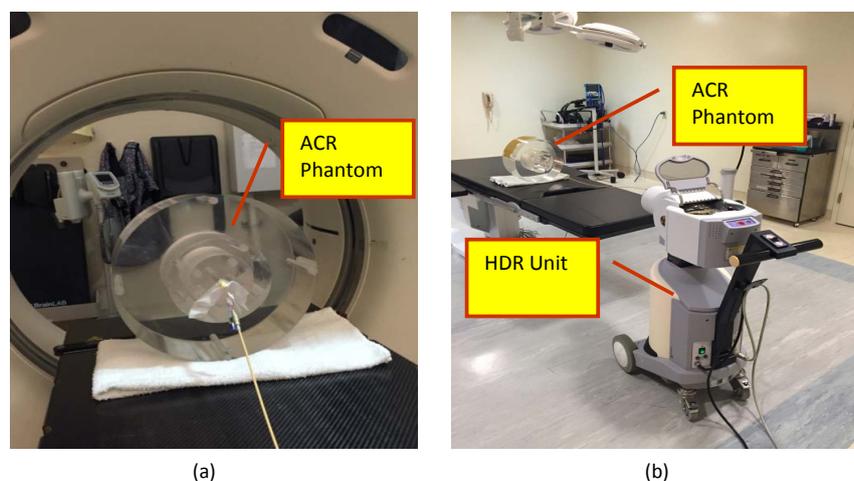


Figure 1. (a) CT scan of the adult abdomen phantom with source position simulator, (b) treatment setup for the phantom.

(<http://nrc-cnrc.github.io/EGSnrc/doc/pirs801-egsinprz.pdf>). DOSRZnrc user code was selected. The adult abdomen phantom was simulated with PMMA materials and the Ir-192 microSelectron spectrum was chosen for the simulation. Uniform isotropically radiating disk of finite size was used and the settings for source simulation were: RMINBM = 0, RBEAM = 0.06, ZSMIN = 5.0 and ZSMAX = 5.35. The Ir-192 Source dimension (diameter and active length) information followed the paper published by Mowlavi *et al.* in 2008 [18]. In total, 10^9 particles were simulated with ECUT = 0.521 MeV and PCUT = 0.01 MeV settings. **Figure 2(a)** depicts the geometry of the simulation. Longitudinal and radius profiles for the simulation were recorded. The 3 mm margin outside of the diameter = 16 cm was assumed to be detector region and used for recording the CT-like device detector signal (here is the deposited dose information) purpose.

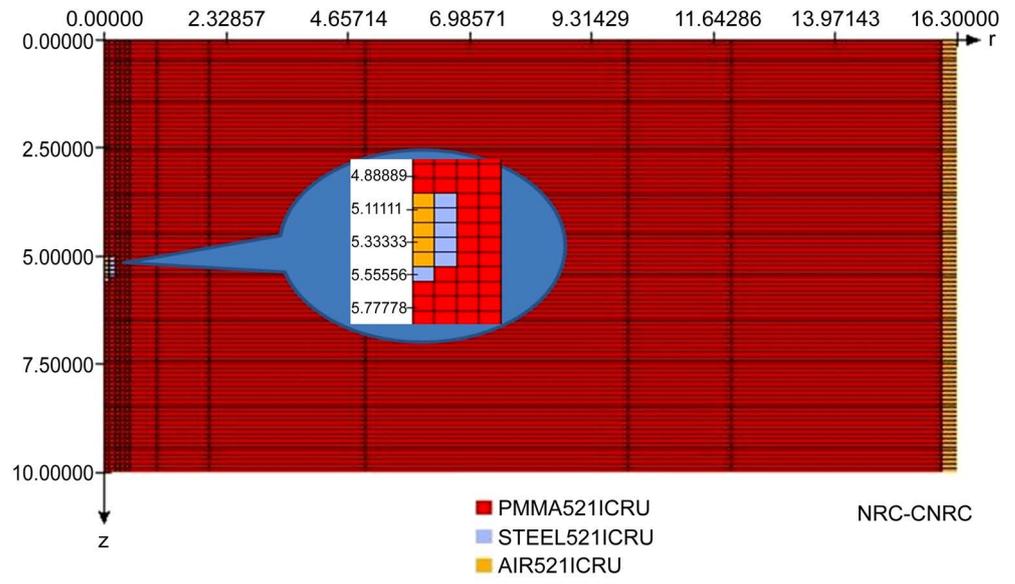
2.3. Post-Processing of the MC Results

In-house Matlab (R2010a, The MathWorks, Natick, MA, USA) code was written to post-process the simulated results. The inverse radon transform (“iradon” function in Matlab) was performed and the 2D CT image was generated. The inverse radon transform will reconstruct the image from the projection data in the 2D array. A de-noising technique with wavelet decomposition at level 5 and soft thresholding (“wavedec2” function in Matlab with N = 5 and wname = “db5”) was performed on original reconstructed 2D image to get better image quality. The wavedec2 function will return the wavelet decomposition of the image at certain level. With the same inverse radon transform, de-noised image showed better reconstructed result.

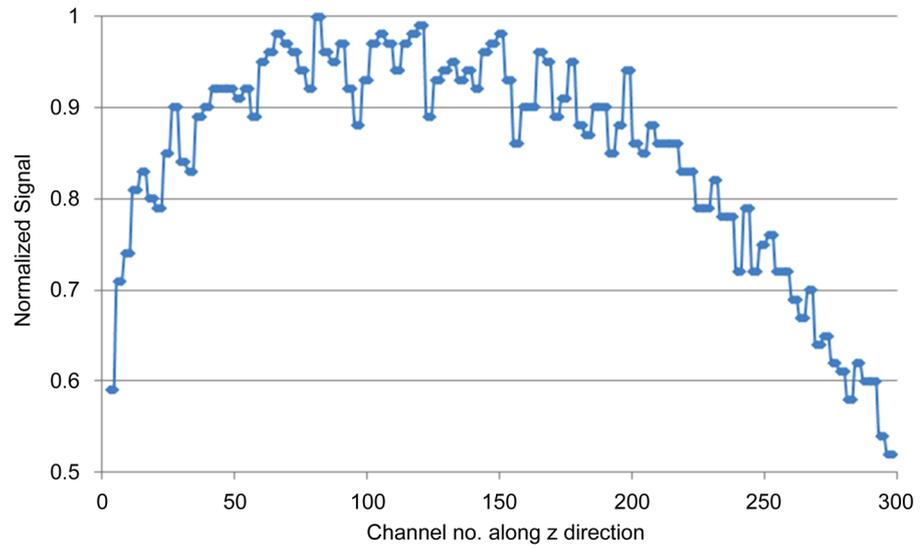
3. Results

3.1. Measurement vs. Monte Carlo Simulation for Absolute Dosimetry

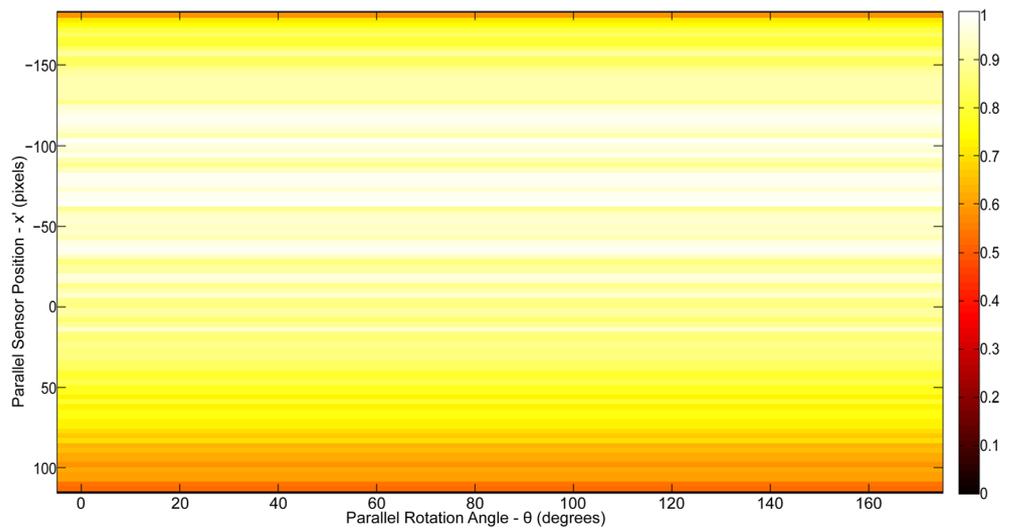
The measured dose rate using the 451 P survey meter was 8.3 R/h (the apparent source activity was 8.11 Ci at the measurement time), which was converted to air dose using the factor 0.877 cGy/R and a treatment time of 332.3 s. The measured dose was 0.67 cGy. If $\pm 6.5\%$ uncertainty was taken into account, the measured result was then 0.67 ± 0.04 cGy. The Monte Carlo simulation result was then calibrated at 2.0 cm away from the phantom center receiving 500 cGy. The calibration factor was 2×10^{13} particles/Gy. The dose was 0.71 ± 0.20 cGy at a radius distance of 16.3 cm, which was 5% higher than the measured result. Factors for the difference between measurements and simulations include the survey meter’s calibration uncertainty and the Monte Carlo simulation’s uncertainty of $\pm 28.0\%$ @ depth 16.3 cm due to the weak signal of the brachytherapy source. However, the measurement was still in the simulation range. Please note that the Monte Carlo uncertainty is independent of the source activity. The source activity was used for deriving the calibration factor for Monte Carlo simulation in order to get absolute dose information. If the source is weaker, the delivery time for the measurement will be longer, but the integral dose will be the same.



(a)



(b)



(c)

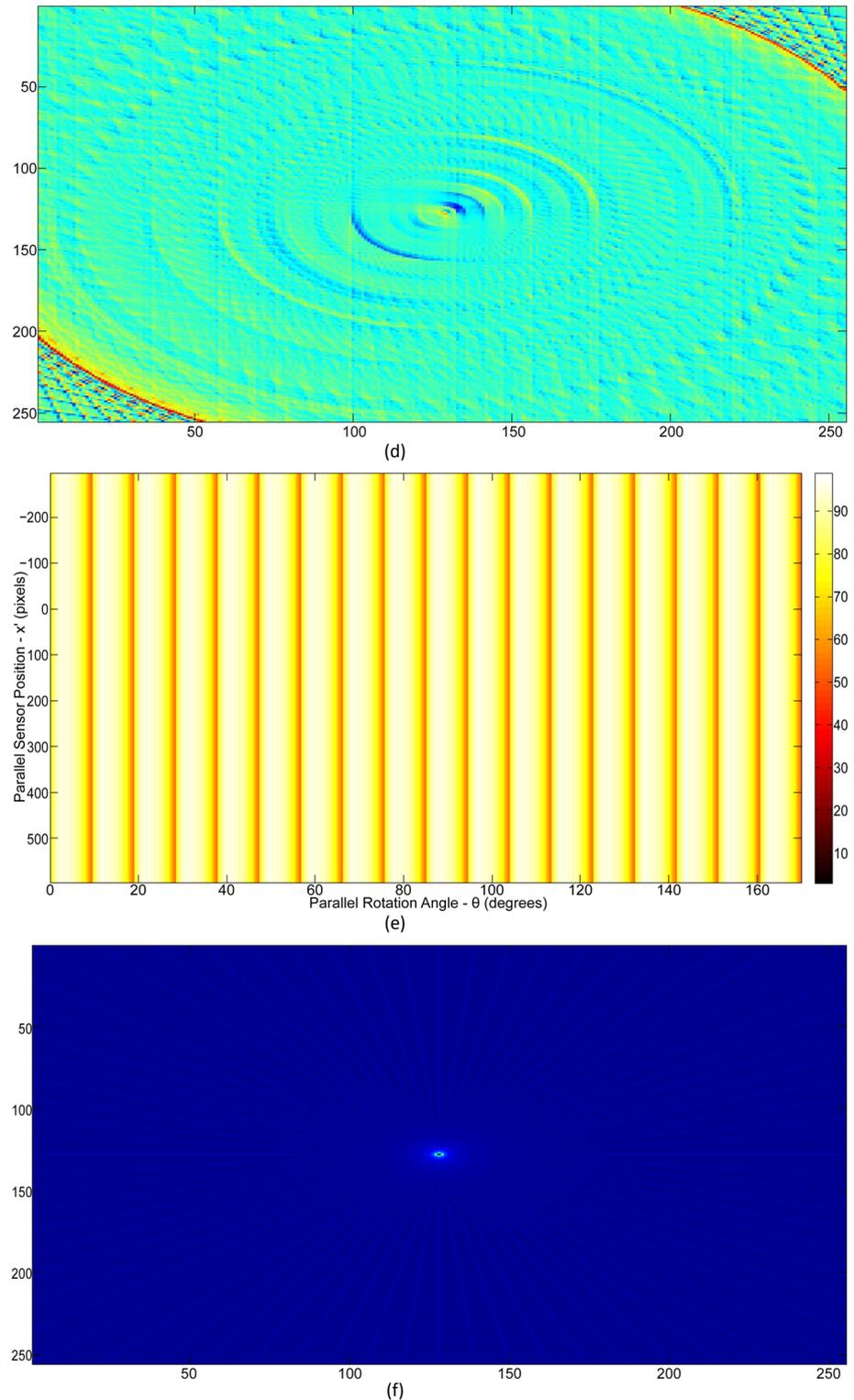


Figure 2. (a) Monte Carlo simulation geometry; (b) Detector response at the radius distance 16.3 cm, the channel number is corresponding to the z direction in (a) and the signal is normalized dose deposited at the region 16.0 cm - 16.3 cm; (c) Raw projection data plot. The parallel rotation angle is corresponding to angle along the z direction in (a) and the parallel sensor position is corresponding to signal response in r the direction. (d) Reconstructed CT image from raw projection (signal c); (e) Projection data plot with de-noising; (f) Reconstructed CT image with de-noising (signal e).

3.2. Source Position, Dimension Comparison and Signal-to-Noise Ratio

Figure 2(b) shows the profile of one normalized signal result at a distance of 16.0 cm to 16.3 cm along the z direction. **Figure 2(c)** shows the original 2D CT sinogram image and **Figure 2(d)** shows the re-constructed image. **Figure 2(e)** shows the de-noised 2D CT sinogram image and **Figure 2(f)** shows the re-constructed image of **Figure 2(e)**. The 2D CT was constructed using a dimension of 255×255 . If 50% of the maximum signal was considered to be source diameter boundary, the source diameter dimension was then 1.3 mm ($2 \times 160 \text{ mm}/255 \text{ pixel} \times 1 \text{ pixel}$), which is comparable to the 1.1 mm diameter reported by Mowlavi *et al.* in 2008 [18]. If we use the de-noised image shown in **Figure 2(d)** for comparison, the signal to noise ratio would be 19.91 db. Histogram plots for **Figure 2(d)** and **Figure 2(f)** show a clear improvement in signal to noise ratios and are presented in **Figure 3(a)** and **Figure 3(b)**.

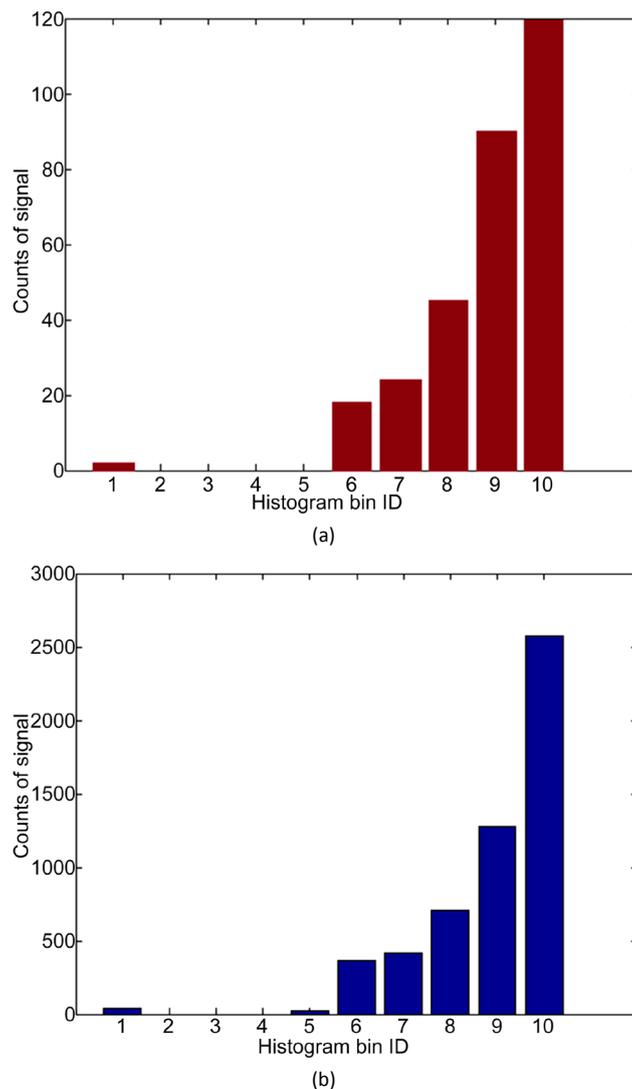


Figure 3. (a) Histogram plot for **Figure 2(d)**, (b) histogram plot for **Figure 2(f)**. A 20-fold improvement was observed with the de-noising technique.

A 2nd Monte Carlo simulation was performed with the source retracted by 5 cm. The results were merged with the first simulation to simulate two source dwell positions. The magnitude of the 2nd simulation was weighted at 30% of the first simulation in order to differentiate the two signals. **Figure 4** shows the dose/particle plot for the simulation of the two differentially weighted dwell positions. The second source center position calculated from the Monte Carlo simulation was 50.2 mm compared to the programmed center position of 50.175 mm. The sub-millimeter accuracy of results clearly illustrates correct source position and relative weight.

4. Discussions and Summary

Even though the signal response is low compared to the prescribed dose ($0.672/500 = 0.13\%$), it is still possible to use the Ir-192 source as a radiation source for CT-like imaging. With proper signal processing, this could offer localization and verification information. Our Monte Carlo simulation results show that the image can offer correct source dose, position, and dimension information despite of relative high uncertainties.

Currently, there are few publications on the use of the Ir-192 source itself as an imaging source. Safavi-Naeini *et al.* designed a BrachyView *in vivo* imaging device for brachytherapy [15] [16] [17], which can achieve sub-millimeter accuracy with an equivalent acquisition time of 0.5 s and their results support the use of the Ir-192 source as an imaging source. However, since BrachyView is an in-body device, its application is limited in other treatment sites, such as with breast HDR. For the real device design, the publication by Duan (2010) showed a

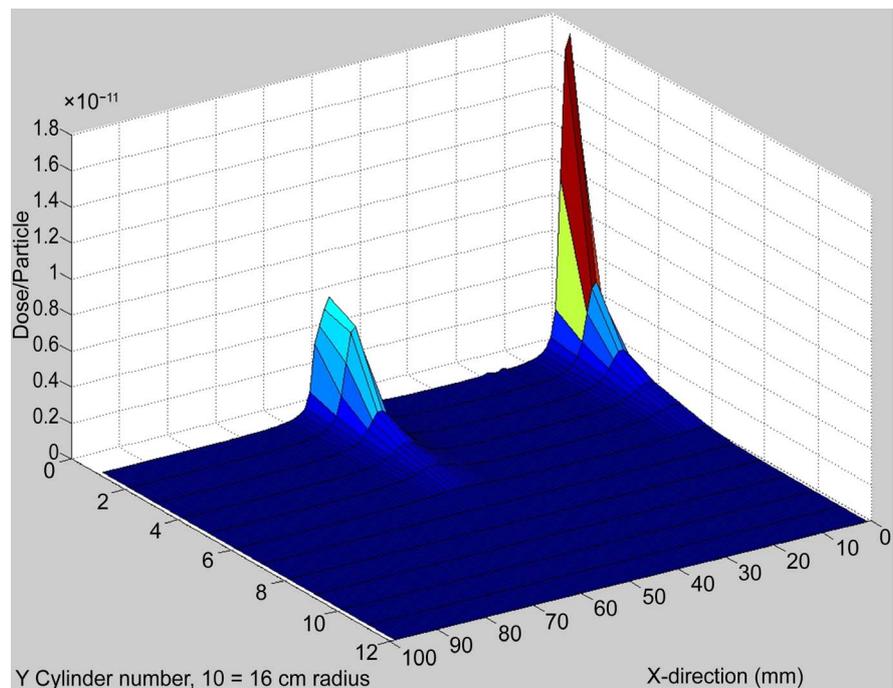


Figure 4. Simulation of two differentially weighted dwell positions. Where X direction is along **Figure 2(a)** z direction and Y direction is along **Figure 2(a)** r direction.

0.5 mm pinhole camera could resolve source position separation as small as 1 mm [19].

Our goal is to design an outside of the body CT-like device and to achieve similar imaging and dose information as was done with BrachyView. Further study is still needed for the detailed design of an out of the body CT-like device for brachytherapy imaging, but this preliminary study provides a proof of a concept for Monte Carlo simulations and also provides guidelines for future device design.

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