Original Article

Dosimetric Comparison of Treatment Plans Computed With Finite Size Pencil Beam and Monte Carlo Algorithms Using the InCise™ Multileaf Collimator-Equipped Cyberknife® System

Kalpani Nisansala Udeni Galpayage Dona[1], Charles Shang[1,2], Theodora Leventouri[1]

[1]Department of Physics, Florida Atlantic University, Boca Raton, [2]South Florida Proton Therapy Institute, Delray Beach, Florida, USA

Abstract

Purpose: InCise™ multileaf collimator (MLC) was introduced for CyberKnife® (CK) Robotic Radiosurgery System (CK-MLC) in 2015, and finite size pencil beam (FSPB) was the only available dose computation algorithm for treatment plans of CK-MLC system. The more advanced Monte Carlo (MC) dose calculation algorithm of InCise™ was initially released in 2017 for the CK Precision™ treatment planning system (TPS) (v1.1) with new graphic processing unit (GPU) platform. GPU based TPS of the CK offers more accurate, faster treatment planning time and intuitive user interface with smart three-dimensional editing tools and fully automated autosegmentation tools. The MC algorithm used in CK TPS simulates the energy deposited by each individual photon and secondary particles to calculate more accurate dose. In the present study, the dose disparities between MC and FSPB algorithms for selected Stereotactic Ablative Radiation Therapy (SABR) CK-MLC treatment plans are quantified. **Materials and Methods:** A total of 80 CK-MLC SABR plans computed with FSPB were retrospectively reviewed and compared with MC computed results, including plans for detached lung cancer (or tumors fully surrounded by lung tissues, $n = 21$), nondetached lung cancer (or tumor touched the chest wall or mediastinum, $n = 23$), intracranial ($n = 21$), and pancreas lesions ($n = 15$). Dosimetric parameters of each planning target volume and major organs at risk (OAR) are compared in terms of normalized percentage deviations (N_{dev}). **Results:** This study revealed an average of 24.4% overestimated D_{95} values in plans using FSPB over MC for detached lung ($n = 21$) and 14.9% for nondetached lung ($n = 23$) lesions. No significant dose differences are found in intracranial (0.3%, $n = 21$) and pancreatic (0.9%, n = 15) cases. Furthermore, no significant differences were found in N_{dev} of OARs. **Conclusion:** In this study, it was found that FSPB overestimates dose to inhomogeneous treatment sites. This indicates, the employment of MC algorithm in CK-MLC-based lung SABR treatment plans is strongly suggested.

Keywords: Conformity Index, CyberKnife, finite size pencil beam, homogeneity index, Monte Carlo, stereotactic ablative radiation therapy, tissue heterogeneity

Received on: 24-07-2019 **Review completed on: 02-11-2019** **Accepted on: 19-11-2019** **Published on: 13-03-2020**

Introduction

InCise™ Multileaf Collimator (MLC) of CyberKnife® (CK) Robotic Radiosurgery System (Accuray Inc, Sunnyvale, CA, USA) with significantly improved treatment efficiency[1,2] was released to clinical use in 2015. An overwhelmed concern was related to the dose discrepancies in the treatment plans computed with the then-available finite size pencil beam (FSPB) algorithm and those computed using industry well-accepted algorithms including the Monte Carlo (MC) dose algorithm.[3]

The shorter treatment delivery time and experience with MC planning encouraged the clinical implementation of CK-MLC to stereotactic radio surgery (SRS) and stereotactic ablative radiation therapy (SABR) taking the existing advantages of the unique dynamic motion management powered by its robotic system.[4,5]

The robotic arm empowered CyberKnife® consists of three types of beam collimator: Discrete fixed collimators

Access this article online

Quick Response Code:

Website: www.jmp.org.in

DOI: 10.4103/jmp.JMP_64_19

Address for correspondence: Ms. Kalpani Nisansala Udeni Galpayage Dona, Department of Physics, Florida Atlantic University, 777 Glades Road, Boca Raton, Florida, USA.
E-mail: gudeni2014@fau.edu

This is an open access journal, and articles are distributed under the terms of the Creative Commons Attribution-NonCommercial-ShareAlike 4.0 License, which allows others to remix, tweak, and build upon the work non-commercially, as long as appropriate credit is given and the new creations are licensed under the identical terms.

For reprints contact: reprints@medknow.com

How to cite this article: Galpayage Dona KN, Shang C, Leventouri T. Dosimetric Comparison of treatment plans computed with finite size pencil beam and monte carlo algorithms using the incise™ multileaf collimator-equipped cyberknife® system. J Med Phys 2020;45:7-15.

(5–50 mm), Iris variable aperture collimators, and InCise MLCs for variably shaped dose-intensity-modulated beamlets.[6-8] The capability of respiration synchronized real-time motion tracking enabled the CK effectively deliver dose to a moving target, such as for lung SABR with high precision.[9-11] Otherwise much larger treatment margins have been used in such treatments.[12]

This Synchrony® Respiratory Tracking System of CK establishes a prediction three-dimensional model of moving target based on a series of paired orthogonal X-ray images correlated with the body surface motion due to respiration. During dose delivery, the synchronization model is periodically verified and adjusted to ensure its accuracy.[11,13] The clinical efficacy and reliability of CK-MLC have been demonstrated in multiple studies.[8,14,15]

The MC algorithm of InCise™ was approved for clinical use in the new graphic processing unit (GPU) platformed Precision™ treatment planning system (TPS) (v1.1) in 2017. In this TPS, FSPB dose calculations are primarily based on the effective path length, while the MC particularly addresses the tissue heterogeneity effects on primary and scatter doses to give a more accurate dose calculation.[10,16,17] It has been widely accepted in the literature that MC shall be used in treatment planning of regions with substantial density variation, such as a tumor in the lungs or near the sinus air cavities.[10,18] Accuracy of dose calculation is an important factor to ensure that adequate dose can be delivered leading to a better clinical outcome in radiation therapy.[19,20]

The main objective of this study is to quantify the dosimetric differences between MC and FSPB dose calculation algorithms for CK-MLC plans for lung SABR and other selected anatomical sites. Initially, the beam commissioning for the MC algorithm was processed for the CK-MLC M6 system using the Accuray Precision™ TPS (v1.1). The dose calculation results report the dosimetric accuracy of MC algorithm for the CK-MLC included in the GPU platformed Precision™ TPS. Furthermore, the selected dose points of organs at risk (OAR) for critical volumes were recorded for both algorithms of each treatment plan and compared. Moreover, the quality of the treatment plans was evaluated by calculating the Radiation Therapy Oncology Group (RTOG) Conformity Index (CI) and Homogeneity Index (HI).

Materials and Methods

The Monte Carlo beam commissioning for CyberKnife Precision treatment planning system

Beam commissioning for the MC dose calculation algorithm with MLC collimators of the CyberKnife® M6 system was processed using the Accuray Precision System. Each secondary collimator type (Fixed, Iris and MLC) of CK requires independent source model.[10] The Gaussian distribution source model was selected for the system with initial full width at half maximum (FWHM) of 1.8 mm and related modeling functions for source distribution, fluence distribution, and the energy spectrum. During further tuning, MC calculated off centre ratio (OCR) and tissue phantom ratio (TPR) were compared with the measured values for all depths and field sizes with 0.2% uncertainty and the source model parameters (FWHM and energy spectrum) were adjusted accordingly. The final FWHM was determined to be 2.0 mm with 7.0 MeV energy spectrum. The computed output factors were compared with the measured values with 0.2% uncertainty. All the MC calculations are subjected to a statistical uncertainty with a trade-off between calculation time and spatial resolution.

Treatment plan selection, dose calculation, plan comparison, quality of treatment plans and evaluation tools

A total of 80 CK-MLC SABR treatment plans from the selected institution that had been computed with FSPB were retrospectively reviewed and recomputed with MC dose calculation algorithm. All the MC dose calculations were performed with 1% uncertainty and resolution was matched with the FSPB plans with same monitor units settings. Treatment plans included detached lung cancer (tumors fully surrounded by lung tissues, n = 21), nondetached lung cancer (tumor touched the chest wall or mediastinum, n = 23), intracranial (n = 21), and pancreas lesions (n = 15). Detached lung cancer was defined when the closest edge of tumor is 0.5 cm away from the chest wall, or mediastinum at its closest view. The prescription doses (Rx) ranged from 5000 to 6000 cGy in 5 consecutive fractions for the lung SABR, 1400–2400 cGy for intracranial single fractional SRS, and 3500–4000 cGy in 5 fractions for pancreas cases. Planning target volume (PTV) D95 coverage in all the treatment plans was 90% or greater.

The FSPB plans were compared with MC computed plans in regard to dose distribution in PTVs and OARs for each treatment site. Recorded dose parameters for each treatment site include dose to 95% (D_{95}) volume of PTV, mean (D_{mean}), prescription dose (Rx), and maximum (D_{max}) dose to the PTV. In addition, maximum dose to 0.03 cc of OAR for each treatment was recorded. Furthermore, selected dose points in OAR for critical volumes were recorded. Mean values for dose parameters were calculated to evaluate the effect of tissue inhomogeneity on dose calculation for MC and FSPB algorithms. The quality of the treatment plans was evaluated by calculating RTOG CI, and the HI.

Statistical analysis

The percentage normalized deviation $\left(N_{dev} = [\text{MC - FSPB}] \times {}^{100\%}\!/_{\text{FSPB}}\right)$ for all dose parameters was calculated to express the percentage changes from MC to FSPB. Paired two sample t-tests with null hypothesis and Wilcoxon signed rank nonparametric test were accomplished with significance level $\alpha = 0.05$ to determine the significant statistical dissimilarity between the MC and FSPB dose calculation algorithms. Spearman correlation coefficient test was carried out to determine the effect of PTV size on the normalized deviation of each dose parameter.

Results

Monte Carlo commissioning for the multileaf collimator

Good agreement was found for FWHM of 2.0 mm with 7.0 MeV energy spectrum between measured and MC calculated OCR/TPR for all the field size ranges from 7.6 mm × 7.7 mm to 115.0 mm × 100.1 mm at all depths. Figure 1a presents the calculated and measured TPRs for all field sizes as a function of depth. The difference between measured and calculated TPR for all the field sizes versus depth are presented in Figure 1b. The MC calculation statistical uncertainties were 0.2% and 0.2% for the smallest and largest field size, respectively.

Figure 2 presents the x (left) and y (right) OCR profiles at 100.0 mm depth (a) for the smallest field size, 7.6 mm × 7.7 mm, and (b) for the largest field size, 115.0 mm × 100.1 mm. The average statistical uncertainty of MC calculation for the smallest and largest fled size at the 100 mm depth is 0.07% and 0.07%, respectively.

MC calculated and measured output factors are compared in Figure 3. The MC calculation had an average statistical uncertainty of 0.2%. The output factor is defined as ratio of the absorbed dose at a particular field size to the absorbed dose by 60 mm fixed collimator size at 800 mm source-axis distance (SAD). All the OF for MLC are measured only at 800 mm SAD using Sun Nuclear Edge detector™.

Dosimetry comparison of Monte Carlo and finite size pencil beam computed dose distributions

The percentage normalized deviation $\left(N_{dev} = [\text{MC - FSPB}] \times {}^{100\%}/_{\text{FSPB}}\right)$ between MC and FSPB dose calculations was calculated for each dose parameter of all 80 treatment plans. The averaged normalized deviations for each PTV dose parameter of all treatment plans are listed in Table 1. The paired sample *t*-test and Wilcoxon signed-rank nonparametric tests were carried out to evaluate the significant differences in dose parameters between MC and FSPB dose calculations. The calculated *P* values and sign rank *P* values are also included in Table 1.

Figure 4 presents the comparison of N_{dev} of dose parameters through the selected treatment sites. FSPB plans of CK-MLC overestimate D_{95} of PTV by an average of 24% compared to the MC computed plans in detached lung cases, and 15% in nondetached lung cases. No significant normalized dose deviations were found for the intracranial or pancreas treatment plans. Intracranial treatment plans have N_{dev} of 0.3% and pancreas treatment plans have N_{dev} of 0.9%. Note that percentage deviation >7% is considered clinically significant.[21]

The largest differences between MC and FSPB dose calculations of the D_{mean} PTV dose parameter were found for the lung treatment plans. Specifically, ~19% for the

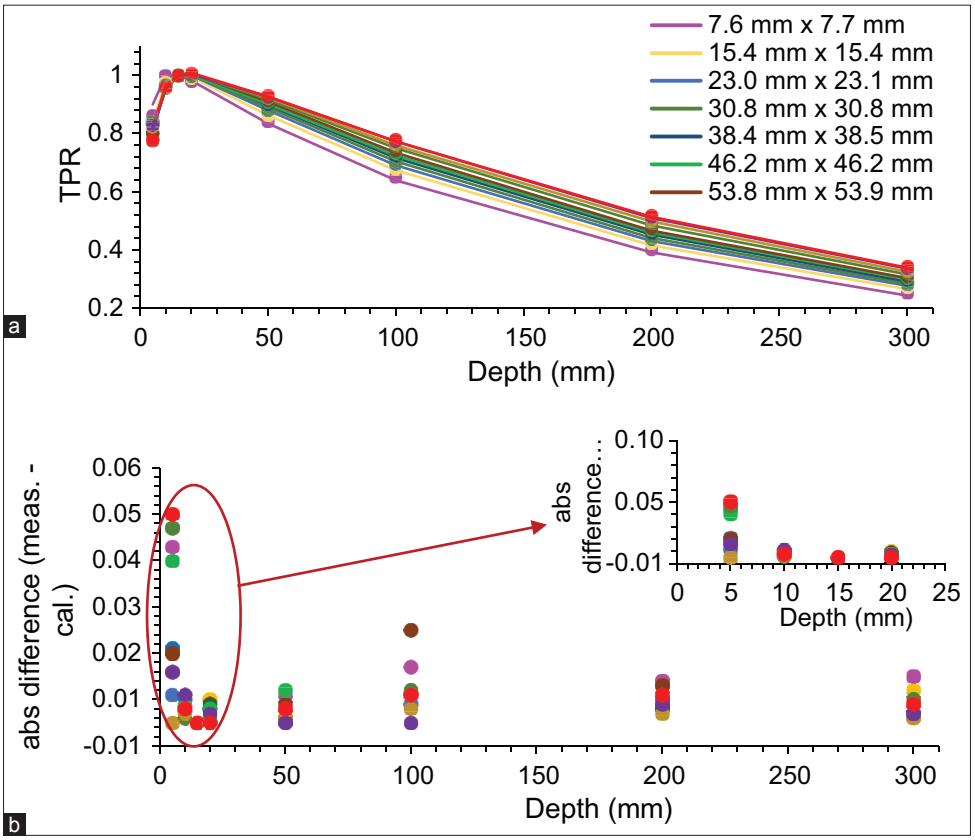

Figure 1: (a) Measured (dots) and Monte Carlo calculated (solid line) tissue phantom ratio graphs for all the field sizes. (b) Difference between measured and calculated tissue phantom ratio values of all field sizes as a function of depth

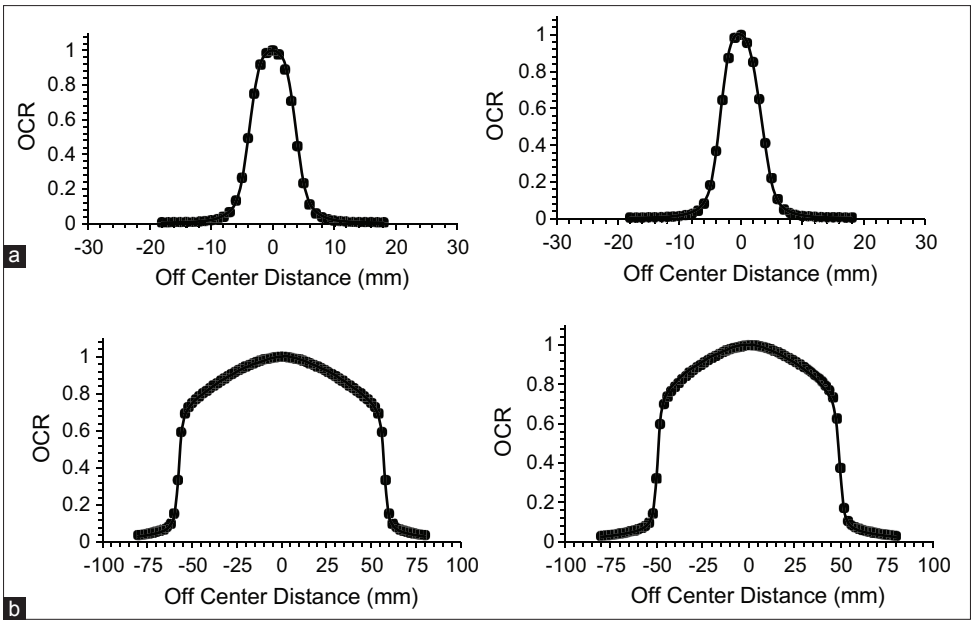

Figure 2: Measured (dots) and Monte Carlo calculated (solid line) off center ratio profiles, x (left) and y (right) profiles (a) for the smallest field size, 7.6 mm × 7.7 mm of multileaf collimator (b) for the largest multileaf collimator field size, 115.0 mm × 100.1 mm at 100 mm depth

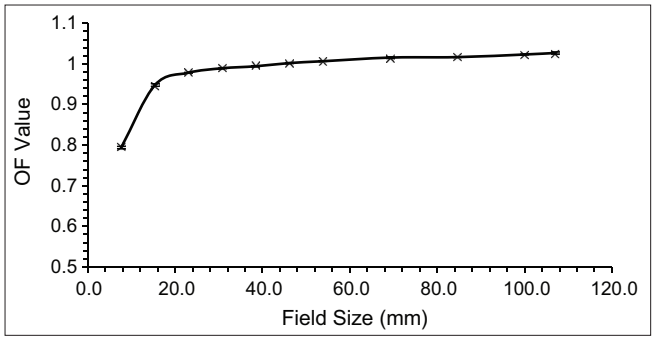

Figure 3: Measured (crosses) and Monte Carlo calculated (solid line) output factors as a function of the field size for the multileaf collimator collimators

detached lung treatment plans and ~12% for the nondetached lung treatment plans. Intracranial (N_{dev} = 0.6%) and pancreas (N_{dev} = 0.7%) treatment plans have no significant differences. The D_{max} of PTV has lower significant differences (<7%) between MC and FSPB dose calculation for all four treatment sites.

Overall, MC dose calculated plans show significantly reduced values (P < 0.00001) for lung treatment plan that are in close agreement with previous studies.[21-23] The lung SABR plans encompass the situation with the largest tissue density heterogeneity among all the groups, especially those for detached lung tumors where the disadvantage of FSPB results to the largest percentage normalized deviation in dose computations as expected.[21,24,25] These differences emphasize the lack of photon scatter correction at low tissue densities in FSPB algorithm where dose calculations are primarily based on the effective path length, while the MC dose calculation considers the heterogeneity effects on scatter dose.[10,16,25]

Table 1: Comparison of the results between dose distribution calculated by Monte Carlo and finite size pencil beam algorithms

Treatment site	PTV dose parameter	*P* (2 tail)	Signed rank *P*	Averaged N_{dev} (%)	SD (%)
Nondetached lung (*n*=23)	D_{95}	<0.00001	<0.00001	−14.9	7
	D_{mean}	<0.00001	<0.00001	−12.4	5
	D_{max}	0.06	0.0003	−4.3	4
Detached lung (*n*=21)	D_{95}	<0.00001	<0.00001	−24.4	12
	D_{mean}	<0.00001	<0.00001	−18.9	9
	D_{max}	0.002	0.004	−4.2	5
Intracranial (*n*=21)	D_{95}	0.5	0.7	−0.3	2
	D_{mean}	0.02	0.03	0.6	1
	D_{max}	<0.00001	<0.00001	5	1
Pancreas (*n*=15)	D_{95}	0.3	0.4	0.9	3
	D_{mean}	0.2	0.2	0.7	2
	D_{max}	<0.00001	<0.00001	6.5	2

SD: Standard deviation, PTV: Planning target volume, SD: Standard deviation

Greater values of the standard deviations were found for all the dose parameters of selected treatment sites, indicating that the data points are broadly distributed. This may be attributed to the small sample size for each treatment site. Similar standard deviations were found in the dose comparison study by Kim *et al*. for the lung and breast treatment sites.[25]

The performance of planning target volume size

Normalized percentage deviations of all dose parameters between MC and FSPB were compared as a function of the PTV size for all treatment plans. The results are presented in Figure 5. Spearman correlation coefficient was carried out to represent the PTV size dependency of the N_{dev} for all treatment plans.

From Figure 5a and b, dose parameters of PTV (D_{95}, D_{mean}, D_{max}) show larger normalized percentage deviation for the smaller PTV size of lung treatment plans. The largest % discrepancy (−53%) was found for PTV volume 17.3 cc [Figure 5b] for the case of detached lung PTV where tissue inhomogeneity is greater compared to the other treatment sites.[21,24,25] For the PTV D_{mean}, the largest discrepancy (42%) was found at detached lung PTV at the same tumor volume 17.3 cc as shown in Figure 5b. The largest difference (8.7%) of the PTV D_{max}, was found at PTV size of 16.5 cc also located at detached lung tumor.

Table 2 lists the mean Spearman correlation coefficient values of all dose parameters of all treatment plans. The RHO values closer to ±1 show perfect relation between the N_{dev} and PTV size and closer to zero indicate that N_{dev} has weaker dependency on PTV size. No significant dependency of Ndev on PTV size were found for pancreas [Figure 5c] and intracranial [Figure 5d] treatment plans. *P* values indicate the statistically significant relation between N_{dev} and PTV size. Our results show that PTV size has weaker dependency on the N_{dev} for all dose parameters of all treatment sites.

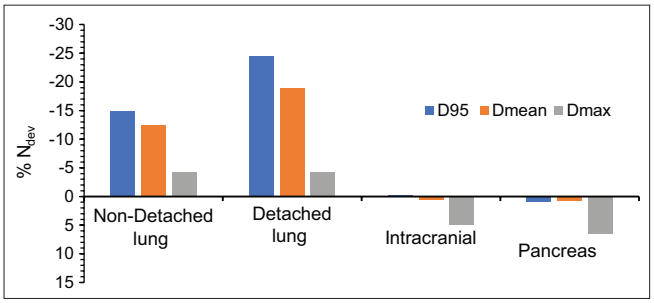

Figure 4: Comparison of the normalized percentage deviation (N_{dev}) of the dose parameters for the selected treatment sites

Radiation Therapy Oncology Group Conformity Index and Homogeneity Index

The RTOG CI and HI were calculated with both dose calculation algorithms by using the following equations:

$$CI = \frac{V_p}{PTV} \quad (1)$$

$$HI = \frac{D_{max}}{D_p} \quad (2)$$

V_p is the volume covered by the prescription isodose line, D_{max} is the maximum dose, and D_p is the prescribed dose.[26,27] The results of CI as a function of the PTV size are plotted in Figure 6. Triangles represent the CI for the treatment plans computed with the MC algorithm, and dots represent the same treatment plans calculated with the FSPB algorithm.

Plans with CI in the range of 1.0 and 2.0 are complying with RTOG protocol. Plans with CI between 0.9–1.0 and 2.0–2.5 are within minor deviations, while plans with CI >2.5 or CI <0.9 have major violations, but nevertheless acceptable.[26-30] CI <1

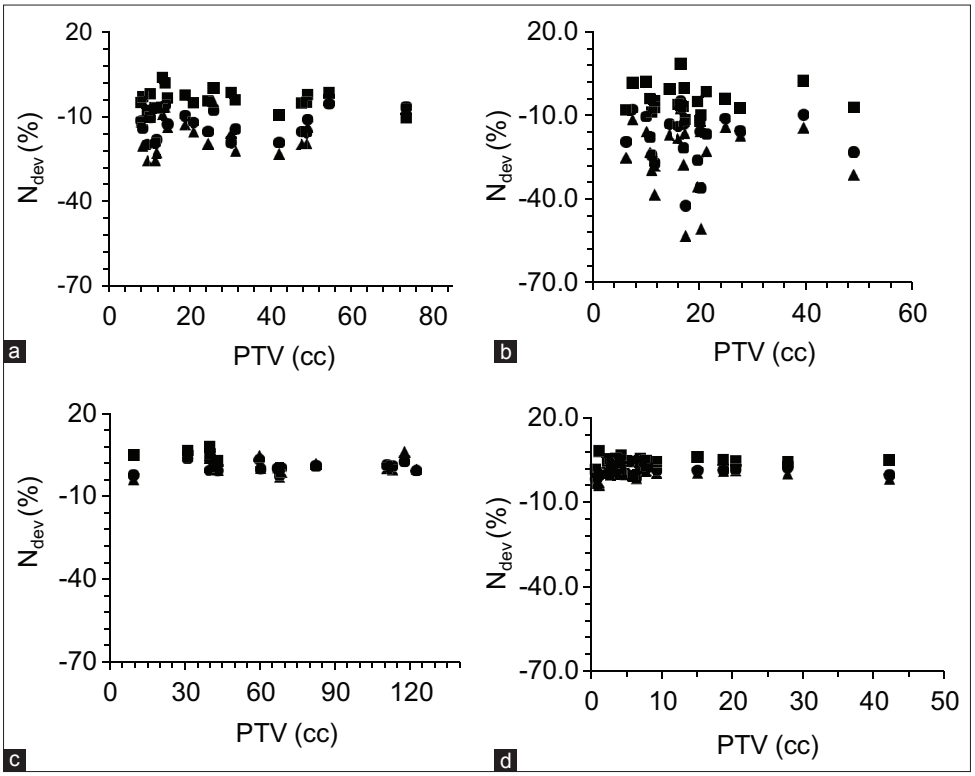

Figure 5: The normalized percentage deviation of the dose parameters as a function of planning target volume size: (a) for nondetached lung treatment plans ($n = 23$) (b) for detached lung treatment plans ($n = 21$), (c) for pancreas treatment plans ($n = 15$), (d) for intracranial treatment plans ($n = 21$). Triangle: D_{95}, Circle: D_{mean}, Square: D_{max}

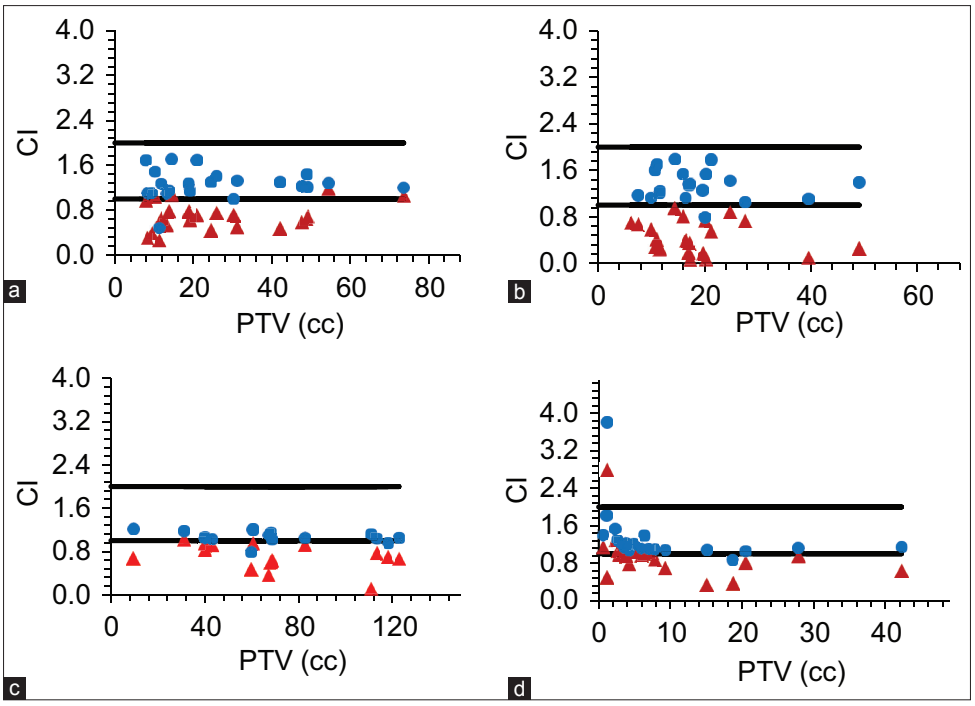

Figure 6: Conformity Index as a function of planning target volume size (a) for nondetached lung treatment plans (b) for detached lung treatment plans (c) for pancreas treatment plans (d) for intracranial treatment plans. The data points that fall within the two horizontal solid lines are considered within the radiation therapy oncology group protocol. Triangles mark the Monte Carlo conformity index and dots mark the finite size pencil beam conformity index

Table 2: Spearman correlation coefficients for all treatment plans

Treatment site	PTV dose parameter	ρ	*P*
Detached lung (*n*=21)	D_{95}	−0.04	0.9
	D_{mean}	−0.03	0.9
	D_{max}	−0.1	0.6
Nondetached lung (*n*=23)	D_{95}	0.2	0.3
	D_{mean}	0.2	0.4
	D_{max}	0.2	0.4
Intracranial (*n*=21)	D_{95}	0.4	0.070
	D_{mean}	0.5	0.02
	D_{max}	−0.02	0.9
Pancreas (*n*=15)	D_{95}	−0.05	0.8
	D_{mean}	0.04	0.9
	D_{max}	0.04	0.9

PTV: Planning target volume

means that the PTV is not covered by the reference dose, and CI >2 means that the entire PTV is covered by the reference dose, but healthy tissues are also included into the high-dose irradiation which is not negligible.[28]

It is found that 71.8% of the MC computed treatment plans had major deviations with CI <0.9 where the largest contribution was found in detached lung cancer treatment plans. 15.4% had no deviation from the RTOG protocol, while minor deviations occurred for 12.8%. A CI_{max} = 2.8 was found in intracranial treatment plans with PTV size of 1.14 cc [Figure 6d]. The CI_{min} = 0.1 was found for detached lung treatment plan [Figure 6b] and pancreas treatment plan [Figure 6c]. For the detached lung plan CI_{min} occurred at three PTV sizes, 13 cc, 14 cc, and 21 cc. For the pancreas treatment plans, the CI_{min} was found at PTV size of 110.9 cc.

Among all treatment plans computed with the FSPB algorithm, 91% fall within RTOG protocol. 7.7% of treatment plans show major deviations while minor deviations occur only for 1.3% of all FSPB computed treatment plans. CI_{max} = 4.2 was found for the detached lung treatment plan [Figure 6b] with PTV size of 6.16 cc. The CI_{min} = 0.5 appeared for the nondetached lung treatment plans with PTV size of 11.4 cc [Figure 6a].

Homogeneity Index

Figure 7 presents the HIs as a function of PTV size for all the treatment plans. All HIs were calculated with both MC and FSPB algorithms. Triangles mark the HI for the MC treatment plans and dots mark the HI for the FSPB treatment plans.

According to the RTOG protocol, treatment plans with HI ≤2.0 are considered to comply with it. Treatment plans with homogeneity indices between 2.0 and 2.5 have minor deviation, while major deviations occur for HI >2.5.[26,31]

As shown in Figure 7 the HI values for all MC calculated and FSPB calculated treatment plans are below 2.0, meaning that all the treatment plans are within the RTOG protocol. The HI_{max} of 1.4 was found for the treatment plans computed with MC. This occurred at the nondetached lung treatment plan for two PTV sizes, 11.8 cc and 54.5 cc. The corresponding HI_{max} for the treatment plans computed with

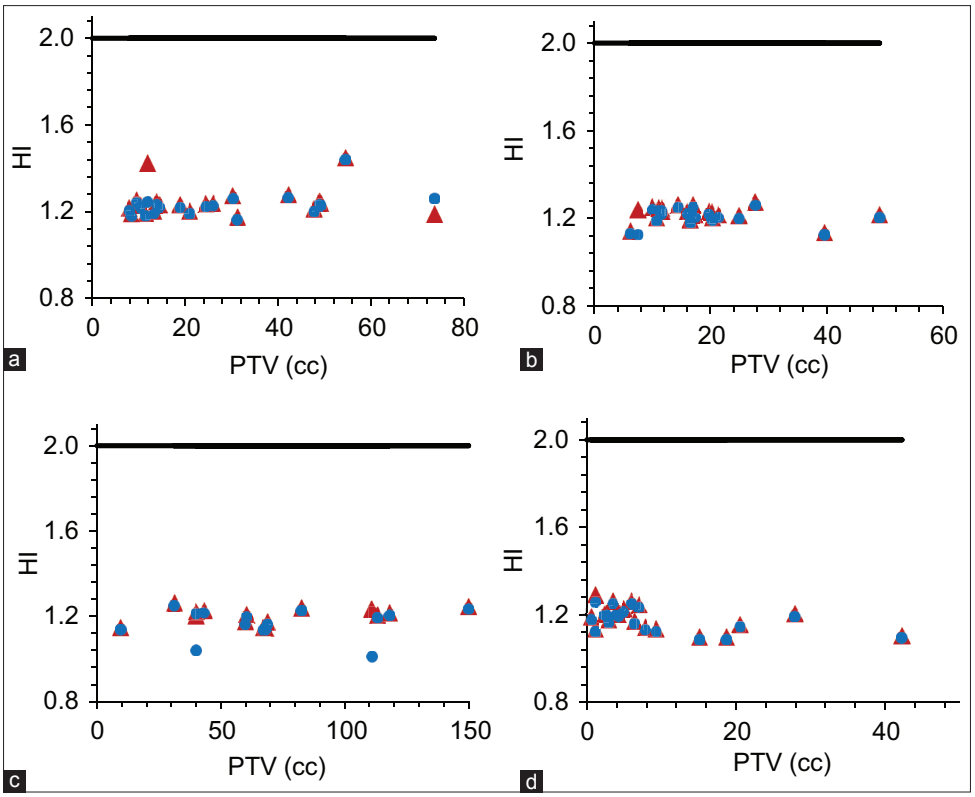

Figure 7: Homogeneity Index as a function of planning target volume size. (a) for nondetached treatment plans (b) detached treatment plans (c) for pancreas treatment plans (d) intracranial treatment plans. Points that are below the solid line (homogeneity index = 2.0) are defined as being within the radiation therapy oncology group protocol. Triangles mark the Monte Carlo homogeneity indexes; dots mark the finite size pencil beam homogeneity indexes

FSPB was also 1.4 and it was found at the same treatment plan, nondetached lung treatment plan with the same PTV size of 54.5 cc. In general, it was found that the HI values do not change significantly with the PTV size for both algorithms.

Discussion

In this research we study the tissue homogeneity effect on dose calculation during the CK-MLC treatment planning. The MC algorithm is well known of its ability to consider homogeneity corrections to deliver an accurate dose calculation at low tissue density medium.[10,16,17] The dosimetric parameters of CK-MLC plans computed with MC and FSPB algorithms are quantitatively compared. From our results, FSPB overestimates dose to the lung treatment plans where PTVs are surrounded by low-density inhomogeneous lungs. This overestimation is insignificant for the intracranial and pancreas treatment plans where the tumor location is in a homogeneous area. The dose deposited in low density medium is affected by the lack of photon scatter correction at low densities and the lack of electron equilibrium in heterogeneous geometries where these factors are considered during MC dose calculation.[32] Thus, our results imply that the MC algorithms provide accurate and optimum dose to the PTV at inhomogeneous treatment sites.

Similar results have been published by multiple studies on different treatment machines.[21,22,33-36] In Chen *et al*. study, PB algorithm overestimates the dose to the PTV significantly by an average of 22.9% for the D_{95} of PTV.

In addition, insignificant dose deviations were found between MC and FSPB for the OARs for all the treatment site which may be attributed to adjustment of the distance from the PTV. Perceptible differences were found only at spinal cord for lung treatment plan. This may be attributed to electron transport corrections at inhomogeneous tissue medium by the MC. Similar findings are published in studies by Zhuang *et al*.[21,34]

From the RTOG CI study, our results show that MC dose calculation has low PTV coverage compared to the FSPB dose calculation algorithm. This is suggesting that MC computed treatment plans are not satisfied with RTOG protocol. Zhung *et al*. also found that MC computed treatment plans are not satisfied with RTOG protocol.[37] All the treatment plans had the ideal RTOG HI meaning that both MC and FSPB provide homogeneous dose to the PTV. Stanley *et al*. also found ideal HI for all 170 treatment plans in their study.[31]

In summary, our CI study suggests that more plan optimization is required to have a superior PTV coverage for MC evaluated CK-MLC plans. This can be done during the treatment panning. However, this study suggests that the MC algorithm provides

more accurate dose calculation in evaluating CK-MLC plans at low tissue density regions.

Conclusion

The findings in this study suggest that FSPB overestimates the dose in the interface region between soft-tissue density lung tumor and low-density lung. This occurs due to the scatter electron equilibrium is built up at low density regions and doses computed by MC algorithm improves accuracy. In CK-MLC-based SABR, the plans for high heterogeneous regions, such as lungs, it is strongly recommended that MC algorithm become standard.

Financial support and sponsorship

Nil.

Conflicts of interest

There are no conflicts of interest.

References

1. Kathriarachchi V, Shang C, Evans G, Leventouri T, Kalantzis G. Will cyberknife m6™ multileaf collimator offer advantages over IRIS™ collimator in prostate SBRT? J Appl Clin Med Phys Open Access 2015;16:442-3.
2. Kathriarachchi V, Shang C, Evans G, Leventouri T, Kalantzis G. Dosimetric and radiobiological comparison of CyberKnife M6™ InCise multileaf collimator over IRIS™ variable collimator in prostate stereotactic body radiation therapy. J Med Phys 2016;41:135-43.
3. Heidorn SC, Kilby W, Fürweger C. Novel monte carlo dose calculation algorithm for robotic radiosurgery with multi leaf collimator: Dosimetric evaluation. Phys Med 2018;55:25-32.
4. Adler JR Jr, Murphy MJ, Chang SD, Hancock SL. Image-guided robotic radiosurgery Neurosurgery 1999;44:1299-306.6.
5. Kuo JS, Yu C, Petrovich Z, Apuzzo ML. The CyberKnife stereotactic radiosurgery system: Description, installation, and an initial evaluation of use and functionality. Neurosurgery 2003;53:1235-9.
6. Asmerom G, Bourne D, Chappelow J, Goggin LM, Heitz R, Jordan P, Kilby W, *et al*. "The design and physical characterization of a multileaf collimator for robotic radiosurgery." Biomedical Physics & Engineering Express 2016;2:017003.
7. Echner GG, Kilby W, Lee M, Earnst E, Sayeh S, Schlaefer A, *et al*. The design, physical properties and clinical utility of an iris collimator for robotic radiosurgery. Phys Med Biol 2009;54:5359-80.
8. Jang S, Lalonde R, Ozhasoglu C, Heron D, Burton S, Huq M. Is the newly released cyberknife InCise multileaf collimator better than circular collimators for treatment of single or multiple brain mets? Int J Radiat Oncol Biol Phys 2015;93:E589.
9. Sayeh S, Wang J, Main WT, Kilby W, Maurer CR. Respiratory motion tracking for robotic radiosurgery. In: Treating Tumors that Move with Respiration. Berlin, Heidelberg: Springer; 2007. p. 15-29.
10. Incoorperated A. Physics Essential Guide: Cyberknife Robotic Radiosurgery System. 11.0X ed. Accuray; 2017.
11. Collins BT, Erickson K, Reichner CA, Collins SP, Gagnon GJ, Dieterich S, *et al*. Radical stereotactic radiosurgery with real-time tumor motion tracking in the treatment of small peripheral lung tumors. Radiat Oncol 2007;2:39.
12. Brown WT, Wu X, Fayad F, Fowler JF, García S, Monterroso MI, *et al*. Application of robotic stereotactic radiotherapy to peripheral stage I non-small cell lung cancer with curative intent. Clin Oncol (R Coll Radiol) 2009;21:623-31.
13. Hoogeman M, Prévost JB, Nuyttens J, Pöll J, Levendag P, Heijmen B. Clinical accuracy of the respiratory tumor tracking system of the cyberknife: Assessment by analysis of log files. Int J Radiat Oncol Biol Phys 2009;74:297-303.
14. Eshleman JS, Berkenstock KG, Singapuri KP, Fuller C. The CyberKnife® M6™ radiosurgery system. J Lanc Gen Hosp. 2013;8:44-49.
15. Jang SY, Lalonde R, Ozhasoglu C, Burton S, Heron D, Huq MS. Dosimetric comparison between cone/Iris-based and InCise MLC-based cyberknife plans for single and multiple brain metastases. J Appl Clin Med Phys 2016;17:184-99.
16. Ma CM, Li JS, Deng J, Fan J. Implementation of Monte Carlo dose calculation for CyberKnife treatment planning. InJournal of Physics: Conference Series 2008;102:012016.
17. Rogers DW, Bielajew AF. Monte Carlo techniques of electron and photon transport for radiation dosimetry. The dosimetry of ionizing radiation 1990;3:427-539.
18. Wilcox EE, Daskalov GM. Accuracy of dose measurements and calculations within and beyond heterogeneous tissues for 6 MV photon fields smaller than 4 cm produced by Cyberknife. Med Phys 2008;35:2259-66.
19. Chetty IJ, Curran B, Cygler JE, DeMarco JJ, Ezzell G, Faddegon BA, *et al*. Report of the AAPM Task Group No. 105: Issues associated with clinical implementation of Monte Carlo-based photon and electron external beam treatment planning. Med Phys 2007;34:4818-53.
20. Wilcox EE, Daskalov GM, Lincoln H, Shumway RC, Kaplan BM, Colasanto JM. Comparison of planned dose distributions calculated by Monte Carlo and Ray-Trace algorithms for the treatment of lung tumors with cyberknife: A preliminary study in 33 patients. Int J Radiat Oncol Biol Phys 2010;77:277-84.
21. Zhuang T, Djemil T, Qi P, Magnelli A, Stephans K, Videtic G, *et al*. Dose calculation differences between Monte Carlo and pencil beam depend on the tumor locations and volumes for lung stereotactic body radiation therapy. J Appl Clin Med Phys 2013;14:4011.
22. Gete E, Teke T, Kwa W. Evaluation of the AAA treatment planning algorithm for SBRT lung treatment: Comparison with monte carlo and homogeneous pencil beam dose calculations. J Med Imaging Radiat Sci 2012;43:26-33.
23. Panettieri V, Wennberg B, Gagliardi G, Duch MA, Ginjaume M, Lax I. SBRT of lung tumours: Monte carlo simulation with penelope of dose distributions including respiratory motion and comparison with different treatment planning systems. Phys Med Biol 2007;52:4265-81.
24. Papanikolaou, Nikos, Battista JJ, Boyer AL, Kappas C, Klein E, *et al*. "Tissue inhomogeneity corrections for megavoltage photon beams." AAPM Task Group 65 2004:1-142.
25. Kim SJ, Kim SK, Kim DH. Comparison of pencil-beam, collapsed-cone and monte-Carlo algorithms in radiotherapy treatment planning for 6-MV photons. J Korean Phys Soc 2015;67:153-8.
26. Feuvret L, Noël G, Mazeron JJ, Bey P. Conformity index: A review. Int J Radiat Oncol Biol Phys 2006;64:333-42.
27. Shaw E, Kline R, Gillin M, Souhami L, Hirschfeld A, Dinapoli R, *et al*. Radiation therapy oncology group: Radiosurgery quality assurance guidelines. Int J Radiat Oncol Biol Phys 1993;27:1231-9.
28. Petkovska S, Tolevska C, Kraleva S, Petreska E. Conformity index for brain cancer patients. In Conference On Medical Physics and Biomedical Engineering 2010. p. 56.
29. Petrova D, Smickovska S, Lazarevska E. Conformity index and homogeneity index of the postoperative whole breast radiotherapy. Open Access Maced J Med Sci 2017;5:736-9.
30. Pemler P, Besserer J, Schneider U, Neuenschwander H. Evaluation of a commercial electron treatment planning system based on Monte Carlo techniques (eMC). Z Med Phys 2006;16:313-29.
31. Stanley J, Breitman K, Dunscombe P, Spencer DP, Lau H. Evaluation of stereotactic radiosurgery conformity indices for 170 target volumes in patients with brain metastases. J Appl Clin Med Phys 2011;12:3449.
32. Akino Y, Das IJ, Cardenes HR, Desrosiers CM. Correlation between target volume and electron transport effects affecting heterogeneity corrections in stereotactic body radiotherapy for lung cancer. J Radiat Res 2014;55:754-60.
33. Sikora M, Muzik J, Söhn M, Weinmann M, Alber M. Monte carlo vs. pencil beam based optimization of stereotactic lung IMRT. Radiat Oncol 2009;4:64.
34. Elcim Y, Dirican B, Yavas O. Dosimetric comparison of pencil beam

and monte carlo algorithms in conformal lung radiotherapy. J Appl Clin Med Phys 2018;19:616-24.
35. Chen A, Li Q, McNair S, Xu K. Target dose characterization by monte carlo versus pencil beam algorithm for lung stereotactic body radiation therapy (SBRT): A preliminary analysis in 20 successfully treated patients. Int J Radiat Oncol Biol Phys 2011;81:S852-3.
36. Kry SF, Alvarez P, Molineu A, Amador C, Galvin J, Followill DS. Algorithms used in heterogeneous dose calculations show systematic differences as measured with the Radiological Physics Center's anthropomorphic thorax phantom used for RTOG credentialing. Int J Radiat Oncol Biol Phys 2013;85:e95-100.
37. Zhung J, Melian E, Rusu I, Nagda S, Sethi A. Evaluation of RTOG guidelines for monte carlo-based Lung SBRT planning. Int J Radiat Oncol Biol Phys 2012;84:S201.