Recommendations for dose calculations of lung cancer treatment plans treated with stereotactic ablative body radiotherapy (SABR)

S Devpura, M S Siddiqui, D Chen, D Liu, H Li, S Kumar, J Gordon, M Ajlouni, B Movsas and I J Chetty

Henry Ford Health System, Detroit MI 48202 USA

Email: ichetty1@hfhs.org

Abstract. The purpose of this study was to systematically evaluate dose distributions computed with 5 different dose algorithms for patients with lung cancers treated using stereotactic ablative body radiotherapy (SABR). Treatment plans for 133 lung cancer patients, initially computed with a 1D-pencil beam (equivalent-path-length, EPL-1D) algorithm, were recalculated with 4 other algorithms commissioned for treatment planning, including 3-D pencil-beam (EPL-3D), anisotropic analytical algorithm (AAA), collapsed cone convolution superposition (CCC), and Monte Carlo (MC). The plan prescription dose was 48 Gy in 4 fractions normalized to the 95% isodose line. Tumors were classified according to location: peripheral tumors surrounded by lung (lung-island, \( N=39 \)), peripheral tumors attached to the rib-cage or chest wall (lung-wall, \( N=44 \)), and centrally-located tumors (lung-central, \( N=50 \)). Relative to the EPL-1D algorithm, PTV D95 and mean dose values computed with the other 4 algorithms were lowest for “lung-island” tumors with smallest field sizes (3-5 cm). On the other hand, the smallest differences were noted for lung-central tumors treated with largest field widths (7-10 cm). Amongst all locations, dose distribution differences were most strongly correlated with tumor size for lung-island tumors. For most cases, convolution/superposition and MC algorithms were in good agreement. Mean lung dose (MLD) values computed with the EPL-1D algorithm were highly correlated with that of the other algorithms (correlation coefficient =0.99). The MLD values were found to be ~10% lower for small lung-island tumors with the model-based (conv/superposition and MC) vs. the correction-based (pencil-beam) algorithms with the model-based algorithms predicting greater low dose spread within the lungs. This study suggests that pencil beam algorithms should be avoided for lung SABR planning. For the most challenging cases, small tumors surrounded entirely by lung tissue (lung-island type), a Monte-Carlo-based algorithm may be warranted.

1. Introduction
Sterotactic body radiation therapy (SBRT) or stereotactic ablative body radiotherapy (SABR) as coined by Loo and Timmerman et al [1] is a common therapeutic option available for patients with medically inoperable lung tumors. McGarry et al [2] and Timmerman et al [3,4] reported an excellent tumor control for early stage lung tumors when treated with SABR. Among the existing treatment planning systems, the Monte Carlo method is one of the most accurate algorithms for lung cancer dose calculations in the SABR setting [5]. In low density tissues, such as lung, the contribution of dose from electrons becomes significant because electrons which deposit dose within the tumor scatter laterally into the surrounding lung tissues carrying dose away from the tumor [6,7]. If the dose...
algorithm does not properly predict electron scattering, significant inaccuracies can result in the dose distributions, as in the case when pencil beam algorithms (employing equivalent path length, EPL corrections) are used.

In this retrospective study, five clinically commissioned dose calculation algorithms were used to compute lung SABR treatment plans for 133 non-small-cell lung cancer patients. The patients were planned and treated using a pencil beam algorithm (with a 1-D equivalent path length heterogeneity correction). We have investigated the dose accuracies in each algorithm by comparing the dose-volume histograms and dose parameters, and analyzed the impact of the field size and location comprehensively.

2. Method and Materials

The lung tumors were categorized by location into three groups as follows: island-type peripheral tumors surrounded by lung tissues (lung-island; \( N = 39 \)), tumors attached to the chest-wall (lung-wall; \( N = 44 \)), and tumors located in the central area (lung-central; \( N = 50 \)). Average equivalent square field sizes (FS, side of equivalent square) were used as a surrogate of tumor length in the beam’s eye view (BEV) and were subdivided into three groups, \( 3 \leq \text{FS} < 5 \), \( 5 \leq \text{FS} < 7 \), and \( 7 \leq \text{FS} < 10 \) cm. These lung patients were originally treated with 6 MV photons using IMRT. The dose regime was 48 Gy in 4 fractions normalized to 95% isodose line. Five clinically commissioned treatment planning systems (TPS) were used. They are BrainLAB iPlan 4.1.2 pencil beam (equivalent path length (EPL-1D)), Varian Eclipse 8.6 pencil beam algorithm (EPL-3D) and anisotropic analytical algorithm [8,9] (AAA), Phillips Pinnacle 8.1 collapsed cone convolution superposition [10,11] (CCC), and BrainLAB iPlan 4.1.2 Monte Carlo (MC) method [12,13,14]. Each of the above algorithms was commissioned using measurements in water phantoms as well as slab-phantoms with low-density, lung-equivalent media [15]. Beam models were within 2%/2 mm agreement vs. measurements in water phantoms and within 3% agreement in slab phantoms with lung-equivalent media [15] for all except the pencil-beam algorithms, where large discrepancies were observed.

For patient treatments IMRT plans were created using the iPlan EPL-1D algorithms. Using these same treatment plans (MUs and other parameters), dose was recomputed using the following algorithms: EPL-3D, AAA, CCC, and MC. To confirm that the beam models were in agreement with each other, uniform density (UD) plans (with densities assigned to 1.0) were first generated. Subsequently, heterogeneity corrected (HC) plans of all the treatment plans were calculated. The dose-volume histograms (DVHs) of the UD plans were among the five algorithms were found to be in agreement within a few percent for all plans. The dose to 95% of the planning target volume (PTV D95) was determined for all the dose calculation algorithms. The PTV D95 values were normalized to EPL-1D method (e.g. MC/EPL-1D). PTV mean doses were also reported. Mean lung dose (MLD) values were computed; the lung volume excluded the ITV from the sum of both lung volumes. The results presented here show the effects of tumor location and the field size in lung SABR-based treatment planning.

3. Results and Discussions

Tables 1 summarizes the PTV D95 (dose covering 95% of the PTV) and mean lung dose results with standard deviations for three field size categories. Note that \( N \) represents the number of patients in a particular category.

| FS (cm) | \( N \) | PTV D95 (%) | MLD (%) |
|--------|------|-------------|--------|
|        |      | EPL-3D | AAA | CCC | MC | EPL-3D | AAA | CCC | MC |
| \( 3 \leq \text{FS} < 5 \) | 50   | 95.3±1.9 | 82.4±5.1 | 82.6±6.2 | 82.3±6.0 | 94.4±7.4 | 90.7±5.6 | 91.8±6.1 | 90.9±5.7 |
| \( 5 \leq \text{FS} < 7 \) | 62   | 95.8±2.1 | 85.3±5.2 | 85.7±6.1 | 85.6±5.8 | 100.5±2.5 | 95.3±2.7 | 96.1±2.4 | 96.1±1.8 |
| \( 7 \leq \text{FS} < 10 \) | 21   | 95.9±1.7 | 90.4±3.7 | 90.6±3.8 | 90.8±3.8 | 102.0±2.8 | 96.1±2.1 | 95.0±2.4 | 97.3±3.2 |
The 3D pencil beam method (EPL-3D) showed approximately the same PTV D95 and PTV mean dose values regardless of the field size (within 4-5%). This illustrates that the EPL-3D algorithm is not accurate enough to estimate electron transport (lateral electron scattering and conditions of charged-particle disequilibrium) necessary for SABR-based dose calculation of lung tumors [16]. When the tumor and associated planning field size are small (< 5 cm) the impact of the dose reduction to the tumor is exacerbated due to the added effect of lateral electronic disequilibrium (resulting in more energy leaving versus being deposited in the tumor), which occurs when the field size is close to the range of the secondary electrons[5,7]. This requires more accurate accounting of electron transport in dose calculations for lung SABR [7]. Pencil beam algorithms lack the capability of account for loss of charge particle equilibrium and heterogeneity of the low density media. AAA and CCC dose values are in good agreement with MC and showed significantly lower doses (PTV D95~20%, PTV mean~12%, and MLD ~10%) for small field sizes. As the tumor (field) sizes increase, the PTV D95, mean dose, and MLD dose differences are reduced to ~10%, ~5%, and 3-4%, respectively.

Table 2. PTV D95 and MLD for EPL-3D, AAA, CCC, and MC algorithms for lung-island, lung-wall and lung-central tumors relative to the EPL-1D (100%) method.

| Location         | N  | PTV D95 (%) | MLD (%) |
|------------------|----|-------------|---------|
|                  |    | EPL-3D      | AAA     | CCC     | MC      | AAA     | CCC     | MC      |
| Lung-island      | 39 | 95.2±2.0    | 81.6±4.4| 81.4±5.8| 97.2±6.1| 92.1±5.1| 92.9±5.6| 92.9±5.3|
| Lung-wall        | 44 | 96.5±1.8    | 86.8±4.9| 87.4±5.6| 98.6±6.7| 94.3±4.8| 94.4±4.3| 94.9±4.9|
| Lung-central     | 50 | 95.2±1.8    | 86.2±5.9| 86.5±6.3| 99.3±4.8| 94.5±3.7| 95.4±3.8| 95.0±3.8|

Table 2 summarizes the PTV D95 and normal MLD dose results for lung-island, lung-wall, and lung-central tumors. The EPL-3D method was not able to capture the dose effects due to heterogeneity of the lung tissue. Thus, the dose values for all the tumor locations were somewhat similar. There were dramatic dosimetric differences in AAA, CCC, and MC algorithms due to location of the lung tumors. AAA and CCC media algorithms can handle the tissue inhomogeneity and secondary electron transport within the lung better than pencil beam methods [9]. Thus, these convolution superposition type algorithms predict the dose in lung media with much better accuracy relative to pencil beam algorithms. The largest dose differences were observed for island type tumors computed with AAA, CCC, and MC algorithms relative to the EPL-based methods. The tumors located in the central area and tumors attached to the rib cage showed similar dose results for AAA, CCC, and MC algorithms.

Table 3. PTV D95 dose for each algorithm for different tumor locations and field sizes relative to the EPL-1D (100%) method.

| Location         | FS (cm) | N  | PTV D95 (%) | AAA (%) | CCC (%) | MC (%) |
|------------------|---------|----|-------------|---------|---------|--------|
|                  |         |    | EPL-3D      | AAA     | CCC     | MC     |
| Lung-island      | 3≤FS<5  | 21 | 95.1±2.1    | 80.2±4.3| 80.0±6.0| 79.7±6.0|
|                  | 5≤FS<7  | 16 | 95.7±1.9    | 83.0±4.3| 82.7±5.4| 83.0±5.1|
|                  | 7≤FS<10 |  2 | 92.8±0.3    | 84.5±0.8| 85.3±0.7| 85.7±1.4|
| Lung-wall        | 3≤FS<5  | 13 | 96.2±1.2    | 84.7±4.4| 85.4±5.4| 84.5±5.3|
|                  | 5≤FS<7  | 22 | 96.4±2.3    | 86.2±4.8| 86.7±5.6| 86.3±5.6|
|                  | 7≤FS<10 |  9 | 97.1±1.1    | 91.4±2.8| 92.1±3.2| 91.9±3.2|
|                  | 3≤FS<5  | 16 | 94.8±1.8    | 83.6±5.5| 83.7±6.1| 84.0±5.7|
| Lung-central     | 5≤FS<7  | 24 | 95.3±2.1    | 86.1±5.8| 86.8±6.5| 86.8±6.1|
|                  | 7≤FS<10 | 10 | 95.4±1.1    | 90.7±3.9| 90.4±3.8| 90.9±3.9|

The combined impact of the location and the field size was reflected in table 3 for PTV D95 and Figure 1, in this case showing the comparison for the PTV mean dose. AAA, CCC, and MC algorithms predicted the lowest dose differences (PTV D95~20%, PTV mean dose~15%, and MLD~10%) for the lung-island tumors with small field sizes. The lung-wall and lung-central tumors with large field (tumor) sizes showed relatively lower dose differences (PTV D95~9%, PTV mean dose~5%, and MLD~4%). However, large lung-island tumors showed 15% PTV D95 dose differences.
and 8-10% PTV mean dose differences for AAA, CCC, and MC relative to EPL-1D. Note that the number of patients in 7≤FS<10 category of the island type tumors is much smaller (N=2).

Figure 1. PTV mean dose vs. average field size for 133 lung cancer patients.

Jones et al also reported algorithms using radiological density scaling do not predict the dose near or within inhomogeneities accurately [6]. Large inaccuracies sometimes noted with radiological-path-length-based algorithms have the potential to impact clinical outcomes, as reported in documents, such as the AAPM Task Group Report No. 85 on tissue heterogeneity corrections [17]. This current study provides evidence of the magnitude of the PTV D95, mean dose and normal lung mean dose differences computed with five algorithms for lung tumors of different sizes and locations. Results are highly suggestive that, in addition to the EPL-1D algorithm being inappropriate for SABR-based lung cancer planning [18], the EPL-3D algorithm is also inaccurate and should not be used for SABR.

4. Conclusion

Amongst all locations, dosimetric differences were most strongly correlated with tumor size for lung-island tumors. Significant changes in tumor dose values for 133 lung cancer patients were observed. As the field size becomes small (< 5 cm), the predicted dose to the PTV with pencil beam algorithms showed unrealistic overestimations. In fact, the actual dose delivered is much less than expected based on convolution-type and MC-based calculations. For “island” type small tumors this under-dosage was in the range of 20-25% of the prescribed dose. For most cases, convolution/superposition and MC algorithms were in good agreement. For the most challenging cases, small tumors surrounded entirely by lung tissue, a Monte-Carlo-based algorithm may be warranted [18]. The inability to account for electron transport and lateral electronic dis-equilibrium in the low-density lung medium, result in inaccurate dose distributions using either EPL-1D or EPL-3D dose algorithms. Such algorithms should be avoided for SABR-based dose calculations of patients with lung cancers.

References

[1] Loo B W, Chang J Y, Dawson L A, Kavanagh B D et al 2011 Practical Rad Onc. 1 38–9
[2] McGarry R C, Papiez L, Williams M et al 2005 Int. J Radiat. Oncol. Biol. Phys. 63(4) 1010-15
[3] Timmerman R, McGarry R, Viannouttes C et al. 2006 J of Clin Oncology 24 4833-39
[4] Timmerman R, Paulus R, Galvin J et al 2010 JAMA 303(11) 1070–76
[5] Chetty J, Curran B, Cygler J E et al 2007 AAPM TG 105 Med. Phys. 34(12) 4818-53
[6] Jones A O, Das I J and Jones F L Jr 2003 Med. Phys. 30(3) 296-300
[7] Das I J, Ding G X and Ahnesjö A 2008 Med. Phys. 35(1) 206-215
[8] Ulmer W and Harder D 1995 Med. Phys. 5 25–30
[9] Sievinen J, Ulmers W and Kaissl W 2005 Varian Medical Systems, Palo Alto, CA
[10] Fippel M 1999 Med. Phys. 26 1466-75
[11] Papanikolau N, Mackie T R, Meger-Wells C et al 1993 Med. Phys. 20(5) 1327-36
[12] Kawrakow I, Dippel M and Friedrich K 1996 Med. Phys. 23(2) 445-457
[13] Fippel M 1999 Med. Phys. 26 1466-75
[14] Fragoso M, Wen N, Kumar S et al 2010 Phys. Med. Biol. 55(16) 4445-64
[15] Ahnesjö A and Aspradakis M M 1999 Phys. Med. Biol. 44 R99–R155
[16] Papanikolau N, Battista J J, Boyer A L et al 2004 Medical Physics, Madison, WI 1-135
[17] Benedict S H, Yenice K M, Followill D et al 2010 AAPM TG 101 Med. Phys. 37 4078-4101