Collapsed cone and analytical anisotropic algorithm dose calculations compared to VMC++ Monte Carlo simulations in clinical cases

F. Hasenbalg¹, H. Neuenschwander², R. Mini¹ and E. J. Born¹
¹ Division of Medical Radiation Physics, Insel Hospital, University of Berne, Switzerland
² Clinic of Radio-Oncology, Lindenhof Hospital, Berne, Switzerland
E-mail: hasenbalg@ams.unibe.ch

Abstract. The purpose of this work was to study and quantify the differences in dose distributions computed with some of the newest dose calculation algorithms available in commercial planning systems. The study was done for clinical cases where large density inhomogeneities were present. Three dose algorithms were used: a pencil beam like algorithm, the anisotropic analytic algorithm (AAA), a convolution superposition algorithm, collapsed cone convolution (CCC) and a Monte Carlo program, voxel Monte Carlo (VMC++). The dose calculation algorithms were compared under static field irradiations at 6 MV and 15 MV using multileaf collimators and hard wedges where necessary. Five clinical cases were studied: three lung and two breast cases. We found that the CCC algorithm performed overall better than AAA compared to VMC++, but AAA remains an attractive option for routine use in the clinic due to its short computation times. Dose differences between the different algorithms for the median value of the planning target volume (PTV) were typically 0.4% (range: 0.0 – 1.4%) in the lung and -1.3% (range: -2.1 – -0.6%) in the breast for the few cases we analysed. As expected, PTV coverage and dose homogeneity turned out to be more critical in the lung than in the breast cases with respect to the accuracy of the dose calculation. This was observed in the dose volume histograms obtained from the Monte Carlo simulations.

1. Introduction
Dose calculation algorithms play a crucial role in modern treatment planning systems (TPS). The ICRU recommends the dose to be delivered with an error less than 5% [1] which implies that each step (machine calibration, patient positioning, dose calculation, etc.) needs to be performed to an accuracy far better than 5%. The necessary accuracy for the dose calculation step should be on the order of 2 – 3% [2]. Semi-analytical algorithms for dose calculations of photon beams like pencil beam convolution, anisotropic analytic algorithm or superposition/convolution are known for their limited accuracy under certain circumstances due to several approximations used. This is particularly evident when media with large density and composition variations (large inhomogeneities) are considered and several studies on this topic have been done in the past (see [3, 4, 5, 6, 7] and references therein; for a review of dose calculations in radiotherapy see [8]). It is expected that Monte Carlo (MC) dose calculations will not show this type of behaviour when enough particles are simulated.

The aim of this study was to compare the latest commercially available dose calculation algorithms, anisotropic analytical algorithm (AAA) and collapsed cone convolution (CCC),...
against Monte Carlo simulations in clinical situations.

2. Materials and methods

The anisotropic analytical algorithm [9] is available in the Eclipse TPS (v. 7.5.22.0, Varian Medical Systems) and it is not clear from the available literature whether it computes dose to water or dose to the medium. The collapsed cone convolution [10] is available in the MasterPlan TPS (v. 1.5, Nucletron) and calculates dose to the medium [11]. Voxel Monte Carlo (VMC++) [12] (National Research Council of Canada) calculates dose to the medium and dose to water. For all the cases studied here, we found that the differences in the MC simulations between dose to water and dose to the medium [13] were negligible in the regions of interest of this work. Therefore, in the rest of this paper we neglect this distinction.

The dose calculation algorithms AAA and VMC++ were configured using commissioning data available at our institution. Commissioning of CCC is done by the vendor based on data provided by the user and does not require to be configured. The measured data was obtained with semiconductor detectors (percentage depth doses and profiles) and with ionisation chambers (output factors) measured in a water phantom. All measurements were done at a source to phantom distance SSD = 90 cm and a source to axis distance SAD = 100 cm. The reference field for output factor measurements was always 10x10 cm$^2$. The grid resolution used for all calculations was 2.5 mm.

2.1. AAA configuration

AAA is a pencil beam convolution/superposition algorithm which uses a multiple–source model to represent clinical beam properties and a patient scatter model represented by density scalable poly–energetic kernels. After configuration, AAA uses only analytical functions which makes an analytical convolution possible and therefore reduces the computational time considerably.

![Measurement vs. AAA](attachment:image1.png)

**Figure 1.** 15 MV percentage depth dose curves (left) and profiles at z = 10 cm (right) for field sizes of 4x4, 10x10, 15x15 and 20x20 cm$^2$. Depth dose curves are normalised to 100% at z = 10 cm and profiles to the corresponding output factors.

The version 7.5.22.0 of AAA comes with a small database of parameters for the different source components for several (mostly Varian) accelerators. Using these default parameters for our Varian 2300 C/D Linac, a very good agreement ($\leq 2\%$) between measurements and calculations was obtained as can be seen for example in the depth dose curves and profiles, at a depth of $z = 10$ cm, for 15 MV for the field sizes of 4x4, 10x10, 15x15 and 20x20 cm$^2$ of figure 1.
2.2. VMC++ configuration

The VMC++ version we used in this study performs a complete simulation of the accelerator head. Our implementation consisted in specifying the correct geometry and material composition of a Varian 2300C/D linear accelerator available at the Insel Hospital in Berne. No essential changes needed to be made for the 6 MV photon beam with respect to the vendor specifications. For 15 MV photons however, we reduced the mass density of the flattening filter, made of tungsten, from 19.3 g/cm$^3$ to 18 g/cm$^3$ to accurately reproduce the measured dose profiles.

Sheik–Bagheri and Rogers [14] have shown that the most sensitive parameters to properly configure a MC beam are the incident electron mean energy, $E$, and the width of the electron beam, $\sigma$, parameterised by a gaussian distribution. Our configuration procedure consisted in running a set of simulations of the treatment head and water phantom varying both parameters, $E$ and $\sigma$ and comparing the dose distributions obtained with measured depth dose curves and symmetrised profiles. Typically, values for the mean energy of 5.5, 6.0, 6.5 MeV for the 6 MV configuration (14.5, 15.0 and 15.5 MeV for the 15 MV configuration) and $\sigma$ values of 0.25, 0.5, 0.75, 1 and 1.25 mm were used in the simulations. The most sensitive cases to variations in $E$ and $\sigma$ were the profiles of the largest measured fields (40x40 cm$^2$) at the depth of dose maximum. The set of parameters giving the smallest differences between measurements and simulations were sorted out and the agreement to the remaining profiles at 5, 10, 20, 30 cm checked (meaning that the percentage differences between simulations and measurements should be between ±2% of the maximum dose). This same procedure was then repeated for the 10x10 cm$^2$ field size and a compromise was reached between the best values obtained for the 40x40 cm$^2$ and the 10x10 cm$^2$ cases. Generally speaking, we selected those values of $E$ and $\sigma$ giving the best match to the 10x10 cm$^2$ data (given its clinical relevance) and still providing an acceptable match to the 40x40 cm$^2$ data. Figure 2 shows the measured data compared to the simulations for 15 MV photons and field sizes of 4x4, 10x10, 15x15 and 20x20 cm$^2$. Similar results were also obtained for the configuration of the 6 MV beam. For 6 MV the selected parameters were $E = 6.0$ and $\sigma = 0.5$ mm (FWHM = 1.175mm) and for 15 MV, $E = 15.0$ and $\sigma = 0.5$ mm.

2.3. Clinical cases

A series of clinical cases was selected from our patient database. Since the focus of this work was to study the main differences of the dose calculation algorithms and Monte Carlo, we selected...
cases with large inhomogeneities, namely lung and breast. Since planning itself was not an issue of this study, we do not judge or make any comments on the quality of the plans. Neither do we make an assessment of the clinical impact of the dose differences.

For each clinical case, a base plan was generated with Helax TMS 6.1B TPS using pencil beam convolution [15]. This was the actual plan used to treat the patients. The anonymised CT images and structures, such as planning target volume and organs at risk (OAR), were exported to MasterPlan TPS for CCC calculations and to Eclipse TPS for AAA calculations. In addition, Eclipse TPS allowed the export of plans to the VMC++ algorithm running on a remote workstation. Special care was taken to have exactly the same plan in all calculation algorithms. The results of all calculations were imported into Eclipse where dose volume histograms (DVHs, differential and cumulative) were computed and isodose lines compared using the “Plan Evaluation” workspace. All computations including the VMC++ simulations were done with an X–Y grid spacing of 2.5 mm and CT slices 10 mm and 5 mm apart for the lung and breast cases, respectively. Details of the plans, the volumes considered and CPU times (VMC++ and CCC: 2.4 GHz Athlon and Pentium CPUs, AAA: 3.2 GHz Pentium CPU) involved are given in table 1. The CPU times for VMC++ are only typical and not optimised for efficiency. Because in this study VMC++ is used as a reference for the comparisons, VMC++ results were calculated to a very high precision (dose uncertainty of less than 5% in those voxels having a dose greater than 50% of $D_{\text{max}}$). All plans were isocentric and the PTVs and OARs drawn by physicians.

| Clinical case | Beam quality, (Wedge angle) (MV, °) | PTV Volume (cm$^3$) | OAR Volume (cm$^3$) | $N_e$/field | VMC++ Prec. (%) | CPU (min) | CCC CPU (min) | AAA CPU (min) |
|---------------|------------------------------------|---------------------|--------------------|------------|----------------|----------|--------------|--------------|
| L1            | 15 (60), 15 (60), 15 (15)          | 204                 | 1400               | $10^7$     | 0.5            | 114      | 12.5         | 1.1          |
| L2            | 6 (15), 15                        | 141                 | 1316               | $10^7$     | 0.4            | 72       | 3.5          | 0.7          |
| L3            | 15, 6                             | 344                 | 996                | $10^7$     | 0.4            | 60       | 3.5          | 0.6          |
| B1            | 6 (30), 6 (30)                     | 513                 | 1175               | $4 \times 10^7$ | 0.5        | 156      | 3.3          | 0.9          |
| B2            | 6 (30), 6 (30)                     | 383                 | 1444               | $4 \times 10^7$ | 0.5        | 168      | 2.8          | 0.7          |

Two dose normalisation methods were used: in the first one, which we called fixed monitor units, we used the same number of monitor units for each field as the original pencil beam plan. This modality, exhibits more clearly the actual differences between the several algorithms and does a fair comparison between them, because the beam intensities are not modified. At a given point, like the prescription point, however, this method results in a different dose for each plan. Therefore, a second normalisation method was considered. In this latter normalisation method, which we called dose contribution, the intensities of each field are varied to obtain a prescribed dose at the prescription point. This procedure is done automatically in Eclipse TPS and in our
Figure 3. Isodose lines at the isocentre plane calculated with PBC (left) and VMC++ (right) for lung case L1. The PTV (white line) is shown in the left figure. Most compromised OAR is the left lung.

Figure 4. Isodose lines at the isocentre plane calculated with CCC (left) and AAA (right) for lung case L1.

In our study, all plans normalised according to this method were normalised to 1 Gy at the isocentre. In the dose contribution method the number of MUs for each beam is not the same for the different algorithms. In our opinion, this second approach better represents what a planner would probably do, having at his disposal more accurate dose distributions. Therefore in our study, both normalisation schemes are used to compute the DVHs and exhibit their differences. In our implementation, VMC++ can not compute the MUs for each field (since backscattering into the monitor chamber is not modelled), therefore the VMC++ relative dose normalisation was done adjusting the weights of each field to the same dose contribution per field at the isocentre as the base plan. This guaranteed a 100% dose at the isocentre in this scheme. For the fixed monitor units normalisation, since the dose distributions of CCC were found to be very
similar to that of VMC++ (see left figures 5 and 9), we required the same dose contribution per field for VMC++ as with CCC at the isocentre. This produced doses different from 100% at the isocentre. It is clear that this is an arbitrary procedure and introduces a bias towards CCC when comparing AAA and CCC with VMC++. This is however not the case when using the dose contribution normalisation.

3. Results and discussion

3.1. Lung cases

The three lung cases selected correspond to: a three field boost (L1) of a lung lesion centrally located in the left lung near the tracheal bifurcation, a typical opposed anterior–posterior field plan (L2) of a lesion in the right upper lobe with infiltration of the thoracic wall and a tangential beam arrangement (L3) of a lesion located in the lung right lower lobe infiltrating the thoracic wall.

Figure 5. PTV differential DVH using fixed monitor units normalisation (left) and dose contribution normalisation (right) for lung case L1.

Figure 6. DVH for the PTV and left lung using fixed monitor units normalisation (left) and dose contribution normalisation (right) for lung case L1.

For illustration purposes, the results of PBC, VMC++, CCC and AAA are shown for the lung case L1 only. Figure 3 (left) shows the PBC dose distribution, the PTV (white line) and the
most compromised OAR, the left lung, at the isocentre plane. Figure 3 (left) was calculated with PBC of MasterPlan including inhomogeneity corrections. The dose distributions of figures 3 – 4 were all normalised to 1 Gy at the isocentre (dose contribution normalisation). As expected, figures 3 – 4 show that the isodose lines obtained with VMC++, CCC and the AAA extend beyond the beam path in the low density regions due to the range of secondary electrons. Small differences between these algorithms can already be seen in these figures and, in spite of the presence of large inhomogeneities, AAA and CCC agree in this clinical case reasonably well (within ± 5% of D_{max}) with the MC simulations.

Nevertheless, the dose distributions for the PTV and OAR structures exhibit differences between the dose algorithms, and this can be appreciated in the differential and cumulative DVHs of figures 5 – 6. Table 2 summarises the results obtained for the three lung cases using several dose volume indices for the PTV such as: median, mean, the minimum dose received by 95% of the volume (D_{95}), the maximum dose received by 5% of the volume (D_{5}) and a measure of the variation of the dose in the PTV, (D_{5} – D_{95}). Since we are not evaluating the quality of the plans nor its clinical impact, no V_{20} or mean lung dose were computed.

Table 2. PTV dose volume indices (median, mean, D_{95}, D_{5} and D_{5}–D_{95}) for the lung cases and three algorithms in the two normalisation schemes used. 100% represents 1 Gy.

| dose volume index | Lung case L1 | Lung case L2 | Lung case L3 |
|-------------------|--------------|--------------|--------------|
|                   | VMC          | CCC          | AAA          | VMC          | CCC          | AAA          |
| fixed monitor units |              |              |              |
| PTV median        | 95.4         | 95.4         | 96.7         | 97.3         | 97.3         | 97.3         | 99.6         | 100.0        | 100.4 |
| PTV mean          | 94.3         | 94.4         | 96.6         | 96.9         | 96.5         | 97.1         | 99.2         | 99.8         | 100.4 |
| PTV D_{95}        | 82.7         | 83.2         | 88.0         | 88.4         | 85.5         | 90.6         | 93.1         | 94.6         | 96.0 |
| PTV D_{5}         | 101.9        | 102.6        | 104.5        | 102.9        | 102.5        | 102.7        | 104.3        | 104.5        | 104.6 |
| PTV D_{5}–D_{95}  | 19.2         | 19.4         | 16.5         | 14.5         | 17.0         | 12.1         | 11.2         | 9.9          | 8.6  |
| dose contribution |              |              |              |
| PTV median        | 97.0         | 97.1         | 96.7         | 97.5         | 97.5         | 97.7         | 101.4        | 101.9        | 101.8 |
| PTV mean          | 95.8         | 95.9         | 96.6         | 97.1         | 96.7         | 97.6         | 101.0        | 101.7        | 101.7 |
| PTV D_{95}        | 84.1         | 84.6         | 88.0         | 88.6         | 85.7         | 91.1         | 94.8         | 96.4         | 97.2 |
| PTV D_{5}         | 103.5        | 104.1        | 104.3        | 103.1        | 102.8        | 103.3        | 106.2        | 106.4        | 106.0 |
| PTV D_{5}–D_{95}  | 19.4         | 19.5         | 16.3         | 14.5         | 17.1         | 12.2         | 11.4         | 10.0         | 8.8  |

In spite of the overall good agreement, differences in the dose distributions of the PTV between AAA, CCC and VMC++ are observed, (see e.g. figure 5 for lung case L1) showing that when the fixed monitor units normalisation is used the medians of the dose histograms can differ up to 1.4% and the low dose regions can differ considerably. Considerable differences are also observed in the low dose regions. While AAA shows significant differences compared to VMC++ in the low dose regions of the PTV, CCC and VMC++ have similar dose volume histograms. AAA also produces a somewhat narrower differential DVH than CCC and VMC++. Using the dose contribution normalisation, the maximum differences in median between AAA and VMC++ change from 1.4% to -0.3% (lung case L1), 0.0% to 0.2% (lung case L2) and 0.8% to 0.4% (lung case L3). One should also note, that the variation of the dose in the PTV, measured by D_{5}–D_{95} tends to be larger for VMC++ and CCC than for AAA in all cases (see table 2). This is also true for the PBC algorithm, which exhibits an even narrower differential histogram for the PTV than AAA calculations. This, quite generally, implies that pencil beam
like algorithms like PBC and AAA tend to give the wrong impression that a good PTV coverage has been achieved when in reality this is not the case.

Figure 7. Isodose lines at the isocentre plane calculated with PBC (left) and VMC++ (right) for breast case B2. The PTV (white line) is shown in the left figure. Most compromised OAR is the right lung.

Figure 8. Isodose lines at the isocentre plane calculated with CCC (left) and AAA (right) for breast case B2.

3.2. Breast cases

We selected two cases for the breast: a typical breast case (B1) treated with opposed tangential fields without axillary and supraclavicular involvement and a second similar but somewhat meagre case (B2). Therefore, the ratio of volumes PTV/OAR (see table 1) is larger for the first case than for the second one. As for the lung, the plans were isocentric and the structures
drawn by physicians. Figures 7 – 8 show the dose distributions obtained with PBC, VMC++, CCC and AAA at the isocentre plane, the PTV and the most compromised OAR, the right lung, for breast case B2. All plans were normalised to 1 Gy at the isocentre (dose contribution normalisation).

As can be seen in figure 9, breast cases have the tendency to produce narrower differential distributions than the lung cases, because less lung tissue is irradiated. Again, the dose contribution normalisation reduces the spread in the mean values of the distributions. VMC++ and CCC exhibit distributions with larger spreads than AAA.

![Figure 9. PTV differential DVH using fixed monitor units normalisation (left) and dose contribution normalisation (right) for breast case B2.](image)

Table 3 summarises the results obtained for the breast cases in terms of dose volume indices for the PTV. Using the dose contribution normalisation, the maximum differences in median between the distributions of VMC++ and AAA was reduced from -2.0% (fixed monitor units) to -0.0% (dose normalisation) for breast case B1 and from -2.1% to -1.0% for breast case B2. In both breast cases CCC gives the largest dose variations in the PTV (D$_{5-95}$) which do not correspond to the VMC++ results and implies some modelling problems for these particular cases.

![Figure 10. DVH for the PTV and right lung using fixed monitor units normalisation (left) and dose contribution normalisation (right) for breast case B2.](image)
cases. Notice however that these dose variations are almost a factor 2 smaller than for the lung cases.

Table 3. PTV dose volume indices (median, mean, \(D_{95}\), \(D_5\) and \(D_5 - D_{95}\)) for the breast cases and three algorithms in the two normalisation schemes used. 100% represents 1 Gy.

| dose volume index | Breast case B1 | Breast case B2 |
|-------------------|----------------|----------------|
| \(PTV_{\text{median}}\) | 101.0 | 100.3 | 99.0 | 100.3 | 99.7 | 98.2 |
| \(PTV_{\text{mean}}\) | 100.8 | 100.0 | 98.7 | 100.1 | 99.1 | 97.7 |
| \(PTV_{D_{95}}\) | 96.3 | 94.4 | 93.3 | 94.9 | 92.8 | 92.2 |
| \(PTV_{D_5}\) | 104.5 | 104.1 | 102.4 | 103.3 | 102.5 | 100.8 |
| \(PTV_{D_5 - D_{95}}\) | 8.2 | 9.7 | 9.1 | 8.4 | 9.7 | 8.6 |

4. Conclusions
Quite generally, the differences in clinical situations between the three dose algorithms studied here are small (within 5% of \(D_{\text{max}}\)) and this agrees with a previous study [7]. Small differences between the algorithms are evident however, in cases with large inhomogeneities such us lung or breast with partial lung irradiation. In our comparative study, we observed that CCC performs rather well compared to MC simulations which we assume to be the best representation of the real dose distributions under the limitations of the present study. The analytical anisotropic algorithm does a reasonable job and its short CPU times make it very attractive for routine use in the clinic.

For a fixed monitor unit normalisation, which represents a “fair” comparison between the algorithms, we found dose discrepancies for the median value of the PTV of typically 0.4% (range: 0.0 – 1.4%) in the lung and -1.3% (range: -2.1 – -0.6%) in the breast for the few cases we analysed. These differences are strongly dependent on individual patient properties such as tumour location, PTV size, beam arrangement, etc. Monte Carlo differential and cumulative DVHs show that for the PTV the dose spreads over a larger range than previously assumed with simple pencil beam type algorithms. Therefore, for the cases here considered, underdose regions occur more frequently than previously thought. Another noticeable difference in the cases we studied, is that, due to the more irregular shapes and localisation sites of the PTVs, lung cases are more critical than breast cases with respect to the accuracy of the dose calculation. This can be seen in the less regular MC differential DVHs of the lung cases compared to the breast cases and in the higher values of \(D_5 - D_{95}\) for the PTV.

Our work clearly shows that the actual practice of relying solely on PBC for dose calculation, particularly for lung cases, may be inappropriate. How the physicians will deal with these facts and which implications this will have for the assessment of radiotherapy treatment outcome remains to be established in future studies.
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