Dose calculation and reporting with a linear Boltzman transport equation solver in vertebral SABR

Nicholas Hardcastle1,2,3 · Jeremy Hughes1 · Shankar Siva1,2 · Tomas Kron1,2,3

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Abstract
Vertebral Stereotactic ablative body radiotherapy (SABR) involves substantial tumour density heterogeneities. We evaluated the impact of a linear Boltzmann transport equation (LBTE) solver dose calculation on vertebral SABR dose distributions. A sequential cohort of 20 patients with vertebral metastases treated with SABR were selected. Treatment plans were initially planned with a convolution style dose calculation algorithm. The plan was copied and recalculated with a LBTE algorithm reporting both dose to water ($D_w$) or dose to medium ($D_m$). Target dose as a function of CT number, and spinal cord dose was compared between algorithms. Compared with a convolution algorithm, there was minimal change in PTV D90% with LBTE. LBTE reporting $D_m$ resulted in reduced GTV D50% by (mean, 95% CI) 2.2% (1.9–2.6%) and reduced Spinal Cord PRV near-maximum dose by 3.0% (2.0–4.1%). LBTE reporting $D_w$ resulted in increased GTV D50% by 2.4% (1.8–3.0%). GTV D50% decreased or increased with increasing CT number with $D_m$ or $D_w$ respectively. LBTE, reporting either $D_m$ or $D_w$ resulted in decreased central spinal cord dose by 8.7% (7.1–10.2%) and 7.2% (5.7–8.8%) respectively. Reported vertebral SABR tumour dose when calculating with an LBTE algorithm depends on tumour density. Spinal cord near-maximum dose was lower when using LBTE algorithm reporting $D_m$, which may result in higher spinal cord doses being delivered than with a convolution style algorithm. Spinal cord central dose was significantly lower with LBTE, potentially reflecting LBTE transport approximations.

Keywords SABR · Vertebra · Spine · Linear Boltzman transport equation · Dosimetry

Introduction
Stereotactic ablative body radiotherapy is becoming standard of care for local treatment of amenable metastases in the oligometastatic paradigm [1–4]. Vertebral metastases are commonly observed in patients with breast and prostate cancer, and can result in significant morbidity if untreated. SABR for vertebral metastases has shown improvement in local control and pain control compared with conventional radiation therapy [2].

Recent improvements in dose calculation algorithms available in commercial treatment planning software such as Monte Carlo methods or linear Boltzmann transport equation (LBTE) solvers, have resulted in improvements in dose calculation accuracy, in particular in regions of heterogeneities that impact on lateral electron spread [5–7]. These algorithms however compute energy deposition in and report dose to the medium of interest [8–10]. This departure from convolution/superposition style algorithms, which report dose to varying densities of water, results in different doses for the same incident fluence due to differences in the electron stopping power between medium and water [8, 10, 11]. This is most dramatically seen in bone, where the electron stopping power difference is up to 10% compared with water [9, 12].

For bone metastases, advanced dose calculation algorithms such as Monte Carlo and LBTE solvers may result in variations in reported dose compared with convolution/superposition algorithms. This may be further impacted by the choice of prescription – coverage of a target that may or
may not be in bone, or prescribing such that target dose coverage is maximised while sitting at organ at risk constraints such as in vertebral SABR. In this technical note, the difference between a convolution/superposition style algorithm and a LBTE solver is determined for vertebral SABR.

**Methods**

Approval to conduct this study was given by the Peter MacCallum Cancer Centre Ethics Committee (17/95R). A retrospective sequential cohort of 20 vertebral SABR cases were selected. The Gross Tumour Volume (GTV) was contoured using a combination of positron emission tomography, magnetic resonance imaging and computed tomography, and the Clinical Target Volume (CTV) was delineated according to the International Spinal Radiosurgery Consensus guidelines [13]. A 2 mm Planning Target Volume (PTV) margin was applied to the CTV. The spinal cord was delineated using MRI, with a 2 mm planning organ-at-risk volume (PRV) margin used for spinal cord dose constraints. The plans delivered either 20 Gy in 1 fraction or 24 Gy in 2 fractions. Vertebral SABR plans were prescribed to maximise the dose to 90% of the PTV whilst meeting the spinal cord PRV and oesophagus near maximum dose constraints. The maximum dose to the GTV was typically boosted to 125% of the prescription dose. These cases were originally planned with either intensity modulated radiation therapy (IMRT) or volumetric modulated arc therapy (VMAT) in the Eclipse treatment planning system (V11.0–13.5) using the Anisotropic Analytic Algorithm (AAA). The plans were all recomputed with AAA v15.5 and a LBTE solver AcurosXB algorithm (v15.5). Previous work has shown agreement between AAA v15.5 and previous versions of AAA (v11.0 and v13.5) [12]. Plans were recomputed retaining the same MLC sequences and monitor units, with the physical material table set to AcurosXB v13.5. Dose calculation was computed at an isotropic resolution of 1.25 mm.

The dose to the targets and spinal cord were evaluated for both algorithms. The median (D50%) and near maximum (D02%) doses to the GTV and dose to 90% (D90%) of the PTV were evaluated. The dose to 0.03 cm$^3$ of the spinal cord PRV was compared between algorithms. The mean dose to a 3 mm diameter circle in the middle of the spinal cord was also evaluated, as a surrogate for the out of field doses and that which may represent a measurement point in quality assurance measurements. All dose metrics were compared using the Wilcoxon signed rank test with a threshold for statistical significance of 0.05. The relationship between GTV Hounsfield Units (HU) (as a surrogate for if tumour is blastic, lytic or a mix of both) and difference in GTV D50% between algorithms was assessed using the Pearson correlation coefficient.

**Results**

Figure 1 shows the dose distribution for the three calculation modes for a lytic and a blastic lesion. The profiles through the anterior–posterior direction indicated by the white line on Fig. 1 are shown in Fig. 2. In regions of tissue or low density bone, there are minimal differences between the AAA and AcurosXB algorithms. The dose to 0.03 cm$^3$ of the spinal cord PRV shows minimal differences between algorithms. The mean dose to a 3 mm diameter circle in the middle of the spinal cord was also evaluated, as a surrogate for the out of field doses and that which may represent a measurement point in quality assurance measurements. All dose metrics were compared using the Wilcoxon signed rank test with a threshold for statistical significance of 0.05. The relationship between GTV Hounsfield Units (HU) (as a surrogate for if tumour is blastic, lytic or a mix of both) and difference in GTV D50% between algorithms was assessed using the Pearson correlation coefficient.

![Fig. 1 Dose distributions for a lytic lesion (top) and blastic lesion for AAA, AXB $D_m$ and AXB $D_w$ algorithms. The white vertical line on the CT image shows the location of the dose profile in Fig. 2. The segmentations displayed are the GTV (red), PTV (vertebral body, cyan and green), and spinal cord PRV](image_url)
between algorithms. The exception to this is in the region in the middle of the spinal cord, where both AXB \( D_m \) and AXB \( D_w \) are lower than AAA. In regions of increased HUs, such as increased density bone, AXB \( D_m \) is decreased relative to AAA, and AXB \( D_w \) is increased relative to AAA. Figure 1a shows a lytic GTV, with the main differences observed in higher density components of the lytic GTV with AXB \( D_m \). Figure 1b shows a blastic GTV, with AXB \( D_m \) lower than AAA by approximately 5%, and AXB \( D_w \) approximately 6% higher than AAA in the GTV.

The change in GTV dose as a function of HU is shown in Fig. 3. As the mean HU of the GTV increases, the median GTV dose calculated with AXB \( D_m \) is increasingly lower than that calculated with AAA, by up to 3.5%. Conversely, as the mean HU of the GTV increases, the median GTV dose calculated with AXB \( D_w \) increases compared with AAA by up to 6%. When comparing AXB \( D_m \) with AXB \( D_w \), up to 8.7% higher GTV D50% is calculated with AXB \( D_m \) compared with AXB \( D_w \).

Figure 4 shows the differences between AAA and AXB reporting dose to medium or water. This data is summarised in Table 1. AXB \( D_m \) resulted in slight decreases in PTV D90%, GTV D50% and D02% compared with AAA. AXB \( D_w \) resulted in slight increases in PTV D90%, but larger increases in GTV D50% and D02% associated with HU differences, as shown in Fig. 3. The shift in opposite directions relative to AAA is reflected in AXB \( D_m \) compared with AXB \( D_w \), where the largest differences in target doses were observed.

The SpinalCord PRV D0.03 cc was on average 3% (0.4 Gy) lower when calculated with AXB \( D_m \) compared with AAA, however there was no statistically significant difference in this metric when calculated with AXB \( D_w \) compared with AAA. On average, the SpinalCord PRV D0.03 cc was 2.8% (0.4 Gy) higher when computed with AXB \( D_w \) compared with AXB \( D_m \). The dose to the centre of the SpinalCord was significantly lower when computed with AXB \( D_m \) and AXB \( D_w \) when compared with AAA (average 8.7%/0.6 Gy and 7.2%/0.5 Gy respectively). Minimal differences were observed for SpinalCord central dose between the two reporting modes for AXB.

**Discussion**

Vertebral SABR presents a challenging dose calculation scenario due to heterogeneities in material, high dose gradients and small fields. As a result, dose calculation with Monte Carlo or LBTE solving algorithms is an attractive option due to its improved dose calculation accuracy. Compared with Type B algorithms however, LBTE algorithm presents a number of key differences which may impact reporting of dose distributions in vertebral SABR.

Reporting dose to medium or water results in substantial differences in reported GTV dose, which depends on the density of the bone in the target. Lytic lesions exhibit minimal difference between AAA and either reporting mode for AXB; however increased HU in blastic lesions results in GTV D50% differences of up to 3.5% with \( D_m \) and 6% with \( D_w \). Differences between AXB \( D_m \) and AXB \( D_w \) are more

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Fig. 2 Dose profiles along the white line in Fig. 1 for the lytic lesion (top) and blastic lesion (bottom). The divergence of the three algorithms from 17 mm in the lytic lesion corresponds to the high density bone around the edge of the vertebral body.

Fig. 3 Difference in GTV median dose between AAA and AXB \( D_m \), AAA and AXB \( D_w \), and AXB \( D_m \) and AXB \( D_w \), as a function of the mean CT number of the GTV. The Pearson correlation coefficient and p value is provided for each series.
significant; GTV D50% was between 2–9% higher when calculating with AXB $D_w$ compared with AXB $D_m$. This is a result of the increased difference in stopping power between bone and water, which is increasing with density of the bone. These differences translate into small reporting differences in PTV coverage metric. Ma et. al. demonstrated in Monte Carlo for a single vertebral lesion a difference between $D_m$ and $D_w$ of 3.5% for soft bone and 11% for compact bone [8]. Zhen et. al. compared AAA and AXB $D_m$ (v10.0.28), and showed increased dose in the vertebral body GTV and PTV with AXB $D_m$ compared with AAA by 1–2%, in opposition to our finding [5]. The cause of this discrepancy is unknown, but may be related to the different versions of the material density table. In v10, the mass density at which voxels were assigned to bone was 1.475 g cm$^{-3}$ compared with 1.100 g cm$^{-3}$ from v11 onwards and in the version used in the current study. Nevertheless, this is still an unexpected result, since dose to voxels assigned to skeletal muscle and cartilage (a significant proportion of the voxels in the vertebral body) should still result in a lower dose with AXB $D_w$ compared with AAA. Usmani et. al. compared Monte Carlo $D_w$ with $D_m$ for IMRT vertebral plans in iPlan [14]. Increases in dose in high density bone were observed with $D_w$ compared with $D_m$. The CTV mean dose was increased by approximately 5% with $D_w$ compared with $D_m$ a similar magnitude to that observed in our study.

The spinal cord PRV near maximum dose is a hard constraint in vertebral SABR planning. During optimisation, the spinal cord PRV is typically taken to the constraint, and the PTV coverage is maximised while meeting this constraint. When using AXB $D_m$, the spinal cord PRV near maximum dose was on average 3% lower than that with AAA. There was no systematic difference however when using AXB $D_w$.

Zifodya et. al. [16] compared AXB (v11.0.3.1) with AAA. The near maximum spinal cord dose was on average 1% higher with $D_w$ compared with $D_m$ [14]. Head and neck cancer radiotherapy presents a similar geometry, with the target volume wrapping around the spinal cord. Munoz-Montplet et. al. [15] compared AXB (v13.0.26) with AAA for 140 head and neck cancer treatment plans. The spinal cord dose D0.2% was on average 1.07 Gy lower with AXB $D_m$ compared with AAA, and 0.18 Gy lower with AXB $D_w$ compared with AAA.
for head and neck plans; AXB $D_m$ resulted in systematically lower mean spinal cord dose and near maximum spinal cord dose compared with AAA, however AXB $D_w$ did not show the same trend, and was much closer to AAA.

The dose at the centre of the spinal cord was significantly lower than AAA when calculated with AXB, reporting either $D_m$ or $D_w$ As demonstrated by Hughes et. al., this is likely a result of energy cut-off used in dose calculation [17]. The AXB algorithm uses 200 keV for electrons and 1 keV for photons. The spinal cord is predominantly in the gradient region of the apertures, therefore selection of the electron cut off energy will have a large impact on the dose in this region. Electron energy is deposited proximal to the field edge, and as a result the dose calculated further from the field edge, in the centre of the cord, is reduced [17]. The selection of energy cut-off may also be the cause of lower near maximum cord doses, due to a resultant sharper dose gradient with AXB. Given the serial nature of the spinal cord with respect to toxicity, the dose at the centre of the spinal cord may be of limited clinical importance. In treatment plan verification measurements however, the dose in the centre of the spinal cord is often measured [17]. Since the more clinically relevant measurement is at the edge of the spinal cord, measurement at the centre of the cord must be acknowledged, but may have limited utility in detecting deliverability of a treatment plan in the context of this systematic discrepancy.

A limitation of this study is that we do not determine the accuracy of AXB reporting $D_m$ or $D_w$ through comparison with measurement. Measurement of dose to bone is challenging due to the use of detectors whose response is calibrated in water. Shaw et. al determined a medium dependent correction factor to apply to microDiamond and radiochromic film dosimeters in a dosimetric audit for vertebral SABR [18]. Notwithstanding the inaccuracies in penumbral dose described above, this work demonstrated high dose calculation accuracy in bone with AXB $D_m$, however AXB $D_w$ was not assessed in this work due to this reporting mode not being used by any audit participants.

The differences in target dosimetry and spinal cord doses we have reported here have implications for multi-centre clinical trials, where there may exist variation between reporting modes between each participating centre. When dose to bone is related to the trial endpoint, such as when the tumour is in bone, or when an expected toxicity is in bone, it is advised that differences in reporting modes are taken into account by trial management committees according to published recommendations [11, 19]. Due to the variability in the difference with tumour density, this is an uncertainty which must be acknowledged when linking bone tumour dose with clinical outcomes. With respect to the dose to spinal cord, we have observed an average difference in reporting the near maximum dose of up to 3%. When an individual clinic is aiming to replicate treatment planning objectives and dose constraints, it is advisable that these are reviewed with respect to the dose calculation algorithms used to derive such objectives and constraints, and what algorithms and reporting modes are available locally. Further, in the case of re-irradiation with vertebral SABR, the uncertainty in dose calculation, and potential differences in dose calculation reporting mode between courses should be considered.

**Conclusion**

Given high dose gradients, and planning process in which organs at risk are taken to tolerance in each plan, dose calculation accuracy in vertebral SABR doses is of critical importance. Use of a LBTE solver algorithm has the potential to improve dose calculation accuracy, however users must be aware of fundamental differences between reporting $D_m$ or $D_w$, and between LBTE solver and convolution/superposition algorithms. Although PTV coverage metrics are minimally impacted, differences in reported GTV doses depend on the density of the tumour, namely the blastic or lytic nature of the tumour. The near maximum dose to spinal cord depends on the reporting medium. Further differences in spinal cord near maximum and central dose arise from limitations in transport modelling in the LBTE solver; caution must be exercised when interpreting near maximum doses and central doses when using AXB $D_m$ or $D_w$.

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**Availability of data and material** Will be available upon reasonable request to the authors.

**Declarations**

**Conflict of interest** Nicholas Hardcastle, Tomas Kron and Shankar Siva receive funding for an unrelated project from Varian Medical Systems.

**Ethical approval** This study received ethics approval from the Peter MacCallum Cancer Centre.

**Informed consent** Informed consent was waived for this study.

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