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IGRT kV-CBCT dose calculations using Virtual Source Models and validated in phantoms using OSL

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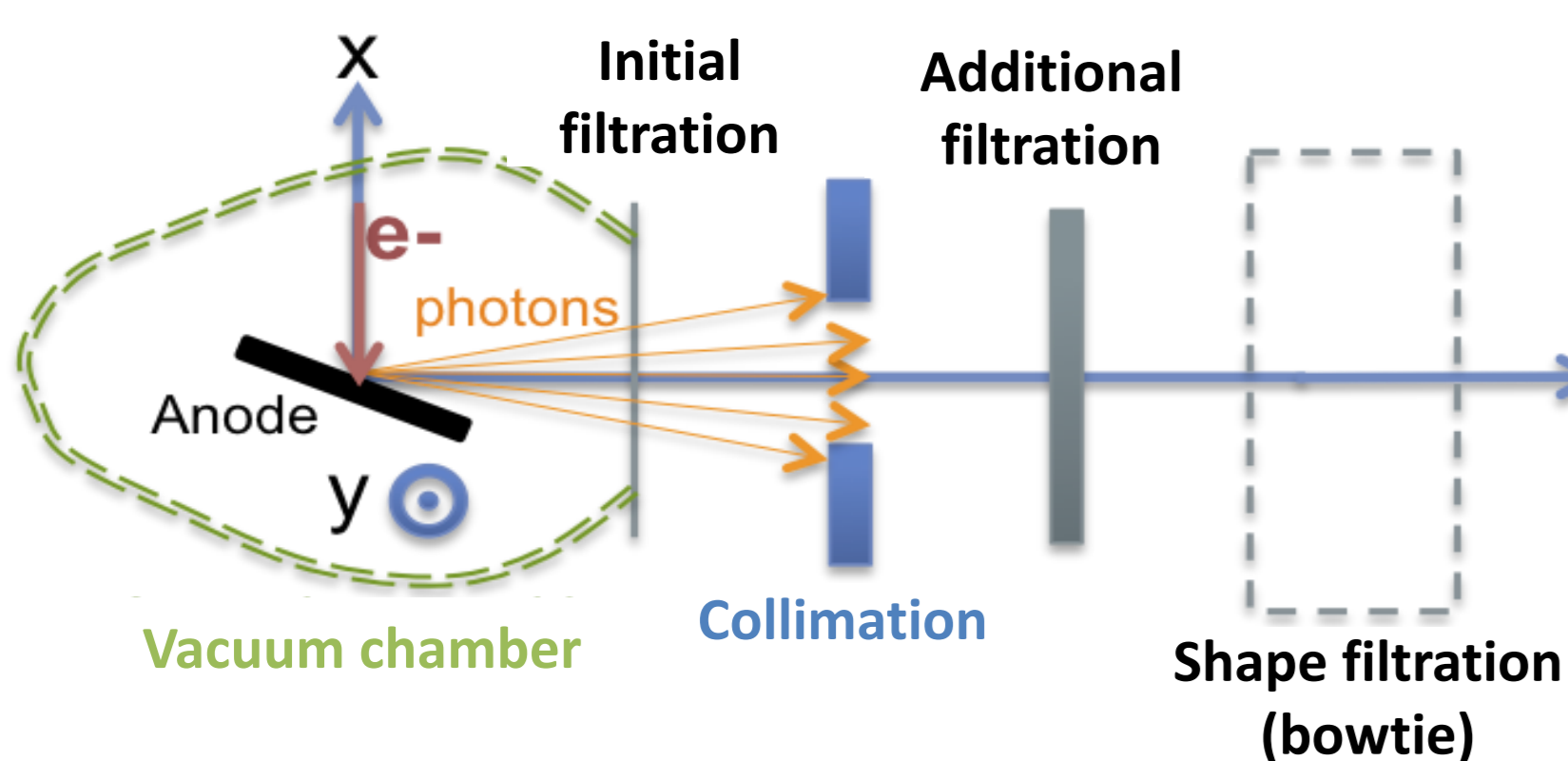
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With the growing use of X-ray imaging equipment in Image-Guided Radiation Therapy (IGRT), the need to evaluate the dose-to-organs delivered by kV-CBCT imaging acquisition increases. This study aims to propose accurate Monte Carlo (MC) calculations of the patient dose-to-organs delivered by two commercially available kV-CBCT systems: the XVI embedded on Elekta VERSA HD linac and the OBI embedded on Varian TrueBeam linac. Simulations were validated using OSL measurements, with dosimeters inserted within different organs in anthropomorphic phantoms.

Energy fluency representation of a X-ray tube

Both XVI and OBI systems have similar X-ray tubes and their energy fluency Φ [1] can therefore be represented in a similar manner.



- Hypothesis 1: Angular diffusion can be neglected**
 \Rightarrow the direction vector is collinear with the direction vector therefore, at a given z position, the fluency depends only on x, y and E .

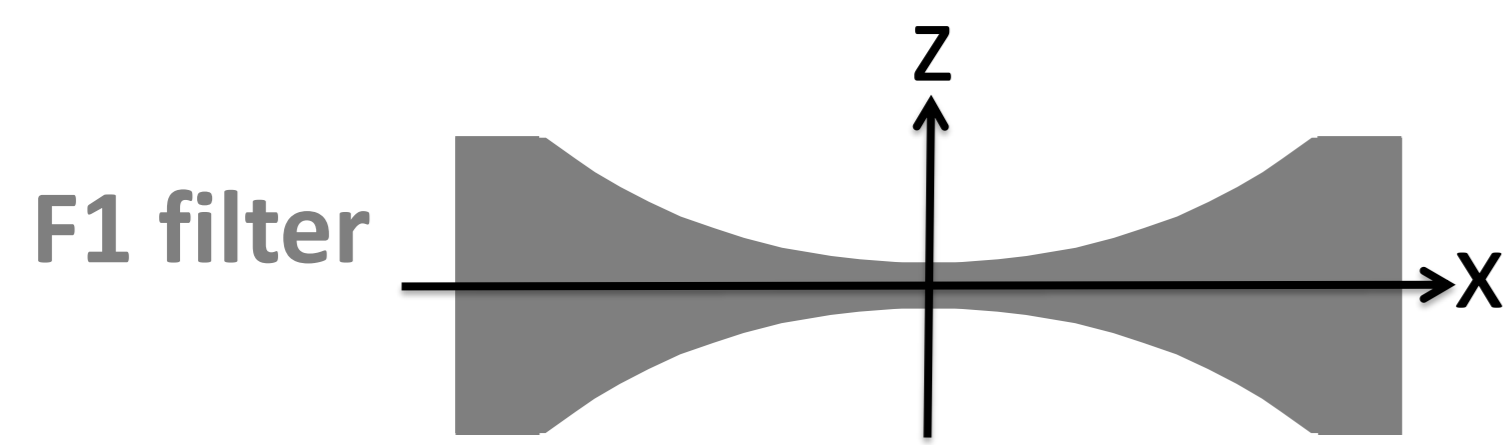
$$\phi(x, y, E) = \phi_0 \cdot p_{x,y}(x, y) \cdot p_{E|x,y}(x, y, E)$$
- Hypothesis 2: Spatial variables are separable (rectangular collimation)**

$$\phi(x, y, E) = \phi_0 \cdot p_x(x) \cdot p_y(y) \cdot p_{E|x,y}(x, y, E)$$
- Hypothesis 3: Before the bowtie filter, the energy spectrum depends only on the x-axis (heel effect from the anode)**

$$\phi(x, y, E) = \phi_0 \cdot p_x(x) \cdot p_y(y) \cdot p_{E|x}(x, E)$$

As the bowtie filter is oriented differently in the two systems, the energy fluence function differs when such filters are used.

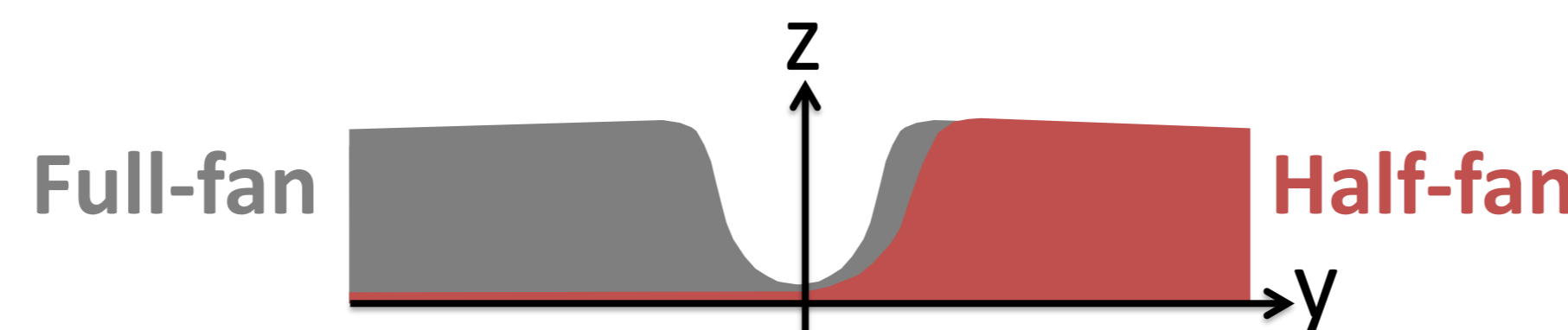
XVI (from Elekta's VERSA HD)
The bowtie thickness is x-dependent



\Rightarrow The spectrum is **x-dependent and y-independent**

$$\phi(x, y, E) = \phi_0 \cdot p_x(x) \cdot p_y(y) \cdot p_{E|x}(x, E)$$

OBI (from Varian's TrueBeam)
Both bowties have y-dependent thicknesses



\Rightarrow The spectrum is both **x and y-dependent**
 \Rightarrow **Hypothesis 4: The x-dependency being smaller is neglected**

$$\phi(x, y, E) \approx \phi_0 \cdot p_x(x) \cdot p_x(y) \cdot p_{E|y}(y, E)$$

Finally, the energy fluency for both systems is represented with two 1-D spatial distributions and by one 2-D distribution, representing the spatial dependency of the energy spectrum. Those distributions are used to describe the energy fluence output of both X-ray tube virtual source models (VSM).

VSM description

Each X-ray tube geometry is implemented in the **Penelope MC** framework. For each {applied voltage, bowtie filter} irradiation condition, a **Phase Space File (PSF)** is calculated. *One should note that each PSF is generated without collimation.*

$$PSF_{Applied\ Voltage, Bowtie, \emptyset\ collimation}$$

In order to construct each VSM, a **collimation** is selected. The estimated energy fluence distribution $\tilde{\phi}$ is extracted from the corresponding PSF while removing any data located outside of that collimation (**Hypothesis 5**).

$$\tilde{\phi}_{Applied\ Voltage, Bowtie, collimation}$$

As the 2-D distribution is susceptible to MC noise, an in-house statistical **smoothing algorithm** based on a Polya tree is used [2].

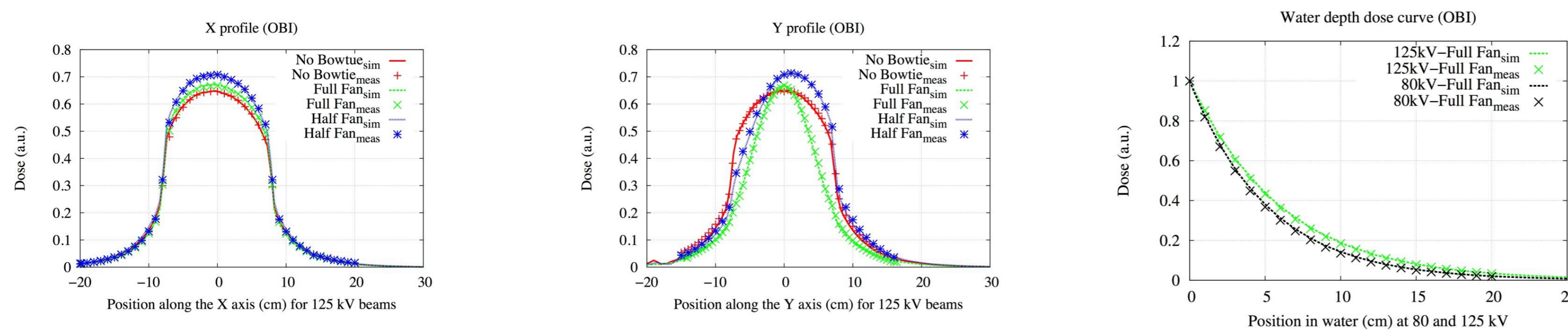
$$VSM_{Applied\ Voltage, Bowtie, collimation}$$

While the noise of MC calculations using PSF is limited by the number of photons it contains, a **VSM** (created from the same PSF) can generate an **infinite number of independent photons** regardless of the PSF size it is based on, therefore enhancing the calculation convergence capability.

In addition, as the OBI system is equipped with a patient-dependent collimation, this VSM generation process enables to use a **single PSF for every possible collimation**.

In water-phantom validation

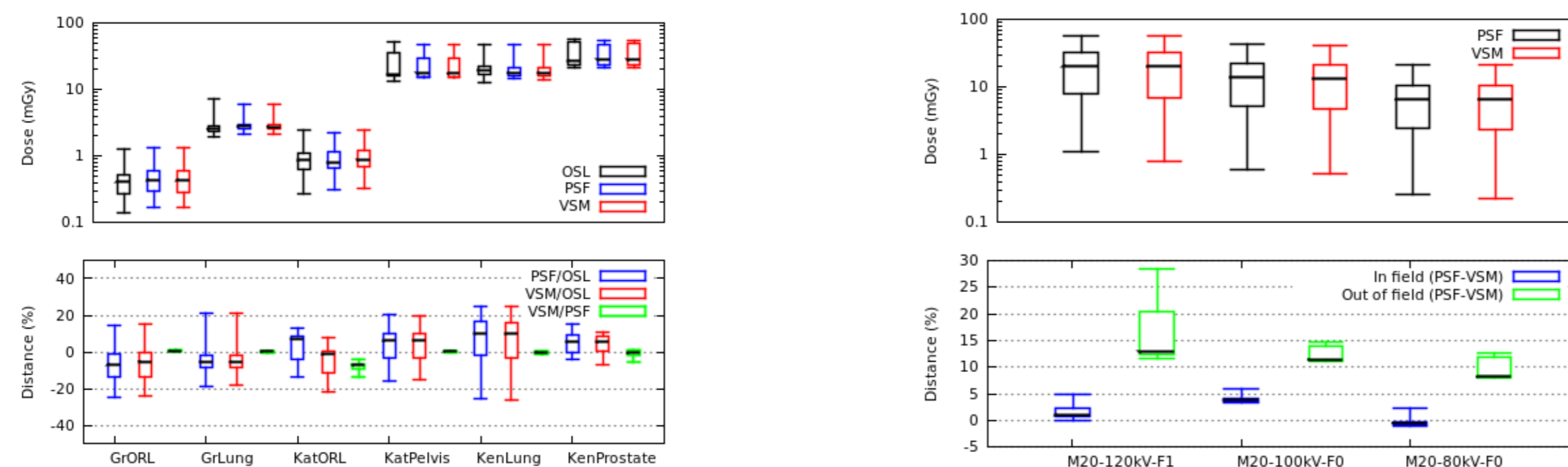
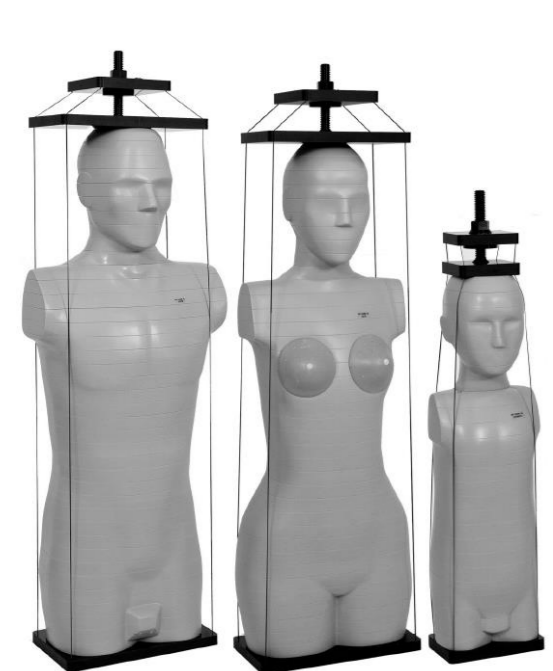
VSMs were validated in a **water phantom** against **ionization chamber** measurements (Farmer, PTW 30013) with profiles for both axis and a depth dose curve for every available filter and at multiple high voltage settings.



Overall, the simulated data passed both $\gamma_{3\%, 3mm}$ and $\gamma_{5\%, 5mm}$ tests with rates respectively higher than 80 and 95%.

Anthropomorphic phantom validation using OSL dosimeters

VSMs were validated against MC data (using PSF directly) and OSL nanodosimeters (Landauer) measurements in three anthropomorphic phantoms (CIRS, ATOM) during kV-CBCT exams performed using XVI [3].



VSM calculations are in very good agreement with other MC calculations using PSF (mean distance below 2%), as well as with OSL measurements (mean distance below 7%). Nevertheless, organ doses highlight that VSM calculation tends to underestimate out-of-field dose by about 20%.

Summary

The output photon distribution of a X-ray tube can be represented with two 1-D and one 2-D distributions as a VSM

- VSM created from a PSF can be smoothed in order to produce an infinite number of independent particles
- Collimation can be specified on the fly when the VSM is created from a PSF

Experimental validation

- In water-phantom calculations using VSM and measurements are in very good agreement (95% of success for $\gamma_{5\%, 5mm}$)
- In anthropomorphic kV-CBCT simulations using VSM are compatible with OSL measurements (mean distance <7%)
- VSM decreases simulation times by a factor of at least 2.6 compared to MC calculations using PSF (in pre-clinical conditions)
- VSM tends to underestimate out-of-field doses by about 20%.

Overall, the use of VSMs seems to be suitable for the estimation of the doses delivered to patients for both OBI and XVI kV-CBCT systems.

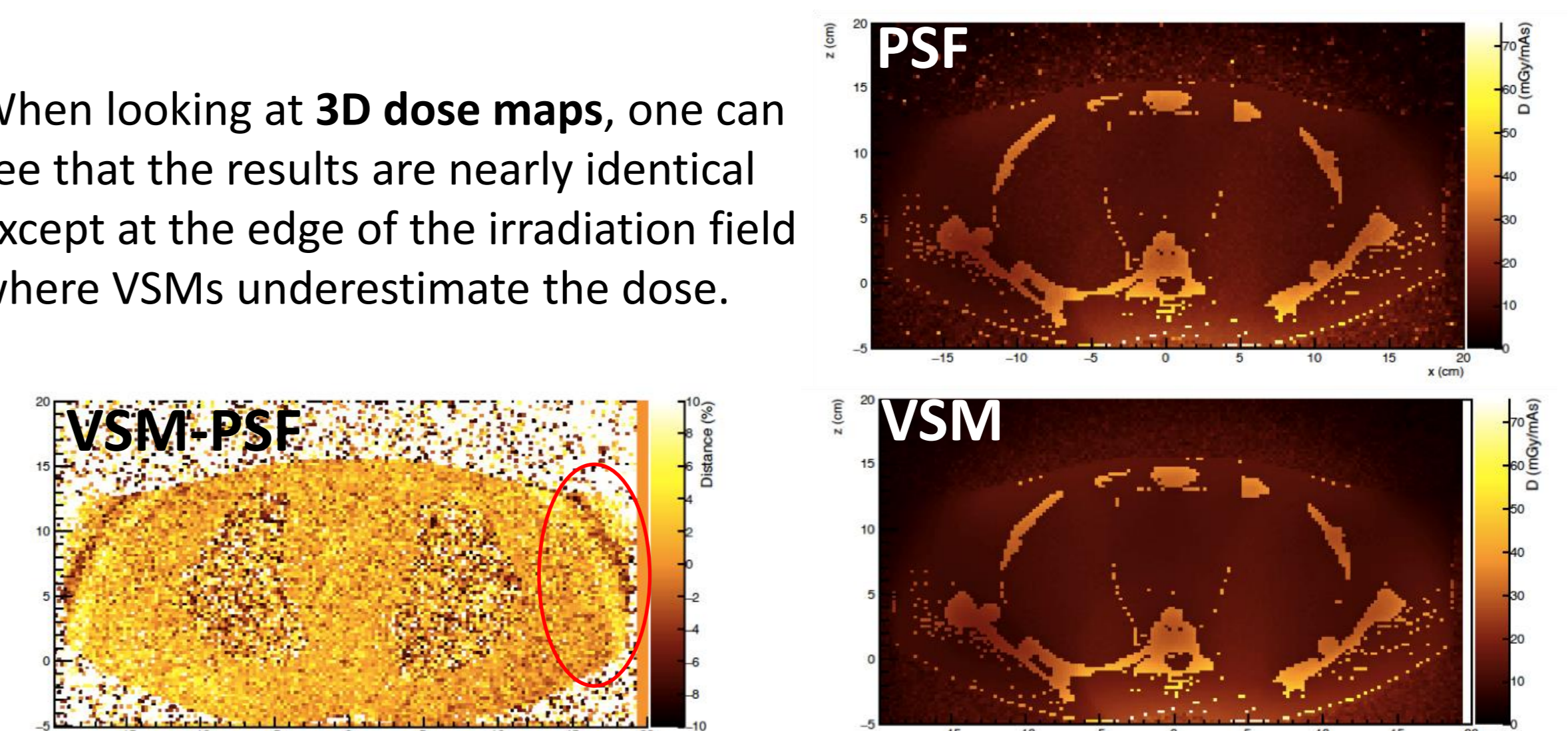
Comparison between VSM and PSF-based calculation

VSM and PSF-based calculations can be compared according their convergence speed. The following table displays the calculation time –saving by using a VSM instead of a PSF in the same conditions (in the Penelope MC framework).

Calculation case	Time saved by using a VSM instead of a PSF
Dose in an air ionization chamber	1,5x10 ³
Dose in a water phantom	11
kV-CBCT of an anthropomorphic phantom	13
kV-CBCT of a patient based on its CT scan	2,6

If VSM calculations are faster than the PSF ones in all four cases, the relative speed decreases inversely to the case complexity to reach a **2.6 time gain for pre-clinical cases**.

When looking at **3D dose maps**, one can see that the results are nearly identical except at the edge of the irradiation field where VSMs underestimate the dose.



References

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