



## PAPER

## Dosimetric evaluation of the Leksell GammaPlan™ Convolution dose calculation algorithm

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G Kollias<sup>5</sup> and P Karaïskos<sup>1,6</sup><sup>1</sup> Medical Physics Laboratory, Medical School, National and Kapodistrian University of Athens, Athens, Greece<sup>2</sup> International Atomic Energy Agency, Vienna, Austria<sup>3</sup> Medical Genesis Care Centre for Radiotherapy, Cromwell Hospital, London, United Kingdom<sup>4</sup> Elekta Instrument AB, Stockholm, Sweden<sup>5</sup> Departments of Medical Physics and Gamma Knife, Hygeia Hospital, Athens, Greece<sup>6</sup> Author to whom any correspondence should be addressed.E-mail: [pkaraisk@med.uoa.gr](mailto:pkaraisk@med.uoa.gr)**Keywords:** Gamma Knife, dosimetry, convolution, Monte Carlo**Abstract**

The dosimetric accuracy of the Leksell GammaPlan Convolution calculation algorithm was evaluated through comparison with corresponding Monte Carlo (MC) dosimetric results. MC simulations were based on generated sector phase space files for the 4 mm, 8 mm and 16 mm collimator sizes, using a previous comprehensive Gamma Knife Perfexion™ source model and validated using film dosimetry. Test cases were designed for the evaluation of the Convolution algorithm involving irradiation of homogeneous and inhomogeneous phantom geometries mimicking clinical cases, with radiation fields created using one sector (single sector), all sectors with the same (single shot) or different (composite shot) collimator sizes. Dose calculations using the Convolution algorithm were found to be in excellent agreement (gamma pass rate greater than 98%, applying 1%/1 mm local dose difference and distance agreement criteria), with corresponding MC calculations, indicating the accuracy of the Convolution algorithm in homogeneous and heterogeneous model geometries. While of minor clinical importance, large deviations were observed for the voxels laying inside air media. The calculated beam on times using the Convolution algorithm were found to increase (up to 7%) relative to the TMR 10 algorithm currently used in clinical practice, especially in a test case mimicking a brain metastasis close to the skull, in excellent agreement with corresponding MC calculations.

**1. Introduction**

The Leksell Gamma Knife® (GK) (Elekta AB, Stockholm, Sweden) is a stereotactic radiosurgery system that has been developed to treat well defined intracranial lesions using multiple non-coplanar photon beams produced by <sup>60</sup>Co radioactive sources (Leksell 1983). In the latest GK models (Icon™ and Perfexion™) the radiation units are identical and consist of a fixed conical collimation system and 192 <sup>60</sup>Co sources equally distributed over eight sectors in a cylindrical configuration (Lindquist and Paddick 2007, Novotny *et al* 2009). Each sector can be moved independently along a conical surface to facilitate alignment of the sources with any of the three available collimation channels, labelled as 4 mm, 8 mm and 16 mm, as well as in the blocked position, allowing the use of composite shots. Treatment planning is performed using the Leksell GammaPlan® (LGP) dedicated treatment planning system which offers two dose calculation algorithms; namely Tissue Maximum Ratio (TMR) and Convolution (Elekta AB 2011a, 2011b). The TMR algorithm is characterized by a straightforward analytical approach, based on the inverse square law and exponential attenuation of gamma rays within the patient geometry. Tissue inhomogeneities are ignored and the algorithm treats the whole head as a uniform water material. There are two available versions of this method (TMR Classic and TMR 10) with slightly different algorithms and different sets of configuration data (Elekta AB 2011a). The Convolution algorithm is based

on collapsed cone and pencil beam convolution methods enabling accurate dose estimation in regions with pronounced tissue inhomogeneities (e.g. lesions in the cavernous sinus) (Elekta AB 2011b).

There are several published dosimetry studies that compare the Convolution with the TMR Classic or TMR 10 results and show increased differences (up to 10%) for specific cases attributed to the presence of tissue inhomogeneities (Nakazawa *et al* 2014, Xu *et al* 2014, Rojas-Villabona *et al* 2016). Despite these findings however, most GK centres still use TMR 10 for dose prescription. This is because corresponding prescription doses have been tested and optimized over the last few decades using water-based algorithms and the dosimetric differences between the water-based and Convolution algorithms need to be better understood before this method can be confidently employed in a clinical setting (Rojas-Villabona *et al* 2016).

The aim of this work is to evaluate the dosimetric accuracy of the LGP Convolution algorithm. Monte Carlo (MC) methods were used to obtain reference dose values due to their proven capability of accurately simulating the radiation transport and energy deposition on patient or phantom geometries. To increase simulation efficiency sector based phase space (PHSP) files were generated for the 4 mm, 8 mm and 16 mm collimation channels using a previous validated comprehensive source model of the GK Perfexion radiation unit (Pappas *et al* 2016, Zoros *et al* 2017). These PHSP files were used as source models for all subsequent MC simulations. Besides increasing simulation efficiency, the use of sector based PHSP source models enables the simulation of target irradiation with multiple composite shots in a single input file by sampling photons from the corresponding PHSP files. The suggested procedure for generating MC reference doses was benchmarked using Gafchromic<sup>TM</sup> film dosimetry. Dosimetric evaluation of the Convolution algorithm was restricted in test cases generated using homogenous and heterogeneous phantom geometries.

## 2. Materials and methods

### 2.1. GK Perfexion phase space generation

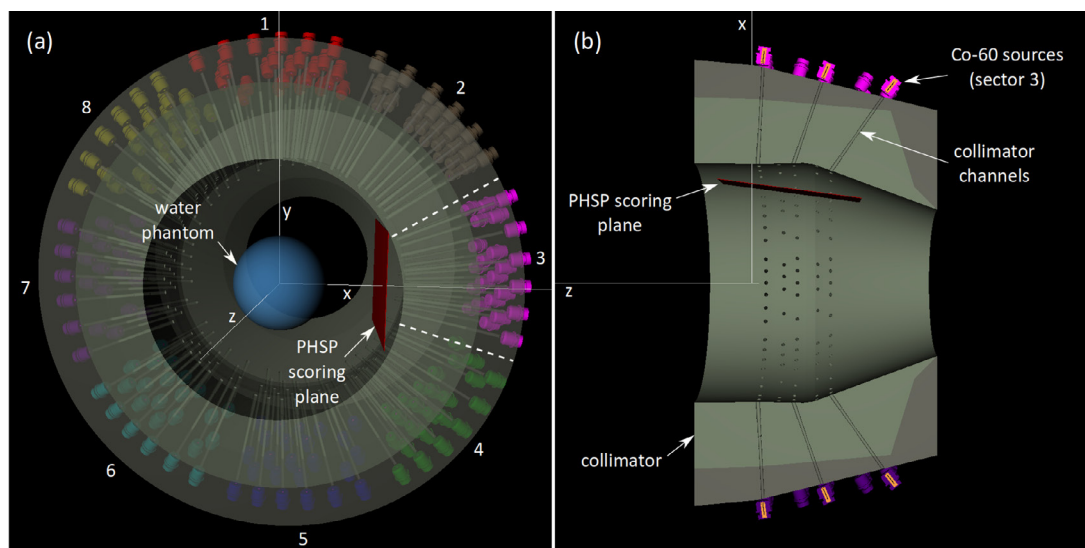
A comprehensive model of the GK Perfexion radiation unit (referred as GK-PFX in the following), developed by our group (Pappas *et al* 2016) based on the C++ class library (egs++) of the EGSnrc MC system (Kawrakow and Rogers 2003, Kawrakow *et al* 2017) was used in this study (see figure 1). In brief, in this GK model all 192 sources are simulated as cylinders of 18 mm height of <sup>60</sup>Co material, emitting photons isotropically with energies 1.1732 and 1.3325 MeV, equiprobably. Each <sup>60</sup>Co cylinder is enveloped in an aluminium bushing and altogether in a stainless-steel capsule. Beta particle emission is neglected as they would not escape source capsules. Each collimator size is modelled as a separate geometry and, consequently, crosstalk of photons between collimation channels was not considered. This assumption is expected to affect dose results by less than 0.15% (Lindquist and Paddick 2007). Detailed information of the full GK-PFX model can be found in Pappas *et al* (2016).

The aforementioned detailed GK-PFX model has been shown to provide accurate dosimetric data in single shot irradiations (Pappas *et al* 2016, Zoros *et al* 2017). However, its use in MC simulations is associated with low efficiency due to the isotropic emission of photons from each <sup>60</sup>Co source. To increase simulation efficiency, photon PHSP data were acquired using a C++ user code developed in house based on the International Atomic Energy Agency (IAEA) PHSP read/write routines available within EGSnrc MC system (Capote and Kawrakow 2006). These routines are called when the traced photons cross a user-defined surface to store their PHSP data into a file using the IAEA format.

While the whole treatment head was simulated, only photons from the third sector were generated, transported and stored using for simplicity a planar PHSP surface (see figure 1). The PHSP surface was situated at the distal end of the collimator and at 17.15 cm from the Radiation Focal Point (RFP) of the radiation unit enabling the simulation of irradiation geometries with peripheral targets (see figure 1). The PHSP planar surface was restricted to outside the collimation system, and therefore photons scattered within the collimator towards the patient without passing the PHSP surface were not stored. The amount of these photons, however, is minimal considering the dimensions and material of the collimation system (Moskvin *et al* 2002). Electron PHSP data were also not stored since their spectrum consists mainly of low energy electrons which are absorbed in the air before reaching the patient (Moskvin *et al* 2002). A total number of  $17 \times 10^{10}$  original photon histories were generated in each MC simulation resulting to one PHSP file for each collimator size. By sampling photons from different PHSP files, irradiation of targets with multiple composite shots can be achieved in a single MC simulation.

### 2.2. Benchmarking of the PSHP based MC calculations

The suggested procedure for generating reference dose data sets based on the sector PHSP source model was validated against corresponding results obtained using MC simulations with the detailed GK-PFX model and Gafchromic films. An irradiation geometry with all sectors blocked except the third sector (single sector) was used for each collimator size. Comparison was performed in terms of relative off axis profiles along the x, y and z axes of the GK-PFX radiation unit. Monte Carlo dosimetry simulations were performed using a mathematical phantom of homogeneous water of 80 mm radius centred at the RFP surrounded by air. Relative off axis profiles



**Figure 1.** (a) Illustration of the GK PFX egs++ model used in the simulations depicting the collimator with its channels, the  $^{60}\text{Co}$  sources inside their bushings as well as their spatial distribution over eight sectors. Different colors have been used for each sector. Sector numbers are also indicated. The planar surface used for storing photon PHSP is shown in red in front of the third sector (delineated). The spherical phantom used for dose profile and output factor scoring is also depicted at the centre of the irradiation unit to showcase the geometrical distance from the PHSP plane. (b) Central cut of the model at the  $xy$ -plane of the LGP coordinate space without the water phantom.

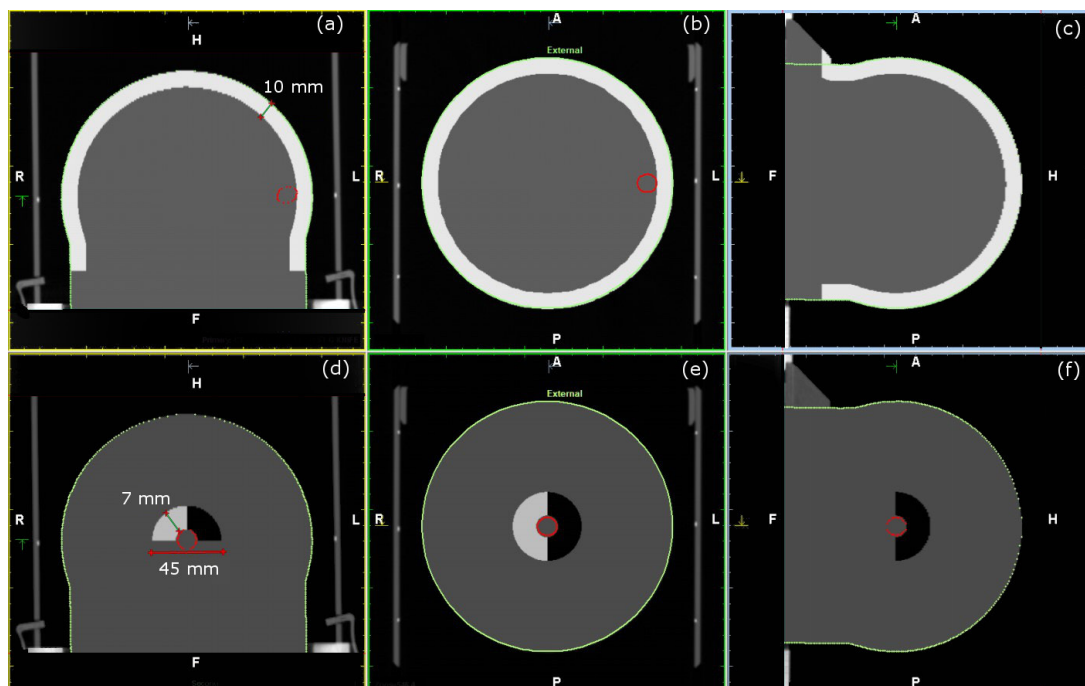
were scored using an array of  $1\text{ mm}^3$  volume cubic voxels. The dose profile statistical uncertainties at 50% of the  $D_{\text{max}}$  were up to 0.65% for the detailed GK-PFX model and  $17 \times 10^{10}$  initial photon histories. Corresponding uncertainties for the PHSP based MC simulations were up to 0.22% using PHSP file recycling.

Film dosimetry was performed following the procedure described in our previous work (Zoros *et al* 2017). In brief, Gafchromic<sup>TM</sup> EBT3 (Lot# 12021402 Ashland Inc., Wayne, NJ) films were cut in  $(110 \times 80)\text{ mm}^2$  pieces and placed on the central slab of the Leksell Gamma Knife<sup>®</sup> Dosimetry Phantom made of Solid Water (called hereafter LGK-SW) (Elekta Instrument AB). Fine (1 mm slice thickness) CT images of the phantom were acquired and imported into the LGP treatment planning system. Treatment plans using only the third sector with the 4 mm, 8 mm and 16 mm collimators, respectively were developed. A dose of 6 Gy at the RFP was prescribed to all treatment plans using the Convolution dose calculation algorithm and exported in DICOM RT Dose format. All irradiations were performed with the phantom situated so that the film's plane coincided with the  $xy$ -plane (axial) except for the relative dose profile measurements along the  $z$ -axis where the phantom was rotated for the film to coincide with the  $xz$ -plane (coronal) of the GK coordinate space. Films were scanned 24 h post irradiation using an EPSON V750 PRO flatbed scanner. RGB positive images of 48-bit depth were obtained using 150 dpi resolution ( $0.169\text{ mm px}^{-1}$ ) and saved as tagged image file format (TIFF) files. Measured pixel values were converted to dose using the calibration data of the used film batch and a triple channel technique was employed in this study to obtain film dosimetry results (Micke *et al* 2011). Each film was registered to the GK coordinate space using four metal pins previously placed into the LGK-SW phantom (Zoros *et al* 2017). An uncertainty of less than 1.5% ( $k = 1$ ) was estimated for the experimentally determined 50% relative off axis profile value.

### 2.3. Test cases for the evaluation of the Convolution algorithm

For the evaluation of the dosimetric accuracy of Convolution three mathematical phantom geometries were generated based on the CT images of the LGK-SW phantom. In detail, the CT images of the LGK-SW phantom were digitally processed in order to identify and replace the Hounsfield Units (HU) of the voxels laying inside the external surface with  $\text{HU} = 0$  (equal to that of water) and produce a homogeneous water phantom. Based on this phantom two inhomogeneous phantoms were constructed: In the first, the HUs of the voxels laying inside a 10 mm thick spherical ring at the periphery of the phantom were set to 1523 to resemble cortical bone (see figures 2(a)–(c)). In the second, the HUs of the voxels laying inside two truncated hemispherical volumes of 7 mm inner radius and 22 mm outer radius were set equal to 1523 and  $-1000$  to resemble those of cortical bone and air, respectively (see figures 2(d)–(f)). The generated CT images were imported into the LGP and registered with the GK coordinate system.

Since in the Perfexion and Icon radiation units each sector can be enabled or disabled independently, the first test involved the irradiation of the homogeneous water phantom centred at RFP using either single sector (i.e. one sector enabled) or single shot (all sectors enabled) irradiation geometries. In total, six treatment plans



**Figure 2.** (a), (d) Coronal, (b), (e) axial and (c), (f) sagittal slices of the two inhomogeneous phantoms generated for the evaluation of the Convolution dose calculation algorithm results. Three materials were defined; water with HU of 0, cortical bone with HU of 1523 and air with HU of  $-1000$ . The localization box used for registering the images of the phantom in the GK coordinate system can be seen.

were developed: three using sector 3 and three using all sectors, for the 4 mm, 8 mm and 16 mm collimators, respectively. For the single shot cases a dose of 24 Gy at 100% was prescribed which for a calibration dose rate of  $3.127 \text{ Gy min}^{-1}$  resulted in 9.45 min, 8.53 min and 7.65 min irradiation times, for the 4 mm, 8 mm and 16 mm collimators, respectively. Dose values were calculated using the Convolution algorithm and exported in DICOM RT format using a spatial resolution of  $1 \text{ mm}^3$ .

The second test case involved the irradiation of a target situated close to the bone inhomogeneity at the periphery of the phantom (see figures 2(a)–(c)), simulating a clinical case of a brain metastasis close to the skull. The third test case involved the irradiation of a target situated at the centre of the phantom and behind bone and air inhomogeneities (see figures 2(d)–(f)) mimicking the treatment of a pituitary adenoma lesion. A single composite shot was used for the irradiation of each target consisting of two 4 mm, four 8 mm and two 16 mm sectors. Convolution was used and a dose of 24 Gy at 50% isodose was prescribed in both test cases which for a calibration dose rate of  $3.132 \text{ Gy min}^{-1}$  resulted to irradiation times of 15.78 min and 16.76 min for the second and third case respectively. Convolution dose values were exported in DICOM RT format using a spatial resolution of  $0.5 \text{ mm}^3$ .

#### 2.4. Generation of Monte Carlo reference dosimetry data sets

The treatment plans of the Convolution test cases were saved in XML format. Software routines were developed using MATLAB® (MathWorks, Natick, MA) to parse the XML plan files and develop corresponding EGSnrc input files for Monte Carlo calculations. A double resolution voxelized geometry was used to describe the phantom geometry based on voxel dimensions and composition retrieved from the corresponding DICOM images. The CT resolution ( $0.4844 \times 0.4844 \text{ mm}^2$  pixel size and 1 mm slice thickness) was used in a cubic region ( $60 \times 60 \times 60 \text{ mm}^3$ ) around the RFP. The rest of the geometry was down-sampled in the  $x$  and  $y$  dimensions by a factor of 8 (i.e.  $3.8752 \times 3.8752 \times 1 \text{ mm}^3$  voxel size) for performance reasons. Three materials were defined; the air, water and cortical bone with densities of  $0.0012 \text{ g cm}^{-3}$ ,  $1 \text{ g cm}^{-3}$  and  $1.92 \text{ g cm}^{-3}$ , respectively, according to ICRU-44 (White *et al* 1989). The fiducial box was not included in the phantom geometry as the LGP dose calculation algorithms do not take it into account.

Photons were sampled from PHSP of collimator  $i$  (4, 8 and 16 mm) with a discrete probability function of:

$$P_i = \frac{n_i \cdot N_i}{\sum_i n_i \cdot N_i}$$

(where  $n_i$  is the number of sectors of collimator  $i$  in the composite shot and  $N_i$  the number of particles stored in the corresponding PHSP file) coming from the fact that the number of photon histories simulated to obtain each PHSP file was the same. The position and direction of each sampled photon was rotated by  $k_i \cdot 2\pi/8$  radians, where  $k_i$ , an integer taking values in the range of 0–7 depending on the shot configuration, was sampled from a homogeneous distribution for each collimator  $i$ . Dose was scored only in the high resolution region.

MC results were converted to absolute dose values by using a scaling factor of:

$$\frac{\dot{D}^{\text{cal}}}{D_{w, 16 \text{ mm}}^{\text{MC, iso}}} \cdot [\text{BOT}]$$

where  $\dot{D}^{\text{cal}}$  is the calibration dose rate of the GK unit taking into account the exponential decay of the  $^{60}\text{Co}$  sources until the plan date,  $D_{w, 16 \text{ mm}}^{\text{MC, iso}}$  is the MC calculated dose at the isocenter of a homogeneous water phantom using a 16 mm collimator shot, and BOT the beam-on-time determined using the Convolution algorithm.

All dosimetry calculations were performed using the egs\_chamber EGSnrc user code (Wulff *et al* 2008). Photon cross sections were taken from the XCOM database (Berger and Hubbell 1987) in all MC simulations. Electron transport was performed using the PRESTA-II algorithm (Kawrakow 2000). The ECUT parameter was set to 521 keV inside the water phantom and to 811 keV elsewhere, while PCUT was set to 1 keV. Other simulation parameters were set to their default values. The photon cross section enhancement (XCSE) variance reduction technique with an enhancement factor of 512 was used to attain reasonably low uncertainties using fewer particle histories (Wulff *et al* 2008).

All MC simulations were performed in a super-computer consisting of 426 computational nodes with ten Ivy Bridge Intel® Xeon® E5 v2 processors per node, which offered a total of 8520 CPU cores (computational threads) clocked at 2.8 GHz. The statistical uncertainty of the reference dosimetry results was better than 0.4% for doses greater than 1% of the maximum dose value. Ignoring the simulation time for scoring the PHSP files, an efficiency gain (defined as the ratio of core hours required to obtain dose results with the same statistical uncertainty) greater than 200 was found for the PHSP simulations depending on the collimators used.

### 3. Results

#### 3.1. GK-PFX PHSP model benchmarking

A number of  $1.3 \times 10^6$  for the 4 mm,  $4 \times 10^6$  for the 8 mm and  $15 \times 10^6$  photons for the 16 mm collimator apertures were stored resulting in PHSP files of 50 MB, 140 MB and 550 MB, respectively. Single sector PHSP, full model MC and experimentally derived relative dose profiles for the 4 mm, 8 mm and 16 mm collimators are plotted as a function of the  $x$ ,  $y$ , and  $z$  coordinate value in figure 3. An excellent agreement between all data sets can be observed. Quantitative comparison of the PHSP dose profiles of each collimator versus corresponding detailed source model results using 1D gamma index analysis (Low *et al* 1998) and 1%/1 mm local dose difference and distance difference gamma criteria (International Atomic Energy Agency 2004, Chung *et al* 2016) revealed a passing rate near 100% for relative dose values greater than 10%. This excellent gamma passing rate was also observed when the PHSP relative dose profiles of each collimator were compared with corresponding EBT3 film dosimetry results using 2%/1 mm local dose difference and distance agreement acceptance criteria due to the higher experimental uncertainties. It must be noted that the observed subtle differences between the presented data sets in the single sector profiles are diminished when corresponding single shot data are compared (data not shown).

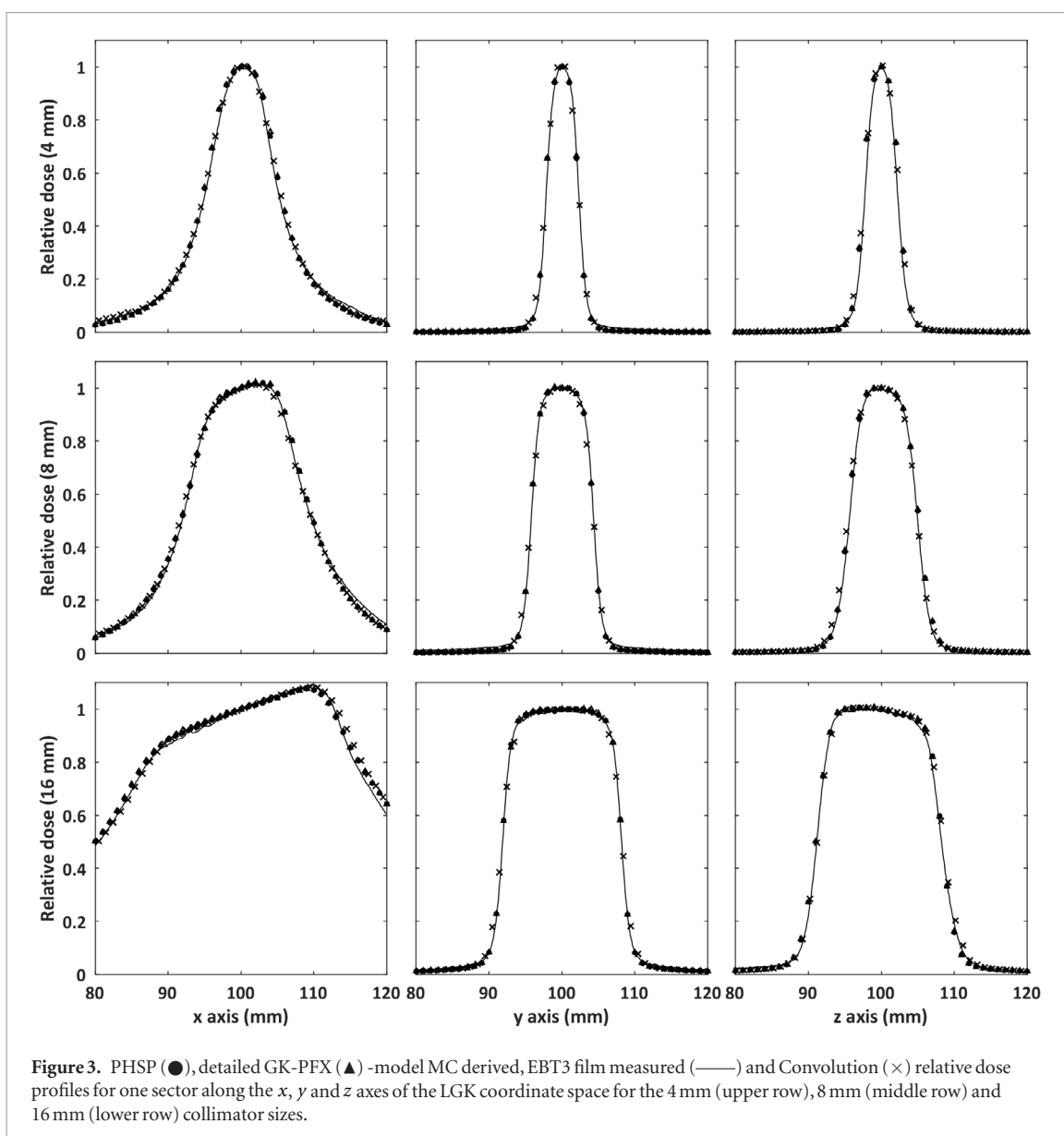
#### 3.2. Convolution dosimetric evaluation results

##### 3.2.1. Homogeneous water test cases

Single sector Convolution relative dose profile data for each collimator size are plotted in figure 3 where an excellent agreement with the corresponding MC (PHSP and detailed source model) and experimental dosimetry results can be observed. Gamma index analysis between Convolution and the MC PHSP relative dosimetry data using 1%/1 mm local dose difference and distance agreement showed a passing rate near 100% for relative doses greater than 10%.

In figure 4, Convolution and corresponding MC PHSP based relative dose profiles for the single shot irradiation geometry are plotted as a function of the  $x$ ,  $y$ , and  $z$  coordinate value, for each collimator size. An excellent agreement between Convolution and MC results can be observed. Gamma index analysis between the Convolution and PHSP MC data sets using 1%/1 mm local dose difference and distance difference criteria revealed an excellent passing rate (nearly 100%) for all collimator sizes and presented relative dose range.

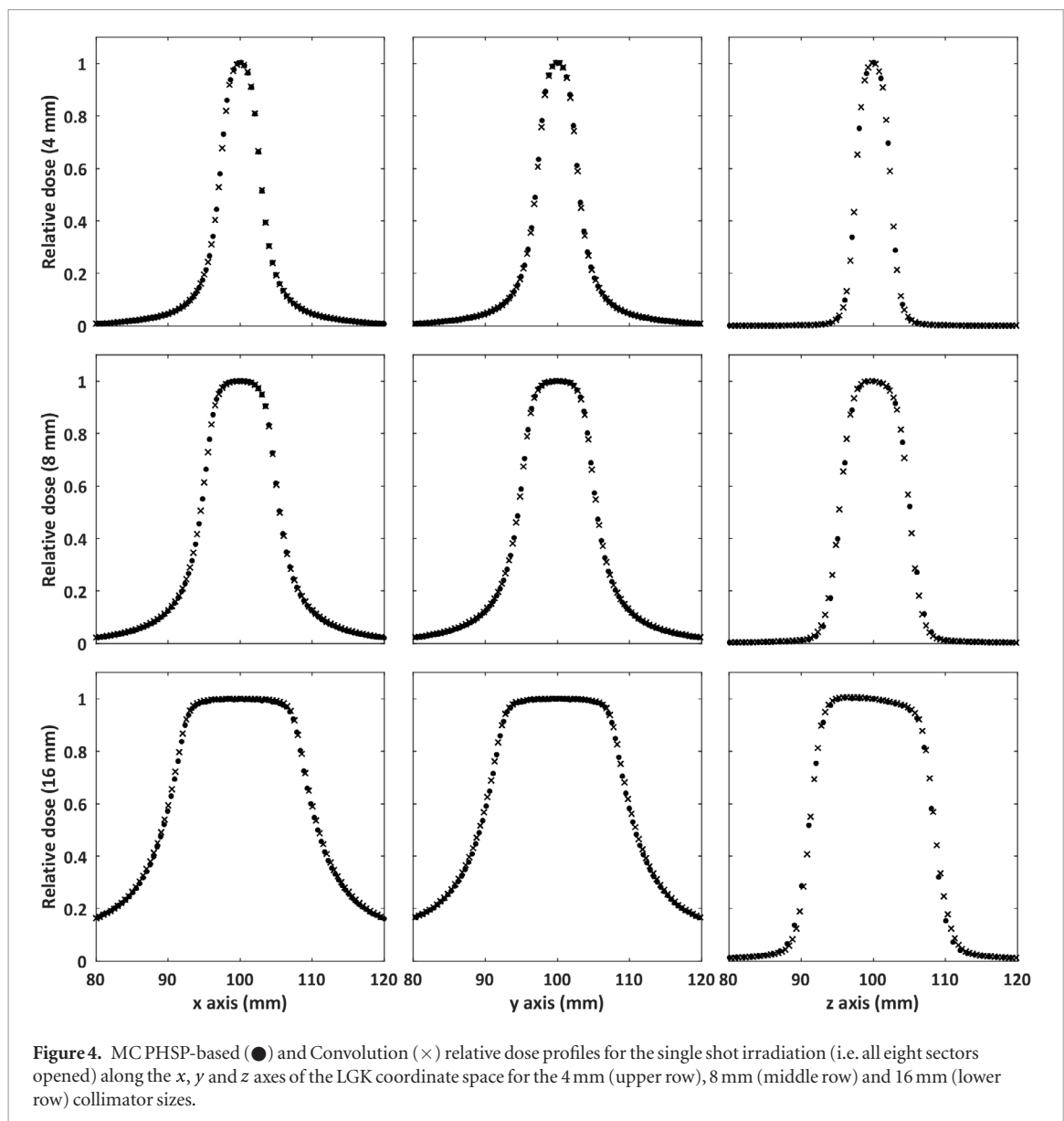
The ratios of the calculated dose rates of the smaller shots (i.e. 4 mm and 8 mm) at their centers divided by the corresponding value of the 16 mm shot are equal to the corresponding output factors (OFs). OFs obtained using the dose rates defined by the Convolution dose calculation algorithm are given in table 1. In the same table the corresponding OFs calculated using the MC PHSP calculated dose data along with the values suggested by the vendor (Elekta AB 2011a) are included for comparison. A fine agreement (less than 1%) between the OFs calculated using the Convolution data and the PHSP MC results can be observed for the 4 mm and 8 mm collimators. The Convolution based OFs were also found in excellent agreement with the corresponding OFs given by the vendor calculated using detailed GK-PFX model MC simulations (Elekta AB 2011a).



### 3.2.2. Inhomogeneous phantom test cases

In figures 5(a), (b), (d)–(f), the central axial and coronal slices of the inhomogeneous phantoms used in test cases 2 and 3 (i.e. those with the peripheral target near skull and with the central target behind bone/air inhomogeneities) are plotted and superimposed with the corresponding Convolution isodose distributions. Corresponding PHSP based isodose results are also plotted for reasons of comparison. Gamma index data were calculated for all phantom voxels having dose values greater than 1% of the maximum dose using 1%/1 mm local dose difference and distance agreement acceptance criteria. Gamma index results are superimposed on the central axial and coronal slices of each phantom using an appropriate colourmap. A histogram of the gamma index values for the 3D scoring cube is presented in figures 5(c) and (f) for each test case respectively.

Excellent agreement between the Convolution and MC isodose results can be observed for all target voxels as well as voxels laying inside the phantom in the second test case (see figures 5(a) and (b)). A gamma pass rate of 99% was attained, evincing perfect agreement between Convolution and MC calculations (see figure 5(c)). For the stringent third test case involving both air and bone inhomogeneities surrounding the target, the Convolution dose calculation algorithm was found capable of predicting accurately the dose to the target, bone and water voxels of the phantom. Accuracy deteriorated for the voxels laying inside the air inhomogeneity where the Convolution algorithm was found to overestimate dose at distances up to ~10 mm and underestimate the dose at air voxels laying at higher distances from the lesion (figures 5(d) and (e)). These comparison results correspond to a gamma pass rate of 93% considering all voxels inside the scoring cube and 98% excluding the voxels of air inhomogeneity due to their minor clinical relevance.

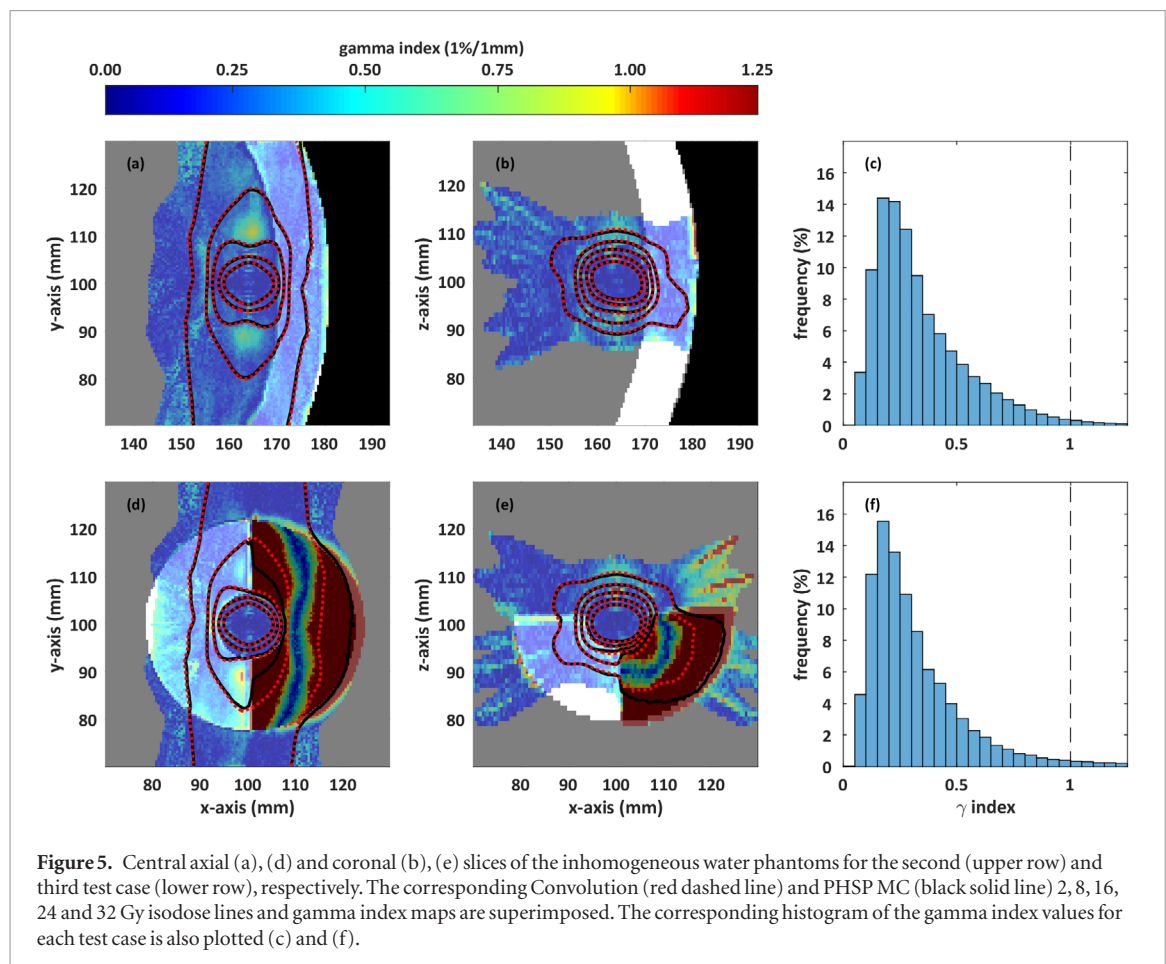


**Table 1.** Output factors for the 4 mm and 8 mm collimators calculated using the MC PHSP and the Convolution single shot data sets. Corresponding values suggested by the vendor are also included for comparison.

Method	OF <sub>4 mm</sub>	OF <sub>8 mm</sub>
MC PHSP	$0.8214 \pm 0.0014$	$0.8930 \pm 0.0017$
Convolution	0.814	0.901
Elekta AB (2011a)	0.814	0.900

## 4. Discussion

In this study, single sector phase space files were obtained for the 4 mm, 8 mm and 16 mm collimators of the Perfexion irradiation unit and benchmarked against corresponding detailed MC simulations and experimental data. An excellent agreement was observed between the PHSP source model based-, the detailed PFX source model based-, MC calculated and the radiochromic film measured single sector dose data. The PHSP files were consequently used to obtain dosimetric data for a homogeneous water voxelized phantom geometry irradiated using a single shot with the 4 mm, 8 mm and 16 mm collimators respectively. Based on these dose data, OFs values were calculated for the 4 mm and 8 mm collimator, respectively (see table 1). The calculated PHSP based OFs were found in excellent agreement ( $<0.3\%$ ) and within statistical uncertainties, with corresponding values reported by our group using the same detailed GK-PFX model and MC simulation code (Pappas *et al* 2016, Zoros *et al* 2017).



**Figure 5.** Central axial (a), (d) and coronal (b), (e) slices of the inhomogeneous water phantoms for the second (upper row) and third test case (lower row), respectively. The corresponding Convolution (red dashed line) and PHSP MC (black solid line) 2, 8, 16, 24 and 32 Gy isodose lines and gamma index maps are superimposed. The corresponding histogram of the gamma index values for each test case is also plotted (c) and (f).

The stored PHSP files were then used to obtain reference dose distributions for the evaluation of the accuracy of the Convolution dose calculation algorithm. Test cases were designed for the evaluation of the Convolution algorithm involving irradiation of homogeneous and inhomogeneous water phantom geometries with radiation fields created using one sector (single sector), all sectors with the same (single shot) or different (composite shot) collimator sizes. An excellent agreement between the Convolution and MC dosimetry results was found for the designed test cases in the homogeneous water phantom. Specifically, the Convolution relative dose profiles agreed with the corresponding MC dosimetry data for all single sector and single shot treatment plans giving a gamma index passing rate of nearly 100% using 1%/1 mm local dose difference and distance to agreement acceptance criteria. The OFs calculated using the Convolution dose results were also found to agree within 1% with the corresponding MC PHSP calculated values and the values suggested by the vendor for dosimetry calculations using the TMR 10 dose calculation algorithm (see table 1). It must be also noted that the prescribed irradiation times using Convolution agreed within 0.3% with the corresponding times prescribed using the TMR 10 dose calculation algorithms. These findings indicate an accurate implementation of the collapsed cone convolution methods in the LGP treatment planning system (Elekta AB 2011b).

For the test case mimicking a brain metastasis close to the skull, results revealed that the Convolution dose algorithm is able to accurately calculate the dose to all water voxels (and therefore the voxels comprising the target) as well as the voxels of the cortical bone inhomogeneity. Similar findings were also observed for the test case mimicking a pituitary adenoma irradiation, except for the voxels laying inside the air inhomogeneity. While of decreased clinical importance, these differences between the Convolution and MC dosimetry data sets could be attributed to pencil beam convolution kernels used in the Convolution dose calculation algorithm (Knöös *et al* 2006, Elekta AB 2011b).

Beam on times (BOT) of 15.78 min and 16.16 min were calculated using the Convolution algorithm for delivering 24 Gy at the periphery of the target in the brain metastasis and pituitary adenoma case, respectively. Given the excellent agreement between the MC and the Convolution calculations to the water voxels comprising a target in the inhomogeneous phantoms, results of this work indicate an accurate implementation of Convolution algorithm in Gamma Knife applications. The TMR 10 beam on times for the same test cases were found equal to 14.72 min and 16.95 min, which are lower by 7% and greater by 1% respectively, with regard to the Convolution BOT values. The increased deviation between the Convolution and TMR 10 BOT results for the brain metastasis case was further investigated by comparing the single shot irradiations at the centre of the homogeneous water

phantom with similar irradiations developed using the phantom with the skull inhomogeneity. It was found that the skull inhomogeneity affects the dose rate at the centre of the phantom by ~5% in agreement with the literature (Moskvin *et al* 2004, Al-Dweri *et al* 2005, Nakazawa *et al* 2014, Xu *et al* 2014, Rojas-Villabona *et al* 2016, Yuan and Machtay 2017, Choi *et al* 2018). The increased deviation of up to 7% in BOT values between Convolution and TMR 10 calculations observed in this study for the target located at the periphery of the phantom, simulating a clinical case of a brain metastasis close to the skull, in excellent agreement with corresponding MC calculations, also supports that the effect of bone inhomogeneity is more pronounced for targets located at the periphery of the head ((Xu *et al* 2014, Rojas-Villabona *et al* 2016) and Convolution algorithm can be more accurately predict dose distributions in such cases.

## 5. Conclusion

Sector PHSP source models were generated using a previously validated GK-PFX detailed source model for the 4 mm, 8 mm and 16 mm field sizes. MC simulations were performed based on the stored PHSP files to calculate reference dose distributions and evaluate the dosimetric accuracy of the Convolution algorithm. Test cases were designed involving irradiation of homogeneous and inhomogeneous water phantom geometries with radiation fields created using one sector (single sector), all sectors with the same (single shot) or different (composite shot) collimator sizes. Dose calculations using the Convolution algorithm were found to be in excellent agreement (gamma pass rate greater than 98%, applying 1%/1 mm local dose difference and distance agreement criteria), indicating the accuracy of the Convolution algorithm in homogeneous and heterogeneous model geometries. While of minor clinical importance large deviations were observed for the voxels laying inside air media. Increased BOT values were found using the Convolution algorithm in the presence of inhomogeneities in agreement with corresponding MC calculations. Further investigation using patient model geometries is required to apply the findings of this study in GK clinical applications.

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