Development of a GPU-accelerated Monte Carlo dose calculation module for nuclear medicine, ARCHER-NM: demonstration for a PET/CT imaging procedure

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Abstract

Objective. This paper describes the development and validation of a GPU-accelerated Monte Carlo (MC) dose computing module dedicated to organ dose calculations of individual patients undergoing nuclear medicine (NM) internal radiation exposures involving PET/CT examination. Approach. This new module extends the more-than-10-years-long ARCHER project that developed a GPU-accelerated MC dose engine by adding dedicated NM source-definition features. To validate the code, we compared dose distributions from the point ion source, including \textsuperscript{18}F, \textsuperscript{11}C, \textsuperscript{15}O, and \textsuperscript{68}Ga, calculated for a water phantom against a well-tested MC code, GATE. To demonstrate the clinical utility and advantage of ARCHER-NM, one set of \textsuperscript{18}F-FDG PET/CT data for an adult male NM patient is calculated using the new code. Radiosensitive organs in the CT dataset are segmented using a CNN-based tool called DeepViewer. The PET image intensity maps are converted to radioactivity distributions to allow for MC radiation transport dose calculations at the voxel level. The dose rate maps and corresponding statistical uncertainties were calculated at the acquisition time of PET image. Main results. The water-phantom results show excellent agreement, suggesting that the radiation physics module in the new NM code is adequate. The dose rate results of the \textsuperscript{18}F-FDG PET imaging patient show that ARCHER-NM’s results agree very well with those of the GATE within $-2.45\%$ to $2.58\%$ (for a total of 28 organs considered in this study). Most impressively, ARCHER-NM obtains such results in 22 s while it takes GATE about 180 min for the same number of $5 \times 10^8$ simulated decay events. Significance. This is the first study presenting GPU-accelerated patient-specific MC internal radiation dose rate calculations for clinically realistic \textsuperscript{18}F-FDG PET/CT imaging case involving autosegmentation of whole-body PET/CT images. This study suggests that the proposed computing tools—ARCHER-NM—are accurate and fast enough for routine internal dosimetry in NM clinics.

1. Introduction

Both positron emission tomography (PET) and computed tomography (CT) are widely used imaging modalities in the diagnosis and monitoring of cancer evolution. The combination of PET and CT, known as PET/CT, enables both anatomical and metabolic imaging of patients, which improves the diagnostic quality and efficiency of radiologists (Kapoor et al. 2004). However, in PET/CT examination procedures, the injection of radiopharmaceuticals results in internal ionizing radiation for patients, which increases the risk of radiation-induced cancer, particularly in younger patients (Huang et al. 2009, Belinato et al. 2017). To quantify such risks, it is critical to calculate the absorbed doses of the patients with enough accuracy and speed acceptable for routine
clinical applications (Einstein et al 2007). Therefore, it is essential to develop a method for internal dosimetry in this field.

Traditionally, the internal dosimetry of a specific patient can be estimated at the organ level via the S-value—the mean absorbed dose in a target organ per radioactivity decay in a source organ calculated in a standard phantom (Loevinger et al 1991, Stabin and Siegel 2003). Based on this method, Andersson et al (Andersson et al 2017) developed an internal dosimetry program IDAC-Dose 2.1 to estimate the absorbed dose for diagnostic nuclear medicine using the S-value of ICRP computational voxel phantoms, and Hu et al (2021) evaluated the absorbed dose for patients undergoing $^{18}$F FDG PET examinations using S-value of Chinese population phantom. But this method assumes homogeneous activity and dose distributions in organs and a generalized geometry; hence, it does not consider patient-specific activity distributions and organ anatomy (Gupta et al 2019). More precise methods to address the internal dosimetry at the voxel level have been developed, including the convolution of dose point kernels (Giapa et al 1995) and the voxel S-value approach (Bolch et al 1999, 2009, Amato et al 2012, 2013a, 2013b). However, these methods usually assume a uniform human tissue material density, neglecting the differences between the lung tissue, soft tissue, and bone (Amato et al 2013a, Moghadam et al 2016). These limitations can lead to certain inaccuracies in the dose calculation.

On the other hand, Monte Carlo (MC) methods have long been a computing tool for internal dose distributions at the voxel level. Direct MC simulations coupled with functional and anatomical imaging, such as PET/CT, are considered the gold standards for patient-specific dose estimation (Zaidi 1999, Zaidi and Xu 2007, Neira et al 2020). At present, there are several studies involving patient-specific internal dose calculation using MC methods. For example, Auditore et al (Auditore et al 2019) calculated the internal dosimetry for Trans-Arterial Radio-Embolization (TARE) therapies using MC software GAMOS and SPECT/CT images. Neira et al (2020) calculated individualized dose distributions in patients undergoing $^{18}$F-FDG PET/CT based on MC software GATE and PET/CT images. Piston et al (Pistone et al 2020) compared the dose-rate results between GATE and GOMOS MC simulation based on the same $^{18}$F-Choline PET/CT patient data. However, MC simulations are notoriously slow due to the massive amount of calculations required to reach an acceptable accuracy, which prevents their clinical use. In the last decade, the use of general-purpose graphics processing units (GPUs) for Monte Carlo radiation transport simulations has emerged (Hisioiny et al 2011, Jia et al 2011, 2014, 2015 Pratx and Xing 2011), bringing impressive parallel computational efficiency to the method that was thought unfit for clinical workflow. For fast dose calculation in radionuclide therapy, Frezza et al (2020) developed and validated a GPU-based MC software irtGUMC, which can complete MC simulation in about 1 min with less than 1% uncertainty based on SPECT/CT images of a patient. However, the dose is not evaluated in radionuclide diagnosis such as PET imaging, which has a different physical process from SPECT due to involving positron annihilation. In addition, they only evaluated the dose of kidney in the specific patient. To evaluate the dose of all the key organs of the whole body in a specific patient, an organ autosegmentation tool is necessary, since the manual segmentations of organs in CT images are very time-consuming.

In this study, we combined GPU-accelerated MC simulations and organ autosegmentation techniques to evaluate rapidly the organ dose for a specific patient based on PET/CT images. In our previous work, a dedicated GPU-based MC code, ARCHER (Su et al 2014), has been validated for a wide range of medical physics applications, such as CT imaging and radiotherapy (Xu et al 2015, Adam et al 2020, Lin et al 2017). Here, ARCHER’s capabilities are extended by adding a new NM module dedicated to internal dosimetry for patients undergoing PET/CT examination involving $^{18}$F-FDG PET/CT imaging. To this end, ARCHER-NM is designed to perform rapid individualized MC dose calculations on the CT image data of the patient, with radioactivity distributions constructed from the PET image data. At the same time, the key radiosensitive organs of the whole body are segmented by an autosegmentation technique. Finally, the organ dose rates are obtained, since PET images are acquired at a specific time. The overall computational flow is shown in figure 1. To the best of our knowledge, this is the first study presenting patient-specific internal radiation dose rate calculations combining GPU-accelerated MC technique and autosegmentation technique of whole-body image dataset for a clinically realistic $^{18}$F-FDG PET/CT imaging case. The comparison of organ dose rates of ARCHER-NM against GATE MC simulations is performed using identical $^{18}$F-FDG PET/CT data. This work aimed to validate and benchmark the ARCHER-NM module for internal dosimetry of PET imaging.

2. Methods

2.1. ARCHER-NM workflow

ARCHER-NM can be divided into five key modules: DICOM module, radionuclide module, phase space module, radiation transport module, and organ autosegmentation module. The total algorithm workflow is following:
The DICOM module parses CT images and PET images. The CT images are resampled to obtain the same size as the PET images. ARCHER-NM is designed to use CT images to construct a density map and PET images construct a radioactivity distribution map. The CT images are converted into mass density and material composition using the Hounsfield unit (HU)-to-density conversion curve (Verhaegen et al. 2005).

In this process, four materials, each having a density specified by the HU, are used for the patient phantom: water, dry air, compact bone (defined by ICRU), and lung (defined by ICPR) (Schneider et al. 2000, Kawrakow and Walters 2006).

The radionuclide module determines the type and number of the decay particles. In ARCHER-NM, primary particles are generated according to branching ratios associated with each radionuclide. The four kinds of radionuclides, including $^{18}\text{F}$, $^{11}\text{C}$, $^{15}\text{O}$, and $^{68}\text{Ga}$, commonly used in PET imaging were simulated in this work. The characteristics of those radionuclides are presented in Table 1. Here, the branching ratios and energies for all kinds of radiations of these radionuclides are taken from ICRP107 (Eckerman and Endo 2008). Some branching radiations are ignored due to the low dose contribution. For example, for $^{68}\text{Ga}$, the decay particles would include $5 \times 10^8 \times 88.91\%$ positrons and a total of $5 \times 10^8 \times 3.5783\%$ photons if the simulated decay event is set to be $5 \times 10^8$.

The phase space module generates the decay particles. The PET images indicate the intensity distribution of radionuclides expressed in Bq/ml (Pistone et al. 2020). The probability density distribution of the decay particles in MC simulations is constructed through a linear conversion of the PET value. Within a voxel, the location of the decay particle is determined by uniform sampling. The energy of the decay particle is determined by the data from ICRP107 (Eckerman and Endo 2008), which provides the energy spectrum of decay particles, including positron, electron, or photon, for each radionuclide. Especially, for the continuous energy spectrum, the rejection sampling method is used. The isotropic angular distribution is defined using Marsaglia method to sample uniformly from the surface of a sphere (Marsaglia 1972).

**Table 1.** Energies and branching ratios of the radiations for radionuclides $^{18}\text{F}$, $^{11}\text{C}$, $^{15}\text{O}$, and $^{68}\text{Ga}$. The data is adapted from ICRP107 (Eckerman and Endo 2008)

| Radionuclides | Radiation | Branching ratio (%) | Energy (MeV) |
|---------------|-----------|---------------------|--------------|
| $^{18}\text{F}$ | Beta + | 96.73 | 0.2498$^a$ |
| $^{11}\text{C}$ | Beta + | 99.77 | 0.3856$^a$ |
| $^{15}\text{O}$ | Beta + | 99.90 | 0.7354$^a$ |
| | Beta + | 88.91 | 0.8293$^a$ |
| $^{68}\text{Ga}$ | Gamma rays | 0.0943 | 1.2611 |
| | | 0.0940 | 0.8058 |
| | | 0.0335 | 0.5785 |

$^a$ Denotes that it is the mean energy of the continuous energy spectrum.
(4) The radiation transport module simulates photon-electron transport inside the patient body. The GPU-based photon-electron transport method has been introduced in detail and validated in our previous work about ARCHER (Su et al 2014, Adam et al 2020). For photons, the photoelectric effects, Compton scattering, and pair production are considered. For electron (or positron), the inelastic and elastic collision and bremsstrahlung interaction are considered. A photon or an electron (or positron) is transported in the human CT phantom until its energy falls below cutoff energy. At this point, the MC transport simulation is terminated, and its remaining energy is deposited locally. Especially, in the endpoint of the positron, two annihilation photons of equal energy (0.511 MeV) are generated in opposite directions. The cutoff energies for photons and electrons (or positron) are set to 10 keV and 200 keV, respectively. The deposited dose is calculated for each voxel, and the corresponding uncertainty is evaluated using the batch statistics error estimation method (Walters et al 2002). The dose and uncertainty maps have the same size and resolution as those of the input CT images. The obtained 3D absorbed dose maps are divided by the total number of decay events in the simulation to deduce dose-per-event maps. Multiplying each dose-per-event voxel by the whole body total radioactivity at the PET scan acquisition time \( (t = t_s) \), we calculate dose rate maps \( \hat{D}(t_s) \) (Gy s\(^{-1}\)) at the acquisition time \( t_s \) according to Pistone et al (2020):

\[
\hat{D}(t_s) = \frac{D}{N_{\text{evts}}} \times A(t_s),
\]

where \( D \) (Gy) is the total absorbed dose, \( N_{\text{evts}} \) is the total number of decay events in the MC simulations, and \( A(t_s) \) (Bq) is the whole body total activity measured in the PET images at the acquisition time \( t_s \).

(5) The organ autosegmentation module generates the organ masks. To evaluate the organ dose of patients undergoing PET/CT examination, corresponding organs must be segmented. However, it would be impractical to manually delineate so many organs from the image dataset. Herein, we used the self-developed deep-learning-based organ autosegmentation software DeepViewer (Wisdom Tech, Hefei, China; http://wisdom-tech.com.cn/) to solve this problem. The accuracy of segmentation has been validated in our previous studies (Peng et al 2020, Wang et al 2020). Specifically, the Dice Similarity Coefficients (DSC) is above 0.9 for most organs such as the brain, lung, heart, liver, and so on. In this study, based on the CT images, 28 key organs of the whole body were segmented. To ensure the accuracy of the segmentation, the segmentation results were checked by a clinical doctor.

ARCHER-NM simulations are executed in a Windows 10 computer system. The hardware includes: (a) GPU—Nvidia Titan RTX Graphics Card with 24 GB memories, and (b) CPU—Intel® Core™ i7-7700 @ 3.60GHz with 16 GB memories.

2.2. GATE setup

In this study, GATE (version 9.0) of Geant4 (version 10.6.2) is used. The CT and PET images used in GATE are the same as those of ARCHER-NM. The nested parametrization and the corresponding navigation algorithm are used to create parametrized volumes. The positron source and photon source are generated by using the UserSpectrum feature, which is similar to the current ARCHER-NM settings. The physical model \( \text{emstandard}_{\text{opt3}} \) is used to process the photon–electron transport. Compared with ARCHER-NM, the difference is that the cutoff range is adopted in GATE to set the final deposit energy of the particles. In this work, the cutoff ranges for photons and electrons (or positron) are set to 1 mm and 0.45 mm (corresponds approximately 200 keV cutoff energy in water), respectively. The \( \text{DoseActor} \) is used to calculate the absorbed dose. GATE simulations are executed in a Linux server with the Intel® Xeon® Gold 5120T CPU @ 2.20 GHz, 28 physical CPU cores, and a total of 64 GB memories. To improve the calculation speed of GATE, 25 processes are used for each simulation experiment of GATE. For multi-processes running, multiple jobs are generated by the job splitter tool which is bundled in GATE source code. The package ‘gatetools’ are used to merge the results of all processes and generate the final dose and uncertainty.

2.3. Simulation experiments

First, we perform the simulation experiments of the point ion sources in water, including \(^{18}\text{F}, ^{11}\text{C}, ^{15}\text{O}, \) and \(^{68}\text{Ga}. \) Here, we create CT images and PET images with a size of \( 51 \times 51 \times 51 \) and a voxel spacing of \( 4 \text{ mm} \times 4 \text{ mm} \times 4 \text{ mm}. \) All voxels in CT images are assigned a value of 0 to simulate the water phantom. All the voxels in PET images are assigned a value of 0 except for the voxel at the center which is assigned a value of 1 as a point source. The number of simulated decay events is \( 5 \times 10^7 \) for both the GATE and ARCHER-NM.

Second, we perform internal radiation dose rate calculations for a whole-body \(^{18}\text{F}-\text{FDG} \) PET imaging patient. One set of PET/CT data for an adult male is collected from the Department of Nuclear Medicine of the First Affiliated Hospital of the University of Science and Technology of China (Hefei, China). This patient is 52
years old, 170 cm in height, and 79 kg in weight. The amount of injected $^{18}$F-FDG is 8.0 mCi. After a 60 min waiting period when the patient has urinated, whole-body static PET scanning was performed on a PET/CT scanner using multiple bed positions, with 180 s per bed position. PET attenuation corrected images were reconstructed (axial plane) using an ordered subset expectation maximization (OSEM) iterative algorithm. The size of PET images is $168 \times 168 \times 219$, and the voxel spacing is $4.063 \text{ mm} \times 4.063 \text{ mm} \times 5 \text{ mm}$. The CT images are resampled to obtain the same size as the PET images. The number of simulated decay events is $5 \times 10^8$ for both GATE and ARCHER-NM.

2.4. Evaluation standard

For the dose distribution calculation of a point ion source in water, the dose-distance curves are compared between GATE and ARCHER-NM. Besides, the average dose is also compared in the dose region of the cube with a side length of 40 mm centered on the point source. For the dose rate calculation of $^{18}$F-FDG PET imaging patients, the average dose rate $\bar{d}(t_i)$ and corresponding uncertainties $\varepsilon$ in the organs are calculated according to Chetty et al (2006):

$$\bar{d}(t_i) = \frac{1}{N} \sum_{i=1}^{N} D_i(t_i)$$

$$\varepsilon = \sqrt{\frac{1}{N} \sum_{i=1}^{N} \varepsilon_i^2},$$

where $N$ is the number of voxels in an organ, $D_i(t_i)$ and $\varepsilon_i$ are the dose rate and uncertainty of voxel $i$ in the acquisition time $t_i$, respectively. Taking GATE as a reference, the relative percent differences $\sigma$ are calculated according to:

$$\sigma = 100 \times \frac{\bar{d}(t_i)_{\text{ARCHER-NM}} - \bar{d}(t_i)_{\text{GATE}}}{\bar{d}(t_i)_{\text{GATE}}}.$$

The dose rate maps of sagittal and coronal slices for GATE and ARCHER-NM are also compared. In addition, the simulation times for GATE and ARCHER-NM are analyzed.

3. Results

3.1. Point ion source

Figure 2 shows the dose-distance curves for the four kinds of point ion sources in water phantom, including $^{18}$F, $^{11}$C, $^{15}$O and $^{68}$Ga. In the dose region of the cube with a side length of 40 mm centered on the radioactive source, the relative percent differences of the average dose between ARCHER-NM and GATE are summarized in table 2, taking GATE as a reference. The excellent dose results from ARCHER-NM suggest that the radiological physics aspects of the new module are modeled correctly. In terms of computing efficiency, the average time of 25 processes is reported in table 2 for each simulation experiment of GATE. It is found that GATE always takes more than 700 seconds to run these MC simulations, while ARCHER-NM takes only less than 3 s for the same number of simulations. In other words, while the dose results are practically identical, ARCHER-NM is almost 300 times faster than GATE for the chosen experiment.

3.2. $^{18}$F-FDG PET imaging

For the dose rate calculation involving the $^{18}$F-FDG PET imaging patient, the minimum time of 25 processes in GATE is about 180 min, while ARCHER-NM takes 22 s. ARCHER-NM is more than 400 times faster than GATE for simulations involving the whole body of the patient anatomy. Figure 3 shows the dose rate maps (single image plane) in sagittal and coronal views for the $^{18}$F-FDG PET imaging patient. The results of GATE and ARCHER-NM are displayed in the first column and second column, respectively. The relative difference maps are displayed in the third column, taking the GATE results as a reference. Here, the dose rates in the air are set to zero. The results indicate that the dose rate distributions are nearly identical between GATE and ARCHER-NM. The high-dose-rate areas appear to localize in the brain, as expected.

The segmentation results of a total of 28 organs are visualized in figure 4, in terms of sagittal and coronal 2D views as well as the 3D views. The total organ segmentation time was approximately 12 min consisting of an autosegmentation time of 7 min and a radiation oncologist’s check-up time of 5 min.

Figure 5 compares the average dose rate for 28 organs of this patient for GATE and ARCHER-NM. It can be seen that the high-dose-rate areas are mainly in the head—including the brain, brainstem, optic chiasm, optic nerve, and pituitary—followed by the bladder and liver. The average dose rate and corresponding statistical uncertainty of these 28 organs are summarized for GATE and ARCHER-NM in table 3. The relative percent
differences are reported, taking GATE results as reference. The dose statistical uncertainties of these 28 organs range from 1.15% to 3.56% for GATE and 0.73% to 3.01% for ARCHER-NM. The relative percent differences for these 28 organs range from $-2.45\%$ to $2.58\%$. The largest dose rate difference occurs in the lung. The average absolute value of relative percent differences for 28 organs is $1.34\%$. Considering the statistical uncertainty, results from ARCHER-NM agree excellently with those from the GATE.

### 4. Discussion

The limitation of this study is that ARCHER-NM provides only dose rate information as the calculations are based on PET images acquired at a certain time. To obtain the total organ dose information for internally deposited radionuclides, we would need multiple PET images at different acquisition time points. Alternatively, of course, it is feasible to use biokinetic models to calculate the total radioactivity distribution (Mattsson et al 2015). Furthermore, although only the radionuclides used in PET imaging are simulated, including $^{18}F$, $^{11}C$, $^{15}O$, $^{68}Ga$. The voxel size of the dose matrix is $4\text{ mm} \times 4\text{ mm} \times 4\text{ mm}$.

Figure 2. The comparison of dose-distance curves between ARCHER-NM and GATE for the four kinds of point ion source in water phantom, including (a) $^{18}F$, (b) $^{11}C$, (c) $^{15}O$, and (d) $^{68}Ga$. The voxel size of the dose matrix is $4\text{ mm} \times 4\text{ mm} \times 4\text{ mm}$.

Table 2. The comparison of the average dose between ARCHER-NM and GATE in the dose region of the cube water phantom with a side length of 40 mm centered on the point source.

| Radionuclides | ARCHER-NM | GATE | Relative differences (%) |
|--------------|-----------|------|--------------------------|
|              | Dose per decay (Gy) | Statistical uncertainty (%) | Calculation time (s) | Dose per decay (Gy) | Statistical uncertainty (%) | Calculation time (s) |
| $^{18}F$     | 1.071E-12 | 0.92 | 2.6                      | 1.059E-12 | 0.89 | 712 | 1.08 |
| $^{11}C$     | 1.533E-12 | 0.90 | 2.4                      | 1.545E-12 | 0.88 | 787 | 0.54 |
| $^{15}O$     | 2.723E-12 | 0.89 | 2.7                      | 2.705e-12 | 0.88 | 956 | 0.85 |
| $^{68}Ga$    | 2.723E-12 | 0.95 | 2.8                      | 2.685E-12 | 0.93 | 880 | 1.44 |
and $^{68}$Ga, ARCHER-NM can be readily extended to more radionuclides used in SPECT, such as $^{177}$Lu and $^{131}$I, that emit gamma and beta radiation. Combined with the organ autosegmentation technique, ARCHER-NM will have the ability to achieve rapid organ dose assessment for nuclear medicine in the future. This work will provide an excellent research direction for patient-specific rapid organ dose evaluation in nuclear medicine.

**Figure 3.** Comparison of dose rate maps (single image plane) in sagittal and coronal views between results from GATE (a), (d) and ARCHER-NM (b), (c) for the $^{18}$F-FDG PET imaging patient. Relative error maps are displayed (c), (f) taking GATE results as reference.

**Figure 4.** Visual display of organ autosegmentation results for the $^{18}$F-FDG PET imaging patients.
Figure 5. Comparison of organ dose rate results showing excellent agreement between GATE and ARCHER-NM for an 18F-FDG PET imaging patient.

Table 3. Comparison of the average dose rate in 28 organs calculated by GATE and ARCHER-NM for an 18F-FDG PET imaging patient. The percent differences are reported taking GATE results as reference.

| Organs          | GATE dose rate (Gy s\(^{-1}\)) | Statistical uncertainty (%) | ARCHER-NM dose rate (Gy s\(^{-1}\)) | Statistical uncertainty (%) | Relative difference (%) |
|-----------------|---------------------------------|-----------------------------|------------------------------------|-----------------------------|-------------------------|
| Bladder         | 7.19E-07                        | 1.38                        | 7.32E-07                          | 0.84                        | 1.86                    |
| Bowel           | 2.43E-07                        | 2.25                        | 2.45E-07                          | 1.78                        | 0.99                    |
| Brain           | 1.00E-06                        | 1.15                        | 1.02E-06                          | 0.78                        | 2.10                    |
| Brain stem      | 8.02E-07                        | 1.23                        | 8.18E-07                          | 0.89                        | 2.07                    |
| Esophagus       | 3.58E-07                        | 1.87                        | 3.61E-07                          | 1.36                        | 0.70                    |
| Left eyeball    | 3.16E-07                        | 1.94                        | 3.20E-07                          | 1.51                        | 1.43                    |
| Right eyeball   | 2.97E-07                        | 2.01                        | 3.00E-07                          | 1.59                        | 1.21                    |
| Heart           | 3.39E-07                        | 1.87                        | 3.43E-07                          | 1.39                        | 1.12                    |
| Left kidney     | 3.78E-07                        | 1.79                        | 3.82E-07                          | 1.28                        | 1.22                    |
| Right kidney    | 3.77E-07                        | 1.77                        | 3.82E-07                          | 1.29                        | 1.25                    |
| Larynx          | 2.59E-07                        | 2.14                        | 2.58E-07                          | 1.57                        | −0.31                   |
| Liver           | 4.13E-07                        | 1.70                        | 4.16E-07                          | 1.21                        | 0.95                    |
| Left lung       | 2.89E-07                        | 3.52                        | 2.82E-07                          | 2.95                        | −2.22                   |
| Right lung      | 2.98E-07                        | 3.56                        | 2.91E-07                          | 3.01                        | −2.45                   |
| Mandible        | 2.13E-07                        | 2.03                        | 2.08E-07                          | 1.67                        | −2.30                   |
| Optic chiasm    | 8.47E-07                        | 1.22                        | 8.55E-07                          | 0.78                        | 0.98                    |
| Left optical nerve | 1.06E-06                     | 1.15                        | 1.07E-06                          | 0.73                        | 0.85                    |
| Right optical nerve | 8.12E-07                   | 1.30                        | 8.32E-07                          | 0.87                        | 2.51                    |
| Pancreas        | 3.44E-07                        | 1.81                        | 3.47E-07                          | 1.35                        | 0.84                    |
| Left parotid    | 2.68E-07                        | 2.16                        | 2.72E-07                          | 1.61                        | 1.61                    |
| Right parotid   | 2.98E-07                        | 2.08                        | 3.05E-07                          | 1.48                        | 1.61                    |
| Pelvis          | 2.46E-07                        | 2.00                        | 2.47E-07                          | 1.50                        | 0.49                    |
| Pituitary       | 5.35E-07                        | 1.45                        | 5.49E-07                          | 1.21                        | 2.58                    |
| Rectum          | 2.35E-07                        | 2.15                        | 2.57E-07                          | 1.73                        | 0.63                    |
| Spinal cord     | 3.21E-07                        | 1.91                        | 3.25E-07                          | 1.43                        | 1.09                    |
| Spleen          | 2.99E-07                        | 1.97                        | 3.02E-07                          | 1.44                        | 1.00                    |
| Stomach         | 2.62E-07                        | 2.06                        | 2.61E-07                          | 1.71                        | −0.27                   |
| Thyroid gland   | 2.40E-07                        | 2.22                        | 2.42E-07                          | 1.62                        | 0.88                    |
5. Conclusion

In this study, a GPU-accelerated MC dose calculation code dedicated to nuclear medicine internal dosimetry is developed and validated. We first validate the code by calculating the dose distribution in a water phantom for some point ion sources, including $^{18}$F, $^{11}$C, $^{15}$O and $^{68}$Ga. Then, the validated code is applied to the calculations of dose rate distributions for an $^{18}$F-FDG PET imaging patient. The results of both cases show that there is an excellent agreement between ARCHER-NM and the widely used MC code GATE. In terms of calculation speed, GATE would need about 180 min to achieve acceptable statistical accuracy for the absorbed dose of $^{18}$F-FDG PET imaging patients, while ARCHER-NM only needs 22 s. ARCHER-NM presents an enormous advantage compared with GATE. Together with the autosegmentation tool, accurate and fast patient-specific organ dose assessment is feasible for routine PET/CT imaging and radionuclide therapy procedures.

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