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FULL PAPER

Effective dose to staff members in a positron emission tomography/CT facility using zirconium-89

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Objective: Positron emission tomography (PET) using zirconium-89 (⁸⁹Zr) is complicated by its complex decay scheme. In this study, we quantified the effective dose from ⁸⁹Zr and compared it with fluorine-18 fludeoxyglucose (¹⁸F-FDG).

Methods: Effective dose distribution in a PET/CT facility in Riyadh was calculated by Monte Carlo simulations using MCNPX. The positron bremsstrahlung, the annihilation photons, the delayed gammas from ⁸⁹Zr and those emissions from ¹⁸F-FDG were modelled in the simulations but low-energy characteristic X-rays were ignored.

Results: On the basis of injected activity, the dose from ⁸⁹Zr was higher than that of ¹⁸F-FDG. However, the dose per scan from ⁸⁹Zr became less than that from ¹⁸F-FDG near the patient, owing to the difference in injected activities. In the corridor and control rooms, the ⁸⁹Zr dose

was much higher than ¹⁸F-FDG, owing to the difference in attenuation by the shielding materials.

Conclusion: The presence of the high-energy photons from ⁸⁹Zr-labelled immuno-PET radiopharmaceuticals causes a significantly higher effective dose than ¹⁸F-FDG to the staff outside the patient room. Conversely, despite the low administered activity of ⁸⁹Zr, it gives rise to a comparable or even lower dose than ¹⁸F-FDG to the staff near the patient. This interesting result raises apparently contradictory implications in the radiation protection considerations of a PET/CT facility.

Advances in knowledge: To the best of our knowledge, radiation exposure to staff and public in the PET/CT unit using ⁸⁹Zr has not been investigated. The ultimate output of this study will lead to the optimal design of the facility for routine use of ⁸⁹Zr.

Positron emission tomography (PET)/CT scanners have quickly become the preferred technology for PET imaging, as the integration of functional PET images with the anatomical visualisation of CT has allowed more accurate diagnosis [1]. However, the doses delivered to patients and staff owing to the procedure should be considered and should be kept as low as reasonably achievable (ALARA) [2–4], *i.e.* the dose of a classified staff member should not exceed one-third of the annual dose limit.

Radiation exposure of technical staff operating PET units with the use of fluorine-18 fludeoxyglucose (¹⁸F-FDG) has been a controversial issue in the use of high-energy photons (511 keV), added to by a sharp increase in the number of scans undertaken using a limited number of PET/CT facilities [5–7]. However, the doses received by staff in PET/CT procedures may vary considerably among facilities depending on the practices, positron emitter used, facility layout, injected activities and patient workload. Lately, several studies [5–7] have been carried out to assess and optimise radiation exposure to staff in the PET/CT unit

when using ¹⁸F-FDG and these studies were in agreement with a slight variation owing to different facilities and unit design. Alsafi et al [5] showed that an average staff dose per patient is 5.7 ± 2.7 and $5.0 \pm 1.7 \mu\text{Sv}$ per procedure for mobile and static facilities, respectively, using ¹⁸F-FDG. The study of Al-Haj et al [6] reported that the estimated dose at the waiting area was $0.8 \mu\text{Sv}$ per procedure. Dalianis et al [7] showed that the staff (nurse) dose during injection was in the range of 2.7–4.0 mSv and the dose to the technologist during positioning was in the range of 3.5–5.0 mSv over 10 months. Al-Haj et al [6] revealed that the estimated doses to staff working in PET/CT were within safe limits but would increase with patient workload and stay time. The –1 annual increase in patient workload necessitates that staff dose estimation be done and the protocol be reviewed for possible decrease in the ¹⁸F-FDG activity to be administered. Elschot et al [8] studied the shielding requirement of a PET/CT facility using Monte Carlo methods with the MCNPX code [9] and the Medical Internal Radiation Dose full body phantom [10]. They found that the self-absorption of ¹⁸F-FDG emission by the

patient's body reduced the dose to the clinical staff by a factor of 0.36—similar to the American Association of Physicists in Medicine (AAPM) recommendation [11], but the recommendation also overestimated the shielding requirements.

Immuno-PET, the combination of PET with monoclonal antibodies (mAbs), is an attractive novel option to improve diagnostic tumour characterisation and guide mAb-based therapy in certain cancers such as metastatic breast cancer [12]. ^{89}Zr has recently drawn significant interest as a promising metallo-radionuclide for use in immuno-PET, owing to favourable decay characteristics [13–15]. Furthermore, ^{89}Zr can be produced in medical cyclotrons through the proton irradiation of an yttrium-89 target [15–18]. When ^{89}Zr PET imaging is commonly used, a higher dose to the patient and staff is more likely to be expected, owing to 511 keV gamma photons from positron annihilation and delayed photons (909 keV at an abundance of 99.9%). To the best of our knowledge, radiation exposure to staff and the public in the PET/CT unit using ^{89}Zr has not been investigated. This necessitates the assessment and optimisation of radiation exposure to staff and the public before ^{89}Zr radionuclide is commonly used.

In this work, we studied an existing PET/CT facility at King Faisal Specialist Hospital and Research Centre, Riyadh, Saudi Arabia, using the ^{89}Zr -labelled mAbs in immuno-PET imaging. The radiation protection shielding of the facility was primarily designed for ^{18}F -FDG. We estimated the effective dose to staff with Monte Carlo techniques. We used MCNPX v. 2.5.0 to simulate the ^{18}F -FDG and ^{89}Zr emissions from a voxelised anthropomorphic phantom. The simulation geometry was derived from the floor plan of the facility, and the phantom was positioned in the examination room where the PET/CT scanner was located and in the injection room. The effective dose was calculated using photon fluence-to-effective dose conversion coefficients. There are many variables affecting the dose during daily clinical work, such as shielding, workload, injected activity, uptake time between injection and imaging, clearance of the tracer and time spent by the staff member in the vicinity of the patient. This study attempted to account for as many factors as possible. We estimated the effective dose from ^{18}F -FDG and ^{89}Zr at different locations throughout the facility and identified some hotspots in it.

MATERIALS AND METHODS

MCNPX

MCNPX is a general-purpose Monte-Carlo package developed by the Los Alamos National Laboratory. It is capable of tracking 34 radiation types, including photons, electrons and positrons, which were of interest to this study. The photon physics takes into account Rayleigh scattering, Compton scattering and the photoelectric effect at energies below 1 MeV. The electron and positron interactions include elastic and inelastic scattering approximated by multiple scattering theory. Charged particles are tracked in an energy grid that is calculated at the beginning of a simulation. The grid determines the particle's track length at each step. Bremsstrahlung production is sampled randomly along the track. In MCNPX, electrons and positrons share the same stopping power. The difference between electron and

positron tracking is the creation of annihilation photons at the end of the positron track when the energy is below the cut-off energy. Since charged particle transport is computationally expensive, MCNPX offers photon-only transport. In this case, all the photon interactions mentioned earlier are still available but electrons are not created after Compton scattering and the photoelectric effect. The thick-target bremsstrahlung model is invoked to generate the bremsstrahlungs instead of tracking the electrons. Furthermore, MCNPX v. 2.5.0 does not have the option for positrons as a source. Therefore, we adopted the following strategy to study the dose to staff members from the positron decays in ^{18}F -FDG and ^{89}Zr .

Simulations

The production of bremsstrahlung X-rays from electrons and positrons are usually of very low energy in low-atomic-number materials. They will be absorbed mostly by the patient. However, some bremsstrahlungs from the high-energy positrons as the radioactive nuclides decay may have some effect in the vicinity of the patient but not outside the scanner and injection rooms. Thus, we characterised the effective dose owing to positron bremsstrahlung production using a spherical phantom 30 cm in diameter. The positron bremsstrahlungs were approximated by those from the electrons of the same energy. This approximation will introduce an error in the range of 10% because of the difference in electron and positron stopping power. MCNPX was run in photon–electron mode. The effective dose was estimated in concentric spheres around the phantom using photon fluence-to-effective dose conversion coefficients.

For the effective dose in the entire facility, only the higher energy photon emissions were modelled in the voxelised anthropomorphic phantom [19]. The photon energies included in the simulations are the 511 keV annihilation photons from ^{18}F -FDG and ^{89}Zr and the 909 keV delayed gamma from ^{89}Zr . Each energy was simulated separately and the results combined according to the photon intensity per radioactive decay, this is summarised in Table 1. In these simulations, MCNPX was run in photon-only mode.

The simulation geometry was derived from the hospital floor plan (Figure 1). The rooms were partitioned by barite concrete. The injection and the scanner rooms were further lined with lead of tenth-value layer thickness. The lead-lined steel doors were approximated by iron with an appropriate lead layer. The PET/CT scanner was not modelled explicitly but the phantom was positioned at the centre of where the couch should be. A second phantom was positioned at the centre of the injection room in an upright posture. The radionuclide distribution in the phantom was assumed to be 60% in the liver, 20% in the bladder and 20% in the muscle. In each case, the photon emission was sampled isotropically. The source positions in the phantom in the scanner room were also plotted in Figure 1.

The estimation of the effective dose to staff was computed in a grid of 10×10 cm using photon fluence-to-effective dose conversion. MCNPX gave the results in Sieverts per hour, assuming one source photon per second. Since the MCNPX mesh tally is superimposed onto the geometry, there are volumes in

Table 1. Summary of emission radiation types and energies from ^{89}Zr and ^{18}F decays.

Radiation type	Energy (keV)	Intensity (%)
^{89}Zr , ϵ : 100%		
* β_+	(902.3) 395.5	22.74
Auger L	1.91	78.6
Auger K	12.7	19.2
Characteristic X-ray l	1.92	2.34
Characteristic X-ray $k\alpha_2$	14.883	14.1
Characteristic X-ray $k\alpha_1$	14.958	27.0
Characteristic X-ray $k\beta_3$	16.726	2.05
Characteristic X-ray $k\beta_1$	16.738	3.97
Characteristic X-ray $k\beta_2$	17.013	0.76
a Annihilation γ	511.0	45.5
a Delayed γ	909.15	99.04
Delayed γ	1620.8	0.073
Delayed γ	1657.3	0.106
Delayed γ	1713.0	0.745
^{18}F , ϵ : 100%		
β_+	(633.5) 249.8	96.73
Auger K	0.52	3.072
Characteristic X-ray $k\alpha_2$	0.525	0.017
Characteristic X-ray $k\alpha_1$	0.525	0.009
a Annihilation γ	511.0	193.46

^{18}F , fluorine-18; ^{89}Zr , zirconium-89.

a Radiation was included in this study.

The data come from Brook Haven National Laboratory (www.nndc.bnl.gov/chart). For the positrons, the end-point energy is listed in brackets and the mean energy without.

Figure 1. Simulation geometry with the Zubal phantom placed in the examination room. The phantom outline was created from plotting 100 000 randomly sampled source positions in it. It was superimposed on the original floor plan. This plot was part of the work to check the correctness of the MCNPX model. Furthermore, the line across the phantom indicates the position of where the dose rate is plotted in Figure 2.

the mesh that contain materials other than air, such as the walls and the tissues of the phantom. In those volumes, the photon fluence is still converted into an effective dose by MCNPX using the conversion coefficients. The dose in any volumes that are not purely air is meaningless. In the analysis of the dose in each room, those meaningless dose values were excluded, although they were plotted in the dose maps as uninterrupted visualisation of the distribution.

One billion histories were tracked in each simulation of 511 keV and 909 keV photons as sources in the phantoms. In the case of the 909 keV simulation, the relative errors in the air-only tallies were below 1% and the larger relative errors (above 5%) occurred in the walls of the holding room (H). For the 511 keV simulation, the relative errors in the air-only tallies were below 1% in all rooms except the holding room, in which the relative errors were between 1% and 5%. They were above 10% in the walls of the holding room. Since all the tally volumes that overlapped the walls were excluded from the analysis, the large relative errors did not affect our results and discussions. Therefore, the relative errors in the ^{18}F -FDG results were at most 5% but generally below 1%, whereas those in the ^{89}Zr results were below 1% after propagation of errors.

Combining monoenergetic results for each radionuclide

For ^{18}F -FDG, the intensity of the 511 keV photons is 193.46% per disintegration. Thus, the mesh result weighted by the intensity is the effective dose per disintegration coming from the annihilation component. Ignoring the low-energy emissions, this is the effective dose from ^{18}F -FDG per disintegration per second, *i.e.* $\text{Sv h}^{-1} \text{Bq}^{-1}$. This is converted into $\text{mSv h}^{-1} \text{MBq}^{-1}$ by multiplying a factor of 1×10^9 . It should be noted that it is a pure coincidence that the factor is the same as the number of histories in each simulation.

For ^{89}Zr , the intensities of annihilation photons and the 909 keV delayed gammas per disintegration are 45.5% and 99.04%, respectively. The effective dose from each component was derived in the same way as that of ^{18}F -FDG. They were summed to give the effective dose per disintegration. Other emissions of low energy and low intensity were ignored in the calculations.

The effective dose per scan using each radionuclide By multiplying appropriate factors that include initial activity, the decay constant and scan time, the dose per scan was derived. For simplicity, both phantoms in the scanner and injection rooms had the same initial activity. The scan time was assumed for 30 min. We calculated the doses for 250 MBq and 500 MBq of ^{18}F -FDG and 75 MBq of ^{89}Zr .

RESULTS AND DISCUSSION

Spherical water phantom

Figures 2–4 summarise the results of the simulations using a spherical phantom 30 cm in diameter. The effective dose rate of ^{89}Zr was 6% higher than that of ^{18}F -FDG (Figure 2). At 1.35 m from the phantom, the ^{89}Zr dose rate was estimated to be $6.04 \times 10^{-2} \mu\text{Sv h}^{-1} \text{MBq}^{-1}$ and that of ^{18}F -FDG was $5.76 \times 10^{-2} \mu\text{Sv h}^{-1} \text{MBq}^{-1}$. The ^{18}F -FDG dose rate is

Figure 2. Effective dose rate in the vicinity of the phantoms. The spherical phantoms were 30 cm in diameter. The Zubal phantom data were averaged over the left- and right-hand sides of the phantom near the pelvis as indicated by the line in Figure 1.

equivalent to $32.0 \mu\text{Sv h}^{-1}$ at 555 MBq, which is comparable with the American Association of Physicists in Medicine (AAPM) value [11] of $30 \mu\text{Sv h}^{-1}$ at 1 m. The estimated dose rates decreased to $1.53 \times 10^{-2} \mu\text{Sv h}^{-1} \text{MBq}^{-1}$ and $1.47 \times 10^{-2} \mu\text{Sv h}^{-1} \text{MBq}^{-1}$ at 2.85 m for ^{89}Zr and ^{18}F -FDG, respectively.

Since the injected activity of a radionuclide depends on its application, this amount varies from radionuclide to radionuclide and from examination to examination. A typical injected activity of ^{18}F -FDG ranges from 250 MBq [20] to 500 MBq [11], whereas the typical injected activity of ^{89}Zr is around 75 MBq [13]. Figure 3 shows the effective dose per scan under those conditions. It is interesting to note that the dose per scan of ^{89}Zr became 34% of the dose from a 250 MBq scan of ^{18}F -FDG and 17% of the ^{18}F -FDG scan at 500 MBq. The derived dose rate constants are $0.131 \mu\text{Sv m}^2 \text{MBq}^{-1} \text{h}^{-1}$ for ^{18}F -FDG and $0.137 \mu\text{Sv m}^2 \text{MBq}^{-1} \text{h}^{-1}$ for ^{89}Zr . Our dose rate constant derived with a spherical phantom is less than the AAPM value [11] of $0.143 \mu\text{Sv m}^2 \text{MBq}^{-1} \text{h}^{-1}$ because the AAPM value assumed a point source, whereas our value came from a finite volume.

Figure 4 shows the contributions by the delayed gamma, the annihilation photons and bremsstrahlungs in the ^{89}Zr dose. Owing to its high energy (909 keV) and the high intensity (99%), the delayed gamma was contributing about 77.5% of the total dose, whereas the annihilation photons were contributing <22.5%. The bremsstrahlungs accounted for the remaining 0.03% using the average positron kinetic energy of 395.5 keV. When the maximum kinetic energy of 902.3 keV was used, the bremsstrahlung dose increased fourfold. Since the positron energy spectrum is skewed towards the lower energies, it is safe to conclude that the bremsstrahlung contribution is only a fraction of 1%. It should be noted that the bremsstrahlung contribution was approximated by the electron bremsstrahlung in this study. The radiative stopping power of positrons is only 46.5% of the electrons' radiative stopping power of the same

Figure 3. Comparing the effective dose per scan from ^{18}F -FDG at 250 MBq and 500 MBq injected activity to 75 MBq of zirconium-89 (^{89}Zr). The scan time was assumed to be 30 min.

Figure 4. Components of the effective dose from zirconium-89 (^{89}Zr). The delayed gamma energy is 909 keV, and the bremsstrahlungs are generated by the positrons of 395.5 keV.

energy for the atomic number equal to 8 [21]. Our approximation using electrons in the simulations overestimated the bremsstrahlung production of the positrons. Thus, the positron bremsstrahlungs do not have any significant effect on the estimated dose. This also confirms that bremsstrahlungs were ignored in the effective dose calculations using the voxelised anthropomorphic phantom.

Zubal phantom

Figure 5a shows the effective dose distribution in the PET/CT facility using ^{89}Zr . A Zubal phantom was placed in the supine position inside the examination room and another one in the upright position in the adjacent injection room. Both of them had the same distribution of radionuclides in the body and the same initial activity for simplicity. A similar calculation was carried out for ^{18}F -FDG. The effective dose is summarised in Table 2. In the control room, the average dose rate from ^{89}Zr is $0.15 \text{ nSv h}^{-1} \text{ MBq}^{-1}$. However, the average is $0.71 \text{ nSv h}^{-1} \text{ MBq}^{-1}$ along the 1.5-m-wide corridor by the rooms. The figure clearly shows that the dose outside the injection room is coming from the patient inside and the patient in the examination room. It highlights a radiation hotspot in the facility. The maximum dose rate in the hotspot is $7.02 \text{ nSv h}^{-1} \text{ MBq}^{-1}$, which is 10 times the average along the corridor. The cause of this hotspot is primarily dictated by the inverse square law—the proximity to the patient inside this small room.

As in the case of the spherical phantom, we estimated the dose contributions from the delayed gammas and the annihilation photons. Inside the examination and injection rooms with the phantoms, the delayed gammas are causing about three times the dose from the annihilation photons (Figure 5b). This is consistent with the simulations with the spherical phantom. Outside those two rooms, the contribution from the delayed gammas increases significantly, owing to the preferential

attenuation of the different energy photons. The shielding design was based primarily on the 511 keV photons. Because the delayed gammas are of a substantially higher energy at 909 keV, they are more penetrating. Together with the fact that the delayed gamma intensity is 99.04%, whereas the annihilation photon intensity is only 45.5%, the dose from the delayed gammas is between 10 and 100 times that of the annihilation photons. In other words, 90–99% of the dose outside the patient rooms is coming from the delayed gammas.

Comparing the dose from ^{18}F -FDG and ^{89}Zr

When compared on the basis of normalised initial activity (Figure 5c and Table 2), i.e. on the basis of per MBq, the dose from ^{18}F -FDG and ^{89}Zr differs by 16–19% inside the examination room and the injection room. In the control room, the ^{89}Zr dose rate ($145 \text{ pSv h}^{-1} \text{ MBq}^{-1}$) is five times that of ^{18}F -FDG ($27 \text{ pSv h}^{-1} \text{ MBq}^{-1}$). The most affected location is the holding room next to the injection room. The ^{89}Zr dose rate ($57 \text{ pSv h}^{-1} \text{ MBq}^{-1}$) is 23 times that of ^{18}F -FDG ($2.5 \text{ pSv h}^{-1} \text{ MBq}^{-1}$), owing to the high energy and high intensity of the delayed gammas and the small size of the injection room.

Comparing the dose using the two different phantoms

In the simulations using the spherical phantom, the radionuclides were assumed to distribute uniformly; they had higher concentrations near the surface than in a real patient. Therefore, the simulation was likely to give a higher dose and dose rate than it should be. Figure 1 plots the dose rates laterally across the Zubal phantom in the examination room. The dose rate was averaged over the left and right sides of the phantom. For the spherical phantom, the dose rate was calculated in concentric spheres around it. At 1 m from the surface, the spherical phantom gave a dose rate of $114 \text{ nSv h}^{-1} \text{ MBq}^{-1}$ from ^{89}Zr , whereas the Zubal phantom gave only $25 \text{ nSv h}^{-1} \text{ MBq}^{-1}$. At the same distance, the ^{18}F -FDG dose rate was $109 \text{ nSv h}^{-1} \text{ MBq}^{-1}$ with the spherical and $21 \text{ nSv h}^{-1} \text{ MBq}^{-1}$ with the Zubal phantom. Since the radionuclides were located mostly deep inside the Zubal phantom in the liver and bladder, the attenuation of the body reduced the dose rate by a factor of 5 compared with the spherical phantom results. Additionally, the different penetration power of the prompt gammas and the annihilation photons resulted in a high ratio of ^{89}Zr dose to the ^{18}F -FDG dose. With the spherical phantom, the dose rate from ^{89}Zr was only 5% higher than the ^{18}F -FDG dose rate. The ratio increased to 1.16 in the examination room and 1.19 in the injection room.

Further considerations and limitations

In the simulations with the Zubal phantom, the biological distribution of the radionuclide was assumed to be 60% in the liver, 20% in the bladder and 20% in the muscle for both ^{18}F -FDG and ^{89}Zr . The distribution depends on the pharmacokinetics of the tagged compound and the disease in the patient. In the case of ^{18}F -FDG, it is circulated through the blood vessels. Our assumed distribution was a reasonable approximation because some glucose accumulates in the liver and its metabolic byproducts will be excreted through the bladder. In the case of ^{89}Zr , the tagged compound, monoclonal antibody, is a protein that does not normally

Figure 5. Dose maps of zirconium-89 (^{89}Zr) in the positron emission tomography (PET)/CT facility showing (a) the overall dose rate per MBq of injected activity, (b) the relative contribution by the delayed γ and the annihilation photons, (c) the relative dose with respect to fluorine-18 fludeoxyglycose (^{18}F -FDG) and on the basis of injected activity and (d) the relative dose on the basis of scan procedure. The rooms in the facility are the holding room (H), the injection room (I), the examination or scanner room (E), the control room (C) and the reading room (R).

accumulate in the liver or the bladder unless the organs are diseased. Thus, the assumed biological distribution does not represent a typical case. The distribution is important when assessing the organ dose to the patient. However, it is less important in the dose estimation to staff. We chose them to be the same for simplicity and to make the distribution inhomogeneous in the phantom. In particular, the radionuclide concentrated more deeply inside the phantom instead of near the surface. The net effect was a lower estimation of the dose to the staff in the vicinity of the patient.

The measurements by Al-Haj et al [6] suggested that the ^{18}F -FDG dose rate at 1 m from a patient with 505 MBq injected activity was $47 \mu\text{Sv h}^{-1}$. This is similar to the value $46 \mu\text{Sv h}^{-1}$ estimated with the AAPM dose rate constant and the self-absorption correction factor for the same injected activity and the same distance. However, the spherical phantom gave $55 \mu\text{Sv h}^{-1}$, whereas the Zubal phantom gave only $11 \mu\text{Sv h}^{-1}$ with the same 505 MBq injected activity and the same distance (1 m), using the data in Figure 2. If 505 MBq of ^{89}Zr were injected, the dose rate at 1 m would be $13 \mu\text{Sv h}^{-1}$. The low dose rate from the Zubal phantom is the consequence of the assumed biological distribution of the radionuclides deep in the body.

These data suggest that the effective dose to the staff from ^{89}Zr compares very favourably to ^{18}F -FDG in the vicinity of the patient because of its lower injected activity (Figure 5d) and the fact that extra administration of ^{89}Zr is not required during follow-up studies when the original injection is mostly accumulated deep in the body.

It had been reported recently that the injected activity of ^{89}Zr could be as low as 37 MBq, and it was still sufficient for follow-up studies up to 5 days later [22]. If this activity was used, the dose per scan in the control room would be reduced to $4.8 \mu\text{Sv}$ and it would be 90% of the ^{18}F -FDG dose per 250 MBq scan. Along the corridor, the dose per scan would be $24 \mu\text{Sv}$, which was 50% of the ^{18}F -FDG dose per scan.

In this study, we did not calculate the dose to the staff during the uptake period explicitly. However, the use of two phantoms, one in the examination room and the other in the injection room, took this into account implicitly—the second phantom in the injection room was waiting for the radionuclide to be taken up by the tumour. During this period, the dose to staff was added to that from the patient in the examination room. Thus, the comparison on the per-scan basis had incorporated some of this extra dose

Table 2. Dose to staff in the positron emission tomography/CT facility

Room	Radionuclide and ratio	Dose			
			Minimum	Maximum	Mean
Examination room	^{18}F	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	0.41	139.76	10.52
		Dose per scan (mSv)	0.08	28.65	2.16
	^{89}Zr	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	0.60	161.24	12.22
		Dose per scan (mSv)	0.04	10.86	0.82
	Ratio of ^{89}Zr to ^{18}F	Dose rate	1.48	1.15	1.16
		Dose per scan	0.49	0.38	0.38
Injection room	^{18}F	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	0.24	759.36	34.86
		Dose scan (mSv)	0.05	155.65	7.14
	^{89}Zr	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	1.26	865.05	41.41
		Dose per scan (mSv)	0.09	58.26	2.79
	Ratio of ^{89}Zr to ^{18}F	Dose rate	5.31	1.14	1.19
		Dose per scan	1.74	0.37	0.39
Corridor—1.5 m outside the rooms	^{18}F	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	0.02	2.64	0.22
		Dose per scan (μSv)	4.30	540.42	44.46
	^{89}Zr	Dose rate ($\text{nSv h}^{-1} \text{MBq}^{-1}$)	0.09	7.02	0.71
		Dose per scan (μSv)	6.11	472.76	47.93
	Ratio of ^{89}Zr to ^{18}F	Dose rate	4.32	2.66	3.28
		Dose per scan	1.42	0.87	1.08
Control room	^{18}F	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	4.29	64.06	26.75
		Dose per scan (μSv)	0.88	13.13	5.48
	^{89}Zr	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	25.03	348.31	145.13
		Dose per scan (μSv)	1.69	23.46	9.77
	Ratio of ^{89}Zr to ^{18}F	Dose rate	5.84	5.44	5.43
		Dose per scan	1.92	1.79	1.78
Reading room	^{18}F	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	7.23	41.06	15.76
		Dose per scan (μSv)	1.48	8.42	3.23
	^{89}Zr	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	40.38	242.26	99.61
		Dose per scan (μSv)	2.72	16.32	6.71
	Ratio of ^{89}Zr to ^{18}F	Dose rate	5.58	5.90	6.32
		Dose per scan	1.83	1.94	2.08
Holding room	^{18}F	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	0.39	11.87	2.51
		Dose per scan (μSv)	0.08	2.43	0.52
	^{89}Zr	Dose rate ($\text{pSv h}^{-1} \text{MBq}^{-1}$)	4.91	141.71	57.48
		Dose per scan (μSv)	0.33	9.54	3.87
	Ratio of ^{89}Zr to ^{18}F	Dose rate	12.45	11.93	22.87
		Dose per scan	4.09	3.92	7.52

^{18}F , fluorine-18; ^{89}Zr , zirconium-89.

The initial activities were assumed to be 250 MBq and 75 MBq for ^{18}F -FDG and ^{89}Zr , respectively. The scan time was 30 min.

although the injected activity was the same as the initial activity at the beginning of the scan. This is also a simplification that future work needs to address. However, the long half-life of ^{89}Zr reduces its activity by <1% during an hour-long uptake, in contrast to the reduction of over 32% in ^{18}F -FDG activity during the same period. In our future work, we will take into account the work load and will also revise the design of the facility for routine use of ^{89}Zr .

CONCLUSION

We have presented our study on the effective dose to staff in an existing PET/CT facility that uses ^{89}Zr labelled mAbs. The facility was originally designed to use ^{18}F -FDG. We simulated the dose with a spherical water phantom and a voxelised anthropomorphic phantom. In the vicinity of the patient, the ^{89}Zr dose rate was 5–19% higher than the ^{18}F -FDG dose rate, depending on the radionuclide distribution in the body. The primary component of the ^{89}Zr dose came from the prompt gamma at 909 keV, owing to its high energy and high intensity. Thus, the higher concentration

of ^{89}Zr inside the body was giving a higher ratio of dose than the same distribution of ^{18}F -FDG. For the same reason, the ^{89}Zr dose rate could be as much as 22 times the ^{18}F -FDG dose rate outside the patient rooms. When the injected activity and scan time were taken into account, the ^{89}Zr dose compared more favourably with that of ^{18}F -FDG. The dose per scan of ^{89}Zr was <40% of the ^{18}F -FDG dose inside the patient rooms, although it was as much as seven times that of ^{18}F -FDG. For routine clinical use of ^{89}Zr , the shielding requirements may have to be amended accordingly. In further study, we will carry out calculations that include the workload and with a suitably modified radiation protection shielding. Furthermore, we will perform simulations with different biological distributions of the radionuclides.

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