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D9.113 - Report from the feasibility study performed in two hospitals

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Content

1. Introduction

The objective of the PODIUM project is to develop a user-friendly online tool to calculate the radiation dose to workers. This is done by combining positioning information from individual staff members using the PODIUM Indoor Positioning System (IPS) based on the Microsoft Kinect 3D camera as well as information on the radiation field and the geometry of the room. The main aim of the Work Package 4 is to validate the tool in hospitals, in particular in interventional radiology/cardiology. In order to accomplish this, the work was divided into two main tasks. The first task is to test the online application in an experimental set-up using clinical X-ray equipment (described in D9.110) and the second was to test the tool during routine or typical clinical interventions.

The goal of this report is to present the work that has been done on the second task, validation in a real clinical environment during patient procedures. The report contains 6 cases. The first case is an extended validation of the experimental set up described in D9.110. In this report, the three selected Monte Carlo calculation codes used in PODIUM: PENELOPE/penEasyIR, MCGPU-IR, MCNPx, as well as, IPP-SE, the new software based on the look-up table approach, are compared with the measurements in D9.110. The other 5 cases are all new measurements made during patient interventions, 2 at Skåne university hospital, Malmö (Sweden) and 3 in St. James's Hospital, Dublin (Ireland).

Figure 1 summarises the dose calculation workflow and the information required to obtain the occupational dose based on the PODIUM approach. The validation of the system consists of comparing the calculated dose with the occupational dose measured with physical dosemeters. Although some of the Monte Carlo codes provide organ doses and effective dose, of course only $H_p(10)$ calculation could be compared with measurements.

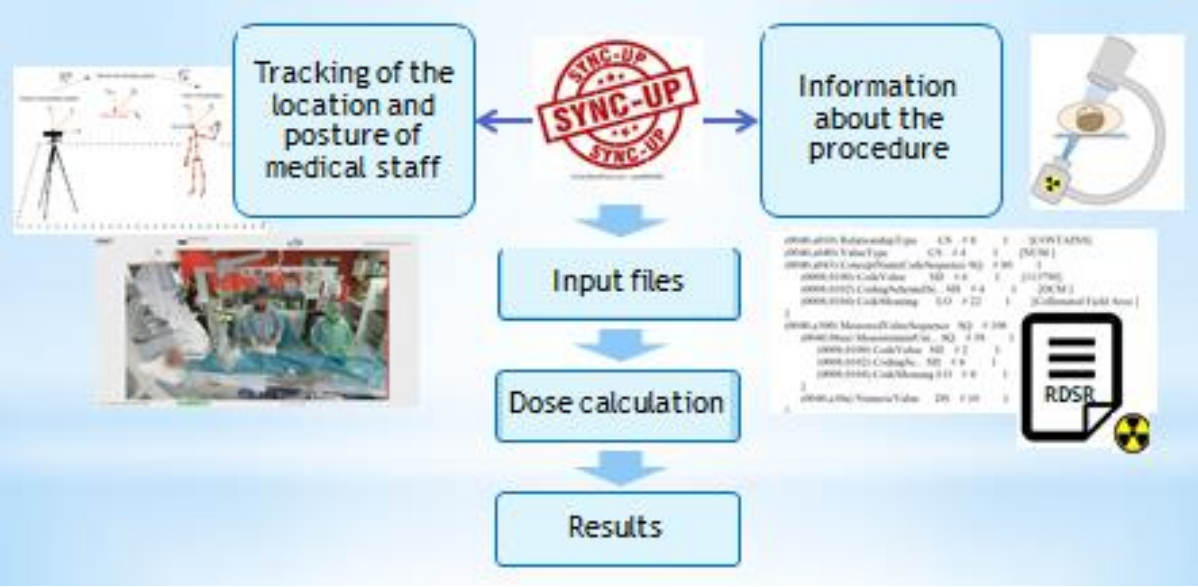


Figure 1: Scheme of the dose calculation workflow.

The next paragraph describes the methodology followed for the validation. The results for each case are reported in detail. The final evaluation is summarized at the end of the report.

2. Methodology

2.1 Tracking of the radiation worker

The tracking of staff was performed using one single Kinect camera except for the last case in Malmö where the two-camera system was tested. The criteria for the selection of the position in the operating room and for the tracking were presented in D9.105.

Firstly, the camera position was calibrated to refer the recorded positions to the isocenter (D9.105). In the case of St. James's Hospital, the camera had to be positioned on the TV monitor and thus, usually, required a specific calibration before each use. The position on the TV monitor was ideal in terms of achieving a good view of the main operator; however the TV monitor is not a fixed item and may be moved during cases affecting the calibration of the camera. For the single camera system, the Kinect calibration software tool was used. The software provides the user with the means to click on a reference point or an approximate location of the isocenter and it returns the coordinates of the point in the camera coordinate system, including correction for the camera rotation. This point is then used to translate the coordinate system from the camera location to the isocenter as the origin.

Although up to 6 persons were tracked, only the main operator data was used for the calculations. The software body ID that is assigned to the main operator 'skeleton' can change during tracking. Therefore, the files that are created for each tracked person (6 files in total) required filtering to locate the main operator body ID. This process was done manually for the first recorded cases based on observers' notes. To automate the process, a tool implemented in Python was developed to identify the body ID of the main operator for the single camera system based on an iterative algorithm that compare the position of the different skeletons across multiple frames. Also, for most cases, data was recorded at 30Hz and subsequently reduced down so that the final CSV file contains data at 1Hz and consists of the co-ordinates of the main operator ID only.

The format of the table of the tracking data is shown in Figure 2. The fields of the table were established in the PODIUM DCA application for both the single camera and the multi-camera tracking systems.

	Timestamp	BodyID	SpineBase_X	SpineBase_Y	SpineBase_Z	SpineBase_inferred	SpineMid_X	SpineMid_Y	SpineMid_Z	SpineMid_inferred	...
0	2019-01-10 09:29:42	3	0.773607	-0.860406	1.378223	1	0.768828	-0.619381	1.393390	0	...
1	2019-01-10 09:29:42	3	0.774552	-0.860633	1.377925	1	0.769946	-0.619716	1.392455	0	...
2	2019-01-10 09:29:42	3	0.774170	-0.860322	1.378723	1	0.769496	-0.619031	1.393489	0	...
3	2019-01-10 09:29:42	3	0.774174	-0.860485	1.378583	1	0.769539	-0.619184	1.393348	0	...
4	2019-01-10 09:29:42	3	0.774288	-0.860513	1.378390	1	0.769680	-0.619272	1.393109	0	...
5	2019-01-10 09:29:42	3	0.774385	-0.860684	1.377998	1	0.769835	-0.619564	1.392627	0	...
6	2019-01-10 09:29:42	3	0.773251	-0.860320	1.379467	1	0.768460	-0.618947	1.394428	0	...
7	2019-01-10 09:29:42	3	0.773241	-0.860341	1.379393	1	0.768445	-0.619026	1.394331	0	...

Figure 2: Screenshot of the recorded tracking data.

2.2 Experimental dose measurements

For the experiment described in D9.110, Case 5, in-house (provided by UPC) calibrated thermoluminescent detectors (TLD) and Thermo Electron Corp. type EPD MK, active personal dosemeters (APDs) were used. The uncertainty of the TLDs is of the order of 10% (k=1) for doses above 70 µSv. This value is calculated considering the uncertainty associated with the repeatability, energy and angular response for the energy range of interest. The uncertainty of the EPD MK reading is of the order of 10% (k=1) according to the manufacturer information, considering energy and angular response and the accuracy for Cs-137.

For the cases involving clinical procedures, each monitored worker was wearing at least one APD. Depending on availability, either the Thermo EPD MK, DMC 3000 or Raysafe were used. The uncertainty of the different APDs is of the order of 10% (k=1).

For case 6 (Malmö), along with the active dosemeters, 35 NaCl (salt) pellets[1] were placed on the apron, arms and head of the physician (Figure 23). The salt pellets provided an estimation of $H_p(10)$ and gave an assessment of the scattered radiation field homogeneity. These dosemeters were used for relative measurements only.

The position of the dosemeters on the operator was only known approximately, so this is an uncertainty for the comparison which is difficult to estimate. In some of the calculations, the influence of the position was evaluated by determining the dose at different points.

2.3 X-ray source input

For each procedure, a Radiation Dose Structured Report (RDSR) file is generated by the X-ray system, anonymised and uploaded to the application for calculation. The RSDR file should have all the information required for the calculation as described in D9.106.

An RDSR contains data on every unique irradiation event on the X-ray system. A detailed file of information on the X-ray primary beam conditions, table positions, angulation and collimation is generated. Even though it is a file with a defined standard and format (.SR), it can be converted into other file formats and the contents can vary (examples in Figure 3 and 4 below). Some manufacturers include fields as standard that are omitted when using other systems. When this was the case, for example for case C and D, in SJH, the information had to be recorded manually or estimated. Some typical items that were found to be missing or incomplete were the field size, and the horizontal movement of the C-arm. It was a challenge for the project team to become fully familiar with the format and contents of different RDSRs. Another aspect that was considered was the synchronization of the timestamp between the RDSR and the PC used for the Kinect. The computer times were either synchronized or a correction was made to ensure that the position of the operator at each second was correctly matched with the irradiation event occurring at that second.

[1] L. Waldner, C. Bernhardsson, Physical and dosimetric properties of NaCl pellets made in-house for the use in prospective optically stimulated luminescence dosimetry applications, Radiation Measurements 119, 52-57, 2018

	T	U	V	W	X	Y	Z	
1	X-Ray Filter Thickness Minimum (mm)	X-Ray Filter Thickness Maximum (mm)	Fluoro Mode	Pulse Rate (pulse/s)	Number of Pulses (number)	KVP (kV)	X-Ray Tube Current (mA)	Exposu
2	0.3	0.3	Pulsed	4	6	77	30.2	
3	0.6	0.6	Pulsed	4	3	73	58.8	
4	0.6	0.6	Pulsed	4	3	73	56	
5	0.6	0.6	Pulsed	4	4	73	55.6	
6	0.6	0.6	Pulsed	4	4	73	57.3	
7	0.6	0.6	Pulsed	4	5	73	54.3	
8	0.3	0.3	Pulsed	4	5	69	16.3	
9	0.6	0.6	Pulsed	4	7	73	18.7	
10	0.6	0.6	Pulsed	4	8	73	41.5	
11	0.6	0.6	Pulsed	4	24	73	41.8	
12	0.6	0.6	Pulsed	4	7	73	41.8	
13	0.6	0.6	Pulsed	4	4	73	40.6	
14	0.6	0.6	Pulsed	4	5	73	41.6	
15	0.6	0.6	Pulsed	4	8	73	44.2	
16	0.6	0.6	Pulsed	4	4	73	85.7	
17	0.6	0.6	Pulsed	4	14	73	40.9	
18	0.6	0.6	Pulsed	4	9	73	27.3	
19	0.6	0.6		2	13	63	224.9	
20	0.3	0.3	Pulsed	4	4	77	19.5	
21	0.6	0.6		2	21	63	226.6	
22	0.6	0.6	Pulsed	4	12	73	24	
23	0.6	0.6	Pulsed	4	232	73	23.4	
24	0.6	0.6	Pulsed	4	50	73	26	
25	0.6	0.6	Pulsed	4	41	73	25.5	
26	0.6	0.6	Pulsed	4	23	73	25.3	
27	0.6	0.6	Pulsed	4	10	73	25.3	
28	0.6	0.6	Pulsed	4	5	73	24	
29	0.6	0.6	Pulsed	4	43	73	26	
30	0.6	0.6	Pulsed	4	18	73	25.9	
31	0.6	0.6	Pulsed	4	39	73	36.9	
32	0.3	0.3	Pulsed	4	4	67	16.8	
33	0.6	0.6	Pulsed	4	18	65	14.3	
34	0.6	0.6	Pulsed	4	5	73	19.1	
35	0.9	0.9		2	12	61	221.3	
36	0.6	0.6	Pulsed	4	40	69	14.6	
37	0.9	0.9		2	11	61	220.3	

Patients | Dose Events

Figure 3: An example of RDSR file in .XLS format.

Accumulated X-Ray Dose Data

Concept Modifier: Acquisition Plane = Plane A (113620, DCM)

Reference Point Definition:
15cm below BeamIsocenter

Dose Area Product Total:
0.00757602493225 Gy.m2

Dose (RP) Total:
1.95415566327387 Gy

Fluoro Dose Area Product Total:
0.00539226933228 Gy.m2

Fluoro Dose (RP) Total:
1.27870216756818 Gy

Total Fluoro Time:
1989 s

Acquisition Dose Area Product Total:
0.00218375559997 Gy.m2

Acquisition Dose (RP) Total:
0.67545349570569 Gy

Total Acquisition Time:
58.8580000000000 s

Total Number of Radiographic Frames:
932 1

Height of System:
1065 mm

Focal Spot to ISO Center:

Figure 4: Example of RDSR file in .DCM format.

2.4 Phantoms used for the Monte Carlo dose calculation

PENELOPE/penEasyIR and MCNPx

With PENELOPE/penEasyIR and MCNPx patient geometry is simulated as a prism (slab) made of soft tissue (ICRU four-component) as described in D9.110, or by a BOMAB like phantom (Figure 5, left) which is more human-shape and can be scaled to the patient size. These simplified phantoms are created with quadric geometries. The type of phantom used is specified in each of the validation cases.

Figure 5: Geometry of BOMAB phantom (left), CIRS Rando phantom (right).

For case 5, PENELOPE/penEasyIR calculations were performed first with a prism phantom and then repeated with the BOMAB. In this case, the BOMAB was considered without arms and legs to emulate the CIRS Rando phantom (Figure 5, green scheme).

The floor and the walls of the room and the operator phantom are not simulated in MCNPx or PENELOPE/penEasyIR. The patient table is replaced by an Al equivalent (according to its attenuation) in the case of MCNPx, but it was not included in PENELOPE/penEasyIR.

MCGPU-IR

MCGPU-IR uses voxel phantoms, the patient and the operator must be the same phantom but they can be scaled differently to adapt their dimensions to the real dimension of the patient and of the worker. For case 5, the patient and the operator were simulated by using a CT-scan of the CIRS Rando phantom (Figure 5 right). The CT scan is shown in figure 6. The floor and the walls of the room and the patient couch were not simulated.

Table 1: Voxel dimensions (Dim) of the CIRS Rando phantom

	Voxels	Dim voxel (cm)	Dim (cm)
X	256	0.15	38.4
Y	256	0.15	38.4
Z	185	0.5	97.5

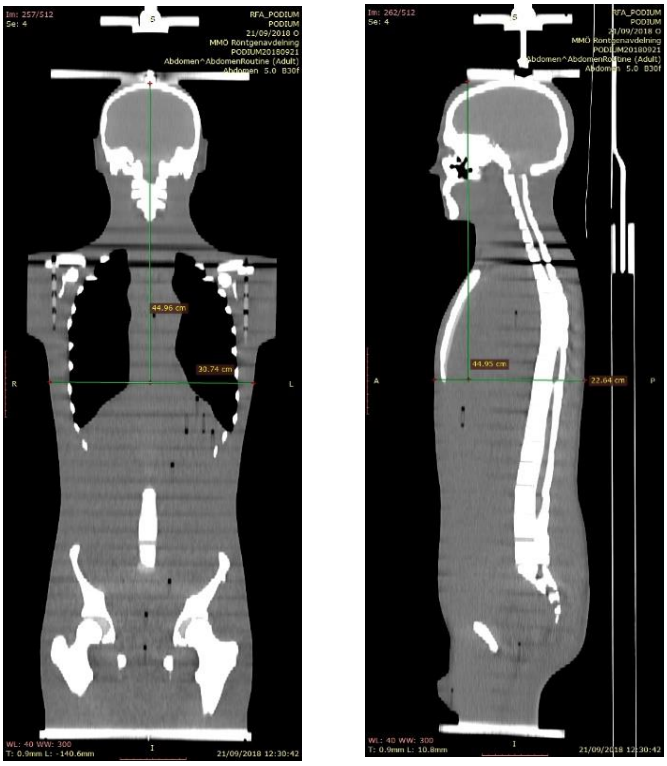


Figure 6: CT-scan of the CIRS Rando phantom.

The voxelized geometry is a box that contains the anthropomorphic phantom made of ICRU soft tissue surrounded by air.

For the other simulated cases, *MCGPU-IR* used the voxelized Rex or Regina phantoms developed by HelmholtzZentrum München - HMGU (Figure 7).

Figure 7: REX and REGINA voxelized phantoms.

The number and dimensions of the considered voxels in each case are:

REX:

```
[SECTION VOXELS HEADER v.2008-04-13]
299      148     348      No. OF VOXELS IN X,Y,Z
0.1775  0.1775  0.484  VOXEL SIZE (cm) ALONG X,Y,Z
```

REGINA:

```
#[SECTION VOXELS HEADER v.2008-04-13]
254      138     222      No. OF VOXELS IN X,Y,Z
0.2137  0.2137  0.8    VOXEL SIZE (cm) ALONG X,Y,Z
```

For PODIUM, the 143 tissues from the original phantoms were reduced to 28: air surrounding; cortical bone; spongious bone; heart; remaining tissues (soft tissue); front skin trunk; back skin trunk; other skin (extremities); blood; muscle; cartilage; lung; oesophagus; thyroid; bladder; liver; bone marrow; breast adipose; breast glandular; colon; stomach; gonads (ovaries & testes); salivary glands; brain; eye lens left; eye lens right. These organs allow calculating the effective dose and the dose equivalent of extremities and of the lens of the eye.

Phantom scaling

In order to adapt the phantom's dimension to the simulated person, the body mass index (BMI) factor defined in equation (1) is applied.

$$BMI = \frac{Weigth\,(kg)}{Heigth(m)^2} \qquad (1)$$

First, the height of the phantom is scaled to adjust exactly to the height of the patient. The two other dimensions (width and depth) are obtaining comparing the BMI values with the waist perimeter. Note that we are assuming the phantom waist as a perfect ellipse. Depending on the irradiated part of the body, another organ could be considered for normalization.

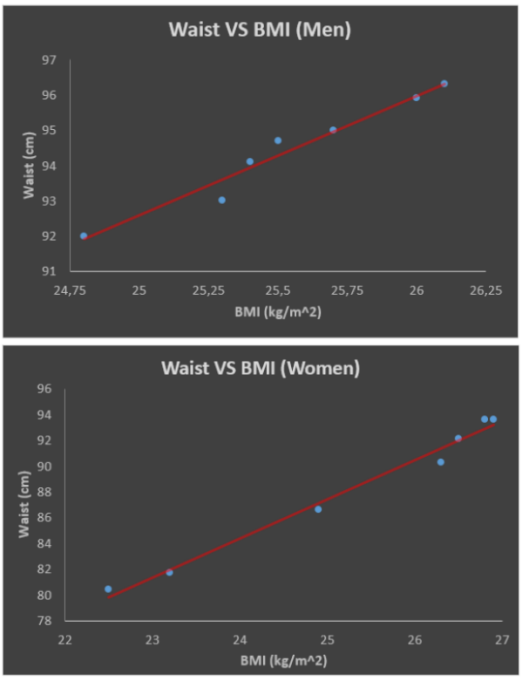


Figure 8: Waist diameter as a function of BMI.

Figure 8 shows the correlation between BMI and waist perimeter for both men and women (data obtained from Wells et al 2007[2]):

The equations for both linear approximations are:

$$waist_{men}(cm) = 3.3602 * BMI_{men}\left(\frac{kg}{m^2}\right) + 8.599 \qquad (2)$$

$$waist_{women}(cm) = 3.0254 * BMI_{women}\left(\frac{kg}{m^2}\right) + 11.787 \qquad (3)$$

The waist perimeter for Rex and for Regina is calculated measuring the two perpendicular dimensions of their waist (Figure 9): waist (Regina) = 72.36 cm; waist (Rex) = 80.85 cm

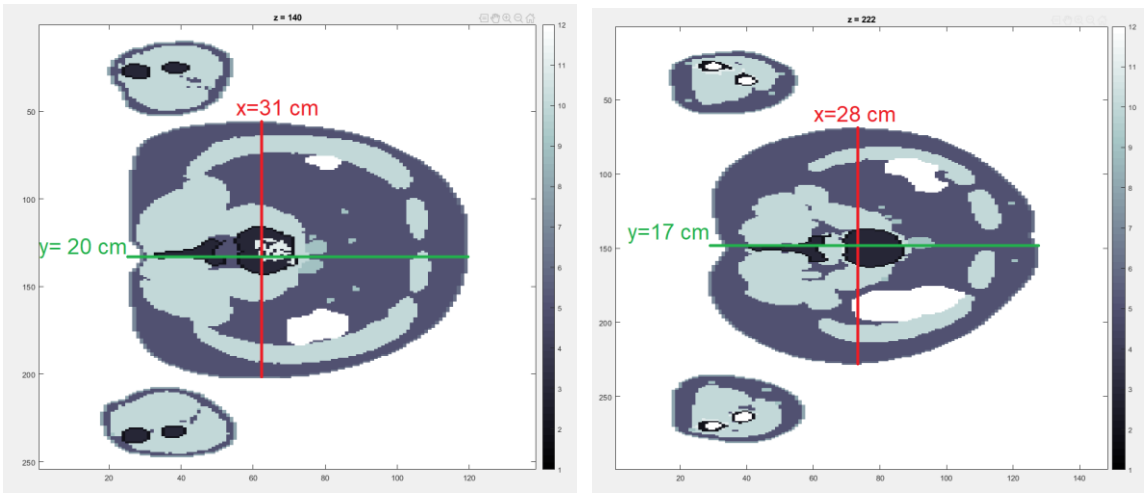


Figure 9: Rex and Regina waist dimensions.

Thus, the scaling values applied to each voxel are:

$$MEN:\ scale_{width} = scale_{depth} = \frac{\left[3.3602 * BMI_{men}\left(\frac{kg}{m^2}\right) + 8.599\right]}{80.85} \qquad (4)$$

$$scale_{heigth} = \frac{Height(cm)}{177.60}$$

$$WOMEN:\ scale_{width} = scale_{depth} = \left[3.0254 * BMI_{women}\left(\frac{kg}{m^2}\right) + 11.787\right]/72.36 \qquad (5)$$

$$scale_{heigth} = \frac{Height(cm)}{168.432}$$

[2] Wells JCK, Treleaven P, Cole TJ. BMI compared with 3-dimensional body shape: the UK National Sizing Survey. Am J Clin Nutr. 2007 Feb;85(2):419-25. doi:10.1093/ajcn/85.2.419

2.5 Dose calculation, Monte Carlo simulations

As described in detail in D 9.110, for all Monte Carlo codes a normalization factor *N* is needed to refer the simulated absorbed dose per history to the real number of emitted photons.

$$N = \frac{entrance\ air\ kerma}{K_a} \tag{6}$$

For each irradiation event (fixed kV, filtration, angulation and field size values), *N* is calculated from the ratio between experimental entrance air kerma and simulated air kerma (K_a) or energy deposited at a point of interest. K_a must be calculated in a point where there is no influence of backscatter in order to simulate same conditions as experimental entrance air kerma value.

In general the experimental entrance air kerma in the point of interest will be calculated following the inverse square law from the dose area product (DAP) value divided by the radiation field-size, both values supplied in the RDSR (see equation 7). As an alternative, the dose at the point of reference reported in the RDSR can also be used. It is worth mentioning that in the energy range of interventional radiology kerma and absorbed dose are equal.

$$entrance\ air\ kerma = \frac{DAP}{field\ size} \cdot \frac{(distance\ source - detector)^2}{(distance\ source - point\ of\ interest)^2} \tag{7}$$

PENELOPE/penEasyIR

PENELOPE/penEasyIR is an adaptation of PENELOPE/penEasy[3][4] developed in the framework of PODIUM project. This version provides personal dose values received by workers according to real procedures in interventional radiology.

In PENELOPE/penEasyIR there is no phantom in operator position. Photon energy fluence is simulated instead at particular points in the air matching operator Kinect joints (Points B in Figure 10). These simulations are obtained with the tally: 'Photon Fluence Point', which uses a variance reduction technique based on detection forcing, as it is described in the User Manual of last penEasy release (available at https://inte.upc.edu/en/downloads/downloads).

With the normalisation factor described in equation 6, $H_p(10)$ received by the operator can be calculated with Equation 8.

$$H_p(10)[\mu Sv] = N \cdot F \cdot \sum_{i=1}^{n} \phi_i^{sim} \cdot \left(\frac{\mu_{tr}}{\rho}\right)_i \cdot E_i \cdot \left(\frac{H_p(10,0^o)}{K_a}\right)_i \tag{8}$$

where

- *N* is the normalization factor.
- *F* is a unit normalization factor $1.602 \cdot 10^{-13}$ [J g kg^{-1}]
- ϕ_i^{sim} is simulated energy fluence, from Tally Photon Fluence Point, for energy region *i* at point B [cm^{-2} eV^{-1} per history].

[3] Sempau, J., A. Badal, and L. Brualla (2011). A PENELOPE-based system for the automated Monte Carlo simulation of clinacs and voxelized geometries—application to far-from-axis fields. Med. Phys. 38, 5887 – 5895. Available at http://dx.doi.org/10.1118/1.3643029.

[4] Salvat, F. (2019). PENELOPE-2018: A Code System for Monte Carlo Simulation of Electron and Photon Transport. OECD Nuclear Energy Agency. Available at http://www.oecd-nea.org.

- $\left(\frac{\mu_{tr}}{\rho}\right)_i$ is mass energy-transfer coefficient for energy region *i* [cm^2 g^{-1}] which at those energies is comparable to $\left(\frac{\mu_{en}}{\rho}\right)$ and can be interpolated from NIST[5].
- E_i is middle energy for energy region *i* [eV].
- $\left(\frac{H_p(10,0º)}{K_a}\right)_i$ is conversion coefficient from air kerma free-in-air to H_p(10,0º) in an ICRU slab for energy region *i*, interpolated from ICRP74[6].

Similarly H_p(0.07) and H_p(3) can be calculated with the same equation 8 but using, respectively, $(H_p(0.07,0º)/K_a)$[6] and $(H_p(3,0º)/K_a)$[7] instead of $(H_p(10,0º)/K_a)$.

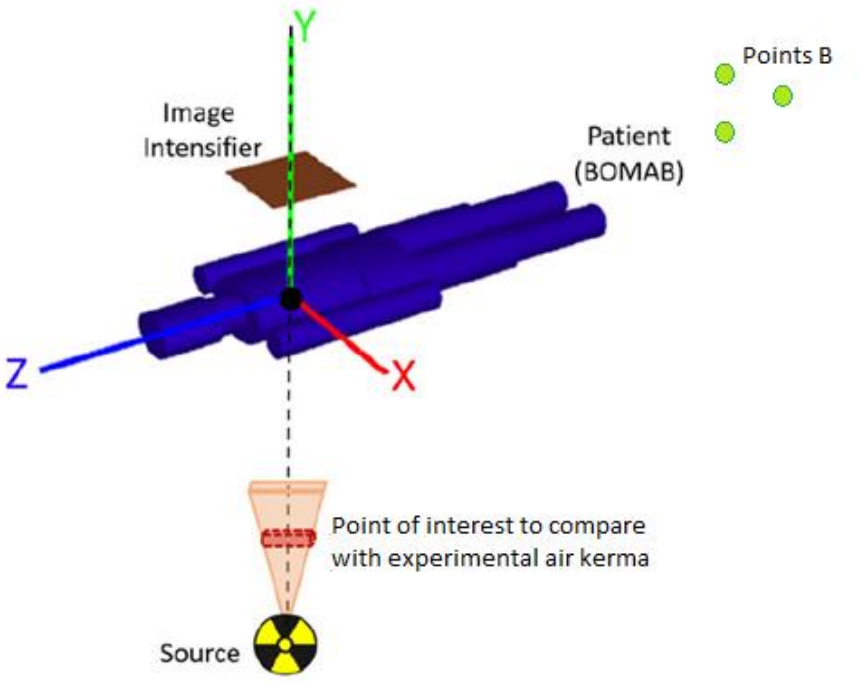


Figure 10: Scheme of geometry used in PENELOPE/penEasyIR simulation.

MCNPx

In MCNPx[8], to calculate the personal dose equivalent H_p(10), the tally F6 was used to calculate the energy deposition in MeV/g per particle in a volume that is behind 10 mm tissue equivalent volume as described in figure 11. This tally estimates energy deposition by integrating the track-length photon flux weighted by photon heating numbers according to equation 9. These numbers represent the average kinetic energy given to electrons along the photon path. Therefore, this tally is approximately

[5] NIST Standard Reference Database 126, last update July 2004 (https://dx.doi.org/10.18434/T4D01F)
[6] ICRP, 1996. Conversion Coefficients for use in Radiological Protection against External Radiation. ICRP Publication 74. Ann. ICRP 26 (3-4).
[7] G. Gualdrini et al.. 2013. Air kerma to Hp(3) conversion coefficients for photons from 10 keV to 10 MeV for photons from 10 keV to 10 MeV, calculated in a cylindrical phantom. Radiation Protection Dosimetry, 154(4): 517-521 . doi:10.1093/rpd/ncs269.
[8] D.B. Pelowitz, Ed., "MCNPX Users Manual Version 2.7.0" LA-CP-11-00438 (2011).

valid only when most of the electrons are trapped in the tallied cells. At the same time, kerma approximation (local energy deposition) was assumed.

$$F6 = \frac{\rho_a}{V_{\rho_g}} \int_V \int_t \int_E H(E)\, \sigma_t\, \varphi(r,E,t)\, dE\ dt\, dV \qquad (9)$$

where ρ_a and ρ_g are, respectively, the atomic and mass (gram) densities and $H(E)$ is the heating response in MeV/g.

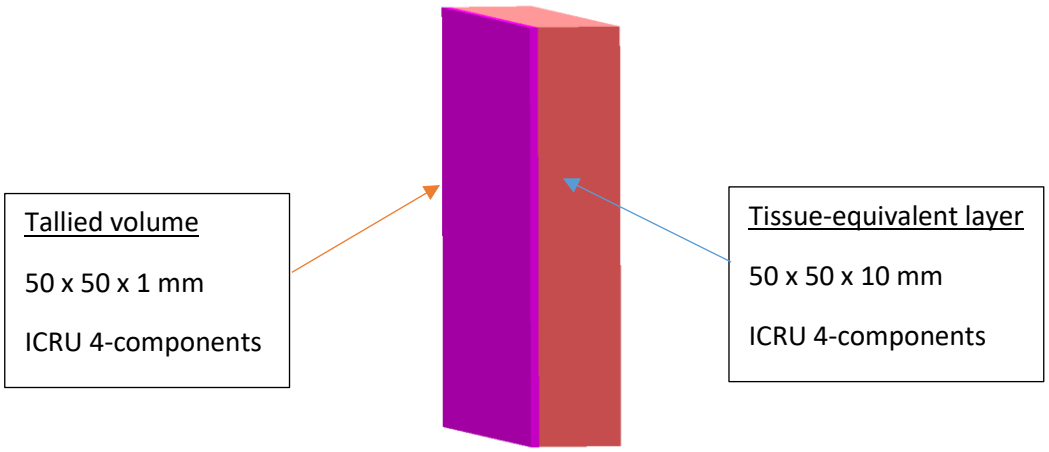


Figure 11: The geometry used to calculate personal dose equivalent in MCNPx.

Finally, the tallied volume is located at the approximate position of the dosimeter (usually on the left-side of the chest). The dosimeter location was obtained by the coordinates of the neck joint from the tracking data as shown in figure 12.

Figure 12: Position for Hp(10) calculation with MCNPx.

An outline of the simulation geometry used in MCNPX simulation is presented in figure 13a.

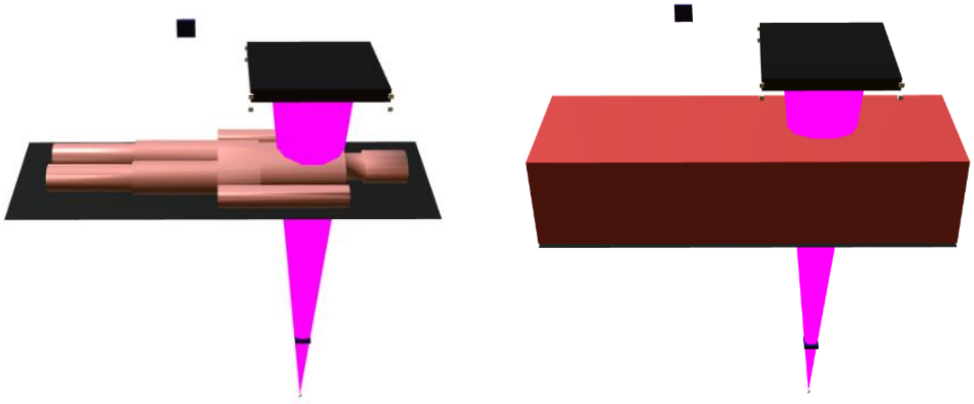


Figure 13a: Scheme of geometry used in MCNPX simulation Left (BOMAB phantom); right (Slab phantom).

The field was simulated as a cone-shaped beam. The field size was taken from the RDSR report and used to scale the beam field of view as shown in figure 13b.

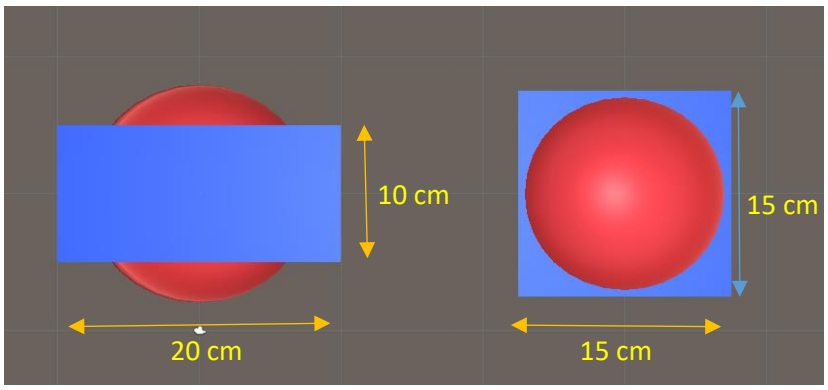


Figure 13b: The field as a cone-shaped beam.

For rectangular fields, the field is simulated with a cone whose projected circle has equivalent area as the rectangle. The KAP meter is simulated with a prism with an area scaled to fit the beam size at 20 cm distance from the source. This simulated KAP meter is used for normalisation purposes.

MCGPU-IR

First of all, the air kerma at the reference point is calculated by simulating the irradiation of a small box of air centered at the correct distance (equation 7). Then, the irradiation event is simulated with the operator located in the position obtained from the camera tracking. MCGPU-IR needs 4 points to locate the operator:

Head; Left Shoulder; Right Shoulder and Hip (labelled as Spine Base in tracking), the position is calculated from the Hip location as shown in Figure 14.

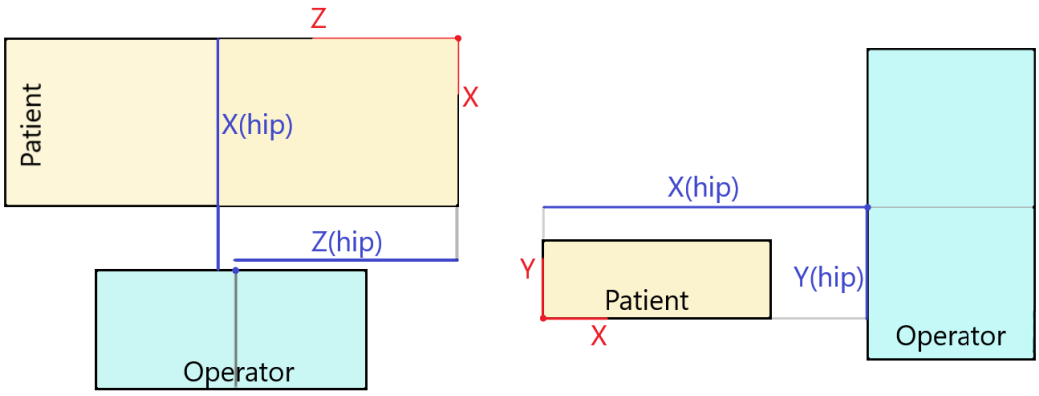


Figure 14: Criteria for positioning the operator using Kinect hip joint location.

The hip position (x, y, z) determines the distance from the origin of coordinates in MCGPU-IR (left foot of the patient) to the middle point in width and height of the operator.

The other three points needed (shoulders and head) are used for the rotations of the operator phantom, bending over the table and not perpendicular to the table. The rotation of the operator is schematically shown in figure 15.

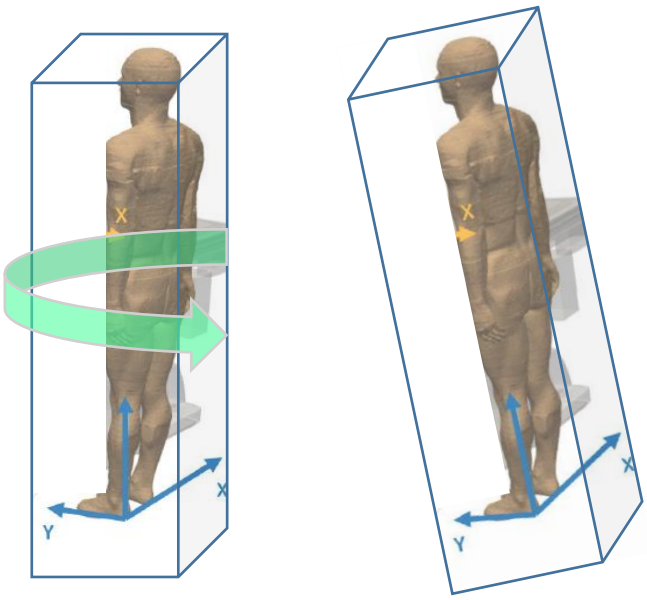


Figure 15: Rotations of the operator.

The results of the simulations provide the absorbed dose in the segmented organs included in the voxelized phantom and the effective dose. $H_p(10)$ is automatically determined at the middle of the chest, at a height of 132 cm for Rex (man) and of 124.87 cm for Regina (women) before scaling, as the mean absorbed dose at eight voxels made of ICRU tissue at a depth of approximately 1 cm. $H_p(10)$ could also be calculated manually at any other point.

Computational facilities

The speed of the calculations will depend on the characteristics of the computer used. The calculation results shown in this report have been obtained with the following machines:

PENELOPE/penEasyIR and MCGPU-IR run in the Argos Cluster from INTE-UPC compounded by the following server machines. PENELOPE/penEasyIR uses CPU and it runs in any available node in the cluster. MCGPU-IR instead, uses GPU and it runs only in one node (called c15) with compatible NVIDIA graphic cards.

Cluster head node:

Argos Main Server

DELL Power Edge 420

2 x 1 Intel® Xeon® E5-2470 v2 (2,40GHz, 10N, caché de 25M, QPI de 8,0GT/s, Turbo) 95W, 1600MHz

4 x 16GB RDIMM, 1600MHz

4 x 2 TB SAS nearline 6 Gb/s 7,2K rpm 3,5"

iDRAC Enterprise

Compute nodes:

from c0 to c11

DELL Power Edge 410

2 x Intel Xeon E5520 Processor (2.26GHz, 8M Cache, 5.86 GT/s QPI, Turbo, HT), 1066MHz Max Memory

8x2GB Dual Rank UDIMMs 1066MHz

1 x 160GB SATA 7200 3,5"

iDRAC Express

c12

DELL Power Edge 420

2 x 1 Intel® Xeon® E5-2470 v2 (2,40GHz, 10N, caché de 25M, QPI de 8,0GT/s, Turbo) 95W, 1600MHz

4 x 16GB RDIMM, 1600MHz

2 x 2 TB SAS nearline 6 Gb/s 7,2K rpm 3,5"

iDRAC Enterprise

c15

DELL Power Edge 720

2 x 1 Intel® Xeon® E5-2670 v3 (2,30GHz, 12N, caché de 30M, Turbo) 2133MHz

4 x 16GB RDIMM, 2133MHz

2 x VGAs NVIDIA GeForce 1080Ti 11GB

iDRAC Enterprise

c13 (basic GPU Node)

Intel Core i7 3820 3,60GHz

16 GB RAM DDR3

1 x VGAs NVIDIA GeForce GTX 780 3GB GDDR5

2 x 2TB HD SATA (RAID1)

Argos Xen server (hosting project Virtual Machines)

DELL Power Edge 710

2 x Intel Xeon E5520 Processor (2.26GHz, 8M Cache, 5.86 GT/s QPI, Turbo, HT), 1066MHz Max Memory

8x2GB Dual Rank UDIMMs 1066MHz

3 x 160GB SATA 7200 3,5" (RAID1 + Hot Spare)

iDRAC Enterprise Edition

MCNPx 2.6 runs at the following computer:

CPU: Intel(R) Xeon(R) CPU E3-1270 v5 @ 3.60GHz
Base speed: 3.60 GHz
Turbo boost: 3.80 GHz
Cores: 4
Logical processors: 8
L1 cache: 256 KB
L2 cache: 1.0 MB
L3 cache: 8.0 MB

Uncertainties

The statistical uncertainty in the calculations is below 2.5% (k=1). In addition, the uncertainties associated with the calculated values of the occupational doses (operational quantities or protection quantities) include the uncertainty associated with the KAP or Kref value indicated in the RSDR. This uncertainty is estimated from quality control surveys and is taken equal to 10% (k=1).

2.6 Dose calculation with the look-up table approach

The look-up table approach of PODIUM is a fast dose calculation method which is meant to provide dose estimations within a few seconds after a procedure. In this approach, we make use of two sets of tables from previously calculated Monte Carlo simulations: the source tables and the Dose Conversion Coefficient (DCC) tables. The source tables contain information about the intensity of the scatter field, while the DCC tables contain dose per fluence coefficients for a series of predefined simplified irradiations, as described in WP2 deliverables. Both the source and DCC tables are integrated within a software, IPP_SE, which is based on the Interactive Posture Program developed in WP2. This software convolutes the C-Arm parameters from the RSDR, defining the scatter field, with the tracking information coming from the camera module, and provides dose estimates for effective doses, (peak) skin doses, eye lens doses, and $H_p(10)$.

IPP_SE can be executed both remotely (on PODIUM's cluster) and locally, making this method of calculation particularly suitable in those operating theaters which, for security reasons, are not directly connected to the internet, or that have unstable internet connection.

In principle, in case the RDSR or the DICOM images will become available in real time, IPP_SE could provide dose estimates immediately after each irradiation event, even during a procedure.

Source tables

The source tables define the intensity of the scatter field generated by the X-Ray tube. The geometry of the simulations used to create the simulations is constituted by:

1. the X-Ray source,
2. a collimator,
3. a KAP meter,
4. a bed,
5. a stretched BOMAB phantom (176 cm height) representing the patient, and
6. an image intensifier.

Each source table corresponds to a set of machine parameters, which are shown in table 2 below.

Table 2: Machine parameters used in the look-up table approach

C-Arm parameter	Range	Number of simulated cases
Primary Angulation	[-75 LAO, 75 RAO] deg	11 cases, every 15 deg
Secondary Angulation	[30CAU, 30 CRA] deg	3 cases, every 30 deg
Field Size	[10, 30] cm	3 cases, every 10 cm
Source-To-Detector Distance	[100, 120] cm	2 cases, every 20 cm
Added Filtration (Copper)	[0, 0.9] mm	4 cases, every 0.3 mm
X-Ray energy spectrum (kVp)	[60, 120] kV	7 cases, every 10 kV
TOTAL		5544, optimized to 4950

In each of these simulations, the intensity of the scatter field is scored in correspondence of the surface of a sphere of 70 cm diameter, centered in the isocenter. The surface of the sphere is subdivided into segments with steps of 5 degrees both in the polar and azimuthal angles. The quantity used to measure the intensity is fluence discriminated per energy, with bins of 10 keV varying from 20 keV to 120 keV. The energy-dependent fluence is scored in each of the 2592 surfaces by averaging fluences over small spheres of 0.5 cm radius. All source simulation was performed by using the MCNP6 Monte Carlo particle transport code. Equation 10 shows the definition of the tally used to calculate the fluence in the small spheres.

$$F4_{face_j,\ Erg_i} = \frac{1}{V}\int_V dV \int_0^{4\pi} d\Omega \int_{Erg_{i-1}}^{Erg_i} dE\ \ \Phi(\boldsymbol{r}, E, \Omega) = \left[\frac{particles}{cm^2}\right] \qquad (10)$$

In this equation, face$_j$ is the jth segment of the large sphere, V is the volume of a small sphere, Ω is the solid angle, Erg$_i$ is the energy bin, Φ is the energy and angular distribution of the fluence within a point **r** of V. This fluence is then normalized by dividing the fluence over the kerma deposited in the KAP meter, which is tallied by means of an F6 tally in equation 9.

$$NF_{face_j,\ Erg_i} = \frac{F4_{face_j,\ Erg_i}}{Kerma} \qquad (11)$$

The number of particles was adjusted depending on the C-Ray energy spectrum in order to obtain uncertainties lower than 20% for each energy bin. Figure 16 shows some examples of the geometry used for some of the source simulations.

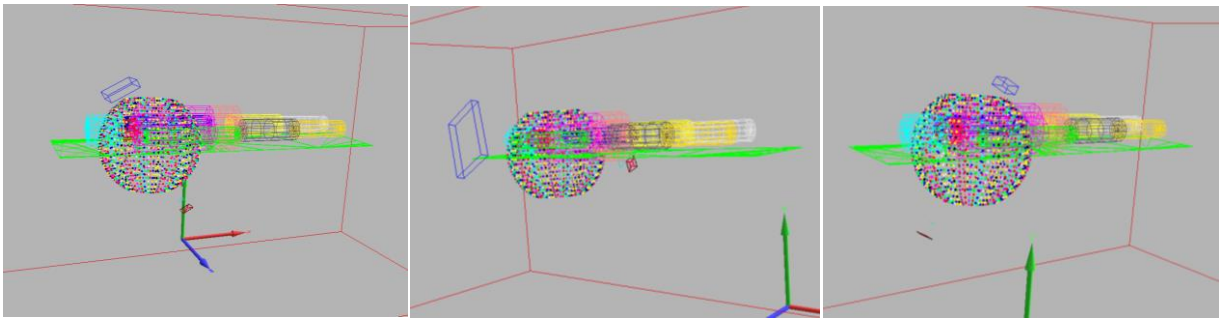


Figure 16: Examples of the geometries simulated for 3 (out of 4950) source configurations.

Dose Conversion Coefficient (DCC) tables

The DCC tables include conversion coefficients of organ and effective doses per fluence (Gy*cm^2), which were calculated using 4 different voxel phantoms. In order to simplify the DCC simulations, these conversion coefficients were calculated by irradiating the phantoms with idealized beam impinging on the phantom body in 6 different locations, named Field Panels (FP), as shown in Figure 17. In these simulations, the FPs uniformly emits mono-energetic beams of photons. In order to simulate a realistic variety of irradiation directions impinging on the phantoms, the uniform beams were rotated both in the vertical and horizontal directions.

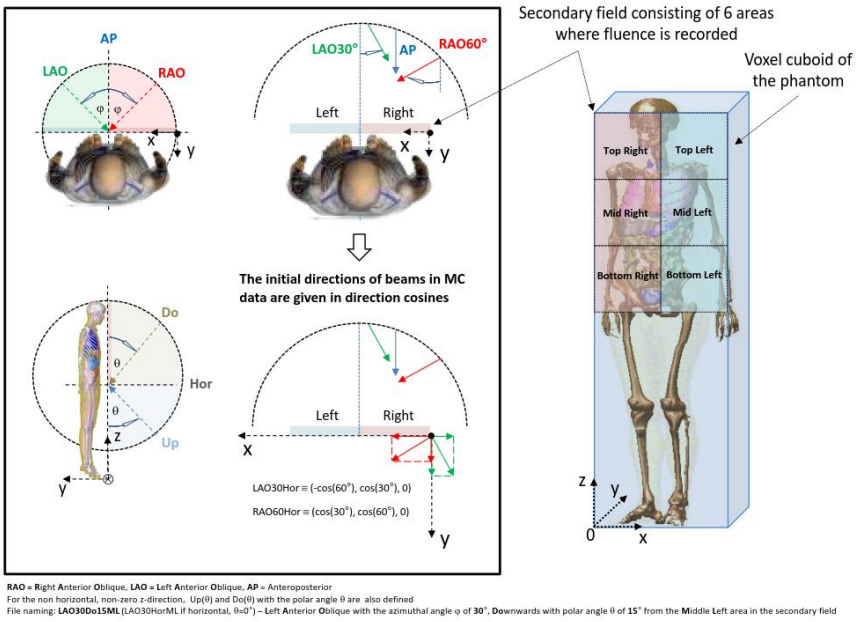


Figure 17: Location of the Field Panels and definition of the vertical and horizontal irradiation directions.

Within WP2, the 4 voxel phantoms used for calculating the DCC tables were selected to represent 3 body types and 2 different postures.

1. Irene2018, representing a small female doctor
2. Donna2018, representing a larger female doctor
3. RAF phantom vertically standing, representing an average Caucasian male doctor
4. RAF phantom leaning forward, representing an average Caucasian male doctor

Figure 18 shows the 4 phantoms.

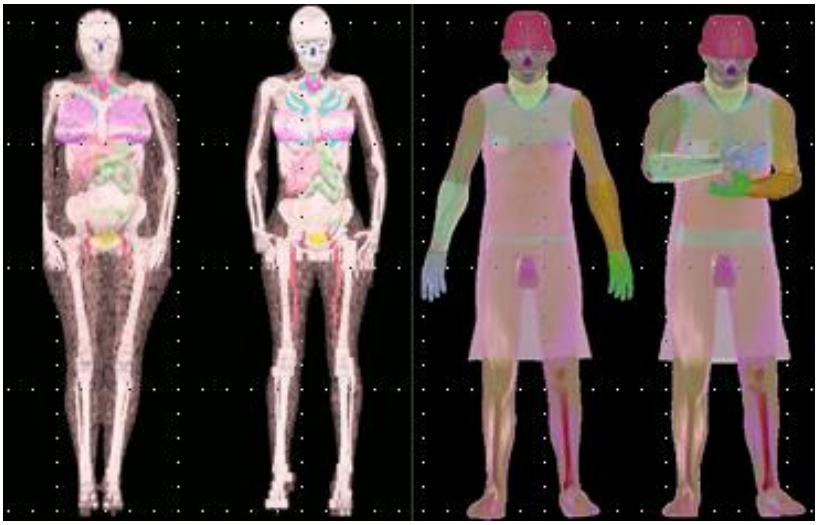


Figure 18: The 4 phantoms used for the DCC simulations. From left to right: Donna2018, Irene2018, RAF phantom standing vertically, RAF phantom leaning forward.

In total, about 2000 different irradiation configurations were considered for each phantom, resulting in 8000 possible sets of dose conversion coefficients. For tallying the organ doses, the F6 tally was used, as indicated in equation (9), and multiplied by the size of the FPs to obtain Dose Conversion Coefficients per Fluence (DCCpF), in units of Gy * cm^2. The number of particles was fixed at 1E8 for all simulations; this number granted statistical uncertainty lower than 10%, for the great majority of the organs contributing to the effective dose (even below the lead garments).

a) IPP_SE

By means of a deterministic algorithm based on ray tracing, IPP_SE convolutes the parameters defining the source simulations with the parameters from the camera. Once the irradiation geometry is generated by IPP_SE, the software performs the convolution in two steps:

1. Cast rays between the isocenter and various locations (the center, the corners, etc.) on each Field Panel, and calculates incidence angles and length of the rays. The incidence angles are used to pick the correct DCCs among the ones available in the look-up tables.
2. Identify the segment intersected on the scatter field and pick the correct fluence normalized over kerma. This fluence is then projected towards the FPs accounting for the inverse square law.

After selecting the correct fluences and dose conversion coefficients by means of the ray tracing algorithm, IPP_SE calculates doses by multiplying the projected fluence for the dose conversion coefficients of each field panel, and by the DAP reading from the RSDR, as shown in equation 11.

$$Organ\ Dose = DAP \sum_{20\ keV}^{120\ keV} \sum_{FP_s} NF_{FP_s,\ Erg_i} DCCpF_{Erg_i} \frac{(radius\ large\ sphere)^2}{(distance\ Isocenter - FP_s)^2} \quad (12)$$

On one hand, the main advantage of IPP_SE is the run-time. Usually an irradiation event can be simulated within a timeframe of 5 to 10 seconds.

On the other hand, the accuracy of the dose estimations is expected to be lower than that of full Monte Carlo simulations. While the combined statistical uncertainty of source and DCC simulation is smaller

25% (k=1), the look-up table methodology is subject to a higher model uncertainty compared to Monte Carlo methods. IPP_SE makes use of a deterministic method (raytracing) to propagate the source simulations, which is not always accurate to simulate the photon transport in an operating theater. One limitation of this approach is the lack of an algorithm to account for objects placed between the isocenter and the doctor which are affecting the scatter field. This is what happens, for example, when using ceiling mounted lead shielding, which are not simulated in the current release of IPP_SE.

Furthermore, due to limited computational power, the number of irradiation configurations for both source and DCC tables cases had to be limited. Configurations that are not explicitly simulated are approximated to the closest simulated configuration. This discretization contributes to lower the accuracy of the look-up table approach. However, new simulations including more configurations for both source and DCC tables will be carried out in the future, which will reduce this part of uncertainty.

By default, IPP_SE performs a weighted average of the fluence over 15 degrees in the vertical and horizontal directions, around the face segment found by the ray tracing algorithm on the surface of the scatter sphere. In normal measurements, averaging fluences by weighting over 5 degrees was found to give the most reliable results with IPP_SE. On the one hand, this algorithm allows to smooth out the fluence outliers found in some faces of the scatter sphere. On the other hand, the smoothing can account for the slight vibrations (1-2 cm in random direction) of the operator body during an irradiation event.

3. Validation: measurements – Monte Carlo simulations

The validations are reported for each case. A description and results from the measurements are followed by a description and results of the simulations. Table 3 summarizes for each case which calculations are reported.

Table 3: A summary of the clinical cases included in the report

Dose calculation	Malmö Hospital (Sweden)			St. James's Hospital (Ireland)		
	Case 5	Case 6	Case 7	Case C	Case D[(1)]	Case 1 Cardiac
PENELOPE/penEasyIR	Yes	Yes	Yes	Yes	No	Yes
MCGPU-IR	Yes	Yes	Yes	No	No	Yes
MCNPX	Yes	Yes	No	Yes	No	Yes
IPP_SE	Yes	No	Yes	Yes	No	No

(1) Dose calculations are still not available because further review of the tracking files is required

3.1 Skåne University Hospital, Malmö CASE 5

3.1.1. Measurement geometry and results

Reference Name: Case 5
Location: Malmö Hospital room 105
Date: 4th November 2018
Clinical Procedure: Phantom measurement

Room dimensions

The dimensions of room 105 and the placement of the Kinect are given below.

KINECT location

In these measurement using phantoms, the Kinect was not used.

Overview of Measurements

Measurements were performed on a static phantom. 2 TLD detectors type TLD 2000C (DC) and 2 Thermo Electron Corp type EPD MK (EMD) for Hp(10) were used for each measurement. The main operator was represented by a CT Torso phantom CTU-41 (Kyoto Kagaku) and the patient phantom was a CIRS Phantom (Adult Male Phantom Model NO. 701). The R100 dose probe (RTI) was used for the measurements in the radiation field.

Exposure field on patient

Two different measurements with the C-arm irradiating the patient on the chest were performed. The first measurements were from 0-degree straight below the phantom and the third measurement was with a 15-degree angle. All machine parameters were taken from the DICOM Radiation Dose Structured Report (RDSR) generated during the exposures. The ceiling mounted lead protection was not used during this case.

Dose results from measurement

The position of the dosemeters for each measurement are given below in figure 19 and the measured results are given in table 4.

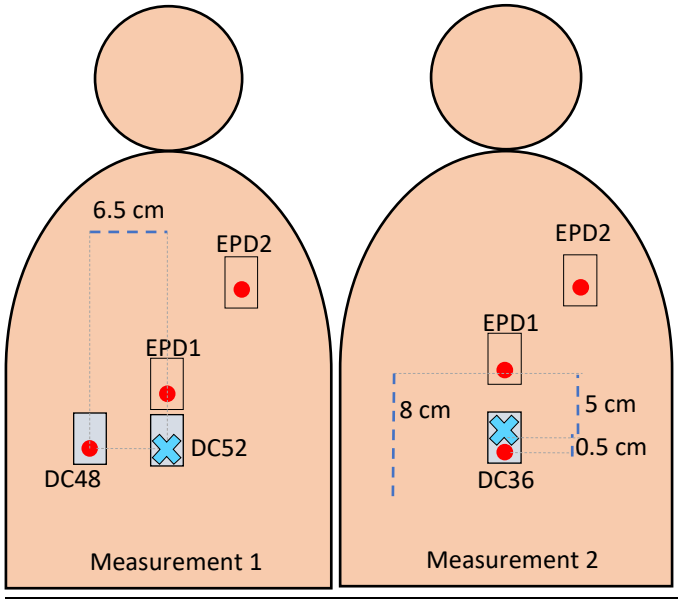


Figure 19: Shows the position of the TLD:s and EPD:s from each separate measurement.

Table 4: Measurement results, $H_p(10)$ µSv

Measurement 1			
EPD1	EPD2	DC48	DC52
73	72	85	134
Measurement 2			
EPD1	EPD2	DC36	
73	63	105	

RDSR Data

The RDSR data for each irradiation event was transferred to an Excel file.

KINECT Data

For this static posture of main operator there was no Kinect data recorded, instead all distances were measured using a calibrated laser rangefinder (Bosch GLM 100 C) with an accuracy of ± 1.5 mm.

3.1.2. Monte Carlo simulation

As described in D 9.110, in a first approach the simulations were carried out with PENELOPE/penEasy[9] code, using a simplified parallelepipedic quadric geometry and the detection forcing tally (Photon Fluence Point approach). Main approximations applied were: The patient phantom was replaced by a prism made of soft tissue (ICRU four-component), the table was replaced by an Al equivalent (according to its attenuation), the floor and the walls of the room were not simulated, the operator phantom was not simulated.

In this report we have performed a more realistic simulation of this case. On the one hand three Monte Carlo codes are used: PENELOPE/penEasyIR, MCNPx and MCGPU-IR and on the other hand, more realistic phantoms are considered.

Monte Carlo geometry

In PENELOPE/penEasyIR simulation the BOMAB phantom without arms is used to simulate the patient (Figure 5). The BOMAB is not scaled since its default dimensions are close to experimental CIRS phantom.

In MCGPU-IR, a CT scan of the RANDO phantom is used (Figure 6). $H_p(10)$ is calculated as the average of 8 voxels at 1 cm inside the operator phantom and centred at the dosemeter position, two voxels for each dimension (which means 0.3 cm for each dimension).

In MCNPX, the main approximations applied were: the patient phantom is replaced by a prism made of soft tissue (ICRU four-component), the table is replaced by an Al equivalent (according to its attenuation). The geometry simulated in MCNPX is shown in figure 20 below.

[9] Salvat, F., 2015. PENELOPE-2014. A code system for Monte Carlo simulation of electron and photon transport. OECD Nuclear Energy Agency. Data Bank NEA/NSC7DOSC3.
Sempau, J., Badal, A., Brualla, L., 2011. A PENELOPE-based system for the automated Monte Carlo simulation of clinacs and voxelized geometries—application to far-fromaxis fields. Med. Phys. 38 (11).

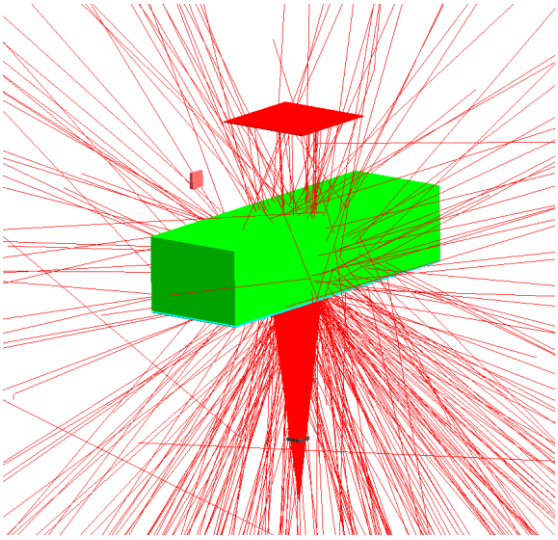


Figure 20: Case 5 experiment 2: MCNPX geometry with visualization of the particle tracks.

Monte Carlo results

For PENELOPE/penEasyIR, $H_p(10)$ is calculated according to equation 8 (section 2.5). But, in this case, as in D 9.110, an additional factor to take into account the angular dependence of the conversion coefficients, $H_p(10,\alpha)/H_p(10,0^o)$, is introduced.

The simulation time was 120 s per event for *Penelope/penEasyIR*, 111 s per event for MCNPx and 40 s per event for MCGPU-IR. The statistical uncertainty was below 2.5% (k=1).

Tables 5 and 6 show the simulated $H_p(10)$ compared with the experimental measurements for each detector and each experiment, including the associated uncertainty (k=1) (Paragraph 2.2). The values are also summarized graphically in figure 21. In the case of Monte Carlo data the uncertainty takes into account the statistical uncertainty and an uncertainty of 10 % (k=1) associated to the air kerma value used in the normalization.

Table 5: Experiment 1

	Experimental [µSv]	**PENELOPE/penEasyIR [µSv]**	**MCGPU-IR [µSv]**	**MCNPx [µSv]**
EPD1	73 ± 8	84 ± 9	80 ± 9	90± 10
EPD2	72 ± 8	80 ± 8	56 ± 6	80 ± 9
DC52	134 ± 13	94 ± 9	75 ± 8	110 ± 12
DC48	85 ± 8	72± 7	96 ± 10	82 ± 9

Table 6: Experiment 2

	Experimental [µSv]	**PENELOPE/penEasyIR [µSv]**	**MCGPU-IR [µSv]**	**MCNPx [µSv]**
EPD1	73 ± 8	56 ±6	54 ± 6	77 ± 9
EPD2	63 ± 7	50±5	40 ± 4	60 ± 7
DC36	103 ± 11	65±7	61 ± 7	78 ± 9

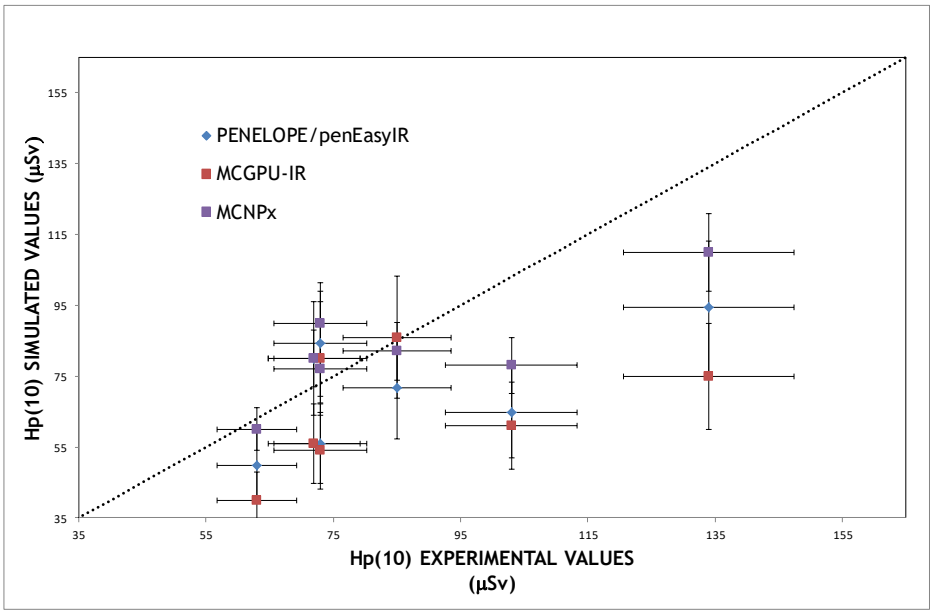


Figure 21: Comparison between experimental and simulated values.

The ratio between simulated and measured values are shown in table 7.

Table 7: Ratio between simulations and measurements

	Ratio: (PENELOPE/penEasyIR)/Experimental			
Experiment	EPD1	EPD2	TLD	TLD
1	1.15	1.11	0.70 (DC52)	0.85 (DC48)
2	0.77	0.79	0.63 (DC36)	-

	Ratio: (MCGPU-IR)/Experimental			
Experiment	EPD1	EPD2	TLD	TLD
1	1.10	0.78	0.56 (DC52)	1.13 (DC48)
2	0.77	0.63	0.59 (DC36)	-

	Ratio: (MCNPx)/Experimental			
Experiment	EPD1	EPD2	TLD	TLD
1	1.23	1.11	0.96 (DC48)	0.82 (DC52)
2	1.05	0.95	0.74 (DC36)	-

IPP_SE dose estimations

Compared to real procedures, where the doctor slightly moves during an irradiation event, in this measurement the position of the phantom is static. For this reason, the non-default option “no smoothing algorithm” of IPP_SE was used. Total execution time was about 10 seconds for both measurements.

For experiment 1, the estimated $H_p(10)$ was 89 uSv, while for experiment 2 it was 51 uSv. Due to the fixed positioning of the dosimeter in the chest of the phantoms of IPP_SE, the $H_p(10)$ calculated with this approach is comparable only to the dose recorded by the EPD1 dosimeter. The ratio between simulated and measured values are shown in table 8.

Table 8: Ratio between simulations and measurements

Experiment	(IPP_SE)/EPD1
1	0.82
2	1.42

3.1.3 Discussion of the results

The procedure was a straightforward static phantom measurement, and the first real case to validate the Monte Carlo simulation. All relevant parameters in this case 5 are known, as they were measured and controlled during the procedure. The limitation of this case is that it is not a clinical case including Kinect movement.

The results are considered satisfactory for the purpose of the PODIUM validation study. The ratio obtained varied between 0.77 and 1.15 for the 4 EPDs measurements for PENELOPE/penEasyIR, 0.63 and 1.10 for MCGPU-IR, 0.95 and 1.23 for MCNPx and 0.82 and 1.42 for IPP_SE. As regards the comparison with the 3 TLDs, the ratios are in the same magnitude. The ratios are in the same range for the four simulation techniques, and reasonable considered the uncertainties.

The uncertainty in the position of the dosemeter in relation to the scoring point in the Monte Carlo simulations may influence these differences. Changing 5 cm the position of the scoring point in the Monte Carlo simulation can produce a 10% difference in the dose. The experiments organized within EURADOS WG12 comparing APDs and passive dosemeters worn at the same time also showed differences of this order and higher.

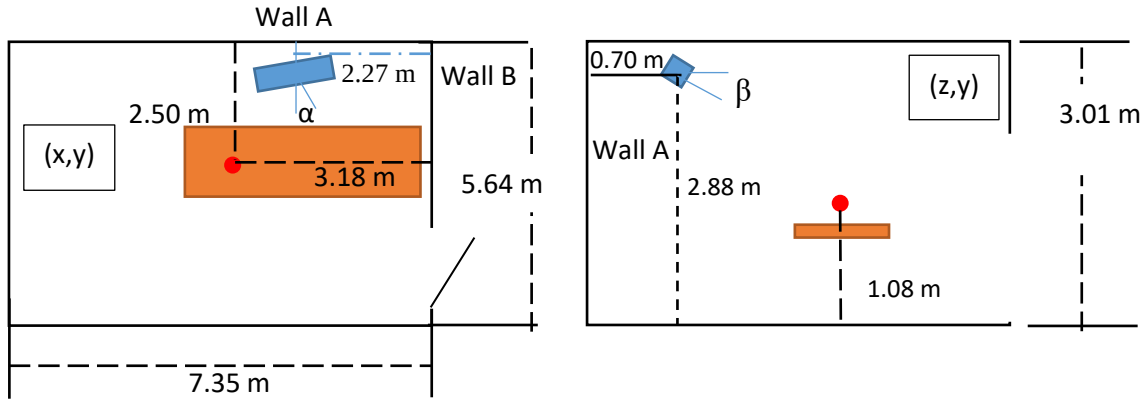
3.2 Skåne University Hospital, Malmö CASE 6

3.2.1 Measurement geometry and results

Reference Name: Case 6
Location: Malmö Hospital room 105
Date: 17th December 2018
Clinical Procedure: Renal artery angiography

Room dimensions

The room dimensions of 105 and the placement of the Kinect are given in the figure below.



Blue: Kinect raised with an α=2° tilted and a β = 32° angle (“looking down ”)
Red: center of rotation

Figure 22: Room dimension and position of the Kinect camera.

KINECT location

The Kinect was ceiling mounted with a tilted angle of 32° (β) downwards, between the top of the Kinect and the ceiling. The roll and yaw angles between the room and the Kinect are important when translating the Kinect coordinate system to the coordinated system of the room.

Overview of measurements

Measurements were performed on a Vascular Surgeon during an uncomplicated Endovascular Lower Limb Angiography procedure. Four Mirion DMC 3000 ($H_p(10)$ were placed in front of lead apron on main operator together with 35 NaCl pellets. The position of the dosemeters and pellets are shown in the figure 23.

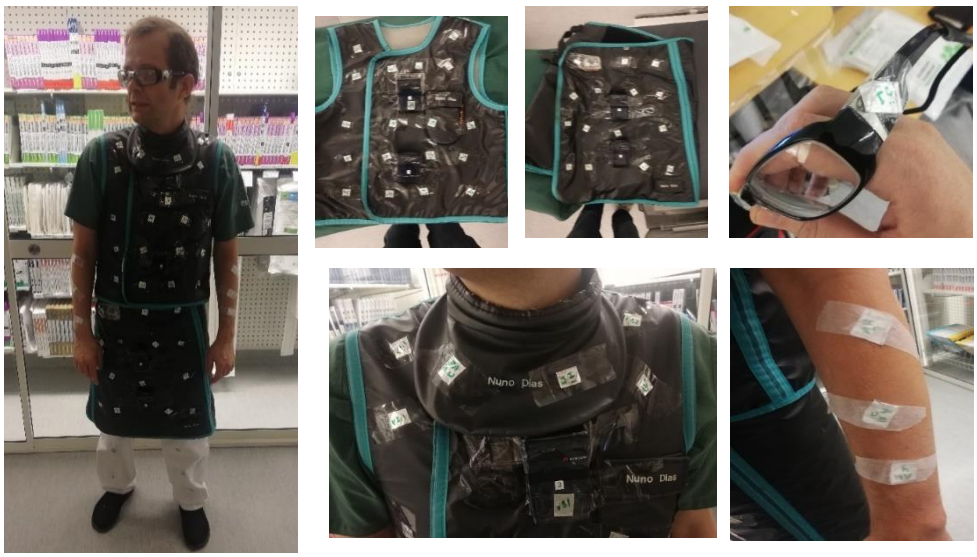


Figure 23. Four Mirion DMC 3000 and 35 salt pellets were placed on the main operator.

Exposure field on patient

Right pelvis was exposed, the radiation field varied between 54 cm^2 and 630 cm^2. The total DAP was 17.2 $Gycm^2$. The treatment contained 186 irradiation events. All machine parameters were taken from the DICOM Radiation Dose Structured Report (RDSR) generated during the procedure.

Ceiling mounted lead protection

The ceiling mounted shielding is kept at the same position close to the detector and the operator is always to some extent shielded.

Dose results from measurement

Mirion dosemeters were located at the mid-sagittal plane of the body and placed over the lead apron of the main operator (figure 24 right). The dosemeter readings were: chest level 7.0 µSv, lower abdominal 30.0 µSv, hip 9.0 µSv and lower leg 2.0 µSv. The cumulated values during the procedure are shown figure 24.

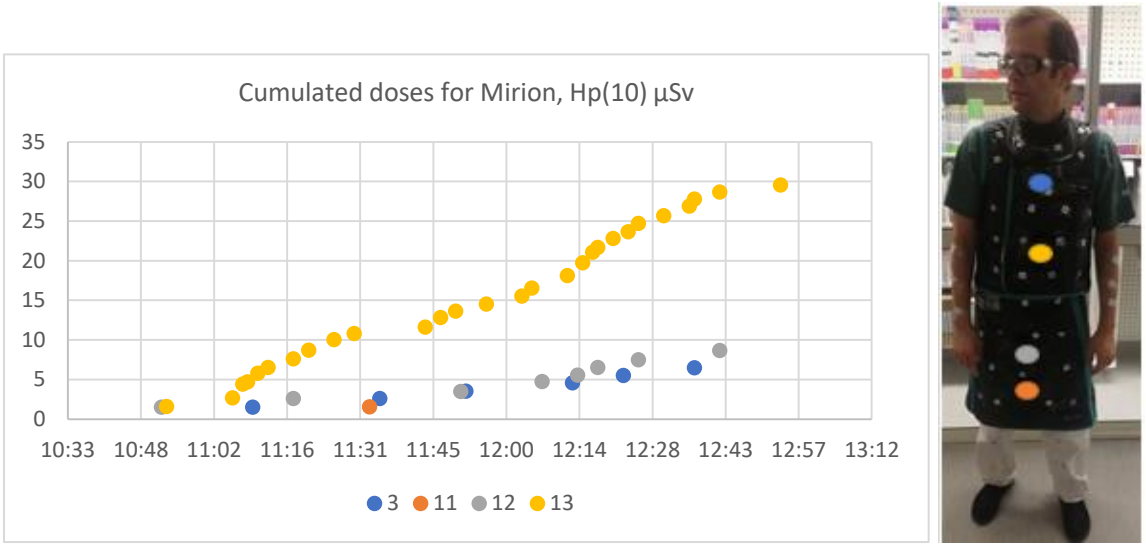


Figure 24. Mirion doses µSv, cumulated time dependent for four different Mirion detectors called 3 (chest), 11 (lower abdominal), 12 (hip) and 13 (lower leg).

The readings including the salt pellets are shown in figure 25. The radiation field reaching the operator is substantially non-homogenous and differ by a factor of about 6.

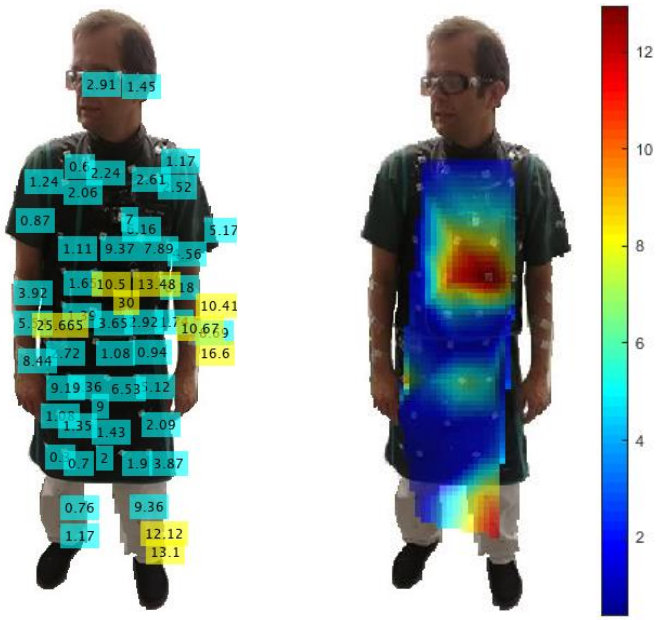


Figure 25. Dose values from measurements. Doses are given in Hp(10) in µSv. Left is the values measured with NaCl pellets (blue) and DMC (yellow). Right is an interpolated color map of the same dose distribution.

RDSR Data

The RDSR data for each irradiation event was transferred to an Excel file. Table height was corrected according to experimental measurement since position supplied in the RDSR file was not correct.

KINECT Data

In order to define the main operator location the CSV files (separate file for each Body ID) that were generated by the KINECT were used. An observer noted during the procedure which Body ID was assigned to the main operator during tracking.

3.2.2 Monte Carlo simulation

The ceiling shielding is not considered in the Monte Carlo calculations.

Monte Carlo simulation geometry

For *Penelope/penEasyIR code a* BOMAB scaled to patient's dimensions (female, weight = 56.2 kg and height = 160 cm) was used in the simulation. The upper part of the body was scaled according to section 2.4. However, with these criteria, the legs of the BOMAB were not realistic. For this reason, BOMAB's legs were modified based on Regina dimensions (Figure 26).

$H_p(10)$ was calculated for each irradiation event at operator's *Neck*, *SpineMid*, *SpineBase*, *LeftShoulder* and *LeftRight* according to kinect's tracking file. These points are located on average in the positions shown in figure 27.

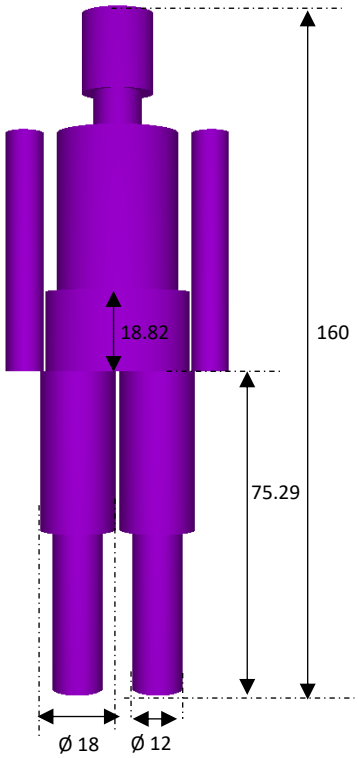


Figure 26: Geometry of BOMAB for case 6 (image obtained with gview3D).

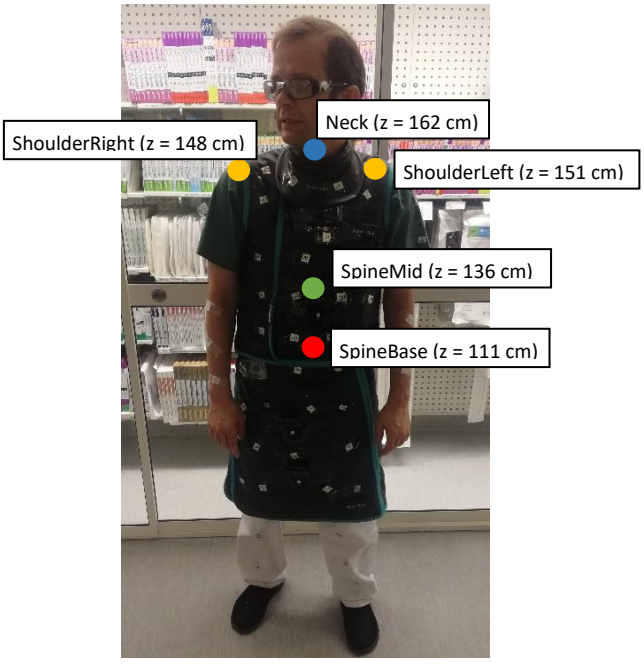


Figure 27: Positions where Penelope/penEasyIR calculation are performed.

For MCGPU-IR, $H_p(10)$ was calculated at several points on the operator chest. The programme also provides organ doses and the effective dose but the results are not reported because at this stage of the project the attenuation of the lead apron and collar are not taken into account.

Since the patient was a woman, the Regina phantom was selected. To simulate the monitored worker a scaled Regina phantom was considered (for MCGPU-IR the same phantom must be used for patient and worker).

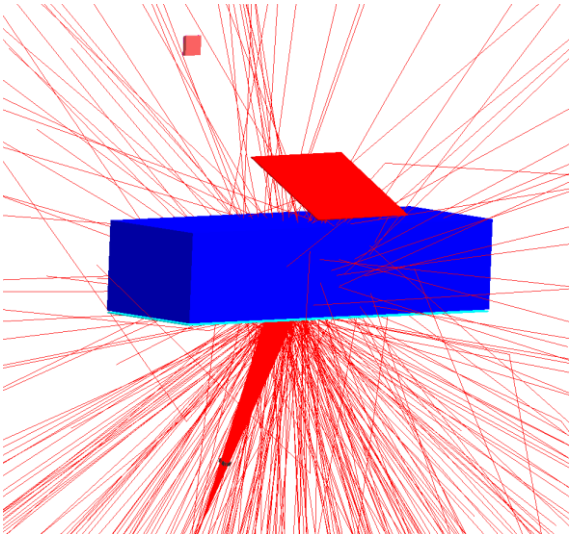


Figure 28: Geometry used in MCNPX simulations of Case 6

For MCNPx, *a* prism scaled to patient's dimensions (female, weight = 56.2 kg and height = 160 cm) was used in the simulation. $H_p(10)$ was calculated for each irradiation event at the location shown in figure 26. This location was derived from the operator's *Neck* joint according to the Kinect's tracking file. The ceiling mounted shielding was not simulated.

Monte Carlo results

186 irradiation events were simulated. The Monte Carlo results cannot be compared with the experiment because the simulations do not consider the shielding.

The simulation time was 120 s per event for *Penelope/penEasyIR.* Table 9 shows the *Penelope/penEasyIR* results (uncertainties with k=1) and figure 27 illustrates the position where $H_p(10)$ is calculated.

Table 9: PENELOPE/penEasyIR results for Case 6

	Adapted BOMAB (µSv)
Neck	47 ± 5
ShoulderRight	32 ± 3
ShoulderLeft	75 ± 8
SpineMid	50 ± 5
SpineBase	62 ± 6

Hp(10) was calculated with *MCGPU-IR* at the positions indicated in Figure 29. The red dot was situated at 123 cm from the floor. The red point was supposed to represent Mirion detector number 13 position. The simulation time was 40 s per event, 7050 s for the complete procedure. The statistical uncertainty was below 2.5% (k=1).

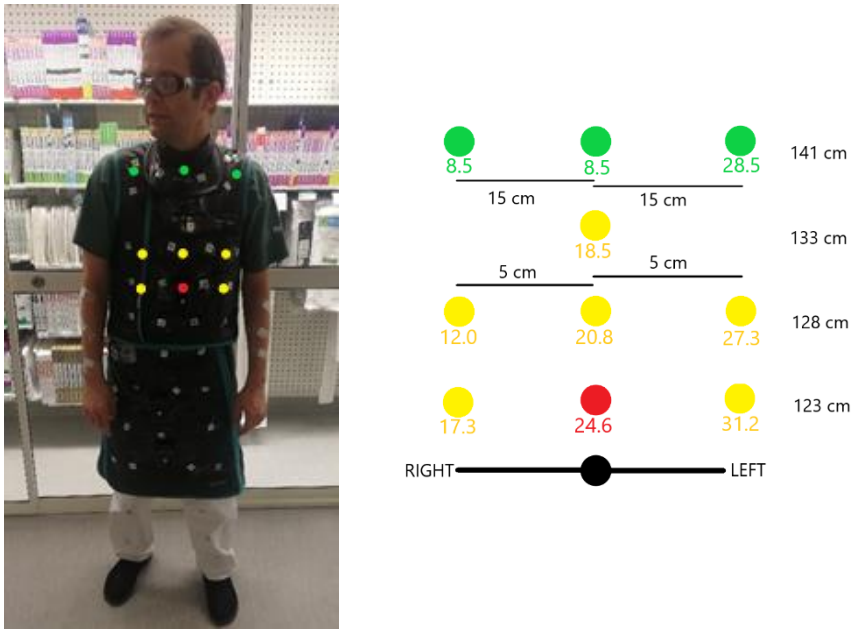


Figure 29: MCGPU-IR results in µSv for case 6 and measurement position.

To have an estimate of the dose distribution, $H_p(10)$ was also calculated at 5 cm to the right and to the left of the red point, 5 cm above these three positions and 10 cm above the red point (yellow dots). Finally, the doses at the shoulder level, 141 cm from the floor, were calculated at three positions (green dots). The calculated doses at these positions are indicated in Figure 29.

With MCNPx, $H_p(10)$ was calculated for each irradiation event at the location shown in figure 30 (red point). This location was derived from the operator's *Neck* joint according to the Kinect's tracking file. The total accumulated dose was 35 µSv. Figure 31 shows the accumulated dose after each event. The statistical uncertainty was below 10% (k=1) for each irradiation event. The simulation run time was 128 s per event for MCNPx.

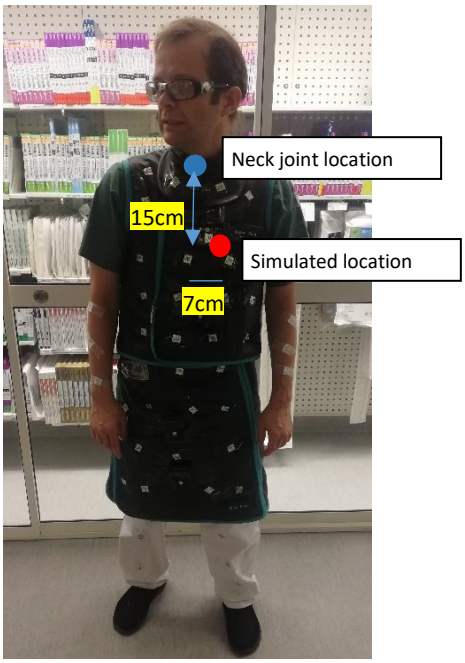


Figure 30: Dosimeter location in MCNPX simulations for Case 6.

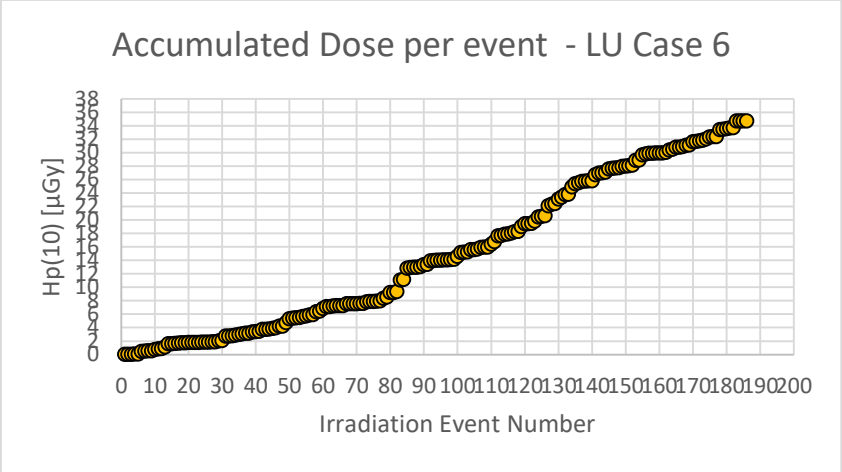


Figure 31: Accumulated dose during case 6 procedure, MCNPX simulations.

3.2.3 Discussion of the results

The procedure was a straightforward renal artery angiography procedure and the ceiling-mounted screen was used. One of the major limitations of the proposed verification is that the ceiling or table mounted shielding nor apron protection were not included in the Monte Carlo simulations. However, the latter do not affect the comparison between calculated and measured values on the surface of the body because the dosemeters were placed over the lead apron.

For the comparison of the three Monte Carlo codes, the position in figure 30 is considered. In this position, the ratio between MCGPU-IR and MCNPx is about 0.8 and the ratio between PENELOPE/penEasyIR (*SpineMid* point) and MCNPx is about 1.4. It is worth reminding that both PENELOPE/penEasyIR and MCNPx do not simulate the operator's body, thus disregarding any attenuation due to its presence . This case highlights the high inhomogeneity of the dose distribution, both the experimental data with salts detectors (Figure 25) and the MCGPUx calculation (Figure 29) show differences of 2 to 10 in 5 cm.

During the procedure there were no movements of the machine isocenter and, the ceiling mounted C-arm shielding was kept at the same position.

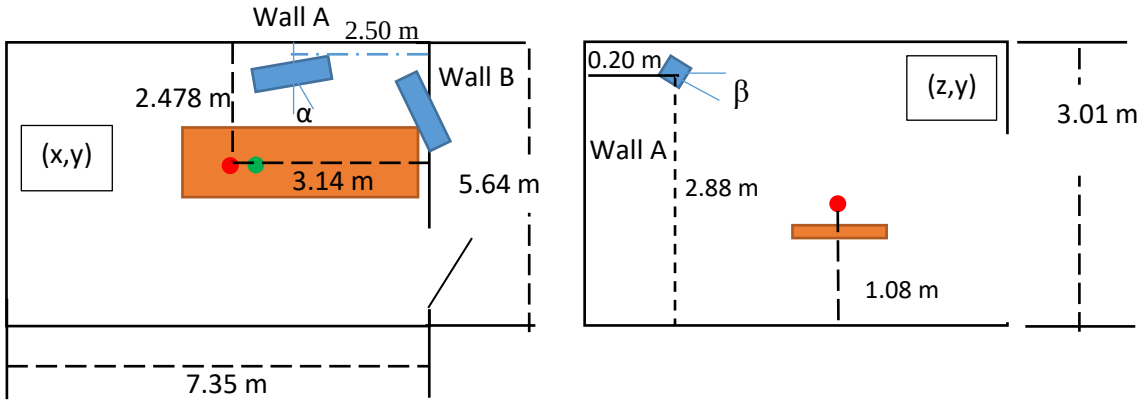
3.3 Skåne University Hospital, Malmö CASE 7

3.3.1 Measurement geometry and results

Reference Name: Case 7
Location: Malmö Hospital room 105
Date: 16th October 2019
Clinical Procedure: angiography of venous malformation around the knee

Room dimensions

The room dimensions of 105 and the placement of the Kinect are given in figure 32.



Blue: Kinect ("looking down")
Green: Plane of calibration during two camera system (27.8 cm between Green and red)
Red: Center of rotation during two camera system 3.14 m from red to wall B

Figure 32: Room dimension and position of the Kinect camera.

KINECT location

This procedure was completed using a more advanced set up with two KINECT cameras ("the 2 camera system). The positions of the cameras are shown in Figure 32.

Overview of Measurements

The main operator wore one RaySafe on the left side pocket above the lead apron. During the measurements no ceiling mounted shielding was used, the table mounted shielding was used.

The procedure was a very low dose procedure and it was observed that the Raysafe only accumulated doses during 7 of the 19 events.

Exposure field on patient

The right leg was exposed with a field size between 325 and 826 cm^2. All machine parameters were taken from the DICOM Radiation Dose Structured Report (RDSR) generated during the procedure.

Ceiling mounted lead protection

The ceiling mounted lead protection was not used during this case.

Dose results from measurement

The procedure was a straightforward angiography of venous malformation around knee procedure with a relatively low radiation exposure.

Figure 33: The relative position of the dosemeter and clavicles.

The RaySafe registered zero dose for 12 of the 19 irradiation events. The dose to the main operator at their chest level over the lead apron (figure 33) was 0.25 µSv (Hp(10)), during the 7 events.

RDSR Data

The RDSR data for each irradiation event was transferred to an Excel file. It contained 19 events, but only those 7 in which the active personal dosemeter recorded doses, could be actually compared with experimental values.

Table height was corrected according to experimental measurement since the position supplied in the RDSR file was referred to a previous procedure. Similarly, according to procedure, table lateral and longitudinal positions were modified in order to point to right knee.

KINECT Data

For the 2-camera system the distances for the Kinect data are given relative a fixed point in the operating room. This point is calibrated before the procedure starts. During this procedure were therefore this point determined and for all irradiation events. The table movement and the isocenter of the machine were also tracked together with relative distance to the Kinect calibrated isocenter.

3.3.2 Monte Carlo simulation

Monte Carlo simulation geometry

For Penelope/penEasyIR code a BOMAB scaled to patient's dimensions according to section 2.4. was used (female, weight = 60 kg and height = 160 cm). Figure 34 a, shows a scheme of the irradiated area in the simulation.

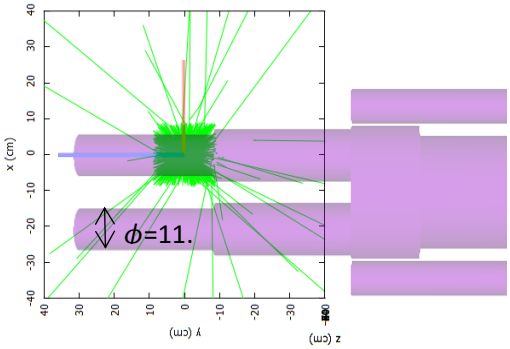


Figure 34a: Penelope/penEasyIR irradiated area.

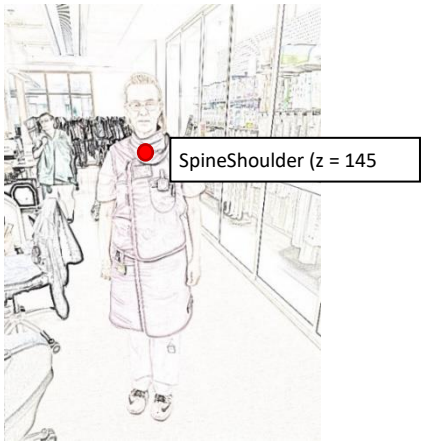


Figure 34b: Hp(10) calculation position.

$H_p(10)$ was calculated for each irradiation event at operator's *SpineShoulder* according to kinect's tracking file (z=145 cm, figure 34 b). This point is higher than Raysafe position according to experimental measurement and more centered.

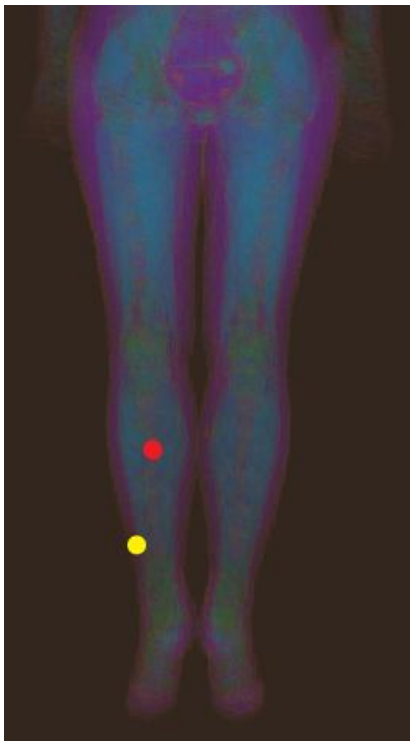


Figure 35 : MCGPU-IR Patient source position.

For MCGPU-IR the female Regina phantom was scaled according to section 2.4, final dimensions: 60.66 cm * 30.02 cm * 160 cm. The RDSR position of the source point is changed from 47.67 cm to 38.54 cm to ensure that the right knee is correctly irradiated in agreement with the radiological image. Figure 35 shows the change in source position, from yellow point to red point.

$H_p(10)$ is simulated in the center of the operator chest, this is not a fixed point in the space because the operator is moving, but in average for the 7 monitored irradiation events. It is located at 131 cm from the floor, a distance comparable to the position of the Raysafe, 125.7 cm.

Monte Carlo results

Simulation time for PENELOPE/penEasyIR was 120 s per event. The statistical uncertainty is below 0.5% (k=1). The cumulative estimated $H_p(10)$ for all 19 events is 0.53 ± 0.05 µSv (k=1). If only those ones reported by Raysafe are taken in consideration, the cumulated $H_p(10)$ is 0.30 ± 0.03 µSv (k=1).

Simulation time for MCGPU-IR was 59 s per event. The statistical uncertainty is below 2.5% (k=1). The cumulative estimated $H_p(10)$ for all 19 events is 0.25 ± 0.03 µSv (k=1). If only those ones reported by Raysafe are taken in consideration, the cumulated $H_p(10)$ is 0.13±0.02 µSv (k=1).

IPP_SE dose calculations

By means of a bash script, IPP_SE was executed with the 19 configurations corresponding to the irradiation events. The RAF phantom was used for simulating the doctor's body. While the axes of the coordinate system were the same of the camera system, the isocenter was transformed to be the origin of the coordinate system in order to match the transformations with the Monte Carlo based calculation methods. The total execution time was 1.5 minutes.

The cumulative estimated $H_p(10)$ was 1.52 µSv, while the estimated effective dose was 0.15 µSv. If restricted to the 7 events which were monitored by RaySafe, the $H_p(10)$ resulted in 0.77 µSv, while effective dose was 0.09 µSv. Figure 36 shows the buildup of dose through the 19 irradiation events, overlaid the corresponding DAP.

Considering that the procedure was relatively low dose and that the doctor moved by less than 30 cm through the various irradiation events, the increases in H_p (10) and effective doses showed to be mostly influenced by the DAP of each event.

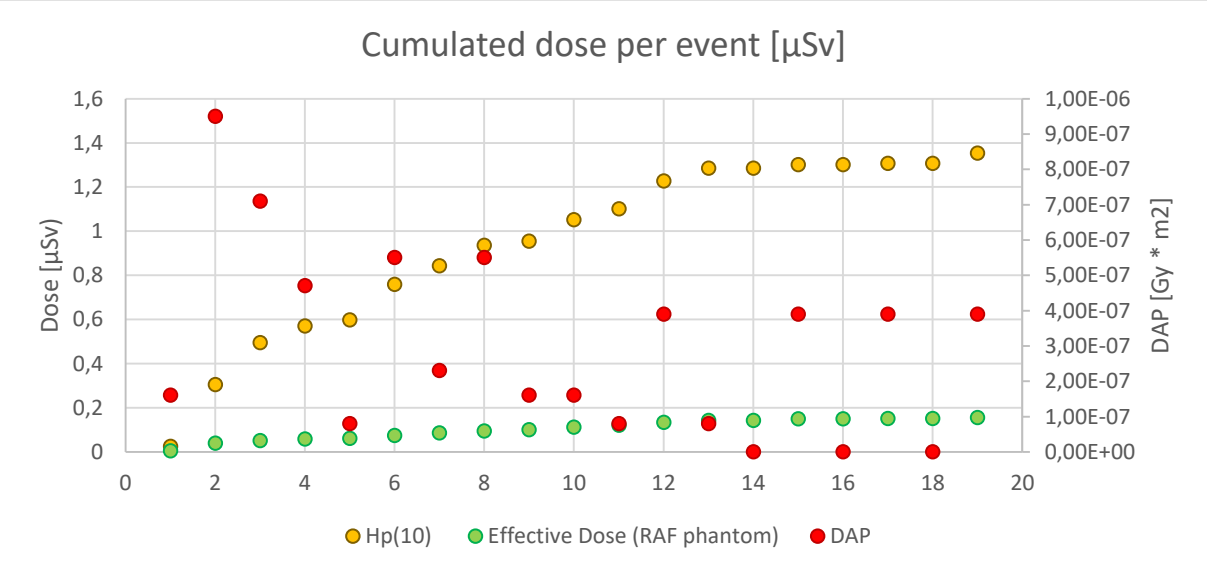


Figure 36: Cumulated dose per irradiation event in LU case 7, estimated by IPP_SE.

3.3.3 Discussion of the results

This case is, in principle, useful for the validation because the ceiling shielding was not used. However, the low dose introduced great uncertainty in the measured values.

The ratio between calculated and measured ranges from 1.2 (PENELOPE/penEasyIR) to 0.52 (MCGPU-IR). IPP_SE seems to provide a higher dose estimate in this case (factor of 2.9 higher than PENELOPE/penEasyIR).

3.4 St. James´s Hospital, CASE C

3.4.1 Measurement geometry and results

Reference Name: Case C
Location: Endovascular Theatre, SJH
Date: 16th July 2019
Clinical Procedure: Angioplasty with Iliac Stenting

Room dimensions

The length and width of the Theatre are shown, along with the approximate location of the KINECT camera within the room in figure 37. The KINECT is mounted on a moving object (Siemens TV monitor) therefore the KINECT can move during the procedure. In order to provide reference measurements for simulation, the KINECT calibration software tool was used to provide reference co-ordinates and distance to (i) the main operator (ii) the edge of the Flat-Panel Detector.

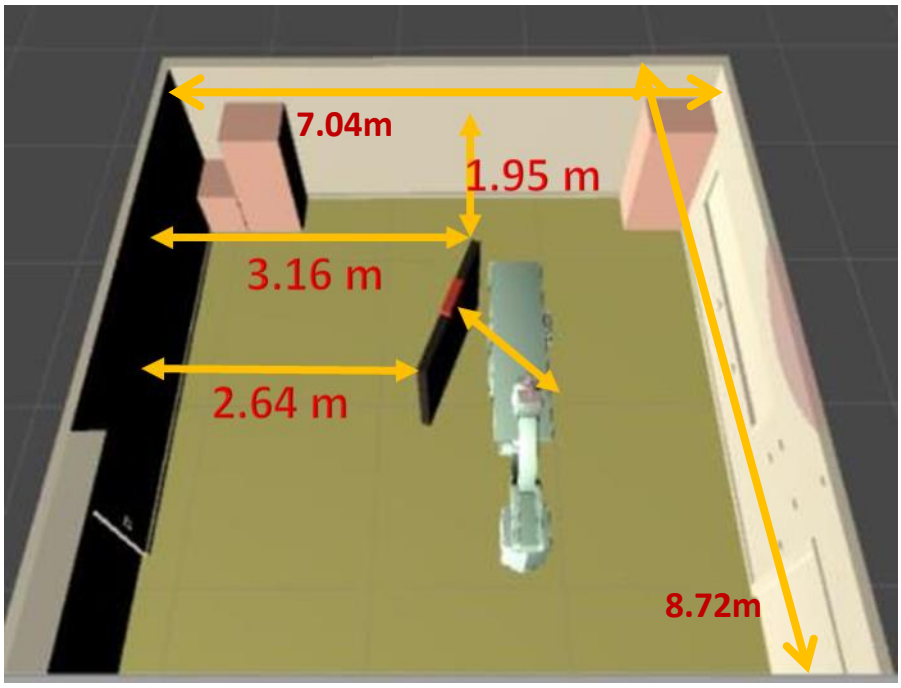


Figure 37: Room dimension.

KINECT location

KINECT calibration co-ordinates were corrected considering TV monitor movement.

Overview of Measurements

Measurements were performed on a Vascular Surgeon during an Endovascular Angioplasty with Iliac Stenting procedure. A Thermo APD was worn over the lead apron on the chest pocket of the main operator (figure 38).

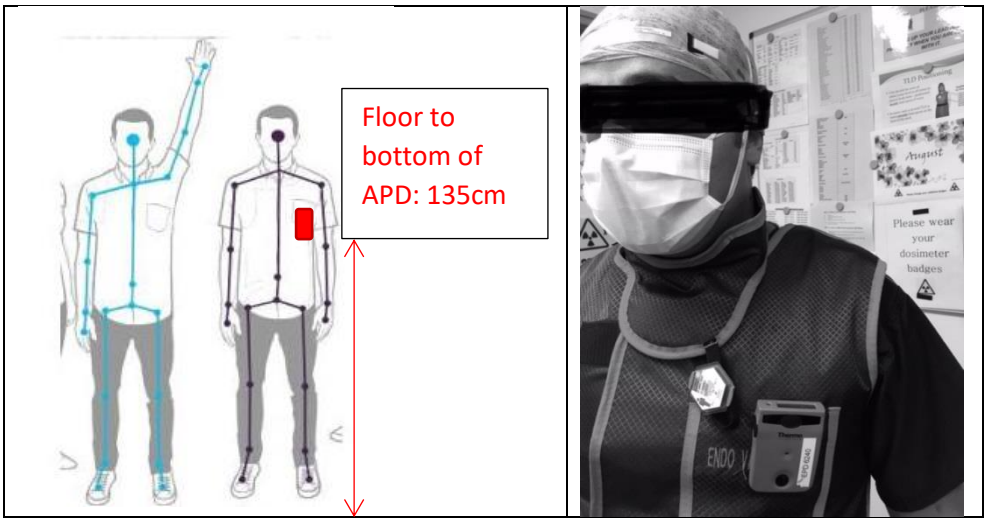


Figure 38: Dosemeter position.

Patient: Male, Average build (approx. 70kg, 177cm). Main Operator: Male, Average build (approx. 70kg, 177cm).

Exposure field on patient

The area imaged was the Abdomen/Pelvis/legs. The field size is not stored in the RDSR of this system which is an anomaly and could not be resolved with the version of RDSR software available during the validation measurements. The field size is estimated. At the start of the procedure, three stored DSAs of pelvis are 25cm x 19cm. Thereafter, the field size is 10cm (width) x 20cm (length) based on 6 stored DSAs.

Ceiling mounted lead protection

The ceiling mounted lead protection was not used during this case.

Dose results from measurement

The dose to the main operator at their chest pocket, over the lead apron was 55 µSv (Hp(10)). The patient DAP was 14.8 Gycm^2 and screening time of around 8mins (during a 45 min procedure). There were 67 irradiation events with a recorded DAP during this procedure. Events with zero DAP were filtered out.

RDSR Data

The RDSR for this case was anonymized and uploaded as an Excel file. In the file, for several rows, 13 of the 67, tracking data were not available for these irradiation events. In some cases the operator was 'lost' with no skeleton tracking, while other staff in the scene continued to be detected and tracked. The events where the operator was 'lost' comprise around 4 Gy cm^2 or 30% of the total exposure. The events were considered for the calculation and the operator position was inferred from his last recorded position.

KINECT Data

The data in the Excel file is the location of the main operator during each irradiation event. A single Excel file was compiled manually (from 6 CSV files generated during the case) based on observations.

The 6 CSV files were used to create a single composite file of Body IDs related to the main operator only, and manually matched one event at a time to the timestamp of each irradiation event. The complete original KINECT files are available (recorded @ 30Hz and then filtered to 1Hz), the largest of which is 15,000 rows with file size of about 14MB.

It was observed that for some events, the skeleton of the main operator was lost. For these cases, previous position is maintained.

3.4.3. Monte Carlo simulation

Monte Carlo simulation geometry

For Penelope/penEasyIR code*, a* BOMAB phantom was used. It was scaled to patient's dimensions following equations presented in section 2.4. $H_p(10)$ was calculated for each irradiation event at operator's *SpineShoulder* and *SpineMid* according to kinect's tracking file. Figure 39 shows the average heights. Usually dose is calculated in the *SpineShoulder* point, but in this case *SpineMid* was closer to experimental height of APD.

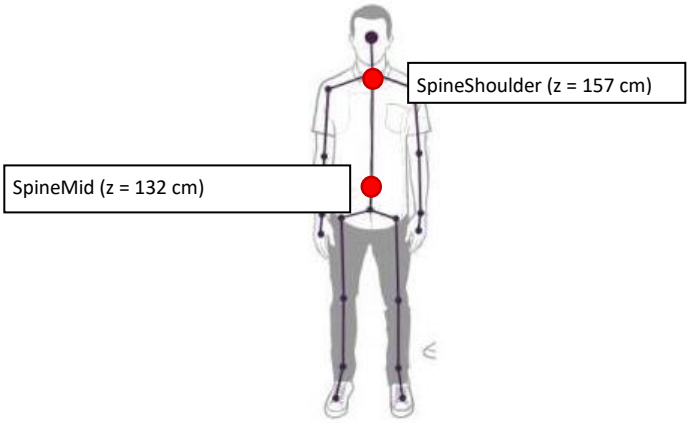


Figure 39: Position of the points of simulation for Penelope/penEasyIR

For MCNPx, the patient was simulated as a prism with the dimensions of 32 x 23 x 100 cm. Hp(10) is calculated at 15 cm from the neck joint downward and 7 cm to the left side as shown in figure 40 (close to the APD dosemeter). The movement of C-arm was not considered in the simulations. However, the movement of the table and the patient was considered in the simulations according to the table increment movement longitudinal and laterally found in the RDSR.

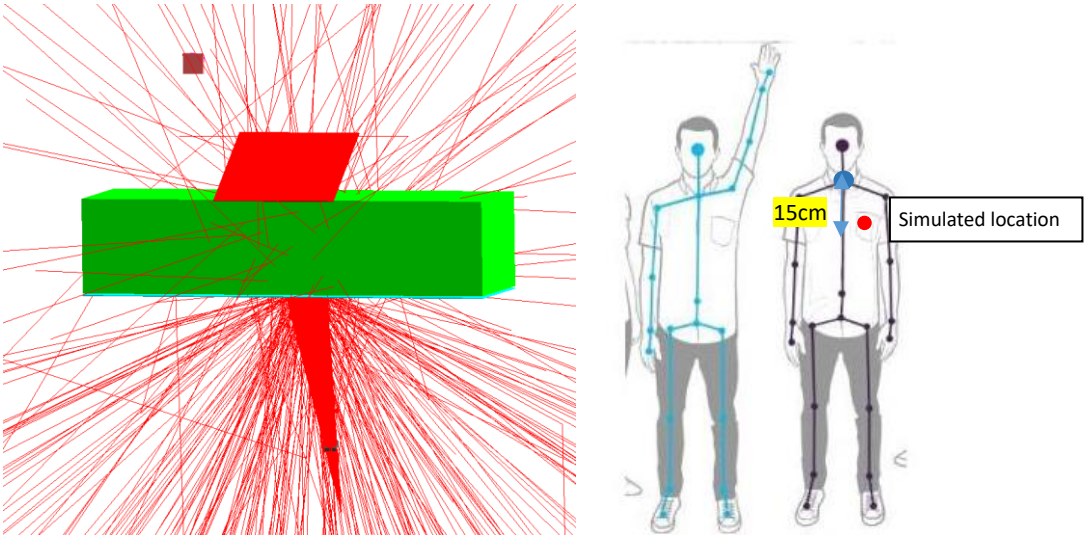


Figure 40: An image of the simulated patient (left), the simulated position on the operator (right).

Results PENELOPE/penEasyIR, MCNPx

Simulation time for Penelope/penEasyIR was 120 s per event. The statistical uncertainty is below 1%. The cumulative estimated $H_p(10)$ is 22 ± 2 µSv (k=1) for the spine shoulder and 18 ± 2 µSv (k=1) for the spine mid joint. In this case probably because the operator was farther from the patient, there is no much difference in dose between the two calculated positions

Simulation time for MCNPX was 120 s per event. The statistical uncertainty is below 5% (k=1). The cumulative estimated $H_p(10)$ is 55 ± 6 µSv.

IPP_SE dose calculations

By means of a bash script, IPP_SE was executed with the 67 configurations corresponding to the irradiation events. Due to the reported size of the main operator, the RAF phantom was used for simulating the doctor's body. While the axes of the coordinate system were the same of the camera system, the isocenter was transformed to be the origin of the coordinate system in order to match the transformations with the Monte Carlo based calculation methods. The total execution time was 3.5 minutes.

The cumulative estimated Hp(10) was 59 µSv, while the estimated effective dose was 14 µSv. Figure 41 shows the buildup of dose, overlaid is the corresponding DAP.

Through the irradiation events, the steadiest increases of cumulated doses correspond to 6 of the 7 angiography acquisitions. Event 51, i.e. the acquisition number 6, did not correspond to an increase of Hp(10). In this occasion, the position of the doctor, who took a safety distance (relatively far from the isocenter), was such that the image intensifier (rotated LAO29) completely shielded the dosemeter. On the other hand, in this event the effective dose increased by a relatively high amount, 0.6 µSv.

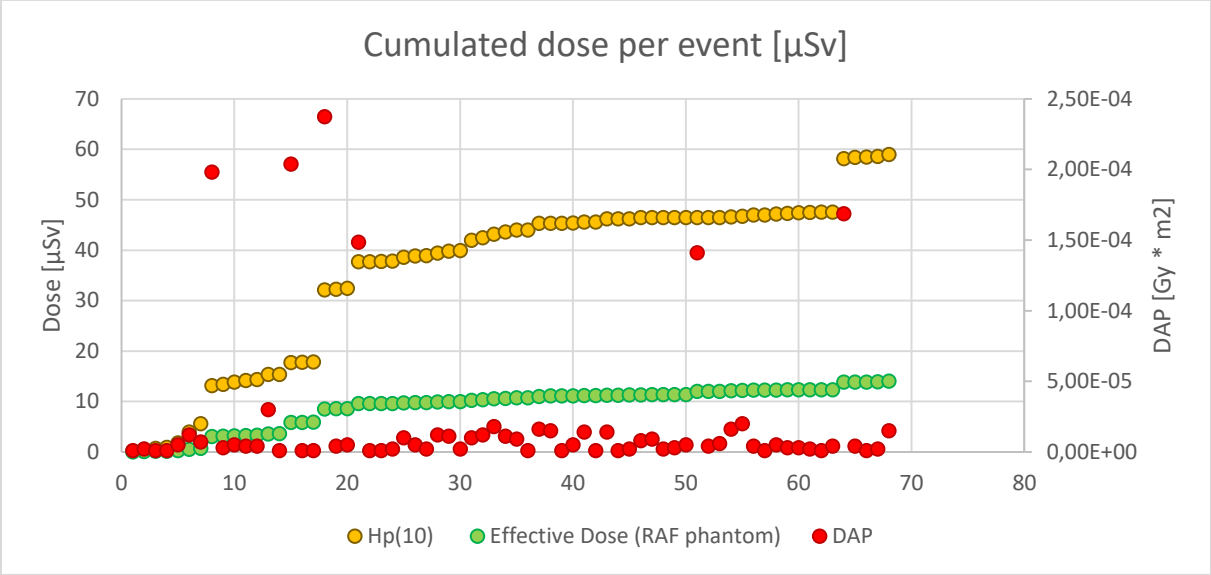


Figure 41: Cumulated dose per irradiation event in SJH case c, estimated by IPP_SE.

3.4.3 Discussion of the results

Tracking of the C-arm position relative to the main operator remains an issue to be solved as the movement of the C-arm is not automatically included in the RDSR, however C-arm movement was more limited in this case (compared to a lower limb angioplasty procedure).

The location of the KINECT on the TV monitor is a compromise and adds another variable to the validation measurements. The calibration procedure provides data when the monitor is moved.

The absence of skeleton tracking KINECT data for some irradiation events is noted. Procedures are performed in a teaching hospital and the role of main operator varies during the procedure. The doctors may swap places many times. At these times, the skeleton tracking may lose track of the main operator and other staff members may be tracked instead. The observer has typically noted the time periods when this loss of tracking occurs. The use of a two-camera system can address much of these problems.

Since the ceiling shielding is not used, this is a good case for the validation of the calculations. There is a very good agreement between MCNPx, IPP_SE and the Thermo measurements with differences within uncertainties. However, this is not the case for the *Penelope/penEasyIR* result which is around 60% below the experimental value. This has to be investigated further.

The results from the simulations with MCGPU_IR are not included because the input data concerning the joints are not very consistent in some tracks. This has major influence here because a complete phantom is simulated. This issue needs further investigation.

3 .5 St. James´s Hospital, CASE D

3.5.1 Measurement geometry and results

Reference Name: Case D
Location: Endovascular Theatre, SJH
Date: 6th August 2019
Clinical Procedure: EVAR

Room dimensions

The length and width of the Theatre are shown, along with the approximate location of the KINECT camera within the room. The KINECT is mounted on a moving object (Siemens TV monitor) therefore the KINECT can move during the procedure. In order to provide reference measurements for simulation, the KINECT calibration software tool was used to provide reference co-ordinates and distance to (i) the main operator (ii) the edge of the Flat-Panel Detector (Figure 42).

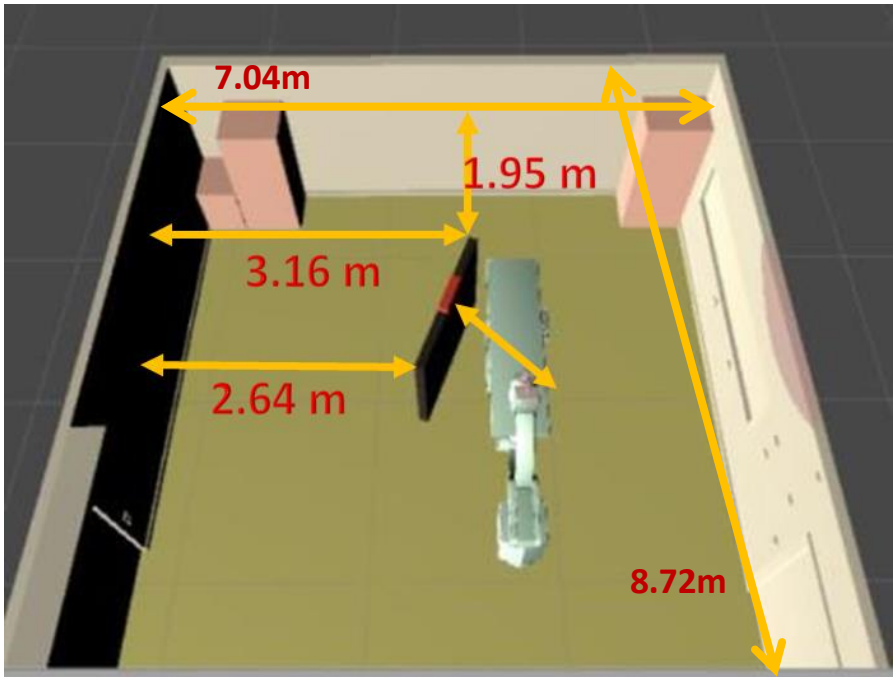


Figure 42: Room dimension and position of the Kinect camera.

KINECT location

KINECT calibration co-ordinates were corrected considering TV monitor movement.

Overview of Measurements

Measurements were performed on a Vascular Surgeon during an Endovascular Aneurysm Repair (EVAR). A Thermo APD was worn over the lead apron on the chest pocket of the main operator and the 2nd operator. An eye dosimeter (PHE headband) was worn by the main operator above the left eye as shown in figure 43.

Figure 43: Dosemeter position.

The patient was an adult male of average build, estimated to be 70 kg with height 170 cm. The operators are both adult males of standard build and height.

Exposure field on patient

Abdomen and Pelvis. The C-arm moved along the patient within a range of approx. 35cm. The field size is not stored in the RDSR of this system which is an anomaly and could not be resolved with the version of RDSR software available during the validation measurements. The field size is estimated. There are six stored DSAs of pelvis with field size of 14cm x 21cm. The following two DSAs are 23cm x 25cm. The final four DSAs are 15cm x 20cm.

Ceiling mounted lead protection

The ceiling mounted lead protection was not used at the beginning of this case. It is noted that during the DSA taken at 13:51 the lead screen was brought in and thereafter left in place. Once in use, it was kept at the same position very close to the detector during all DSA events. A representative setup of how the ceiling screen was positioned is shown in figure 44.

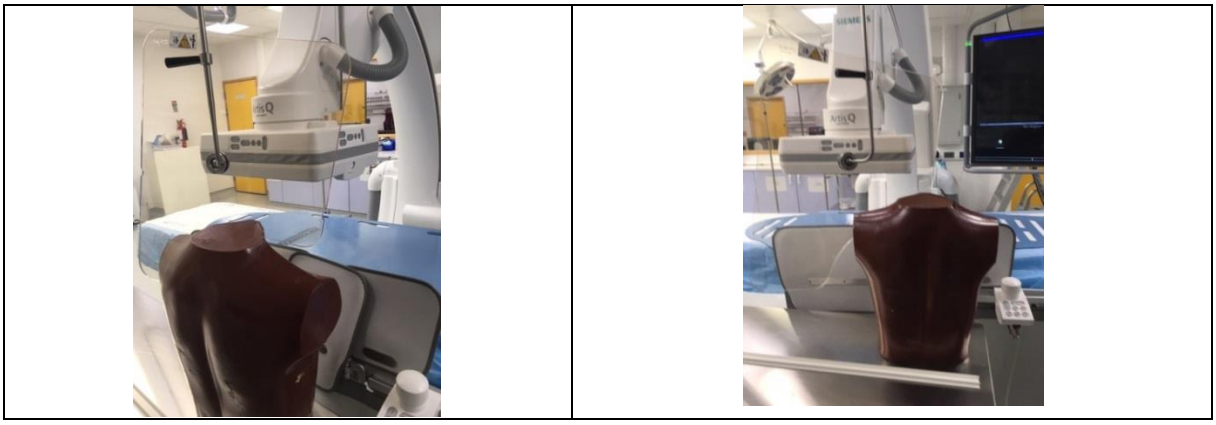


Figure 44: The position ceiling mounted lead protection.

Dose results from measurement

The dose to the main operator at their chest pocket, over the lead apron was 63 µSv (Hp(10)). The dose to the 2nd operator in the same location, was 25 µSv (Hp(10)). The eye dosimeter was accidentally discarded along with the surgeons disposable scrub cap before it could be collected and it could not

be located despite best efforts. This is a risk with dosimeters worn on a disposable cap and the process for collecting headband dosimeters has been improved.

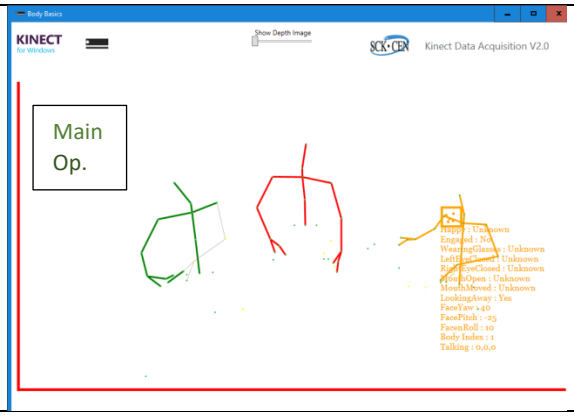
The patient DAP was 29 Gycm^2 and screening time of around 41 mins (during a 2hr 20min procedure).There were 206 events with a DAP recorded for this procedure. Events with zero DAP were filtered out.

RDSR Data

The RDSR for this procedure was anonymized and uploaded as an Excel file. In the file, several rows are coloured grey, as no clear tracking data was available for these irradiation events. In some cases the operator was 'lost' with no skeleton tracking, while other staff in the scene continued to be detected and tracked. Events with zero DAP have been filtered out.

KINECT Data

The data in the KINECT excel file should contain the location of the main operator during each irradiation event. To achieve this a single Excel file is required (based on searching the 6 CSV files generated during the case) to identify the body ID of the main operator only, and then matched one event at a time to the timestamp of each irradiation event. There is a very large amount of data over a 2.5 hr recording. Manually matching one RDSR event with one timestamp in the correct Body ID file takes around 5-6 minutes. There are 206 events in this RDSR which would result in around 1,000–1,200 mins (16-20 hrs) to compile the KINECT file for the main operator. Therefore to address this limitation an automated algorithm was developed by SCK to parse through all the files, locate the main operator in each CSV file and generate one composite file.



Skeleton tracking image from 12:47:33 showing Main Operator as GREEN (ID=2)

Skeleton tracking image from 13:10:05 showing Main Operator as PURPLE (ID=4)

Figure 45: Images of the tracking of the skeleton.

3.5.2 Monte Carlo simulation

Monte Carlo simulations have not yet been completed for this complex case and work is continuing on the input data required for the calculations.

3.5.3 Limitations of this validation measurement

The C-arm moved around 35 cm along the length of the patient, mainly imaging the abdomen area. Tracking of the C-arm position relative to the main operator remains as a problem to be solved.

The location of the KINECT on the TV monitor is a compromise and adds another variable to the validation measurements. For this case, the TV monitor was moved closer to the main operator around

15mins after screening started and thereafter remained in a static position. This was taken into account by carrying out a calibration measurement after the KINECT moved.

3.6 St. James´s Hospital, CARDIAC CASE 1

3.6.1 Measurement geometry and results

Reference Name: CARDIAC CASE 1
Location: Philips Bi-plane Cardiac Cath Lab Theatre, SJH
Date: 1st October 2019
Clinical Procedure: PCI (Percutaneous Coronary Intervention)

Room dimensions

The length and width of the Cath Lab are shown in figure 46, along with the approximate location of the KINECT camera within the room. The KINECT is mounted on a moving object (Philips TV monitor) therefore the KINECT can move during the procedure as shown in figure 47.

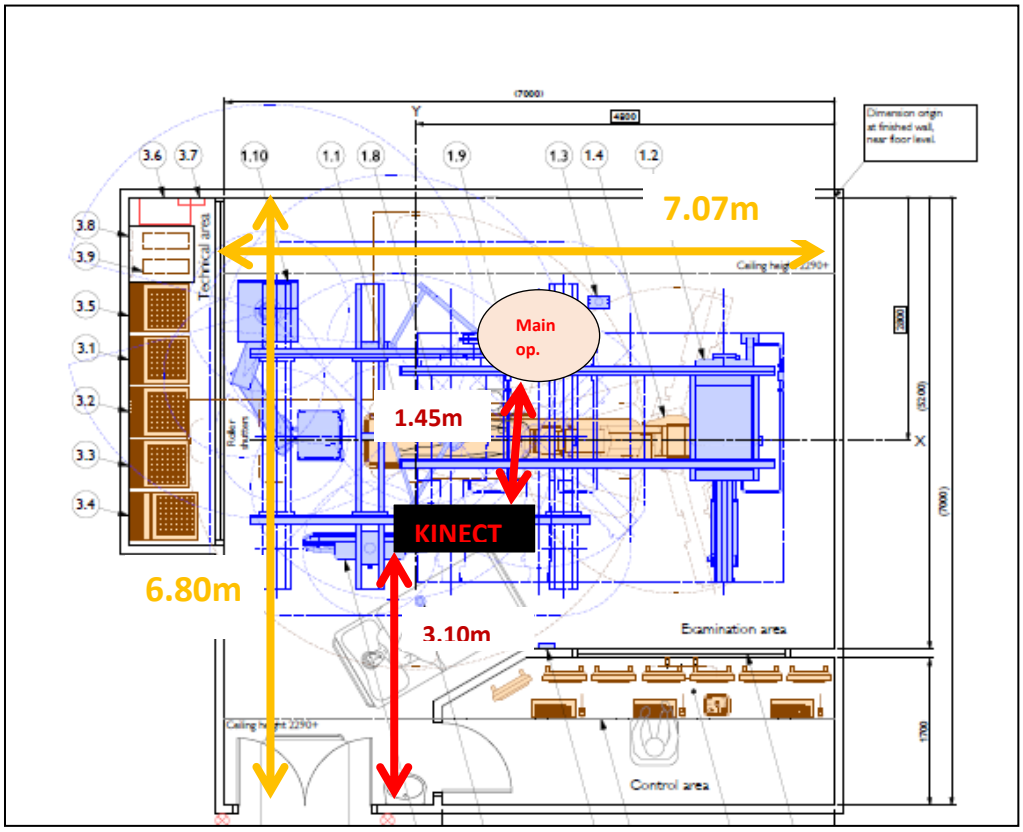


Figure 46: Room dimensions of the Cath Lab.

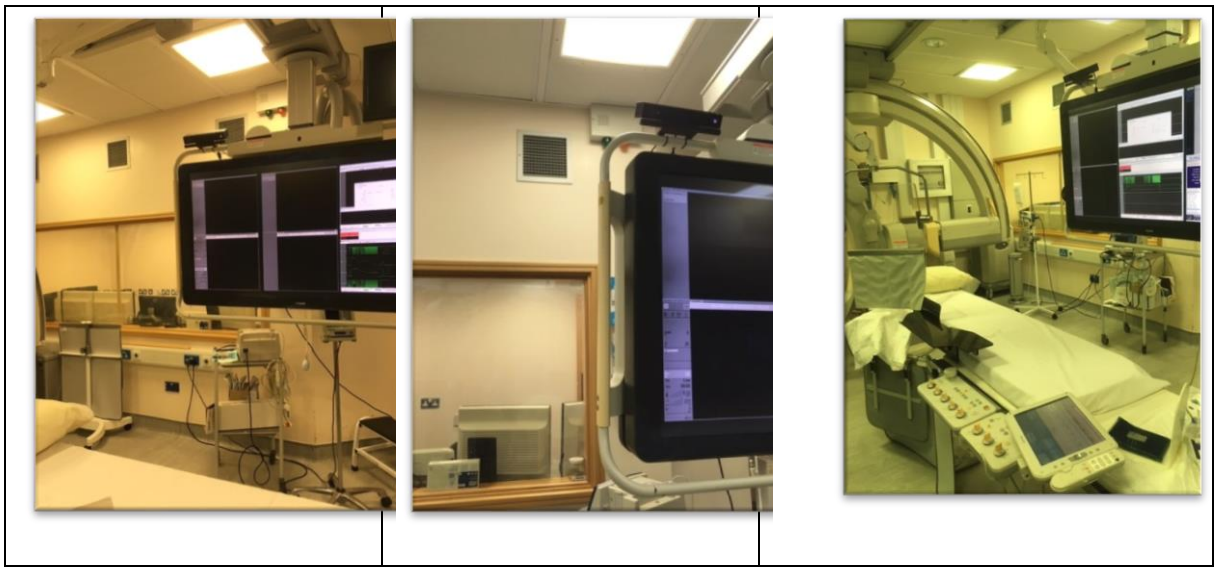


Figure 47: The position of the Kinect.

KINECT calibration co-ordinates from position on TV monitor

In order to provide reference measurements for simulation, the KINECT calibration software tool was used to provide reference co-ordinates and distance to (i) the main operator (ii) the edge of the Flat-Panel Detector.

Overview of Measurements

Measurements were performed on an Interventional Cardiologist during a PCI procedure. A Thermo APD was worn over the lead apron at the chest of the main operator, positioned slightly on his right, there is no pocket on this lead apron to easily attach, figure 48.

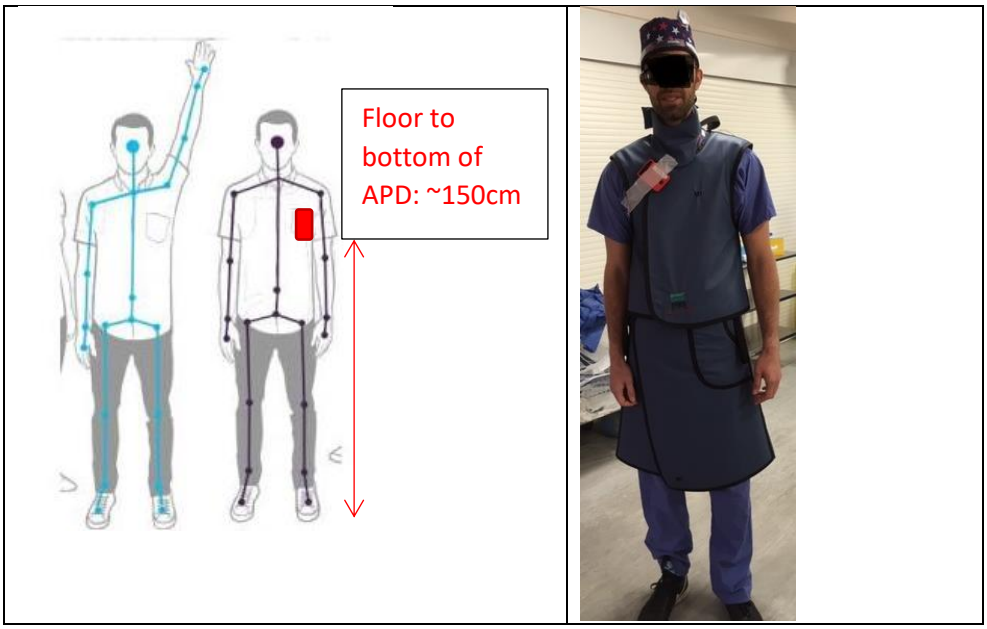


Figure 48: The position of the dosemeters.

The main operator is a male, tall build (193 cm). The typical floor-to-table height for this operator is approximately 95 cm. The Philips system also references the Table Height Position (mm) to the floor, and the mean Table Height for this case taken from the RDSR is 96.4cm.

Exposure field on patient

The field size for this procedure can be calculated using the data in the RDSR (shutter positions). The The patient was a male, average build (approx. 70kg, 177cm). The patient DAP was 76 Gycm^2 and screening time of around 33 mins (during a 1 h procedure).

Ceiling mounted lead protection

The ceiling mounted lead protection (of 0.5mm Pb as shown in figure 50 below) was always positioned in front of the main operator during this case as shown in the depth-map image below. Data showing typical movement of the screen in this Cath Lab were recorded on a Meta-Motion IMU. However the exact position is not easily tracked and this information could not yet be used in simulations.

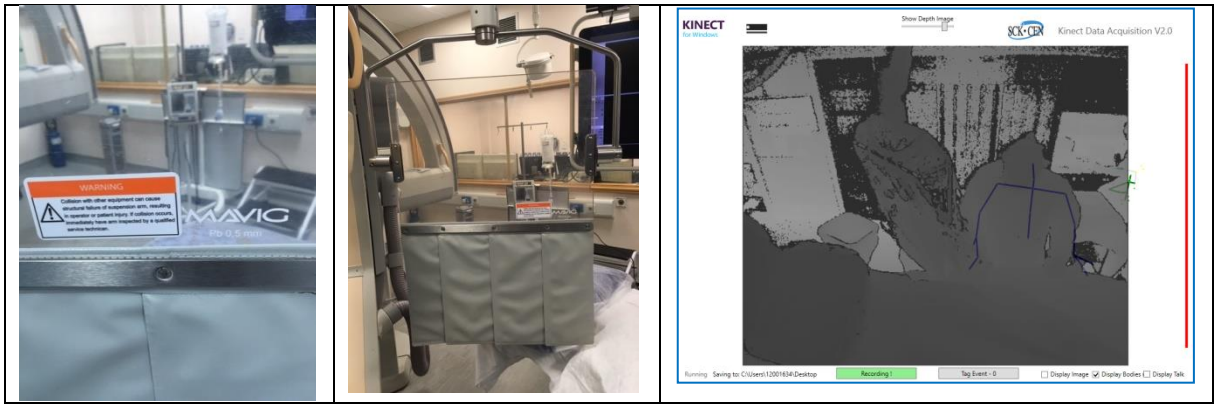


Figure 49: Lead shielding used in the Cath Lab.

Dose results from measurement

The dose to the main operator at their chest, over the lead apron was 31 µSv (Hp(10)).

RDSR Data

The RDSR for this procedure was anonymized and uploaded as an Excel file. It includes 163 irradiation events.

KINECT Data

The raw data in the Excel files (ID 3 and 5) have been filtered and transformed to automatically compile a single composite file of Body IDs related to the main operator only, and related to the timestamp of each irradiation event. The complete original KINECT files are available (@ 30Hz), the largest of which is about 90 MB. Subsequently, the file was filtered to store the data for the main operator @ 1 Hz.

3.6.2. Monte Carlo simulation

The RDSR position of the source point is changed to ensure that the primary beam is located at the heart.

During the procedure, the system included wedges to reduce the patient dose but this cannot be taken into account by the Monte Carlo codes. However, using the recorded DAP value in the normalization partially overcomes this problem because the DAP is also reduced by the attenuation in the filtration.

Monte Carlo simulation geometry

For Penelope/penEasyIR code, *a* BOMAB phantom was used. It was scaled to patient's dimensions following equations in section 2.4. $H_p(10)$ was calculated for each irradiation event at operator's

SpineShoulder joint according to kinect's tracking file (z=157.5) and *Shoulder Right* (z=150). These points (figure 50) are the closest ones to experimental measurement.

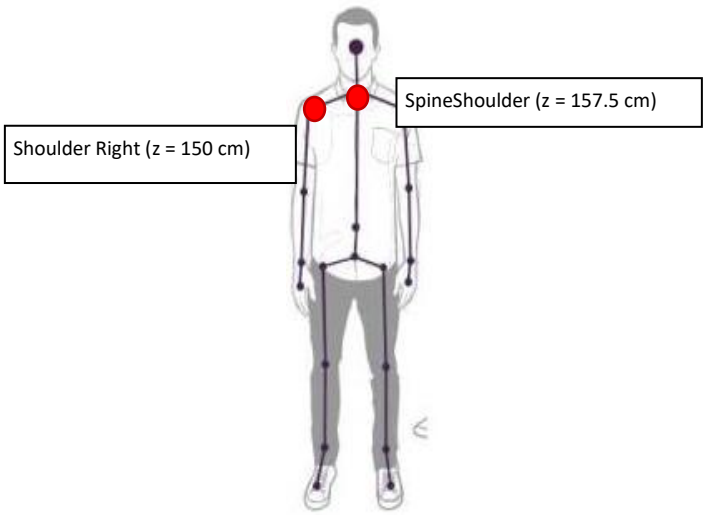


Figure 50: Positions where Penelope/penEasyIR calculation are performed for case 1 cardiac.

For MCGPU-IR the male REX phantom was scaled according to section 2.4. $H_p(10)$ is simulated in the center of the operator chest at 131 cm from the floor.

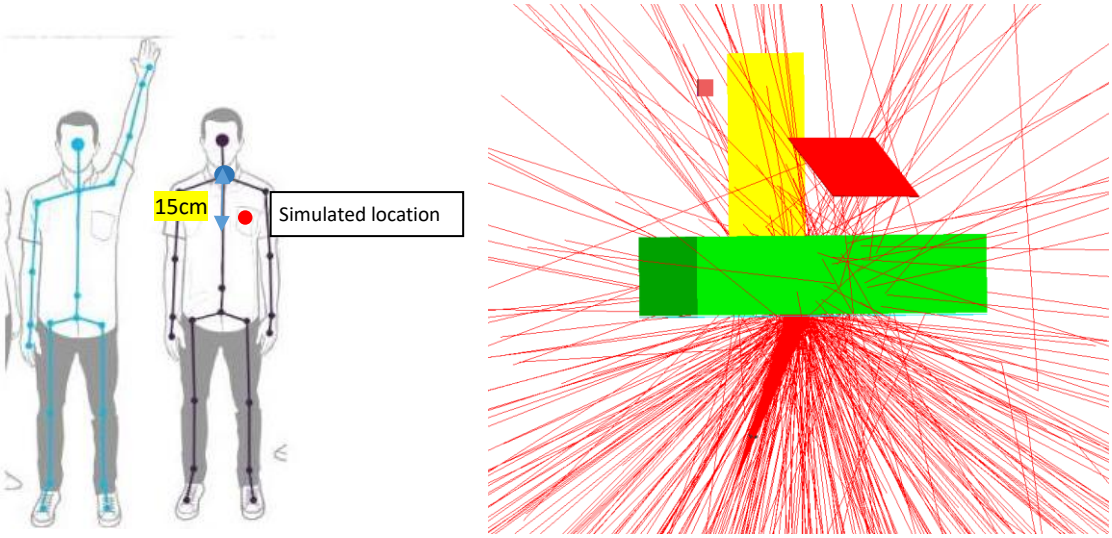


Figure 51: Positions where MCNPx calculation are performed for case 1 cardio.

For MCNPx, the patient was simulated as a prism with the dimensions of 32 x 23 x 100 cm. The dosimeter is located at 15 cm from the neck joint downward and 7 cm to the left side as shown in figure 52. The movement of C-arm was not considered in the simulations. However, the movement of the table and the patient was considered in the simulations according to the table increment movement longitudinal and laterally found in the RDSR. The simulations were performed with and without the ceiling mounted shield.

Monte Carlo results

The 163 irradiation events were simulated using PENELOPE/penEasyIR code with 2-min simulation time per irradiation and a statistical uncertainty lower than 1%. The simulated accumulated dose without considering the shielding is 670 ± 67 µSv at the spine shoulder and 475 ± 47 µSv at the right shoulder (with k=1). One can observe a difference of 30% between the two selected calculation positions.

For MCNPx, 163 irradiation events were simulated with. 2.4 min simulation time per irradiation and a statistical uncertainty of 5% (k=1). The simulated accumulated dose without considering the shielding is 655 ± 97 µSv at the dosimeter location.

The 163 irradiation events were simulated using MCGPU-IR code with 25 seconds simulation time per irradiation event, and about 1 h for the whole procedure. The statistical uncertainty is of 1.8% (k=1). The simulated accumulated dose without considering the shielding is 252 ± 28 µSv (k=1).

3.6.3 Discussion of the results

One of the major limitations to validate the calculations is that the results do not consider the attenuation of the shielding because its precise position is not known. However, using a picture of the procedure, the shielding was introduced in the PENELOPE/penEasyIR simulation geometry, at a fixed location for the complete procedure. The coordinates of the shielding were: X=0, Y=35 cm, Z=20cm with a rotation of 35º. The dimensions of the shielding are 35 cm*70 cm and the thickness 0.05 cm Pb. Similarly, in MCNPx, as the shield position was not tracked, an approximate position and orientation was assumed. The shield dimensions considered are 350 x 700 x 0.5 mm with lead equivalent material and located also at X=0 , Y= 35 cm, Z = 20 cm in the coordinate system where the iso-center is the origin. The shield was also oriented at 35° on Z axis. With the shielding in this position, Hp(10) is 105 ± 10 µSv (k=1) at the spine shoulder and 90 ± 9 µSv (k=1) at the right shoulder. Thus, the shielding would have attenuated 85% – 90%. If this attenuation is applied to the unshielded Hp(10) calculated with MCGPU-IR is about 53-40 µSv. The APD reading is 31 µSv, which is lower than the simulated values, but considered acceptable considering all the hypothesis and simplifications considered. The differences between the two Monte Carlo codes are higher than in other cases. For MCNPx, Simulated accumulated dose for all 163 events is 109 ± 17 µSv with the shield at the same position.

4. Summary and discussion

This report covers 6 different measurements to validate the system.

The first – Case 5 – was also included in the previous report regarding task 4.1 but is included to show possible differences when using three different simulation methods using different Monte Carlo simulation techniques (PENELOPE/penEasyIR, MCNPx, MCGPU-IR). The first two simulation methods use simplified phantoms to simulate the patient and any operator phantom is simulated. The third, MCGPU-IR, is faster and uses a voxel-based phantom that simulates both the patient and the operator. The ratios between the result using PENELOPE/penEasyIR and MCGPU-IR varied between 0.75 and 1.43. The ratio between measured and simulated values using PENELOPE/penEasyIR varied between 0.63 and 1.16 and corresponding values for MCGPU-IR are 0.56 and 1.10. The geometrical description of the problem for MCNPx is similar to PENELOPE/penEasyIR and therefore the results were closer to PENELOPE/penEasyIR results than to MCGPU-IR results.

The first measurement during a clinical intervention – Case 6 – showed greater differences than Case 5 between simulation methods. But in this case, the motion tracking system is also in use adding to the complexity of the simulations compared to Case 5. Comparison with experimental values were not possible since ceiling shielding used during procedure was not simulated.

For the validation Case C, the ceiling mounted lead shielding is not used and this is a good case for the validation of the calculations. There is a very good agreement between MCNPx, IPP_SE and the Thermo measurements however, this is not the case for the Penelope/penEasyIR and this has to be investigated further.

Results from the simulation of a cardiac procedure (Cardiac case 1) show significant differences between simulated and measured values. This is due to the fact that the ceiling mounted shielding was in use during the full case and the simulations do not take this into account. If simulation codes are compared, PENELOPE/penEasyIR and MCGPU-IR shows a ratio about 0.75-1.05 without shielding.

The other cases – Case 7 and Case D – shows the difficulties when comparing measured and simulated values in some complex situations. That is, in Case 7, the dose to staff during one treatment is very low and the dose measurement exhibit particularly high uncertainties. Therefore, it is hard to draw firm conclusions from this case. The tracking data and RDSR from the Case D were complex as the main operator body ID changed many times over the course of a long procedure with many irradiation events. Work will continue on simulating this case.

The results obtained from the different simulation strategies - PENELOPE/penEasyIR**,** MCGPU-IR, MNCPx and IPP_SE needs to be further investigated. However, the results given in this report may indicate the variability using the different simulations.

These first validations in the clinic serve as a valuable input to the improvement of the system. It is obvious that the input values, both from the X-ray machine and Kinect, must be collected automatically. The definition of the isocenter - the center of the irradiation field - must be automatically identified. Improvements are also needed in terms of data collection from Kinect and matching to irradiation events. The issues to be solved are elaborated more in detail in report D9.114 “Report summarizing the experimental and clinical findings when using the online dosimetry application”.