

Monte Carlo study of a X-ray examination, the basis and the effects of the external radiation shields

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Abstract

Syventävä aineopintojen labra. Eunice-phantomi, annoslaskennan simulointi etc

Keywords

BodyPhantom — Eunice — ImpactMC — Dose map

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Introduction

The radiological examination provides valuable information about the health of the patient and plays an important role in helping a doctor to make an accurate diagnosis. However, such an examination exposes the patient to the radiation, thus the benefit of the examination and the possible harm of the radiation dose must be cooperatively taken into account at a planning stage.

It has been shown, that the most common and cost-effective examination in conventional radiography is the radiological examination. (Helasvuo 2013, Speets et al. 2006, Veldkamp et al. 2009, McAdams et al, 2006). Still, the effects of the external shields to reduce the dose are yet to be studied. One of possible approaches for such investigation is simulating the situation and combining the produced situational possibilities to investigate the differences.

The most essential part of this approach is the choice of the simulation engine. The ImpactMC software has been shown correctly reproducing the situation (VIITTEET), thus it was selected as a simulation creation tool. In order to avoid the exposure of the patients to the dose, those analysis are done using the Eunice bodyfantom (VIITE). By using the phantom, it is possible to not only simulate the situation, but also to verify the simulation by direct measuring.

The aim of this paper is to step by step describe the process and make a preliminary conclusion about the effect of the external shields to the dose.

Methods

This section is separated into 3 steps:

1. Imaging the phantom and preprocessing
2. ImpactMC simulation
3. Analysis of the results

1. Methods and theory, preprocessing

This section covers all the processing needed to be done before the simulation.

1.1 CT imaging principle

Let's start with the small snap of theory about Computed Tomography (CT) imaging. The CT-imaging bases on ability of X-ray to pass through the object without significant change of the ray direction, and proportional to signal weakening while passing through material. In other words, there are two assumed postulates:

1. X-ray direction doesn't change while passing through the object
2. The X-ray attenuation proportionally depends on the density of the passed material.

Those assumptions result in situation illustrated in Fig. 1

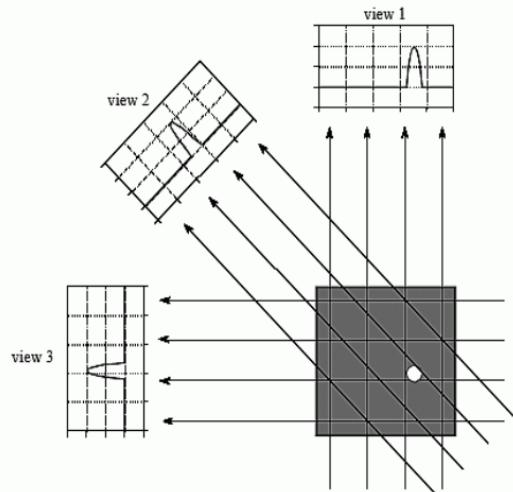


Figure 1. The imaging principle of CT.

The next step is the inverse problem of combining the multiple directional scans into forming the image. The simple version of reconstruction called "Backprojection" is illustrated in Fig. 2.

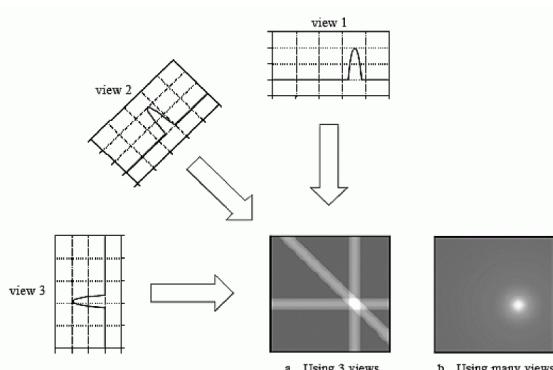


Figure 2. The reconstruction principle of CT.

Nowadays, the reconstruction algorithms are not that simple, there are many filterings applied in order to improve the image quality. In more detail, the reconstruction algorithms are described in the course "Inverse problems" by Samuli Siltanen, Math department at the University of Helsinki.

1.2 Eunice phantom

1.2.1 The structure

The Eunice phantom is a combination of plastic disks with holes for dose measurement wires. (Fig. 3) The consistence of each disk is different and reproduces the organ densities.

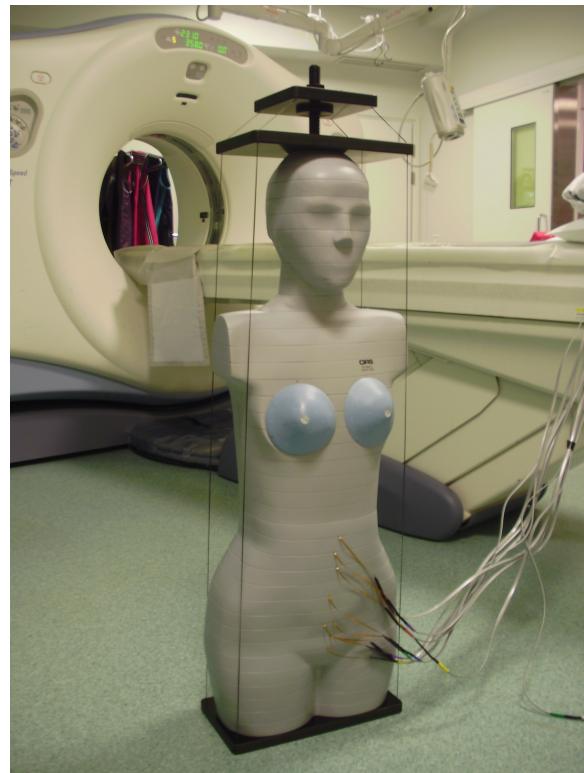


Figure 3. A photo of the Eunice phantom.

In Figure 3 on the right half some wires are visible. Those wires are connecting the MOSFET dosimeters inside the phantom to computer. In this laboratory work this possibility was not used. However those dosimeters inside the phantom provide a measurable connection between the simulation results and reality's doses.

1.2.2 In general

The first step is to obtain the image of the object of the investigation, in other words to get a 3D image of the phantom. The best results for the studies are produced at 129kV (Annan viiteet), however at equipment available at HUS, such sequence is not an option. The closest options are 120kV and 140 kV. Thus, for the analysis the 120kV sequence has been chosen.

The obtained image is illustrated in Fig. 4. (Thanks to Timo Paasonen for measurement.) The dimensions of the image are 512-512-377, with voxel size of 1.5625mm-1.5625mm-2.5mm.

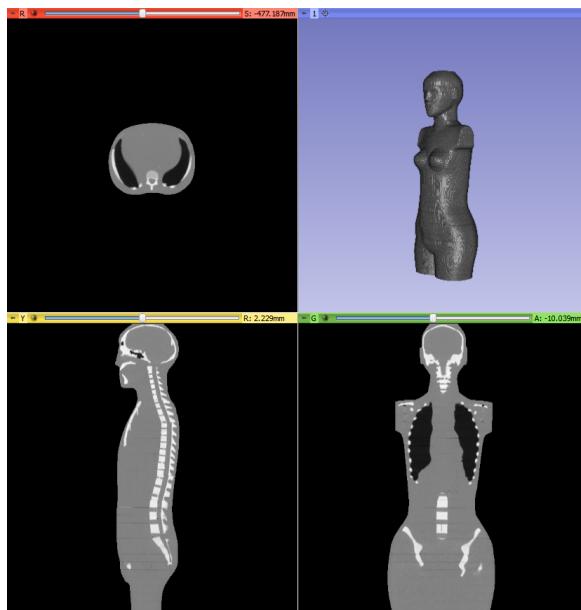


Figure 4. The structure of Eunice phantom visualized as 3D image with axial (red), sagittal (yellow) and coronal (green) slices.

1.3 CT imaging direction

The radiation dose effect is tissue dependable (VIITTEET), thus the imaging direction is chosen to be posterior-anterior. (Fig. 5) The benefit of such selection is minimization of the dose to baby or to a radiation sensitive organ.



Figure 5. The CT-imaging session.

1.4 Types of external shields

This paper analysis the four types of external shields, based on their height. The Eunice phantom consists of 38 disks, so the types of shields are defined accordingly: A (disks 21-38), B (23-38), C (24-38), D (25-38) and K (loose D). (Fig.6.)

2. Methods and theory, the simulation

The general idea of the simulation is simple:

1. Using spectrum create n-amount of particles
2. Pass them through the object
3. Record the result

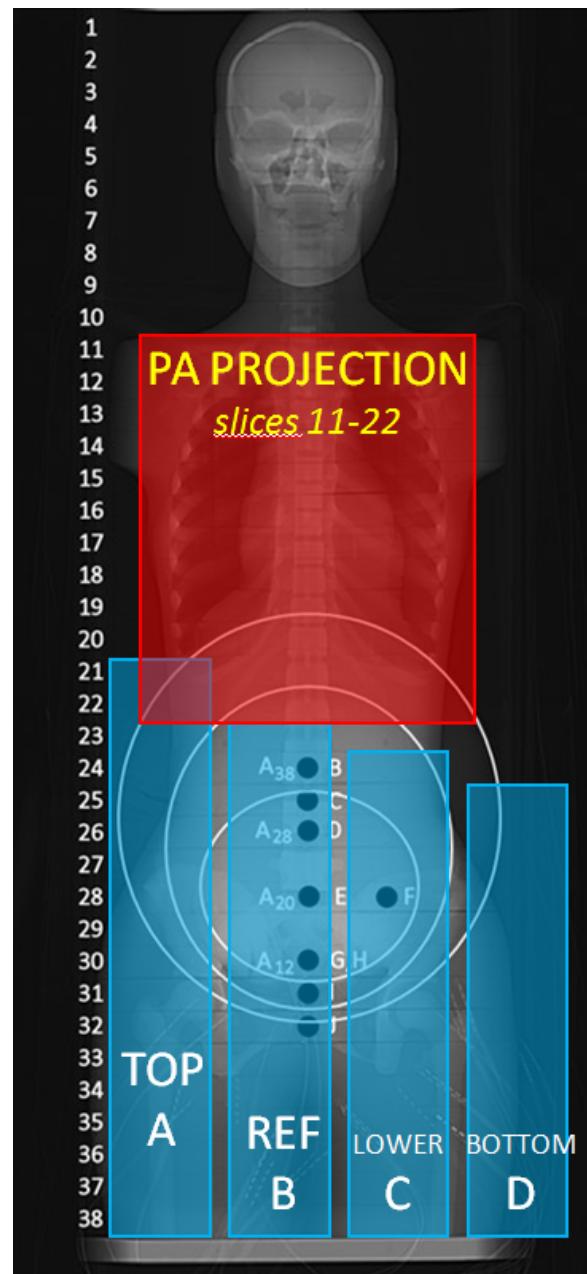


Figure 6. Types of shields

However, at a closer look, several questions like "What spectrum?", "How many particles?" and "How do they interact?" arise. This section will try to answer to those questions

2.1 Spectrum

The model for spectrum is obtained from XXXXXX, and it's probability density function is illustrated in Fig. 7.

2.2 Number of simulation particles

In the literature, the most common limitation for the simulation is the processing time, not the amount of particles.(VIITTEET) However, such approach is not practical, since it limits the reproducibility on another machine.



Figure 7. The probability density function of X-ray particles for 120kV energy.

For this reason, we evaluated at a range from 10^8 to 10^{11} and observed the results in Fig 8.

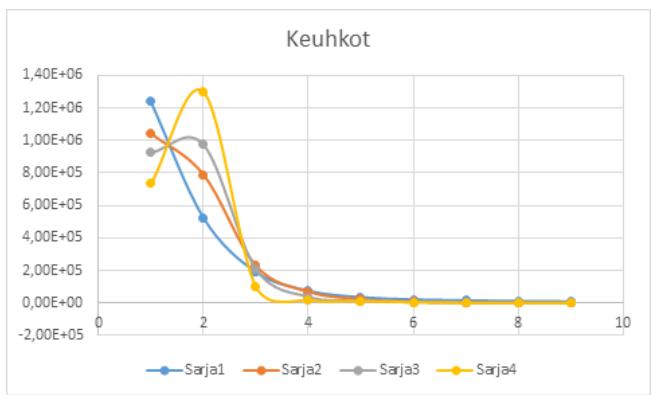


Figure 8. Normalized dose distributions from 10^8 to 10^{11} in the chest.

Paivita kuva kun saadaan uudet laskut valmiiksi...
[Huom. 1e8,5e8, harmaa on 1e9, kultainen 1e10.]

From this image we can clearly conclude that the structure of the curve doesn't significantly change from the 10^{11} , thus that amount of particles we use in following simulations.

2.3 Physics of interaction

X-rays interact with human body in a three possible way: Photoelectric absorption, Compton scattering and Rayleigh scattering.

2.3.1 Photoelectric absorption

In photoelectric absorption the low energy photons interacts with an electron removing it from the shell. The dose dependency is following [VIITE: <https://radiopaedia.org/articles/photoelectric-effect>]:

$$PEA \sim \frac{Z^3 \rho}{E^3} \quad (1)$$

where PEA is photoelectric absorption, ρ is physical density of the attenuating materia, Z is the medium of the attenuating

materia atomic numbers on the photon route, and E is the photons energy.

Therefore in comparison of situation with and without shields, the Equation 1 shows that the absorption into shield will be larger than to the air, therefore the dose hitting the patient will be lower. However in this laboratory examination the shields are placed on the edges of the affecting dose. Which returns the situation to the question of benefits in dose reduction compared to practicality and misplacement effects.

2.3.2 Compton scattering

In a Compton scattering a portion of photon energy is absorbed and a new photon is produced with reduced energy. [VIITE: <http://www.sprawls.org/ppmi2/INTERACT/>] The process is directly proportional to electron and materia densities and inversely proportional to photon energy. [viite: <https://radiopaedia.org/articles/compton-effect>].

Thus, the shields have only a minor effect basing on number of photons reduction, but otherway the dose affecting factors remain the same.

2.3.3 Rayleigh scattering

Rayleigh scattering explains the situations of a protected areas still getting some dose. The principle is that incoming beam (photons) can scatter changing the direction of movement. The remaining intensity of the beam is described with the equation [VIITE: Seinfeld and Pandis, Atmospheric Chemistry and Physics, 2nd Edition, John Wiley and Sons, New Jersey 2006, Chapter 15.1.1]:

$$I = I_0 \frac{1 + \cos^2 \theta}{2R^2} \left(\frac{2\pi}{\lambda} \right)^4 \left(\frac{n^2 - 1}{n^2 + 2} \right) \left(\frac{d}{2} \right)^6 \quad (2)$$

where λ is wavelength, d is the distance between scattering materia's components, n is refractive index, and R is the distance to the particle at the start where the intensity I_0 is measured.

2.4 Step-by-step guide

For some strange reason the GPU version works well only with folders located on the desktop. Thus:

1. Create root folder on desktop (any name) and place all data inside. Especially phantom's dicm-images should be placed inside **Input** folder.

The situation is illustrated in image:

The content of Figure 9 can be classified into 4 blocks:

1. **dicm**-The folder with phantom data in dicom format (Red)
2. Saving details
 - This eliminates the need to manually set parameters at each launch.
 - And allows to resume work from the stop point.
3. Settings
 - Density Conversion settings (more detailed described

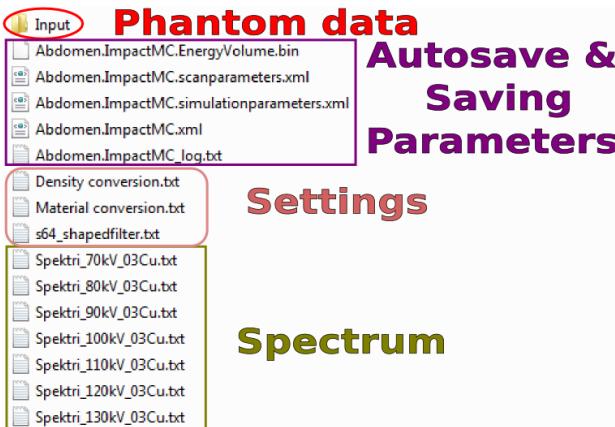


Figure 9. The project root folder locating on the Desktop

in files of Figure 9, part "Settings")

Practically since we assume that shield is plumbum, the transformation term into plumbum is added.

- Material Conversion table
- shapefilter

4. Spectrum tables for each energy

- The simulation is meant to be done on several energies

Table 1. Material conversion settings

Variable	Parameter
Air	-900
H2O	200
Bone	3000
Lead	64535

2. Now we can launch the GUI version of ImpactMC.

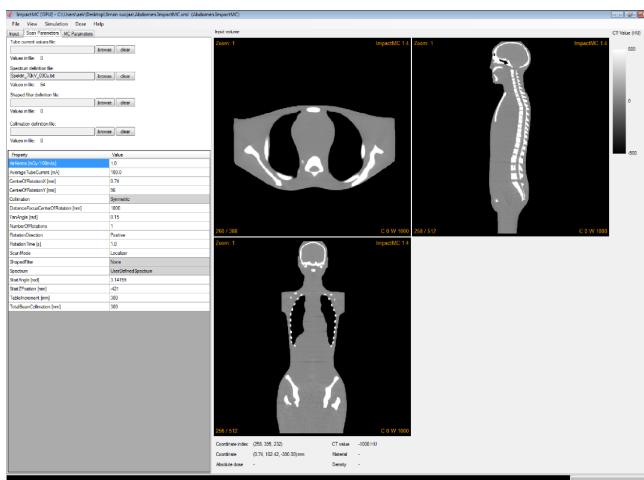


Figure 10. The starting project window

This graphical user interface consists of 2 blocks: Visualization on the right, and the tabbed parameter setting form

on the left. The transfer between MRI image of phantom and dose simulation is performed through the menu options on the top.

Note: This software produces report message about each completed task and tilts in case some action is taken inside it during the simulation, so waiting for the report is highly recommended.

When the process ends it pop ups the Figure 11.

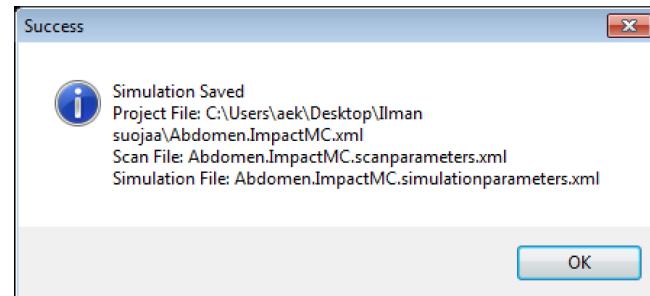


Figure 11. The message window when the simulation has ended. Before getting this message, it might be unwise to touch the computer.

Then by selecting from the upper menu "View" → "Dose" an image similar to Figure 12 is shown.

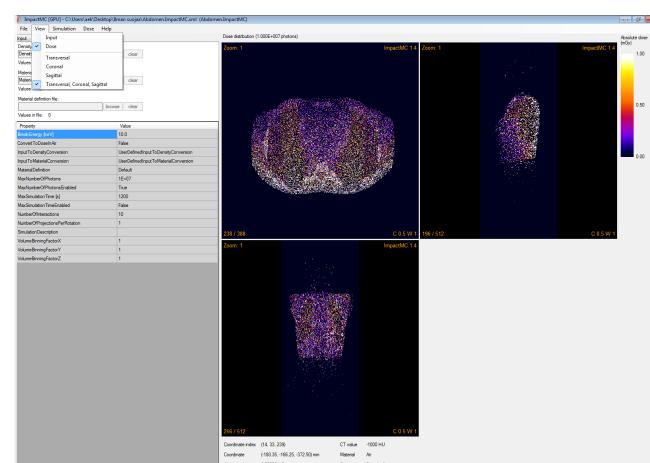


Figure 12. From upper menu select View → Dose.

This is the result wanted to obtain from the Impact MC software, now moving to analysis.

3. Methods and theory, the analysis

In this section we briefly describe the methodology used in result analysis. The main guidelines of wanted results are the doses in organs and dose distribution maps.

3.1 Organ doses

The first step in calculating organ doses is locating those organs in measured CT image. In this paper it is done manually following the guidelines presented in the Eunice phantom documentation (VIITE).

This operation is called volumetric analysis. The volumetric analysis means that each object is segmented from the context, and investigated separately. Such approach provides more detail information about the organ, and allows to investigate maximal accrued doses in a regions.

3.2 Voxel-based analysis

This approach is sometimes called histogrammic analysis. The idea is to investigate the image on a voxel size, and create a histograms based on a suitable binning. There are several goals of such analysis:

1. Analyze the dose distribution.
 - (a) Easy check if the simulation obeys common sense rules.
 - (b) Verifying how the organ dose is formed. = Verifying that the dose is semi-equally distributed over the organ, and not formed as a pack of high and low dose values.
2. Investigate the dose distribution behavior over the different imaging energies.

3.3 Change of the 3D dosemap according to shield

Another possibility is to analyze the effect of external shields to the dose distribution. This analysis is similar to analysis methods presented in 3.1, except that in this case we are interested in the difference between shielded simulations in the same organ.

4. Results

The aims of this paper are to illustrate the process, to make some conclusion about external shields effect on dose and to visualize the predicted effect of the increase in the imaging energy on a dose. Current section describes the last two.

4.1 Relative doses, no shields

The obtained simulation dose image is illustrated in Figure 13

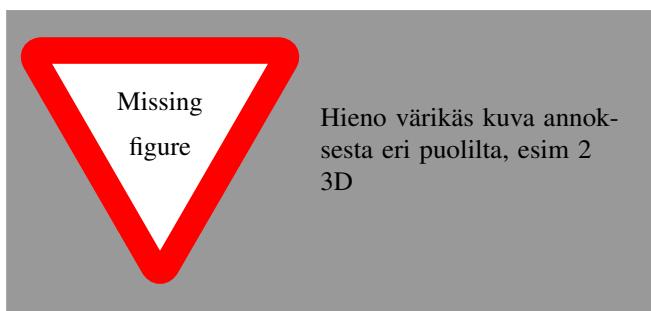


Figure 13. The simulated dose distribution over the phantom.

Figure 14 demonstrates the voxel intensity (the absorbed dose) variation over the energy in lungs.

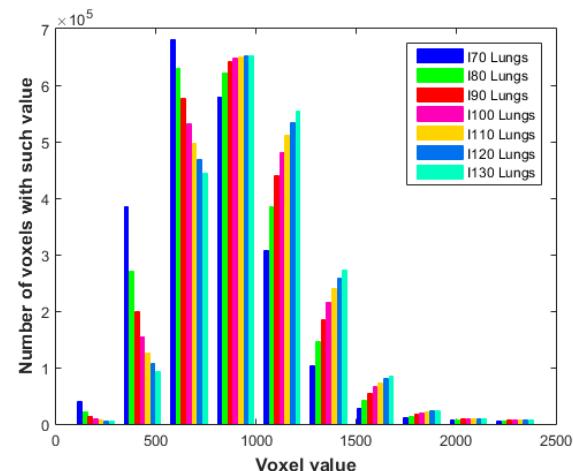


Figure 14. The histogrammatical distribution of the dose in lungs depending on imaging energy.

As it can be noticed the increase of imaging energy results in higher dose. This conclusion makes sense.

In addition we can observe similar analysis of other organs, t.ex. liver (Figure 15), kidney (Figure 16), breasts (Figure 17).

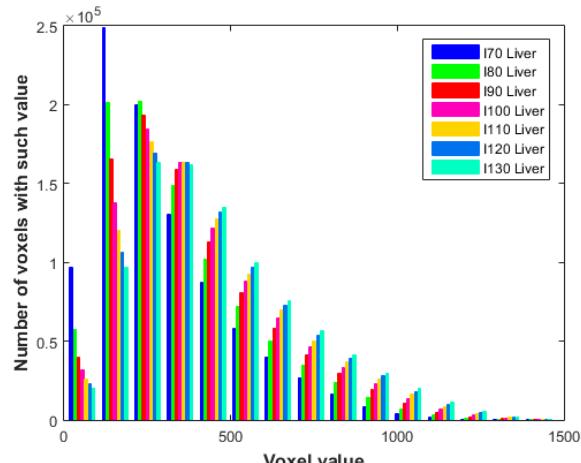


Figure 15. The histogrammatical distribution of the dose in liver depending on imaging energy.

As it can be noticed all of those organs have similar shapes of histograms, but with smaller values, since due to geometry of measurement/simulation they are further from the beam.

4.2 Relative doses, effect of shields

And now let's evaluate the effect of the shields on the dose.

At first let's observe the similar situation with the shields. In this particular case we use B-styled shield which provides us the following image. (Figure ??)

We can notice that the pattern of the non-shielded version and shielded differs. The pattern of non-shielded is gaussian, meanwhile the pattern of the shielded one is significantly lowered for low intensities while remaining similar at top intensities. The top intensities are caused by the beam, which

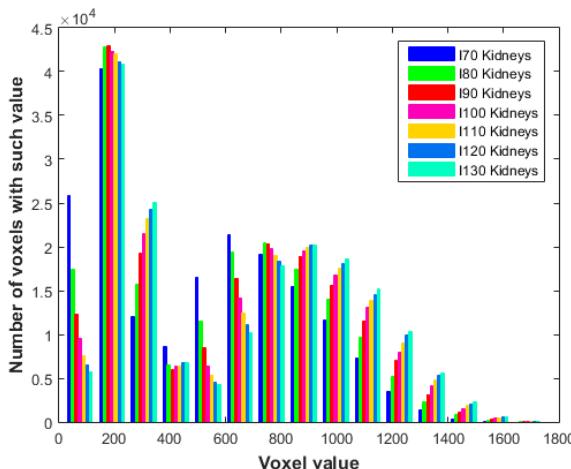


Figure 16. The histogrammical distribution of the dose in kidney depending on imaging energy.

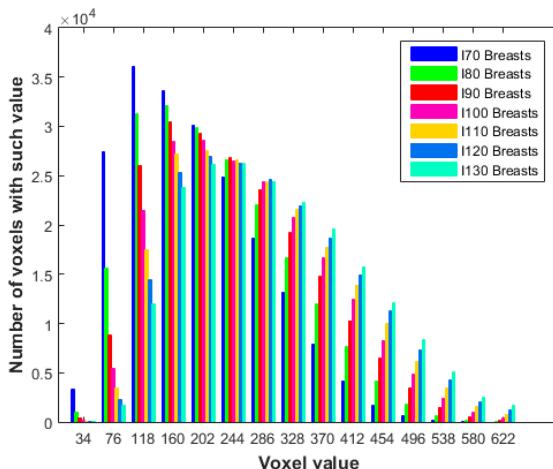


Figure 17. The histogrammical distribution of the dose in breasts depending on imaging energy.

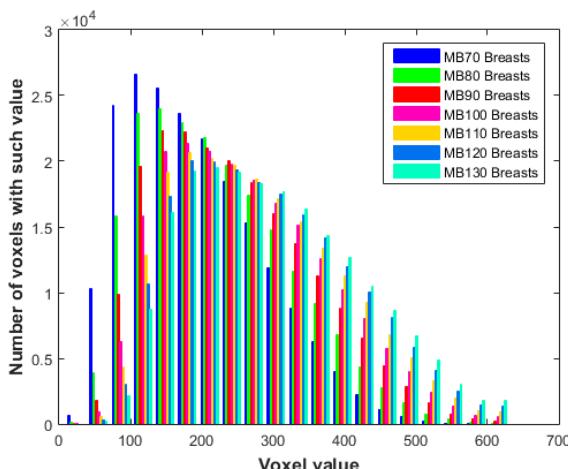


Figure 18. B shield over intensity, lungs.

is not interacting with the shields, but the secondary radiation is filtered out by the shield.

Voiko olettaa että ilmassa beamin leveäminen on likimain nolla? Jos voi, niin mallissa on ongelma.

The next subject of investigation, is the effect of the shield type on the dose. Since the imaging commonly [VIITE] occurs at 120 keV [??????], [VIITE], we investigate all shield types presented in Figure 6 at that energy. The Figures 19 – 22 illustrate the results.

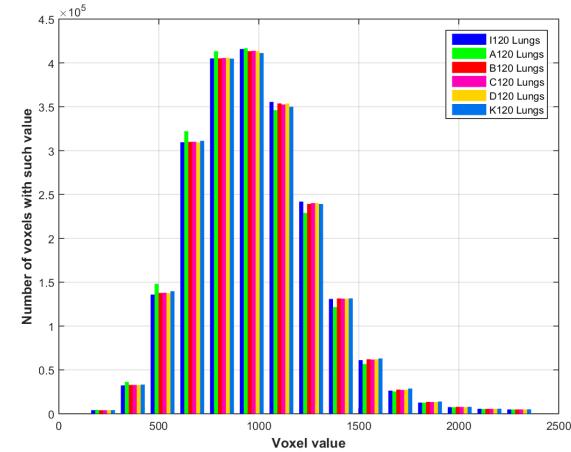


Figure 19. B shield over intensity, lungs.

Breasts

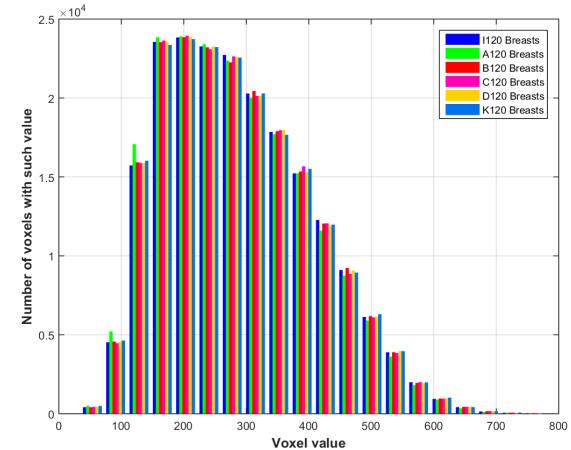


Figure 20. B shield over intensity, breasts.

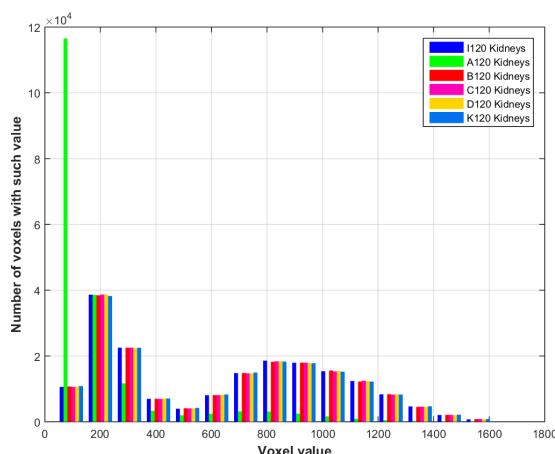
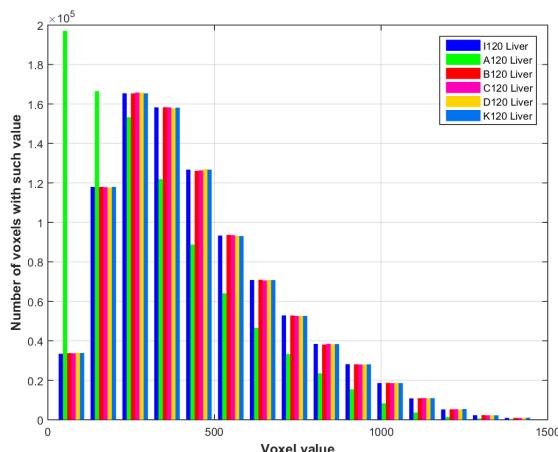
Kidneys

Liver

From the results we can notice that shields of type B – D have similar effect as the lack of shields at all. The type A shield lowers the dose for kidneys, but remaining same for the rest observed organs. In next section we will analyze the dosimetrical effect of this difference.

Table 2. Relative dose table

kV	A	B	C	D	No shield	Loose D
70						
80						
90						
100						
110						
120						
130						

**Figure 21.** B shield over intensity, kidneys.**Figure 22.** B shield over intensity, Liver.

4.3 Absolute doses

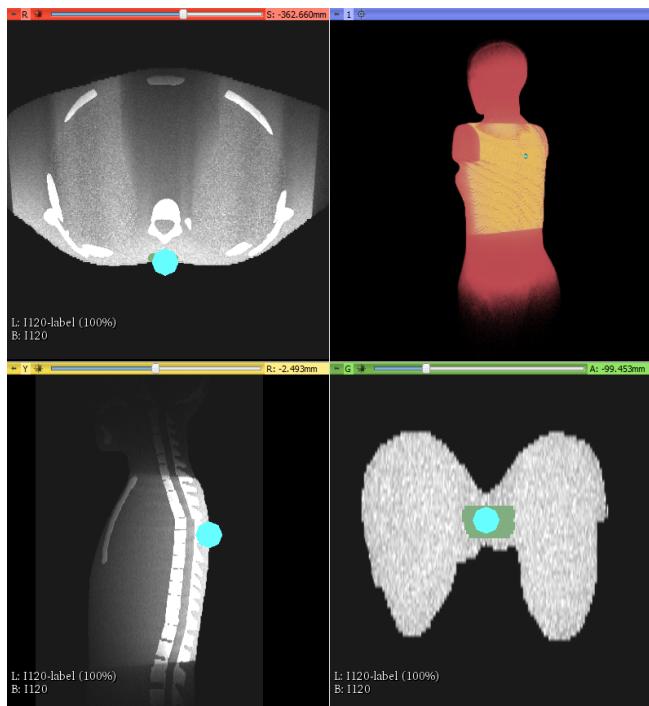
4.3.1 Linkage of relative dose to absolute

Perustele jotenkin että pisteen valitaan teoreettisessä pisteenä sädenipun osumisesta, että häviö matkalla on pieni, häiriöiden takia maalataan pieni alue valitun pisteen paikoissa, ja se alue saa annokseksi kuvausenergian, jolloin koko matriisi vain jaetaan skaalausrivolla, niin saadaan annokset

ka, mihin
nattaa
ata?

In order to obtain a prediction about the dose, we approxi-

mate the dose on the skin at perpendicular between beam and back to be 0.2 mG. (Figure 23)

**Figure 23.** The dose definition principle from 120 kV sequence without shields. The teal dot illustrates the theoretical point of definition. However, due to noise, it is better to define an area, from which choose the intensity value.

The blue dot in following image illustrates the dose reference point. However to minimize the effect of the background noise, we define the area (green) and calculate the statistics over it.

Table 3. Dose dependency on intensity definition statistics

Variable	Parameter
Max	2153
StdDev	202.593

From the Table 4 we can approximate the intensity value

to be

$$\text{Intensity} = \text{Max} - (\text{StdDev})/2 = 2051.704 \quad (3)$$

This provides us Equation 4:

$$\begin{aligned} 2051.704 \text{ intensity} &= 0.2 \text{ mGy} \\ \Leftrightarrow 1 \text{ intensity} &= 9.747997... * 10^{-5} \text{ mGy} \\ &\approx 9.748 * 10^{-9} \text{ Gy} = 9.748 \text{ nGy} \end{aligned} \quad (4)$$

4.3.2 Effects on histograms in conversion between the relative and absolute doses

The change of relative dose to absolute has no effect on histogrammatical images. For example, in chest, the histogrammatical approach produces Fig.24.

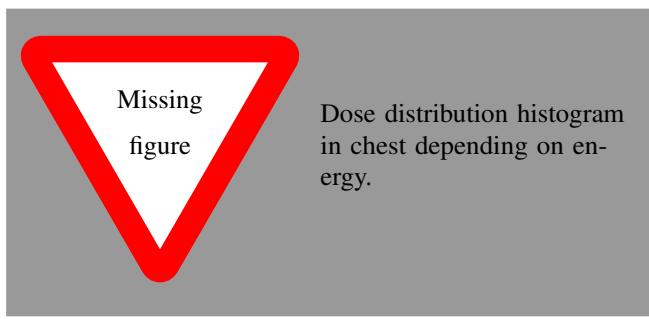


Figure 24. Dose distribution histogram in chest depending on energy.

This figure is clearly same, except scaled y-axis, as the relative dose one.

4.3.3 Absolute Dose

Taulukko absoluuttisista annoksista

5. Discussion

This paper presented a detailic information about the laboratory work of using simulation in computer tomography prototyping. All measurements and simulations were successful. The achieved results are in *yhteensopivat* with the technical description of simulation software and theory predictions. In addition, the effects of the shields were analysed, *and the results suggest that *****. However, due to lack of statiscal significance such theory is only providing a guideline to be investigated.

6. Future improvements

Previous sections described the process of this laboratory work and the expected results. Thus, in this section the possible improvements will be *opened*.

6.1 Free Atlas segmentation

In the current version the organs were manually defined based on [VIITE, Eunice ohje]. Such approach is not precise and not repeatable, thus the usage of digital phantom is recommended. There are several over the internet, but for the start taking a look at ***, ***, ***, *** might be worth it.

6.2 Variance of the model

Each measurement of this laboratory work is done less than five times, thus we are not taking into account the variation of the ImpactMC model. For the illustration of the process/pipeline the performed analysis is enough, but for making the scientifically meaningful analysis the simulation dataset amount should be at least doubled.

6.3 Overall

Overall, this laboratory work achieved the expectations. The ImpactMC produced truthworthful results of the dose distributions, verification and ***** *jatka tänne jotaain yleistä.... *****

Acknowledgments

Attachments

Structure of this paper

Fig.26.

No Shield relative / absolute doses

Listaa tänne kaikki kuvat suurkokoisina

Fig.26.

Table 4. Absolute dose table

kV	A	B	C	D	No shield	Loose D
70						
80						
90						
100						
110						
120						
130						

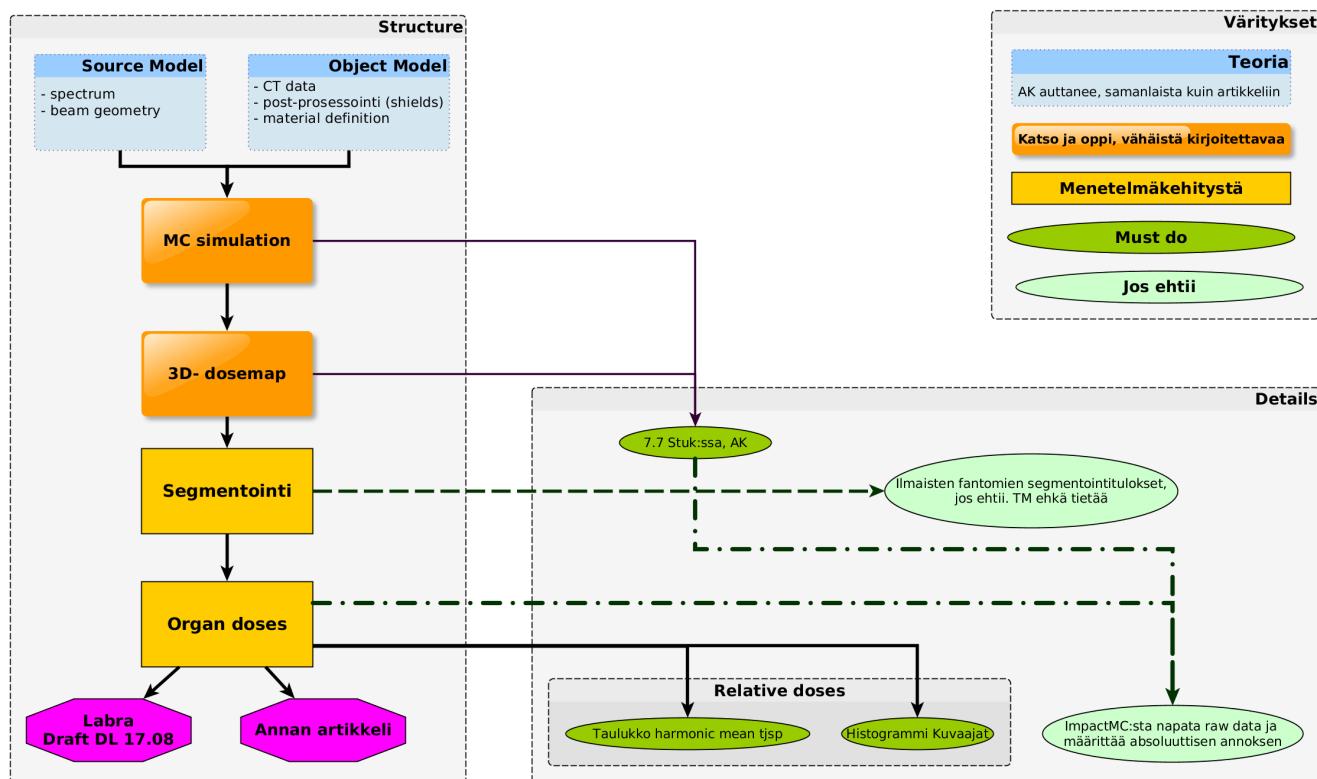


Figure 25. Plan graph

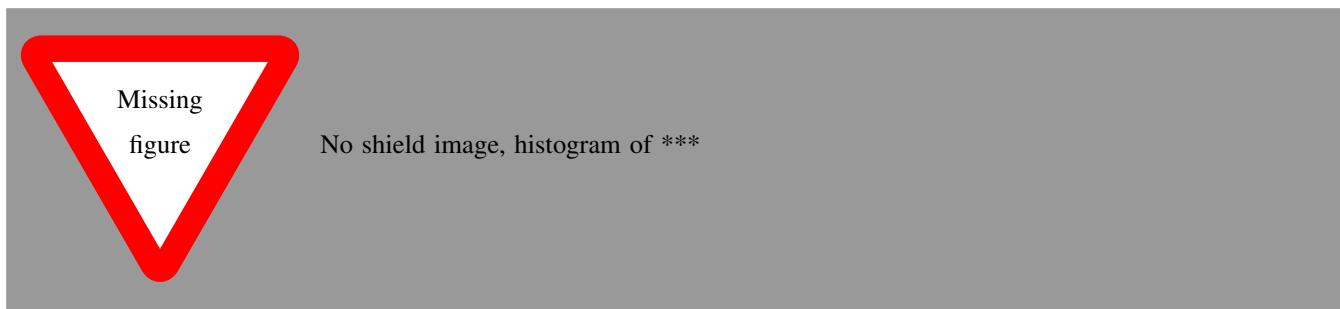


Figure 26. Plan graph