How do we measure dose and estimate risk?

Christoph Hoeschen*[a], Dieter Regulla [a], Helmut Schlattl [a], Nina Petoussi-Henss [a], Wei Bo Li [a], Maria Zankl [a]
[a]Department of Medical Radiation Physics and Diagnostics, Helmholtz Zentrum München, Ingolstädter Landstr.1, 85764 Neuherberg, Germany

ABSTRACT

Radiation exposure due to medical imaging is a topic of emerging importance. In Europe this topic has been dealt with for a long time and in other countries it is getting more and more important and it gets an aspect of public interest in the latest years. This is mainly true due to the fact that the average dose per person in developed countries is increasing rapidly since threedimensional imaging is getting more and more available and useful for diagnosis. This paper introduces the most common dose quantities used in medical radiation exposure characterization, discusses usual ways for determination of such quantities as well as some considerations how these values are linked to radiation risk estimation. For this last aspect the paper will refer to the linear non threshold theory for an imaging application.

Keywords: dose quantities, risk estimates, dose measurements, simulation tools

1. INTRODUCTION

Projectional radiography and even more threedimensional (3D) medical imaging technologies like CT are of great importance for the diagnosis of diseases and therefore for the patient. The digital x-ray imaging technologies and the possibilities of fast and large volume data aquisition in Multislice-CT and more and more also in nuclear medical imaging technologies like SPECT or PET have changed the clinical praxis to a large extend. Diagnosis got faster, provides more accurate and more detailed information about diseases and their reasons. However, this is also the reason for the largest component of exposure to ionizing radiation that the average population is facing today. Regarding to various investigations [1-3] the mean effective dose per person and year in the developed countries has been increasing steadily and is today in the range between 0.4 and 4 mSv and is mainly due to exposures from CT, angiographic and interventional investigations. A single patient will be exposed to effective doses between about a few mSv up to more than 100 mSv. Effective doses below 100 mSv are classified as low dose applications according to BEIR VII [4]. However, doses in the range between 5 and 50 mSv was the lowest range of exposures for investigated persons of the atomic bomb survivor studies. Today, there is a wide scale of discussions and investigations about the harm of low doses to the human being [5-8]. Although this discussion is not yet finalized, at the moment the linear non threshold model ("LNT") assuming a dose proportional risk increasing with the dose beginning with the first bit of ionizing radiation is used for prospective risk assessment as part of the general protection scenario. This is in accordance with the newest recommendations of the international bodies like ICRP ("International Commission on Radiological Protection") [9] or the BEIR report.

Using the figures from Germany as examples, the tendency of exposures due to medical imaging can be highlighted. Studies about the exposure of the population in Germany to ionizing radiation in 1997 [10] and 2003 [11] show, that the number of investigations using CT technology is increased and the average dose per CT examination is not reduced at least. The frequency has been increased so that instead of four to five percent of all investigations in 1997 were CT investigations in 2003 already six percent of all diagnostic investigations were CT examinations. Therefore CT became responsible for more than 50% of the dose applied in medical diagnosis. This trend which is demonstrated in figure 1 is obviously ongoing. The report for 2006, provided by the legal authorities in 2008 reports 7% of all investigations being CT and these result in 56% of the total dose distribution. It corresponds with the development of CT technology and the spreading of new powerful Multislice-CT systems which allow new kinds of investigations due to their fast acquisition modes. This results in a rise of the mean radiation exposure per person and year due to medical diagnostic procedures although the dose per investigation within projectional radiography could be reduced.

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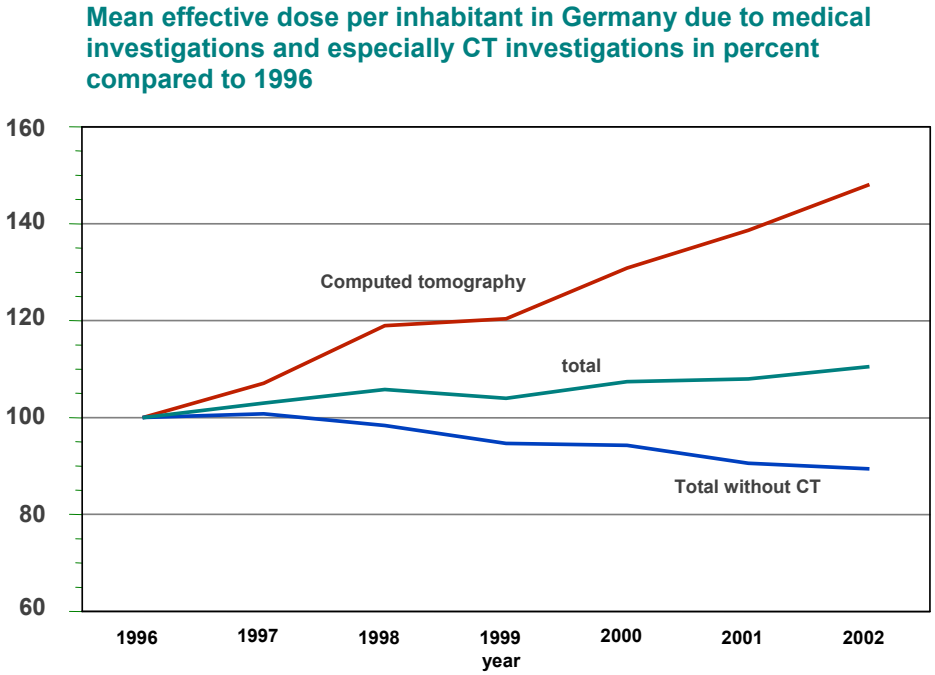

Figure 1: development of exposure to the population due to medical investigations according to data from the Bundesamt für Strahlenschutz, Germany

For Germany, this means that medical radiation exposure is by far the greatest single component of radiation exposure of the population summing up to as much as the overall radiation exposure coming from natural sources in the average in Germany. A similar trend is obviously true in other developed countries. In the USA for example the value for doses resulting from medical exposures have been reported to being increased from 0.5 mSv to now at least 3 mSv. This increase is regarding to the new white paper [12] also mainly caused by CT examinations.

A more detailed discussion about contributions of various types of investigations can be found in the paper by D. Regulla in this proceedings volume. This paper will instead deal with the question how exposure of medical diagnostic procedures can be described, how it can be determined and what can be said about the relation between determined exposure and possible risk related to this exposure.

2. DESCRIPTION OF EXPOSURE

To quantify exposure to ionizing radiation the most relevant values in a physicists' point of view are dose values. There are various kinds of dose values that have to be used in a correct way. So they are defined and described briefly in the following

2.1 Physics-based dose values

From the physics' point of view there are various measures for dose values.

2.1.1 Ion-dose

The ion-dose is measuring variable used in dosimetry for ionizing radiation. The ion-dose describes the charge load in an infinitely small amount of mass due to the ionization process in air caused by the ionizing radiation. The unit of the ion-dose is coulomb divided by kilogram (C/kg). 1 coulomb divided by kilogram is the ion-dose generated assuming ions of a preceding sign with a total electric charge of 1 C in air with a mass of 1 kg caused by ionizing radiation of energy flux

density constant in space. In former times the unit roentgen (symbol: R) was accepted. 1 roentgen was equal to 258 μC/kg.

This dose refers only to the ionization process and not to the energy imparted into the mass. It is therefore strongly depending on the material investigated and it is not providing inside into the whole picture of radiation effects.

2.1.2 Absorbed dose

With this measure the problem of the missing parts of the picture can be avoided since absorbed dose (D) is defined as the amount of energy from an ionizing radiation deposited in a mass of some material. According to ICRU (International Commission on Radiation Units) it is defined again at a certain point as described by:

$$D = \frac{\Delta\varepsilon}{\Delta m}$$

with an increase of energy ($\Delta\varepsilon$) in an infinitely small amount of mass (Δm) in the point of interest. The SI unit is Gray (Gy) which corresponds to J/kg.

2.1.3 Kinetic energy released in mass (KERMA)

The absorbed dose cannot always be measured easily in practice so that for measuring purposes another quantity is often used. This quantity is kinetic energy released per unit mass. This is the total kinetic energy transferred as charged particles per unit mass. For x-ray energies typical for medical diagnosis, the charged particles are absorbed very close to the interaction, thus for most organs Kerma is a good estimate of the absorbed dose. The SI unit is Gray (Gy) as above.

$$K = \frac{\Delta\varepsilon_{Kin}}{\Delta m} = \frac{1}{\rho} \cdot \frac{\Delta\varepsilon_{Kin}}{\Delta V}$$

2.1.4 Activity

Activity is a measure for characterizing nuclear decay of radioactivity as it is used in nuclear medical imaging. It describes the number of decays per second for a radioactive reaction. The SI unit is Becquerel (Bq) corresponding to s^{-1}.

2.2 Biology-based dose values

The descriptors of exposure relying on physical properties as the dose values described in the last subchapter do not explain all relevant properties of ionizing radiation in terms of determine risk and documenting exposure to a patient or medical staff. To do so an estimate of the biological relevance is necessary. This will have to take into account the damage per unit dose that a specific kind of ionizing radiation is causing as well as the different sensitivity of organs.

2.2.1 Equivalent dose and organ dose

The first aspect is dealt with using the concept of equivalent dose which is calculated as the product of the absorbed dose and a weighting factor w_R specific for each type of radiation and for some kinds of radiation even The equivalent dose is therefore calculated as

$$H = w_R \cdot D$$

The equivalent dose is therefore as a first approach calculated for a point mass. However, in radiation protection practice and in medical imaging optimization it is in most times averaged over organs (organ doses) or at least areas in an organ. The rationale behind this is the assumption that the linear non threshold theory is valid. To mark that this dose is a biologically relevant feature of exposure its SI unit is Sievert [Sv] which is still [J/kg].

2.2.2 Effective dose

The efecctive dose has been implemented to gain a risk related value for a partial body irradiation of a population based on the radiation risk evaluated for dose values basically in the range of a few hundred mSv. It is therefore defined as the sum of equivalent doses H_T, of the organs and tissues (organ doses) of a human being which are radiosensitive and therefore critical for radiation protection, each weighted by tissue weighting factors w_T. Its SI unit is again the Sievert.

The weighting factors take into account that the risk for stochastic (and deterministic) radiation caused damage is different for different organs and tissues. The effective dose is therefore:

$$E = \sum w_T \cdot H_T$$

Note: The effective dose is **not** a risk descriptor for an individual person but for an average person referring to an average population.

2.3 Radiation protection and its standard dose descriptors

For radiation protection there are two different approaches to describe the exposure.

2.3.1 Personal dose values

The personal dose values that are used for radiation protection purposes are used for limiting the exposure either due to various medical exposures or occupational doses for staff members. These values are mostly based on effective doses or surface doses. To estimate such doses equivalent personal doses are measured and calculated using dosimeters and conversion factors.

2.3.2 Local dose values

Sometimes personal dose measurements are not available. In this case local dose values in the area where someone might be exposed to ionizing radiation are used which can be measured by local dosimeters and these measured values are then used for calculating the personal dose values.

3. EXPOSURE DETERMINATION IN MEDICAL DIAGNOSIS

The determination of exposures in medical diagnosis is most often a two step process. First of all, there are some measurements and then there is the calculation of values relevant for the patient exposures from such measurements.

3.1 Measurements

Which kind of measurements are performed really depend on the kind of examination that is under investigation. In projectional radiography, entrance surface dose, the dose area product and the detector dose might be quantities that are measured.

3.1.1 Entrance surface dose

The entrance surface dose is the dose at the entrance surface of the radiation beam on the patient. It includes the backscatter of the patient and is sometimes also referred to as the entrance skin dose. It can be measured by placing small dosimeters as TLDs or photoluminescence dosimeters on the patient surface where they are not expected to violate the diagnostic procedure. The incident dose would be a value measured at the same position but without the patient not to include backscatter.

3.1.2 Dose area product

The dose area product is as it is described by its name the product of the dose and the irradiated area. This value is determined without backscatter. It is assumed to be a constant along the beam from source to detector (without the patient, filters, shutters and grid). It is therefore measured close to the source to avoid any influence on the imaging features of the system and to be able to use a small detector chamber.

3.1.3 Detector dose

The dose measured at the detector is typically used for the automatic exposure control. However it can also be used together with the knowledge about the used beam quality and some patient parameters to estimate the dose for the patient.

3.1.4 Computed tomography dose index and dose length product

The dose length product (DLP) is still a physical dose value used to describe dose from CT investigations. It is the product of the average absorbed dose over a certain volume multiplied by the scan length or more exactly the exposed z-range of the scan [13]. However it cannot be measured directly in today's scanners.

To calculate the DLP there is a specific procedure using standardized values as the so called "Computed Tomography Dose Index" values. They have typically special subindices indicating different meanings. This will be explained in the following:

The parameter that is typically measured is the normalized CTDI free in air. This value is intended to give a value that is equivalent to the dose within the nominal width of the slice assuming that the whole radiation is located within a rectangular profile with the nominal width. This is visualized in figure 2.

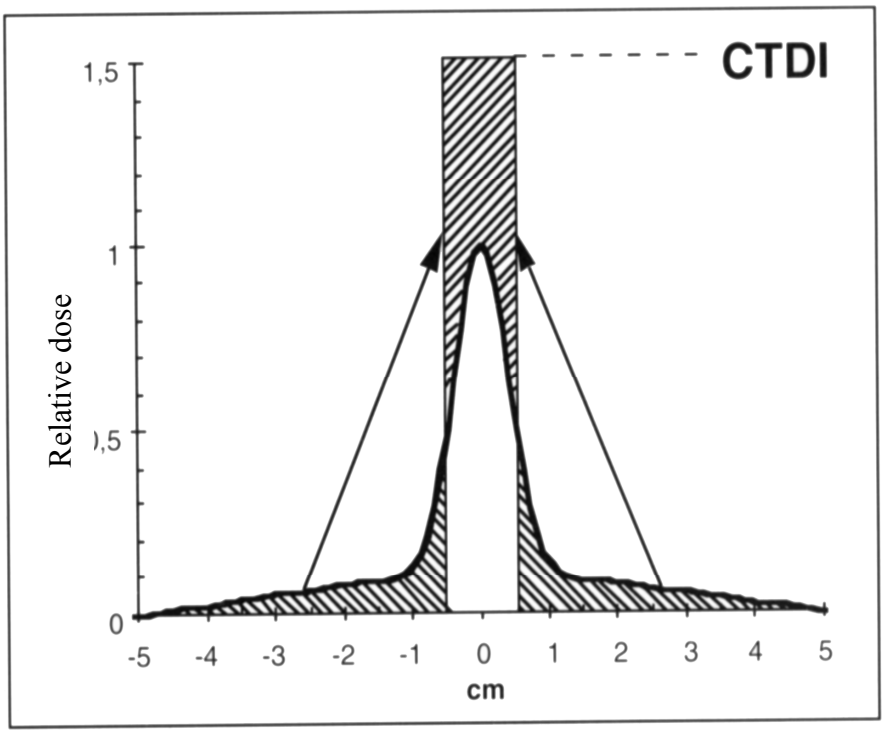

Figure 2: schematic presentation of the meaning of the normalized CTDI free in air.

This value is measured by using a phantom and implementing dosimeters in the center and in a certain distance from the center and to calculate an average value according to given calculation procedures. The dose value is normalized to the applied mAs. It is a system specific parameter which is not giving a value for a dose of an examination or the dose requirements of a system. Different phantoms made of PMMA are used to refer to the different situations for scanning different anatomy. This difference is mainly realized by different phantom diameters (16 cm diameter for head investigations 32 cm diameter for body investigations). Since with multislice CT the collimation is increasing the length of the phantom and the corresponding dosimeters becomes important. This is today typically indicated by further subindices. Multiplying this first index with the used charge and a beam quality correction one gets to a CTDI which is indicating the dose along the scan axis free in air.

The CT dose index that is called volume CTDI ($CTDI_{vol}$) describes the average local dose for the patient within the volume of investigation given in mGy.

3.1.5 Administered activity

As the administered activity we define the activity that was injected to a patient for a radiopharmaceutical investigation. There are various ways to determine that. One would be to measure the activity in the syringe before and after the application. Another method is using the count rates along the body to estimate the activity after using a calibration factor.

3.2 Calculations

3.2.1 Monte-Carlo based simulation

Monte Carlo techniques using computational models of the human body are applied for so called "numerical dosimetry". The mean organ doses are calculated using such simulations following the paths along which photons or particles travel through and object and deposit energy. They are then normalised to a measurable dose quantity. In the past, many dosimetric studies have been based upon schematic representations of the human body where the shape of the body and its internal organs are described by relatively simple geometric bodies [14, 15]. A number of groups have simulated x-ray examinations on mathematical phantoms and used such Monte Carlo techniques to calculate organ dose conversion coefficients [16-22]. During the last two decades, voxel models were introduced that are derived mostly from images of real persons. Among other laboratories, the Helmholtz Zentrum München, formerly known as GSF developed 10 voxel models of individuals of different stature and ages: 2 paediatric ones, 4 male and 4 female adult models [23-25]. It could be shown that the schematic organ shapes of the purely mathematical phantoms presented an over-simplification, having an influence on the resulting dose coefficients [26, 27].

All voxel phantoms constructed so far by our working group were based on computed tomographic (CT) image data of living patients with exception of the 8-week old baby. All patients were scanned with a large number of contiguous axial slices. The volume elements are the pixels of such slices, multiplied by the thickness of the slice. The single slice images are stacked, resulting in a three-dimensional array of voxels. In the primary image data, each pixel has a value that is characteristic of the attenuation of x-rays of a specific radiation quality typically described as grey values.

The property coded by the grey values is related to the tissue composition, not to an anatomical position. That means the anatomical boundaries between organs and tissues needs to be determined on the basis of the grey values and grey value contrasts that have to be combined suitably with anatomical knowledge by the user. Then each voxel is assigned to an organ or tissue to which the corresponding voxel belongs. This process of assigning each voxel to a specific organ is called segmentation. There are only a small number of tissues that separate well from their neighbourhood by grey value contrast. For CT, these are mainly the bones and the lungs. However, the grey values of many individual organs usually are inhomogeneous, and the grey value ranges of various organs largely overlap. Therefore, organs that are in close contact to each other cannot be separated automatically on the basis of their grey values, so far. The result of the segmentation is a three-dimensional array of organ identification numbers. Examples of voxel models are given in Figure 3. The next step is to assign to each organ certain tissue properties, such as elemental composition, density, and cross-sections for radiation interactions with the material. Then the voxel models can be combined with suitable computer programs simulating the radiation transport e.g. in the form of Monte-Carlo simulations in material.

There is an obvious variability among the dose data for the individual voxel models, due to differences in stature and individual anatomical features. Since the currently existing voxel models range from very slim persons to large and heavy persons, the dose values published so far give a dose range in which an individual distribution of organ doses might be expected to lie, together with an indication of the magnitude of dose differences to be expected between individual persons. Furthermore, it is believed that such calculations can be used to roughly estimate the doses to an individual by selecting those for the voxel model fitting best to the person under consideration.

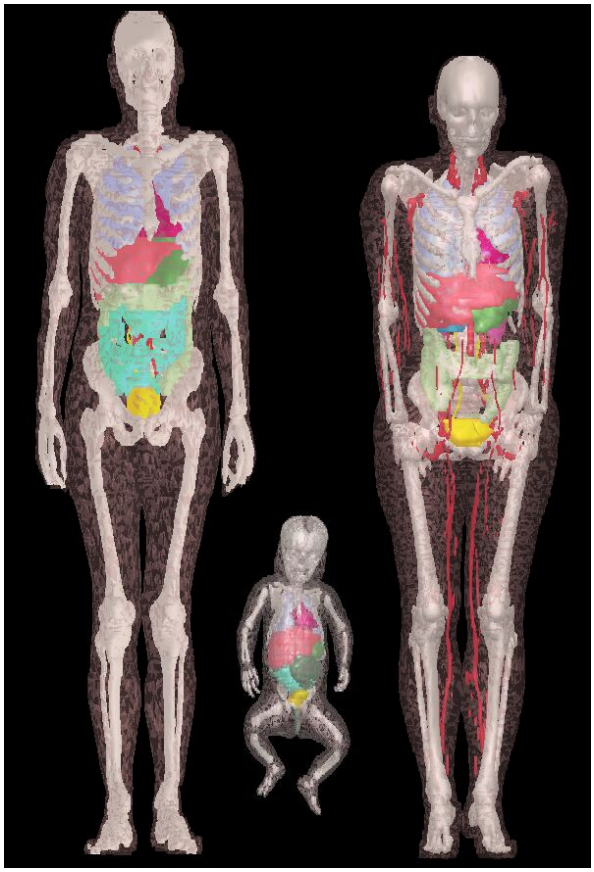

Figure 3: Three voxel phantoms segmented in our working group – Golem (left), Baby (middle), and Laura (right).

3.2.2 Effective dose in CT

The effective dose can be calculated from the scan parameters and some system specific components. The basic principle is shown in the following table. In the first column the needed or calculated parameter is described, its corresponding abbreviation is shown in column 2, while in column 3 the usable unit is specified. To achieve reliable results it is necessary to look for values in correct unities.

Current	I	mA
X time	T	S
= charge	Q	mAs
X normalized CTDI in air	${}_{n}CTDI_{A}$	mGy/mAs
X correction for beam quality	k_V	1
= dose at axis free in air	$CTDI_A$	mGy
X collimation (# rows x thickness of rows)	H	Cm
X number of rotations	N	1
= dose length product (free in air)	DLP_A	mGy cm
X conversion factor	f_{av}	mSv/(mGy cm)
X system correction factor	k_{CT}	1
= effective dose	E	mSv

There is a number of software programs calculating effective dose or even organ dose values for different scan parameters like e.g. [28, 29]. Today, it is important that these programs are able to cover the specialities of multislice CT dosimetry.

To gain organ doses and in the last step to calculate effective doses from DLP a conversion factor is used as it can be seen in the calculation scheme. This is depending on the area that has been exposed. It has to be stated here again, the use of effective dose is valuable as long as the values are not taken as absolute values determining exactly the risk of a person [30, 31].

3.2.3 Specific absorbed fractions

To calculate the doses in nuclear medical imaging or general due to internal expositions it is necessary to gather information about various aspects of the exposure:

- Amount and path of intake
- Temporal distribution of the radionuclide within the body
- Nuclear decay data of the radionuclide considered
- Absorbed fractions
- Mass of target tissues

Then one can calculate the doses to the different tissues as

$$D_T = \sum_S A_S \cdot S_i(T \leftarrow S)$$

with A_S as the activities in the different source organs and S_i as the specific effective energy of a radioactive decay. This S value is a combination of the radioactive decay data and the specific absorbed fraction as defined by the energy absorbed in the target organ divided by the energy emitted from the source organ and this coefficient divided by the organ mass of the target organ. [32, 33]. These SAFs can be derived by Monte Carlo simulations as well.

3.2.4 Biokinetic modeling

Besides the physical knowledge about transfer factors it is necessary to know about the biodistribution of the radionuclides. To gain that biokinetic models have to be derived specific for each radiopharmaceutical. The variability of the patient specific biokinetics is anyhow large [34].

4. RISK ESTIMATION

4.1 Data base for risk evaluation

To evaluate the risk due to ionizing radiation it is common practice to use the Linear-non-threshold model assuming that each photon or particle can induce an effect on the cells, which can be correctly repaired or not, and the total risk depends on the number of cell effects. This model is mainly based on data from epidemiological large groups of people exposed with their whole body to quite high doses of ionizing radiation. The main data sets are still those of the Japanese atom bomb survivors and the corresponding life span studies (LSS) [35] as well as the studies about the Mayak workers [36].

4.2 Models

There are different models about how low dose exposures might influence the body of the exposed person. These are summarized in figure 4 from [37]:

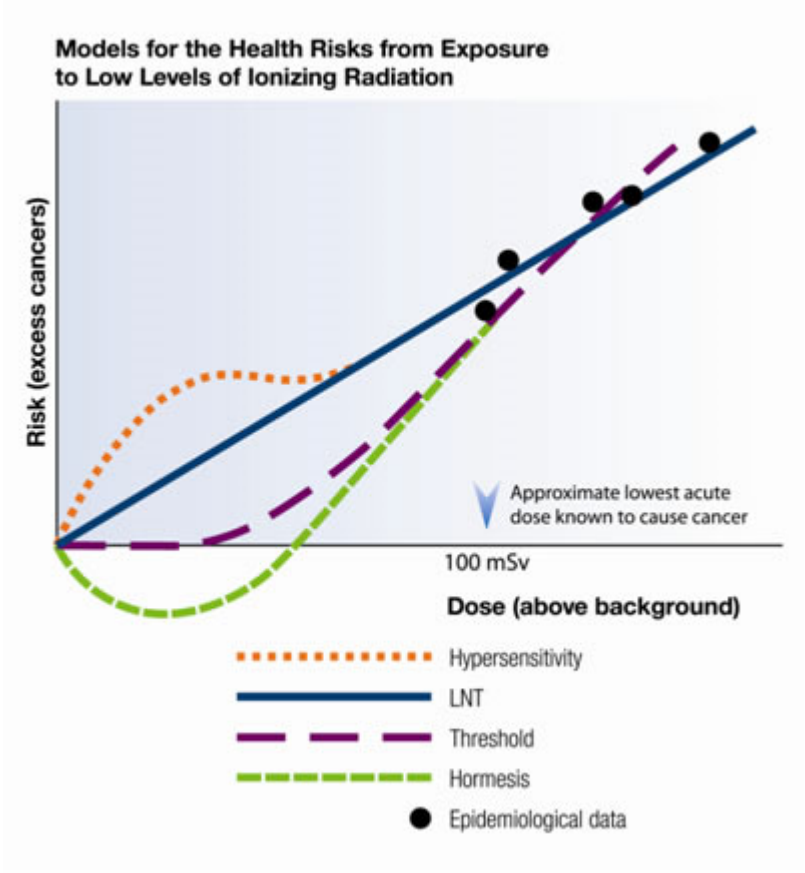

Figure 4: various models for the risk associated with doses lower than those for which defined radiation damages and especially cancer risks could be determined. Taken from [37]

Besides the model corresponding to the linear non threshold theory there are also theories assuming a higher risk for lower doses, as well as those assuming thresholds also for stochastic radiation risks and therefore lower risks for lower exposures. This is especially important for repeated medical imaging procedures. There is even a model assuming that low doses might be beneficial for human beings since the body could be stimulated for an advanced repair activity.

4.3 Application in medical diagnosis

The best way to quantify the exposure of a single patient undergoing a specific investigation is to try to determine the dose for each single organ which has been exposed due to the investigation. The average dose absorbed in each organ is called the organ dose. As it can be easily understood, organ doses cannot be measured directly in general. To a certain extend it is possible to calculate the organ doses as it will be described below. The description of the radiation exposure would in this approach be done by a large number of exposure values for the specific organs. Assuming that the exposure in an organ would be uniform (which is not the case in medical diagnosis), or broadening the linear non threshold theory also to the assumption that it corresponds to the same risk if in the same organ two neighboured cells are hit or if those cells are distributed widely this is correct and very valuable to characterize the radiation exposure and it also allows a certain estimate about the risk related with an investigation, but it is not easy to compare the values for different investigations or even different equipment. One single value would be desirable for such a comparison.

The value typically used for this purpose today is the effective dose. This dose value was introduced to find a way of estimating the risk of exposures of large groups (e.g.workers in certain applications) related to exposures not affecting the whole body. It is not a useful quantity to describe the risk of a single person due to a single exposure.

Another aspect in risk estimates for medical imaging is that the age at exposure plays an important role as well as the gender of the exposed person. But in addition it is also worth to be mentioned that we expose in medical applications often regions which are ill or damaged. This might change their radiosensitivity. In addition, there is an individual

radiosensitivity due to genetic predisposal which would be necessary to know for individual risk estimates. The question of estimating risks remains unsolved. The effective dose is nevertheless an important tool for optimising techniques in medical radiation imaging.

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