Minimum Dose Calculation for Different Imaging Tasks in Digital Projection Radiography

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ABSTRACT

In diagnostic radiology it is a common interest of patient and medical staff to keep the exposure as low as reasonably achievable (ALARA). In spite of this task being well known there is no consensus about how low the exposure for a specific diagnostic examination can be. The methods presented in this paper allow for the mathematical determination of the lowest exposure necessary to perform basic, well defined imaging tasks in digital radiography. The model quantifies how different demands on the result influence the necessary exposure. Dependences on parameters describing the beam quality used and the detector are implemented into the model. A strong rise of the necessary exposure to detect a certain contrast of lower amplitude or with higher certainty was determined. The effects of a change of the energy of the irradiation are as a first step investigated via the connected change of the transmission of a main absorber. By specification of the specimen to be observed the result is connected to beam energies which can be correlated with the energy dependent response of a realistic detector system. The calculations give basic information about the best exposure in a simplified view of patient dimensions and diagnostic needs. Especially in pediatric radiology optimized adaptations to the patient and the clinical question are expected to take great effects due to the great variations of patient sizes.

Keywords: digital radiography, optimization, pediatrics, minimal exposure

1. INTRODUCTION

Medical applications of X-ray imaging techniques are in use since the discovery of X-rays in the late 19th century. Dependences of the exposures on the patients' features and the clinical question have mainly been developed from daily experience [1, 2]. The custom setups and adjustments vary with progress in technology and from site to site. With the introduction of digital techniques which largely compensate deterioration from over- or underexposure the variations between different x-ray departments have become larger. In spite of the gaining power of computer algorithms to improve the visibility of structures in images there is limited information present in each raw picture. Image processing has to keep as much of this information as possible while improving its visibility compared to the background. The diagnostically evaluable content of a radiograph is in the first place depending on the exposure parameters used. Choosing these parameters should ideally be following the principle "as low as reasonably achievable" (ALARA): only the amount of radiation should be applied that is necessary for the diagnostic task to be performed. The calculations presented here are aiming for the task dependent quantification of minimal exposures necessary in digital projection radiography. We start from a very basic imaging task, the detection of a contrast in an imaging system reduced to a source, a homogeneous specimen and a single detector without spatial resolution. This model already provides information about the dependence of the minimal exposure on variables like the desired certainty of the result or the overall transmission of the radiation applied to the specimen. Further modifications implemented in our model include energy dependence properties of a realistic detector.

The work presented is part of a project aiming for the optimization of the relation between image quality and dose in pediatric thoracic projection radiography. Taking the patients features directly into account for changing exposure settings is especially useful due to the great variation of patient sizes and their higher sensitivity to radiation [3] compared to adults.

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2. METHODS

In a simple and general approach to digital projection radiography we reduce the imaging system to a source, a homogeneous specimen, and single detector without spatial resolution.

The source shall send $\tilde{n}$ photons in one exposure through a main absorber of the energy dependent attenuation (1-*p*) and an ideal detector shall count the transmitted photons (*t*). The measured transmission of the system $\tilde{p}$ is given by

$$\tilde{p} = \frac{t}{\tilde{n}} . \tag{1}$$

As the basic imaging task we chose the discrimination of a change of absorption by an additional or left out thin absorber (attenuation (1-*c*)) added in the beam line. We arbitrarily define the maximal error δ of the result for $\tilde{p}$ to be half the difference of the transmission values with or without the "contrast" absorber

$$\delta = \tfrac{1}{2}(p - p \cdot c) . \tag{2}$$

The setup is shown in figure 1.

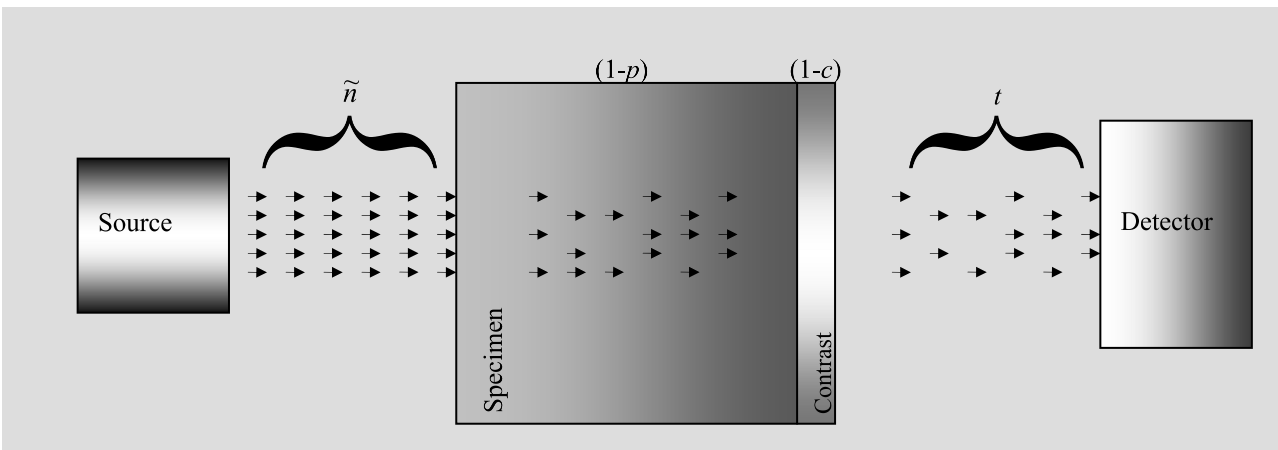

Fig. 1: The core model of a source sending of $\tilde{n}$ photons through a specimen of attenuation (1-*p*), possibly a thin additional absorber of attenuation (1-*c*), and a detector for the registration of the *t* transmitted photons.

The probability $\mathrm{P}(\tilde{n})$ for $\tilde{p}$ being in the range between $p - \delta$ and $p + \delta$ is the same as the sum of the probabilities for the count value t to be in a range between a lower limit $dwn(\tilde{n}, p, \delta)$ and a upper limit $up(\tilde{n}, p, \delta)$. Here $dwn(\tilde{n}, p, \delta)$ represents the closest integer larger than $\tilde{n} \cdot (p - \delta)$ and $up(\tilde{n}, p, \delta)$ represents the closest integer lower than $\tilde{n} \cdot (p + \delta)$. $\mathrm{P}(\tilde{n})$ is therefore calculated by

$$\mathrm{P}(\tilde{n}) = \mathrm{P}(\tilde{p} \in [p - \delta; p + \delta]) = \sum_{i=dwn(\tilde{n},p,\delta)}^{up(\tilde{n},p,\delta)} \binom{\tilde{n}}{i} p^{i} (1 - p)^{\tilde{n}-i} . \tag{3}$$

In connection with the rise of $\mathrm{P}(\tilde{n})$ with $\tilde{n}$ the minimum number of photons *n* can be determined that is necessary to gain an estimate of p in the range between $p - \delta$ and $p + \delta$ with a certainty higher than a chosen level *s*:

$$n = Min\{\tilde{n} \,|\, \mathrm{P}(\tilde{n}) \geq \mathrm{s}\} . \tag{4}$$

n is a measure of exposure. Its value depends on c (or δ), p and s. It can be found numerically using mathematical computer software [4]. This core model allows for the calculation of dependences of the *n* on accuracy and precision.

In order to find the energy dependence for n we now assume that *p* and *c* describe absorbers of only one material. The two attenuations of those absorbers are calculated from only one common energy dependent attenuation coefficient

$\mu(E)$ and two different thicknesses. The absorption behavior of different materials is provided by different organizations [5]. We define the thickness of the main layer x and a constant k so that the thickness of the contrast layer equals $\frac{x}{k}$. The transmission will fall exponentially with the thickness of the material:

$$p = e^{-(\mu * x)} \tag{5}$$

and

$$c = e^{-\left(\mu * \frac{x}{k}\right)}. \tag{6}$$

This leads to the connection of c and p for all possible values of μ and therefore for all energies which is

$$c = p^{\frac{1}{k}}. \tag{7}$$

The constant k can be found at one energy level from the relation between c and p. It connects c to the values of p over the whole range of possible absorption. This simplification enables us to use p as an energy measure, which allows for the identification of the energy dependence of n. By the choice of the same material for main absorber and contrast absorber we avoided the necessity to include the ratio between two different attenuation coefficients into the calculation. For the use of different materials the change could be absorbed into a modified definition of k, too, but only if the ratio of their different attenuation coefficients can be taken to be independent from the energy.

For a certain choice of material and thickness of the absorbers p is directly connected to the photon energy for a monoenergetic x-ray source. The connection can also be directly calculated for x-ray spectra of realistic diagnostic X-ray sources. These connections allow the inclusion of the dependence of the detector sensitivity on different X-ray spectra.

3. RESULTS

3.1 Minimal Dose Dependences on Desired Certainty

The minimal number of photons required for the imaging task defined above is dependent on the certainty level s of the desired gain of information. For even the highest number of photons sent through an absorber the measured absorption $\tilde{p}$ will not be in the range defined by the maximal error δ around p with 100% certainty. If the desired certainty is set to one, one would need an infinitely high number of photons n or dose: the exponential rise of n with s is bent up against that limit. Equally it is bent down close to s=0%. Fig. 2 shows the dependence of n on the level of security s with the main absorption fixed at p=50% and the contrast fixed at c=1%.

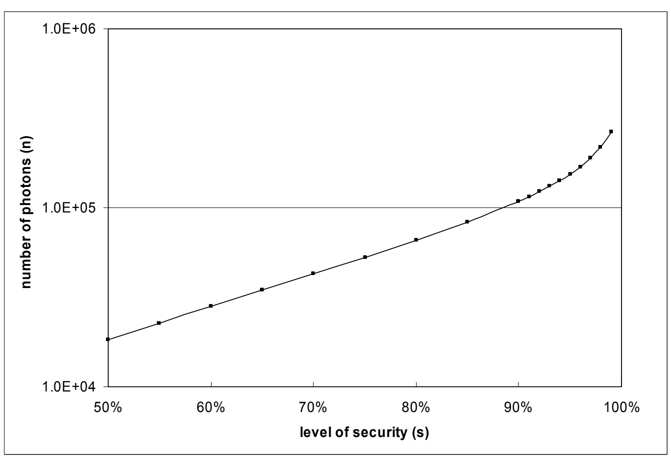

Fig. 2: Exponential and over exponential rise of input photons n needed with the sought-after level of significance s for the correct determination of the main absorption of p=50% with a maximal deviation of c=1%.

The dependence of *n* on the transmission of the contrast *c* defining the width of the aiming interval shows similar behavior. If the transmission attributable to the contrast reaches 100% the highest number of photons will not make the detection possible (Fig. 3).

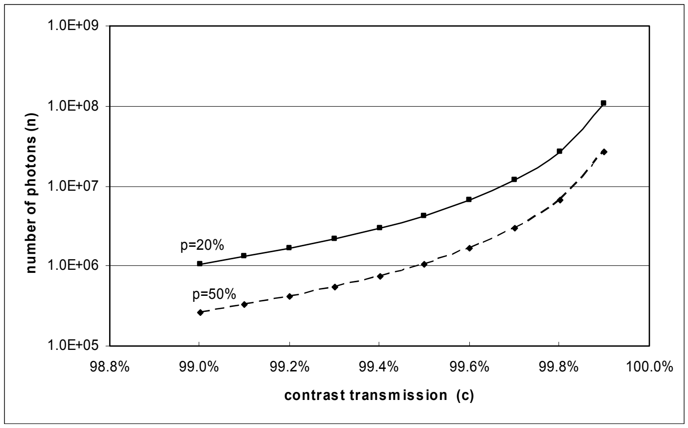

Fig. 3: Over exponential rise of *n* with the transmission *c* of the desired contrast resolution approaching 100% (s=99%).

3.2 Mutual Dependence of Minimal Dose and Energy

In order to get quantitative data about the change of *n* defined as above with an energy related figure the relation between *p* and *c* is fixed at a certain level of *p* through equation (7). The connection represents a certain thickness ratio between a main absorber and a layer of the same material defining the contrast one wants to resolve. Fig. 4 shows the change of the contrast transmission with main absorber transmission resulting from equation (7) for an exemplary of start value of *c*=1% at *p*=50% which leads to a thickness ratio of 1/69.
For the same input values Fig 5 and Fig. 6 show the dependence of *n* versus *p*. One finds a minimum for the exposure at *p*≈20%. A change in the energy of the input photons leading to a transmission of *p* just below 4.5% or just above 55% leads to a rise of *n* by a factor of 1.5 above the minimal possible value.
The change of the thickness ratio chosen to define the contrast resolution changes the values for *n*. In spite of that there is no change in the shape of the curve and especially not in the position of the minimum. For example if the start value for *c* at p=50% is set to 0.1% instead of 1% then *n* rises roughly by a factor of 100.

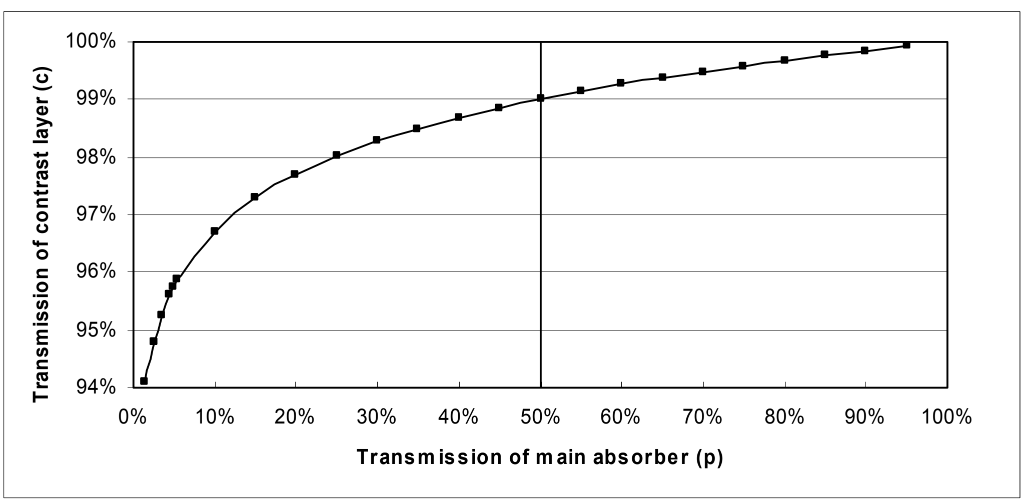

Fig. 4: Coupled change of transmission of the main absorber p and the contrast absorber c representing homogeneous bodies of different thicknesses. The contrast layer thickness is 1/69 of the thickness of the main absorber in the data shown (k=69).

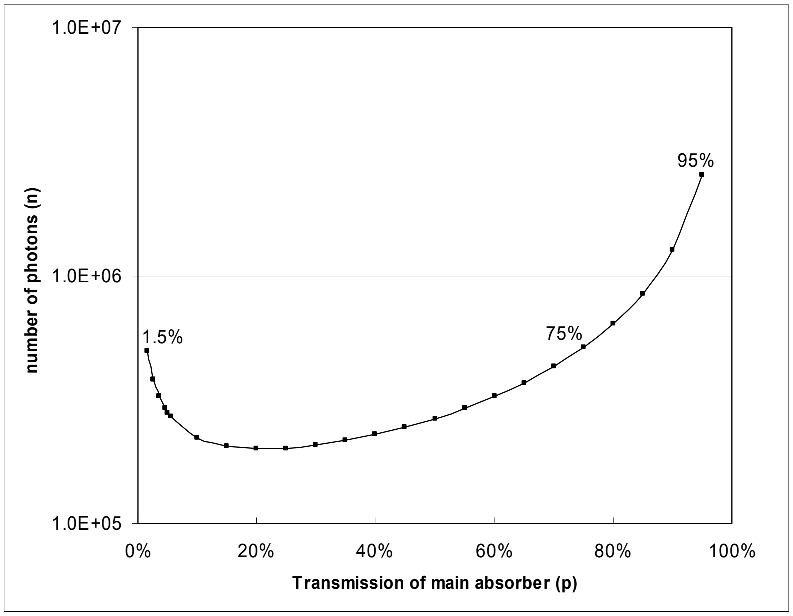

Fig. 5: Logarithmic plot of the number of photons *n* needed for the discrimination of a contrast transmission *c* behind a main absorber of transmission *p* connected by a ratio of thicknesses of 1:69.

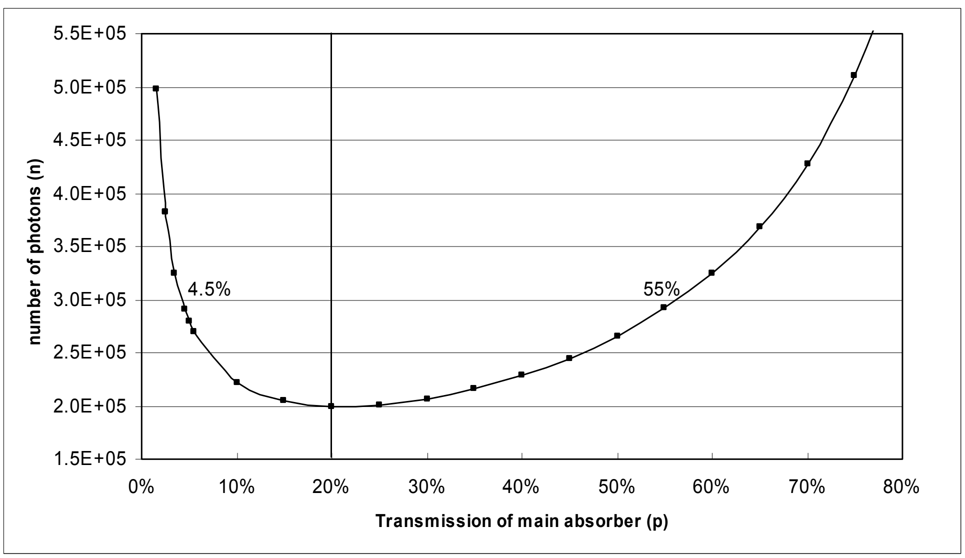

Fig. 6: Number of photons *n* needed for the discrimination of a contrast transmission *c* behind a main absorber of transmission *p* connected by a ratio of thicknesses of 1:69.

To include the dependence between transmission and energy it is crucial to specify the material of the specimen under examination. The energy dependence of the absorption of a large area detector can be multiplied directly with the result. As an example Figure 7 shows the energy dependence of n for monoenergetic exposure of specimens of 2cm and 4cm of soft tissue (ICRP) respectively connected with the detection by a CsI screen of 0.2mm thickness. There is a shift to the efficiency of the low energetic photons with lower thickness of the specimen.

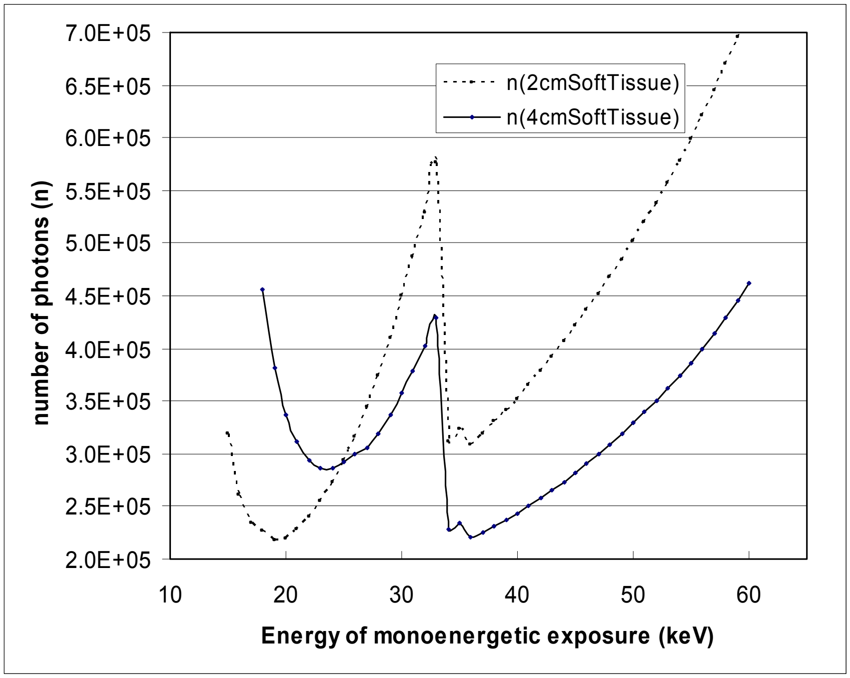

Fig. 7: Minimum number of monoenergetic photons necessary to perform the described imaging task at a specimen of 2cm and 4cm soft tissue and detection by a 0.2mm CsI-screen.

4. DISCUSSION

The work presented here shows how the exposure needed for simple large area contrast detection can be directly calculated in a simplified model of a radiographic imaging system.

There is a strong dependence on the desired certainty of the result. The fact that one gets pretty safe information with reasonable expense but have to spend indefinitely more to reduce the last bit of uncertainty is mathematically reproduced. Also it is quantified how much harder it is to detect a smaller contrast behind a thicker absorber. The results allow for a justification about how low ALARA should get.

We give an example of how to introduce realistic parameters into the model: if the number of photons needed for a safe decision in a contrast experiment is to be minimized then the energy applied is to be adapted to the thickness of the specimen. This result is found including absorption properties of human tissue and as a measure of its spectral sensitivity of the detector. Both are included by multiplication onto previous results. In order to achieve conclusions in dose considerations further weighting is necessary including realistic x-ray spectra, absorption data of real patients and conversion coefficients for organ doses.

Our model is limited to the image generation. We try to examine questions connected with what information can be expected from an image depending on how it is generated. So far there is no model of observer performance for human or computer aided evaluation included.

5. OUTLOOK

The system is open for further modification and extension. Especially features of spatial resolution could be included. The basic question here would be how different the number of photons hitting two adjacent detector elements needs to be so that they can be attributed to differences in the respective absorbing parts of the specimen. A following modification could be to allow for the consideration of pixel to pixel variability and the information gain by evaluation of more than one pixel.

In order to find optimal exposure conditions for x-ray radiography it is necessary to quantify the absorption behavior of real patients and include the results into the calculation and to apply realistic x-ray spectra. Additionally conversion coefficients to calculate organ doses have to be included as the final goal is to minimize possible detriment from medical x-ray radiation.

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