Stephan Rosendahl* and Ludwig Büermann

Dynamic determination of equivalent CT source models for personalized dosimetry

Abstract: With improvements in CT technology, the need for reliable patient-specific dosimetry increased in the recent years. The accuracy of Monte-Carlo simulations for absolute dose estimation is related to scanner specific information on the X-ray spectra of the scanner as well as the form filter geometries and compositions. In this work a mobile measurement setup is developed, which allows both to determine the X-ray spectra and equivalent form filter of a specific scanner from just one helical scan in less than 2 minutes.

Keywords: CT, Personalized Dosimetry, X-ray Spectra, Bow-Tie Filter

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1 Introduction

CT dosimetry is currently based on the quantities *volume computed tomography index* ($CTDI_{vol}$) and *dose-length product* (DLP). The underlying quantity CTDI was developed as early as in the 1980’s by Shope et al [1]. Estimates on the effective dose E are generally derived from $E = c \cdot DLP$ where c is a conversion factor calculated for standard-sized patient models. This method is neither patient- nor scanner-specific and can result in significant derivations of the estimated and the real effective dose of the patient [2]. Furthermore, recent developments in CT technology like cone-beam CT or current modulation as well as modern applications (e.g. tissue perfusion studies) require new concepts for accurate dose estimations [2]. One approach to personalized CT dosimetry uses Monte Carlo calculations to determine the effective dose on an individual basis with the specific object geometry obtained from DICOM data after scanning. Unfortunately, detailed information on the X-ray spectra and form filter geometries (bow-tie filters) of CT scanners which are needed for the simulations are typically unknown. But it has already been shown in the literature that equivalent source models and bow-tie filter geometries can be generated and embedded into MC simulations to calculate the dose [3-4]. Up to now, these concepts have been neither validated for absolute dose accuracy nor optimized for practice in a clinical environment. In the framework of the EMPIR project 15HLT05 “*Metrology for multi-modality imaging of impaired tissue perfusion*” [5], a mobile setup for measuring aluminum half-value layers and attenuation curves based on the COBRA method, established by Boone et al. [6], is developed. For tissue perfusion studies using computed tomography personalized dosimetry is of crucial importance since the usage of dynamic CT images in perfusion studies is related to a non-negligible dose applied to the patient. With the mobile equipment, all relevant information on the X-ray spectra as well as the bow-tie filter attenuation is collected within standard helical scan modes in less than 2 minutes’ scan time. Commercially available simulation software allows in a second step fast dose calculation with filtered spectra and equivalent form filter geometries as input parameters. In order to verify the accuracy of the procedure, the dose in different anthropomorphic phantoms will be measured and simulated in the course of the project and the values are compared for different CT scanner types.

In chapter 2 the mobile setup as well as the measurement procedure and the analysis code are presented. In chapter 3 the measurement results for a GE Optima 660 CT will be presented before a final discussion and an outlook is given in chapter 4.

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2 Materials and methods

Since standard spectroscopy with germanium detectors is not practical for CT scanners in clinical environment, the measurement of aluminium half value layers is a feasible alternative and allows for good approximation of the photon fluence spectra. The measured ratio κ_{Meas} of the attenuated air kerma K_x to the unattenuated air kerma K_0 with

$$\kappa_{Meas} = \frac{K_x^{Meas}}{K_0^{Meas}} \qquad (1)$$

as function of increasing aluminium thickness x_{Al} is compared to calculated values to estimate an Al-equivalent filtering d_{Al}:

$$\kappa_{Calc} = \frac{K_x^{Calc}(x_{Al})}{K_0^{Calc}} = \frac{\int \Phi_E\, e^{-\mu_{Al}(E)\cdot d_{Al}}\, e^{-\mu_{Al}(E)\cdot x_{Al}}\, E\left(\frac{\mu_{en}}{\rho}\right)_{air} dE}{\int \Phi_E\, e^{-\mu_{Al}(E)\cdot d_{Al}}\, E\left(\frac{\mu_{en}}{\rho}\right)_{air} dE} \qquad (2)$$

The Al-equivalent filtering represents the thickness at the centre of the bow-tie where the filtration is minimal and includes also any filtration from the x-ray source itself. The details of the method can be found in [6-7]. Typically, these measurements are performed in service mode with the source placed at the 6 o'clock position. For clinical purpose this is not feasible since it is time consuming and the service mode is usually not available. Hence, different methods have been proposed to perform them dynamically in standard patient mode [8-9].

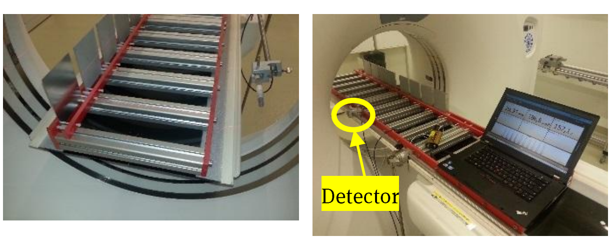

Figure 1: Experimental setup for equivalent source determination.

In this work, a different technique is used: Aluminium plates of different thickness (1 mm to 20 mm) are placed on a holding frame on the patient table as shown in figure 1. The detector is positioned inside the gantry on the iso-plane $y = z = 0$ but of the centre at $x \approx 250$mm. During a helical scan the plates are moved through the gantry and the detector (Radcal ionisation chamber type 10x6-0.6CT with Radcal ACCU Gold readout) measures the intensity as a function of time. Two examples of such waveforms with and without Al filtration are shown in figure 2 bottom and top, respectively. The two large peaks mark the points where the source and the detector are at closest ($y = 0$, at 9 o'clock in figure 1) and one rotation of the gantry with a rotation time of 1 second is fulfilled. In between the source passes the opposite point

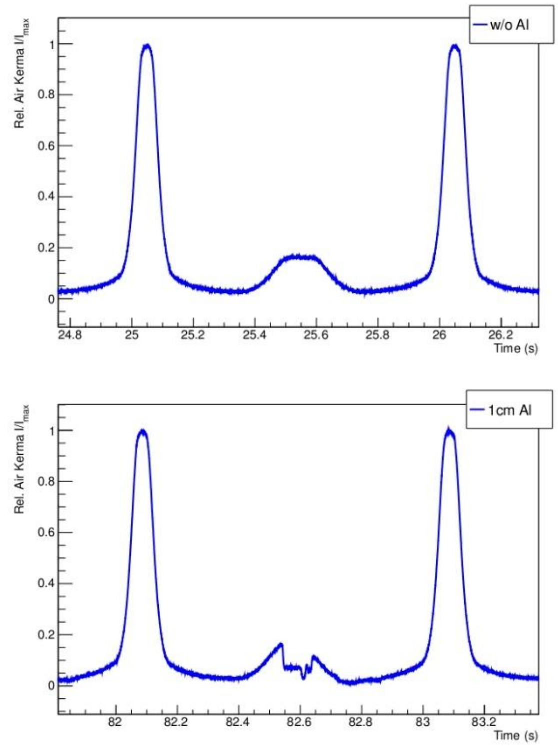

Figure 2: Time-resolved relative air kerma without aluminum filtering (top) and with 1cm Al filtering (bottom).

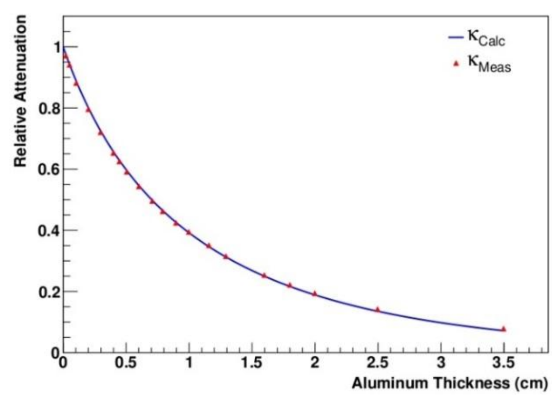

Figure 3: Relative attenuation κ as function of the thickness of the aluminum plates. In this example a measurement at 120kV (red points) is compared with the calculation (blue line).

(also $y = 0$, at 3 o'clock) which is the interesting point for the HVL analysis. The intensity in that position is decreased when the aluminium is passing the beam as one can see in figure 2, bottom. From the relative attenuation, the Al-equivalent filtration thickness can be calculated, according to [7]. In total 9 plates, have been scanned in one series and for each plate 5 rotations have been performed with a total scan time of less than 2 minutes. Custom analysis software generates the attenuation curve as a function of the thickness of the plates (see figure 3) by finding the region of interest and extracting the attenuation values after binning and averaging over several measurements.

Beside the HVL information, also the attenuation profile of the bow-tie filter is recorded as well. Without any aluminium or patient table in the beam, the intensity profile is defined by the attenuation characteristics of the bow-tie filter. By using the COBRA method an attenuation profile as function of the fan angle for the used filter can be calculated. From this, an equivalent bow-tie filter of any material can be calculated, using the methods described in [6-7].

The hard- and software has been tested at a scanner of type CT, GE Optima CT 660. The measurements have been performed at different voltages of 80-140kV in helical mode with a collimation of 40 mm and a velocity of 20.62 mm/rotation (Pitch 0.516:1). The CT is equipped with two bow-tie filters, a large one, used for adult thorax scans and a small one, typically used for adult head scans and paediatric applications.

3 Results

For the MC simulation of the patient dose, the estimated X-ray spectra are of crucial importance. Nevertheless, the value for the calculated Al-equivalent thickness can be used to check the reproducibility of the method. Measurements have been performed at different days, with completely re-mounting the equipment. In table 1 the average values for the Al-equivalent filtration thickness is shown for the two bow-tie filters. The measurements at 80kV have been performed with higher current to get higher signal rates for large Al filtrations. For the large filter an Al-equivalent filtration of 10.8 mm to 11.5 mm have been estimated, depending on the anode voltage the standard deviation has been calculated to be less than 4%. For the small bow-tie filter, the filtration has been determined to 7.7 mm to 7.9 mm with standard deviations of less than 3%.

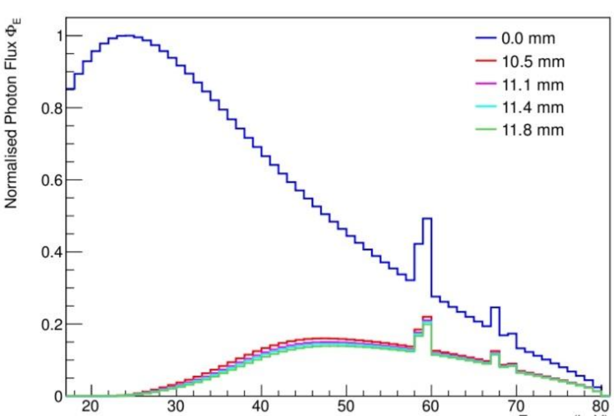

Figure 4: Normalized fluence spectra for 80kV anode voltage of the PTB research CT. Comparison of different Al-equivalent filtrations between 10.5 mm to 11.0 mm obtained from the measurements related to an unfiltered input spectrum.

Table 1: Average Al-equivalent filtration values for different anode voltages from several dynamic HVL measurements at the GE Optima CT 660 using the large bow-tie filter (top) and the small bow-tie (bottom).

Voltage (kV)	Al-equivalent filtration (mm)	Standard deviation (%)	Number of measurements
80 (250mA)	11.1	3.9	6
100 (150mA)	10.8	3.6	6
120 (150mA)	11.2	2.4	6
140 (150mA)	11.5	2.6	6
80 (250mA)	7.9	1.6	6
100 (150mA)	7.7	2.9	6
120 (150mA)	7.9	2.0	6
140 (150mA)	7.9	2.9	6

The resulting spectra for 120kV are shown in figure 4, comparing different Al filtrations with unfiltered spectra. The initial unfiltered spectrum which is used for the HVL calculations is shown in blue, while the filtered spectra which should be close to the real ones are shown in red, magenta, light blue and green. The influence of the uncertainty in the determination in the shape of the fluence spectrum to the dose simulation is currently investigated as part of the work package of the project.

For the generation of equivalent bow-tie models the attenuation curve is calculated from the intensity curve without any aluminium as shown in the top of figure 2, by applying the COBRA method. In figure 5 the attenuation curve for the large filter, extracted from a measurement with 120kV is shown as function of the opening fan angle θ. In red, the data from the dynamic measurement is plotted, obtained with the COBRA method. In the centre, the attenuation is minimal compared to the edges, where it is maximal. The results from the dynamic method (red triangles) are consistent with those from static measurements [7]. The uncertainties are evaluated as standard deviations of the mean of several measurements. With the attenuation curve an equivalent bow-tie filter from any material can be constructed according to the methods in [7]..

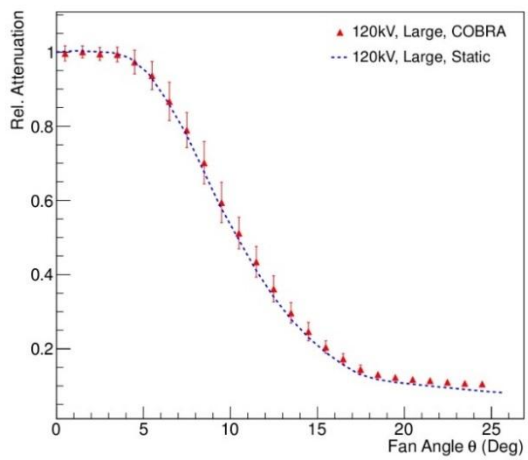

Figure 5: Relative attenuation curves obtained from static measurements (blue dots, from [7]) and by use of the COBRA method (red triangles, this work.

4 Discussion

In the framework of the EMPIR project, a mobile system for the application in clinical environment has been developed which allows for fast measurement of aluminum half value layer thickness to calculate the X-ray spectra for different anode voltages. Short preparation and measurement times and the minimal amount of hardware that is needed are the big advantages of the system. Custom software has been developed which allows for fast analysis of the data within few minutes to obtain the spectra. In addition, the data is used for the construction of equivalent bow-tie geometries using the COBRA method. Within one measurement series both HVL measurements as well as COBRA data are collected without the need of rearranging the system. The reproducibility of the approach has been tested by comparing the estimated values for the equivalent aluminum thickness to be less than 4%. In addition the implementation of the COBRA algorithm into the software is tested by comparing the attenuation curves with the ones obtained from static measurements.

The recent phase of the project includes the application of the method to several different CT scanners in clinical environments and the validation of the method by comparing dose simulations in (anthropomorphic) phantoms with absolute measurements of the dose within the different phantoms. These aspects are treated in collaboration with project partners from Radiation and Nuclear Safety Authority, Finnland (STUK) and the Hospital District of Helsinki and Uusimaa (HUS).

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Author's Statement

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