Physica Medica 32 (2016) 59–66

Contents lists available at ScienceDirect

Physica Medica

journal homepage: http://www.physicamedica.com

Original Paper

Non-invasive experimental determination of a CT source model

Babak Alikhani *, Ludwig Büermann

Physikalisch-Technische Bundesanstalt (PTB), D-38116 Braunschweig, Germany

A R T I C L E I N F O

Article history:
Received 23 April 2015
Received in revised form 15 September 2015
Accepted 22 September 2015
Available online 24 October 2015

Keywords:
CT
Photon fluence spectra
Bow tie filter
Ionization chamber
Solid state detector

A B S T R A C T

Non-invasive methods to determine equivalent X-ray source models of a CT scanner are presented. A high-precision technique called TRIC ("Time Resolved Integrated Charge") was developed and used to characterize the bow tie filters (BT) of the CT scanner installed at Physikalisch-Technische Bundesanstalt (PTB). Aluminum (Al) and polymethyl methacrylate (PMMA) equivalent thicknesses of the BT filters at all tube high voltages were evaluated, assuming that those consist of only one material. Thereby two different dose probes were used, a solid state detector and an ionization chamber, the former characterized by a significant and the latter by an almost negligible energy dependence of the air kerma response. A method was developed to correct for the energy dependence of the solid state dose probe. Next, a two-component material was assumed and equivalent BT filters were evaluated. The latter method was also applied using the known real BT filter materials and compared with the shape of the real BT filters. Finally, the results obtained by the TRIC method were compared with those obtained by using the so-called COBRA method ("Characterization Of Bow tie Relative Attenuation"), the latter being more suitable for measurements in a clinical environment.

Introduction

For the past two decades there have been remarkable improvements made in the technology of computed tomography (CT) scanners entailing a noticeable increase in the frequency of advanced CT examinations worldwide and causing a still rising interest in the radiation dose in CT [1]. Although the CT technology developed very fast, driven by new X-ray tubes, spiral CT scanning and advanced multi-row detector technologies, the radiation dose determination in CT is still mainly based on the computed tomography dose index (CTDI) concept [2] introduced only a few years after the development of the first CT scanner in the 1970s. In addition, modern scanners provide a variety of new dose reduction technologies such as automatic exposure control and tube current modulation combined with new iterative reconstruction techniques. Chen et al. [3] developed a Monte Carlo software package which provides fast on-site calculation of three-dimensional dose distributions in the CT suit which makes it a practical tool for any type of CT-specific application. The calculated dose distributions can be simply normalized to the air kerma output of the X-ray tube measured free-in-air in the center of the gantry. Monte Carlo simulations of this kind need some basic input data like the scanner geometry and the CT source model. The CT source model is mainly characterized by the initial X-ray spectrum and the subsequent filtrations including the bow-tie-shaped form filters. Unfortunately, some of the necessary input data are proprietary information and usually not provided by the manufacturers. To overcome these restrictions some efforts have been made in recent years to develop methods which allow the determination of so-called "equivalent X-ray source models" which consist of an energy spectrum and filtration description that are based wholly on measured values [4–8]. First measurements and evaluations of this kind were performed in 2009 by Turner et al. [4]. Boone [5] developed a theory of a simple method for characterizing the bow tie filter (BT) attenuation profile in CT scanners. While Boone presented simulations of his theory, McKenney et al. [6] verified the method experimentally which they denoted as the "Characterization Of Bow tie Relative Attenuation (COBRA) method". Similar methods were also used by McMillan et al. [7] to develop and validate a source model for a kilovoltage cone-beam CT and Whiting et al. [8] to measure bow tie profiles in CT scanners using a real-time dosimeter.

The purpose of the current study is to further improve and validate these experimental methods in order to generate equivalent X-ray source models. Using the half-value layer measurements, the initial X-ray spectra were characterized. The most essential tool for the measurements of the bow tie attenuation profile is a real-time dose probe. A high-precision technique called "Time Resolved Integrated Charge (TRIC)" is developed and used to measure air kerma rate profiles along the two BT filters. Up to now, how the results of equivalent BT filter determinations are influenced by the energy dependence of the response of the dose probe has never been studied, which is discussed in this work. Furthermore a method is presented to correct for the energy response of the dose probe. The

* Corresponding author. Physikalisch-Technische Bundesanstalt (PTB), D-38116 Braunschweig, Germany. Tel.: +49 531 592 6213; fax: +49 531 592 6205.
E-mail address: babak.alikhani@ptb.de (B. Alikhani).

http://dx.doi.org/10.1016/j.ejmp.2015.09.006
1120-1797/

procedure is similar to the algorithm as used previously by McKenney et al. [9]. Next, a two-component material is assumed and equivalent BT filters were evaluated. The latter method is also applied using the known real BT filter materials and compared with the shape of the real BT filters. Finally, the results obtained by the TRIC method were compared with those obtained by using the COBRA method, the latter being more suitable for measurements in a clinical environment without the availability of the service mode.

Materials and methods

CT scanner

The scanner installed at PTB used in this work is a 64-slice CT of the type GE Optima 660 CT and is manufactured by the General Electric (GE) Company. The opening diameter of the CT gantry is 700 mm. The distance between the X-ray tube focus and the isocenter of the gantry is 541 mm. The X-ray source is designed for Optima 660 CT systems with the four tube high voltages of 80, 100, 120 and 140 kV. The tube current varies from 10 mA to 600 mA. The X-ray tube anode angle is 7 degrees. The collimation system can be set at six different widths at the isocenter from 1.25 mm to 40 mm ($a = 1.25, 2.5, 5, 10, 20, 40$ mm). The CT scanner contains two BT filters (small and large) which are fixed between the X-ray tube and the collimation system [10,11]. Proprietary information like collimator material and geometry, BT filter shape and material is made available to the authors by a non-disclosure agreement.

Real-time dose probes

Two different types of detectors were used as real-time dose probes: An ionization chamber (IC) of the type Radcal 10 × 6 - 0.6 CT with a sensitive volume of 0.6 cm^3 and an Si-based solid state detector (SSD) of the type RTI CT Dose Profiler with a dimension of 1 mm^3. As radiation hits these isotropic probes from all directions the dose probes register the dose values and send the information to the data acquisition system. The responses of both detectors in terms of air kerma free-in-air were measured at the calibration X-ray facilities of PTB at which the free-air ionization chambers [12] as primary air kerma standards are operating. Narrow spectrum X-radiation qualities generated with tube high voltages between 20 kV and 300 kV were used for these measurements. The resulting air kerma responses were plotted against the air kerma weighted mean energies of the radiation qualities, and the results are shown in Fig. 1. Because the widths of the used X-ray spectra are very narrow, it can be assumed that the plotted curves show the response as a function $R(E)$ of monoenergetic photons instead of the mean energies. From Fig. 1 it is obvious that the IC reflects an almost energy-independent air kerma response in the relevant photon energy range from about 40 keV to 140 keV, whereas the corresponding response of the Si-based SSD decreases significantly by about a factor of 5. However, the advantages of the SSD are its high detection efficiency and its high spatial resolution of about 1 mm compared with the IC which has a length of 21 mm and a diameter of 9 mm. But the IC is well suited for the purpose of this work.

X-ray spectra

In order to characterize the X-ray spectra generated at X-ray tube peak voltages (kVp) of 80, 100, 120 and 140 kV, half-value layer (HVL) measurements were performed. For that purpose, the CT scanner was operated in the service mode. The CT gantry was fixed at the 6 o'clock position and the Radcal IC was placed in the gantry plane directly above the X-ray tube. This chamber is well suited for air kerma measurements of this kind because it almost does not change its response when exposed to the radiation qualities filtered with

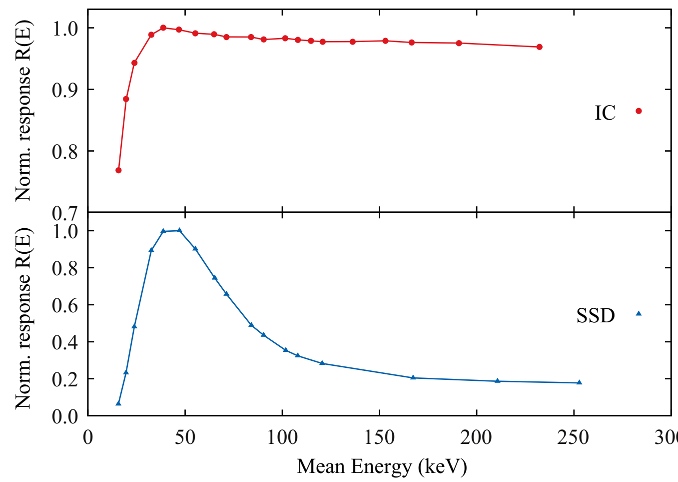

Figure 1. Normalized air kerma response $R(E)$ as a function of the mean energy of the ISO narrow-spectrum series [13] and the special narrow radiation qualities used at PTB (PTB AS series) for the IC (top) and the SSD (bottom).

aluminum layers of different thicknesses. Charge measurements were performed at a tube current of 150 mA and an exposure time of 1 s. The width of the X-ray beam was collimated by $a = 40$ mm. The first and second HVLs for aluminum (HVL_1 and HVL_2) were determined by increasing the thickness of aluminum layers until the measured charge in the attenuated beams is equal to one-half and one-quarter of the charge measured without aluminum layers. Using the program "SpekCalc" [14], photon fluence spectra were generated for a tungsten anode of anode angle 7 degrees without any added filtration except for the air which was present in the measurements between the x-ray source and the point of measurement. Next aluminum layers of increasing thicknesses were added to the generated non-filtered spectra until the evaluated values of HVL_1 and HVL_2 were almost identical with those obtained by the measurements. The obtained spectra were used as an approach to the real initial spectra of the scanner with an aluminum-equivalent inherent filtration. In order to validate the correspondence of these spectra with the real ones, the air kerma attenuations by passing through different aluminum layers with known thicknesses were calculated and compared with measurements.

Experimental setup for the air kerma profile measurements

The characterization of the BT filters is based on the time-resolved integration of the charge (TRIC) signals. Thereby, the charge signals produced in the detectors are collected by an electrometer and digitized using a multimeter. The derivation of the charges correlates to the air kerma profiles.

The measurements were performed in the service mode of the CT scanner. The rotation of the gantry was stopped and the X-ray tube was parked at the 12 o'clock position. The patient support was located outside the gantry. Using the IC and the SSD by the kVp of 80, 100, 120 and 140 kV, the air kerma profiles with and without the BT filters as a function of the fan angle θ were measured. The detectors were fixed at a movable table, which shifted them continuously along the X-axis during an irradiation time of $\Delta t = (8.62 \pm 0.06)$ s a distance of $x_{total} = (500 \pm 1)$ mm. The distance x_{total} is symmetric around the isocenter of the scanner and corresponds to the fan angle θ approximately between –25 degrees and +25 degrees. While the IC and the SSD were moved along the x-axis, the produced charges were collected and read out by the data acquisition system. The data acquisition system consisted of a Keithley 616 electrometer and a Keithley 2000 multimeter. The electrome-