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Dual energy CT

Extracting atomic numbers and electron densities from a dual source dual energy CT scanner: Experiments and a simulation model

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ABSTRACT

Background and purpose: Dual energy CT (DECT) imaging can provide both the electron density ρ_e and effective atomic number Z_{eff}, thus facilitating tissue type identification. This paper investigates the accuracy of a dual source DECT scanner by means of measurements and simulations. Previous simulation work suggested improved Monte Carlo dose calculation accuracy when compared to single energy CT for low energy photon brachytherapy, but lacked validation. As such, we aim to validate our DECT simulation model in this work.
Materials and methods: A cylindrical phantom containing tissue mimicking inserts was scanned with a second generation dual source scanner (SOMATOM Definition FLASH) to obtain Z_{eff} and ρ_e. A model of the scanner was designed in ImaSim, a CT simulation program, and was used to simulate the experiment.
Results: Accuracy of measured Z_{eff} (labelled Z) was found to vary from −10% to 10% from low to high Z tissue substitutes while the accuracy on ρ_e from DECT was about 2.5%. Our simulation reproduced the experiments within ±5% for both Z and ρ_e.
Conclusions: A clinical DECT scanner was able to extract Z and ρ_e of tissue substitutes. Our simulation tool replicates the experiments within a reasonable accuracy.
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A computed tomography (CT) scan provides a measurement of the photon linear attenuation coefficients μ of the scanned object, often expressed as Hounsfield Units (HU). As the HU of a given material varies with photon energy, the measured values are specific to the tube potential and filtration used and may not be directly employed to perform dose calculations in another photon energy range. Furthermore, the attenuation coefficient is a function of both medium density and elemental composition. Thus, there can be several materials with the same HU having different densities and elemental compositions, complicating the use of CT for identifying material properties in applications such as dose calculation [1].

By measuring the linear attenuation coefficient at two different tube potentials, dual energy CT (DECT) provides a means to decompose μ in two components, namely the electron density ρ_e and effective atomic number Z (we call Z the value provided by DECT and Z_{eff} the expected quantity), thus facilitating tissue type identification. The potential benefits of using DECT in radiotherapy have been investigated in the context of brachytherapy dose calculations [2], proton stopping power ratio estimation [3,4] as well as kV and MV photon dose calculations [5,6]. Other studies investigated optimal filtration of the high and low kVp photon spectra for image quality improvement [7,8]. Several of these studies are based on simulations rather than direct measurements, as DECT scanners are only now becoming clinically available. Goodsitt et al. have recently reported on the accuracy of experimentally derived atomic numbers provided by a DECT scanner based on the single source, rapid kVp switching design [9].

In this work a second generation dual source DECT scanner is investigated. In this design, two pairs of X-ray tube and detector array rotate around the patient, simultaneously acquiring a high and low kVp image. We report on the accuracy of Z and ρ_e obtained from measurements of a standard electron density calibration phantom using the algorithm of Bazalova et al. [5,6], adapted from Torikoshi et al. [10]. In addition to measurements, a simulation model of the DECT scanner has been designed. By comparing results from simulation and measurements we aim to validate the use of the simulation tool for DECT research.

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A recent simulation study by our group explored the use of DECT as an alternative to SECT based dosimetry in low dose rate brachytherapy dose calculations [11], which are highly sensitive to tissue composition assignment [12]. Calculated dose distributions from simulated CT images segmented into tissue composition and density using a SECT technique compared to a DECT technique proved more accurate using DECT. As our simulations [11] supported the use of DECT imaging for low energy photon dose calculations, validating our findings using a scanner readily available to the clinic is the focus of this manuscript.

Materials and methods

Experiment

A cylindrical RMI 465 phantom (Gammex Inc., Middleton, WI) was scanned at a second generation dual source CT scanner (SOMATOM Definition FLASH, Siemens Healthcare, Forchheim, Germany) operated in dual energy mode. The scanner has two X-ray tube and detector array pairs, their respective fields of view being 34 cm and 50 cm [13]. Tube potentials were 80 kVp and 140 kVp, with additional Sn filtration for the high kVp (140 kVp/Sn). Exposures of 900 and 348 mAs were used to minimize noise. Images were reconstructed at the scanner with a very smooth filter (B10f), again to reduce noise. The geometry of the phantom inserts is shown in Fig. 1a and material properties are listed in Table 1. The effective atomic number Z_{eff} reported in Table 1 is calculated according to the following equation:

$$Z_{\mathrm{eff}} = \left(\frac{\sum_i f_i \rho_{\mathrm{e,i}} Z_i^{3.3}}{\sum_i f_i \rho_{\mathrm{e,i}}} \right)^{1/3.3} \quad (1)$$

where f_i is the mass fraction of element i. In this paper we call Z_{eff} the quantity derived from Eq. (1) and Z the quantity provided by the DECT algorithm.

From the CT images the atomic number Z and the electron density relative to water, $\rho_{\mathrm{e}}/\rho_{\mathrm{e,w}}$, were obtained [5,6]. The CT scanner spectra required by the algorithm are shown in Fig. 1b and were generated with SpekCalc [14–16] by adjusting manufacturer nominal tube filtration to match measured half value layers (HVLs). The HVL of the scanner was measured at 80, 100, 120 and 140 kVp using a localization technique [17]. The CT scanner Gd_2O_2S detector response was also accounted for in the analysis [18]. Prior to analysis, three slices (thickness = 3 mm) were averaged and the 512 × 512 image matrices were rebinned to 256 × 256 to reduce noise and reduce calculation times.

Simulation

We simulated the experiment using ImaSim [19], a program using SpekCalc, which calculates X-ray projection images using ray tracing for a fan beam CT geometry. The attenuation coefficients of the tissue mimicking inserts were generated with the NIST XCOM database [20] based on the compositions found in Watanabe et al. [21]. The bowtie filters of the CT scanner were not modelled. The same spectra and detector response as mentioned above were used and images were reconstructed with a Shepp Logan filter [22]. A cupping artefact correction was applied; the correction results in uniform HU values when applied to projections of a uniform water cylinder of the same radius as the cylindrical phantom. To match attenuation coefficients generated from ImaSim simulations to those calculated using the spectrum and detector response of the CT scanner the following equation was used:

$$\frac{\mu}{\mu_{\mathrm{water}}} = \frac{\mathrm{HU}}{A} + B \quad (2)$$

where A and B are obtained by fitting [2]. The parameter A took the value 967.7 at 80 kVp and 986.4 at 140 kVp/Sn while B was unity for both spectra.

Results

The measured HVL are found in Table 2 and are compared to the nominal values provided by the CT scanner manufacturer. As it was not possible to operate the scanner in a parked position with the Sn filter in place, the HVL for the 140 kVp/Sn beam could not be measured. It was necessary to add 0.15 mm Ti in SpekCalc to the filtration reported by the manufacturer to match the modelled and measured HVL.

Fig. 2 presents the discrepancies between the attenuation coefficients obtained from scans of the RMI phantom at 80 kVp and 140 kVp/Sn and those obtained from ImaSim with the same photon spectra. The measured μ are obtained from Eq. (2) using A = 1000 and B = 1 while those from ImaSim use the A and B obtained from the fitting procedure. The HU used in Eq. (2) are the averages from a circular ROI positioned at each insert. The standard deviation

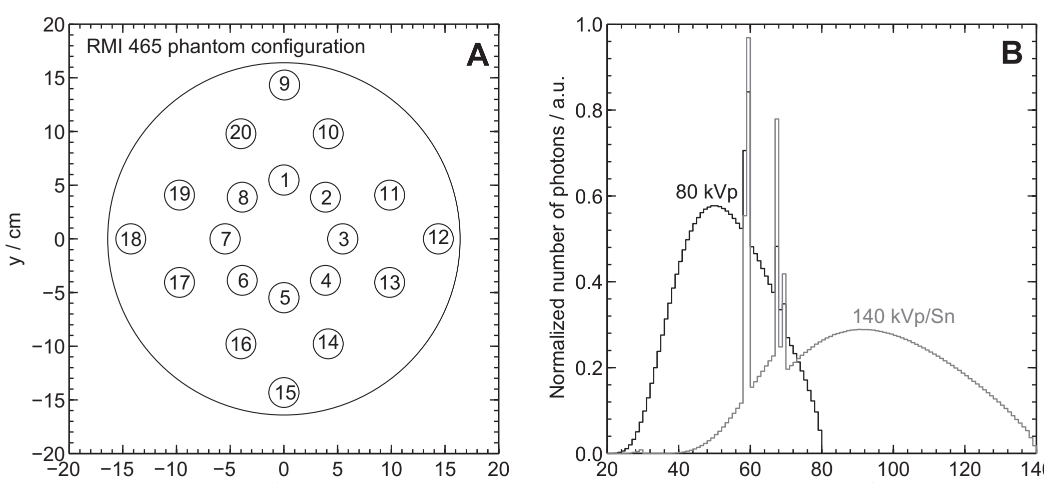

Fig. 1. (A) Configuration of the cylindrical RMI 465 phantom used in the experiments and simulations. The correspondence between insert number and material type can be found in Table 1. (B) CT scanner photon spectra used in the DECT algorithm and ImaSim simulations.