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Effective atomic number estimation using kV-MV dual-energy source in LINAC

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ABSTRACT

Dual-energy computed tomography (DECT) imaging can measure the effective atomic number (EAN) as well as the electron density, and thus its adoption may improve dose calculations in brachytherapy and external photon/particle therapy. An expanded energy gap in dual-energy sources is expected to yield more accurate EAN estimations than conventional DECT systems, which typically span less than 100 kV. The aim of this paper is to assess a larger energy gap DECT by using a linear accelerator (LINAC) radiotherapy system with a kV X-ray imaging device, which are combined to provide X-rays in both the kV- and MV-energy ranges. Traditionally, the EAN is determined by parameterising the Hounsfield Unit; however, this is difficult in a kV-MV DECT due to different uncertainties in the reconstructed attenuation coefficient at each end of the energy spectrum. To overcome this problem, we included a new calibration step to produce the most likely linear attenuation coefficients, based upon the X-ray spectrum. To determine the X-ray spectrum, Monte Carlo calculations using GEANT4 were performed. Then the images were calibrated using information from eight inserts of known materials in a CIRS phantom (CIRS Inc., Norfolk, VA). Agreement between the estimated and empirical EANs in these inserts was within 11%. Validation was subsequently performed with the CatPhan500 phantom (The Phantom Laboratory, Salem). The estimated EAN for seven inserts agreed with the empirical values to within 3%. Accordingly, it can be concluded that, given properly reconstructed images based upon a well-determined X-ray spectrum, kV-MV DECT provides an excellent prediction for the EAN.

1. Introduction

Dual energy computed tomography (DECT) is becoming increasingly accessible in clinical applications, and can be used to perform energy- and material-selective reconstructions. It can be applied to remove beam hardening [1–4] and can reduce the metal artefacts in reconstructed images [5,6]. Owing to the method's strong capacity to discriminate between different materials, DECT is used in advanced material imaging, based on iterative reconstruction algorithms in the image domain [7]. As a reliable linear relationship exists between the energy-subtracted Hounsfield Unit (HU) and the electron density in DECT scanners contrary to the case of single energy CT [8,9], a large improvement can be achieved through DECT when the HU is transformed into the electron density relative to water. This can improve the accuracy of patient dose calculations in radiation therapy treatment planning [10–13].

The DECT may bring an additional benefit to the dose calculation through the estimation of the effective atomic number (EAN) distribution in the human body [14,15]. The conventional dose calculations in radiation therapy allow only 'dose-in-water', which means that the subject is assumed to be composed of water. However, photon-electron interactions approximately depend on the atomic number; therefore, the conventional calculation contains errors. This is particularly pronounced in low-energy applications such as brachytherapy, due to the strong atomic number dependence of the photoelectric effect in the corresponding energy domains [16]. Additionally, hadronic interactions in particle therapy are prone to incorrect estimation when the EAN is not well determined [17–19]. In the case of proton therapy, a proton computed tomography is investigated to substitute or at least complement X-ray CT for more accurate proton therapy treatment planning. Another method is to use DECT. DECT provides reasonably accurate estimations of the EAN and relative electron density, which can be used in the classification of tissue type and the calculation of the stopping power ratio, thereby improving dose

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calculations, not only for proton therapy, but also for heavier ion therapy. Various approaches to derive the EAN were proposed, including the tissue substitute method (TSM) and the reference tissue method (RTM) [20]. The TSM is based on Rutherford parameterisation of the linear attenuation coefficients in order to derive a relation between the EAN and the ratio of the linear attenuation coefficients measured at the DECT scanner, whereas the RTM is based on the derivation of stoichiometric fit parameters for human tissues using reference compositions. Besides these, a generalized fit to the stoichiometric calibration of a dual-energy CT was proposed, by employing the proper orthogonal decomposition or Karhunen–Loeve expansion [21]. All these methods rely, however, on a parameterisation of the measured ratio of linear attenuation coefficients at low and high X-ray energies in the DECT scanner, discarding any knowledge of the X-ray spectrum.

In order to improve the EAN estimation performance in DECT, one can increase the energy gap between the X-ray sources used. In fact, the kV-MV DECT was found to improve the accuracy of the stopping power ratio estimation substantially compared to the kV-kV or MV-MV DECT methods, especially when the HU number uncertainties and artifacts such as imaging noise and beam hardening effects were considered [18]. Similarly, it is expected that the larger energy gap in the kV-MV DECT should improve the EAN estimation compared to current DECT systems.

In this study, we have used a kV-MV dual imaging source system, composed of a LINAC and a kV X-ray image-guided system mounted within the LINAC. Using both the kV and MV CT images of phantoms, we assessed the accuracy of the predicted EAN distributions. While a parameterisation method similar to those mentioned above can be used in the determination of the EAN, large uncertainties are likely to be found due to the different relative uncertainties in the HU numbers between the kV and MV X-ray energies. Instead, we revisited the spectral method [12,16,22], which parameterises the linear attenuation coefficient based on the atomic number using the source X-ray energy spectra. While other authors have suggested that the spectral method is inferior to the TSM and RTM for EAN estimation [20], we show that adding a calibration step based upon the X-ray spectrum for the evaluation of the total mean attenuation coefficient permits kV-MV DECT to accurately and precisely measure EANs using the spectral method.

2. Materials and methods

2.1. Estimating the EAN with dual-energy CT

The total attenuation coefficient μ varies with the electron density ρ_e, atomic number Z, and photon energy E. For the range of energies considered in radiotherapy, μ can be written as the sum of the attenuation coefficients for the individual interaction processes contributing to the attenuation, namely photo-electric ionization, Compton scattering, and electron–positron pair production:

$$\mu(E,Z,\rho_e) = \mu^{\text{phot}}(E,Z,\rho_e) + \mu^{\text{Comp}}(E,\rho_e) + \mu^{\text{pair}}(E,Z,\rho_e). \quad (1)$$

The attenuation coefficient due to photo-electric ionization, $\mu^{\text{phot}}(E,Z,\rho_e)$, can be approximated by [23],

$$\mu^{\text{phot}}(E,Z,\rho_e) = 3.45\times10^{-6}\rho_e(1+0.0082Z)\frac{Z^3}{E^3}, \quad (2)$$

provided the photon energy is well above the absorption K-edge for the element Z. Also, the attenuation coefficient due to electron–positron pair production, $\mu^{\text{pair}}(E,Z,\rho_e)$, can be approximated [24] as,

$$\mu^{\text{pair}}(E,Z,\rho_e) = 0.2545\rho_e(E-2.332)\frac{Z}{137}. \quad (3)$$

Unlike the above two processes, the attenuation coefficient due to Compton scattering, $\mu^{\text{Comp}}(E,\rho_e)$, is described theoretically by the Klein–Nishina formula [25] with the incoherent scattering function $S_0(E,Z)$,

$$\mu^{\text{Comp}}(E,Z,\rho_e) = 2S_0(E,Z)\pi\rho_e\left[\frac{1+E}{E^2}\left\{\frac{2(1+E)}{1+2E}-\frac{\ln(1+2E)}{E}\right\} + \frac{\ln(1+2E)}{2E}-\frac{1+3E}{(1+2E)^2}\right], \quad (4)$$

where the incoherent scattering function can be regarded as a correction factor due to a photon-bound electron collision, instead of the photon-free electron collision typically assumed in the Klein–Nishina theory. Although this function depends on both Z and E, the dependences are sufficiently small to be neglected in the ranges considered in this work.

If the spectrum is well known, then based upon the fractions of the incident photon energies, we can obtain the mean attenuation coefficient,

$$\langle\mu(Z,\rho_e)\rangle = \sum_i \alpha(E_i)\{\mu^{\text{phot}}(E_i,Z,\rho_e) + \mu^{\text{Comp}}(E_i,Z,\rho_e) + \mu^{\text{pair}}(E_i,Z,\rho_e)\}. \quad (5)$$

Here, $\alpha(E_i)$ refers to the fractional size of the i-th photon energy bin in the spectrum considered. From Eqs. (2)–(4), the total mean attenuation coefficient is a linear function of the electron density ρ_e. Accordingly, the ratio of two different photon energies is independent of ρ_e,

$$\frac{\langle\mu(Z,\rho_e)\rangle_L}{\langle\mu(Z,\rho_e)\rangle_H} = f(Z), \quad (6)$$

where $\langle\mu(Z,\rho_e)\rangle_L$ is the total mean attenuation coefficient for low-energy photons, while $\langle\mu(Z,\rho_e)\rangle_H$ is that for high-energy photons. As the ratio of the total mean attenuation coefficients between the different energies depends only on the atomic number Z, a reconstruction of the total mean attenuation coefficient using two different X-ray energy sources can be used to recover the EAN. The value of $f(Z)$ can be evaluated by simulating $\langle\mu(Z,\rho_e)\rangle$ of each element for the low- and high-energy X-ray spectra, which are themselves derived from a Monte Carlo calculation.

2.2. Calibration of total mean attenuation coefficient

In principle, CT reconstruction recovers X-ray attenuation coefficients voxel-by-voxel based upon the assumption that the incident photon spectrum is mono-energetic. In reality, the broad X-ray energy spectrum in most CT scanners yields a non-linear relationship between the energy depositions and reconstructed values for the attenuation coefficient. Furthermore, scattered photons and beam hardening, both of which are not rigorously considered in conventional CT reconstruction, produce discrepancies between the measured attenuation coefficient values and those expected from the materials constituting the object. To remove these uncertainties, we calibrate the reconstructed attenuation coefficients to map onto the expected ones, using a calibration phantom made of known materials. First, the total mean attenuation coefficients for some inserts in the phantom is estimated for two different X-ray energy sources by assuming that the spectra are invariant everywhere in the phantom. Then, the calibration function $\xi(\mu^{\text{recon.}})$ is defined as a function of the reconstructed attenuation coefficient satisfying the relationship

$$\langle\mu^{\text{calib.}}(Z,\rho_e)\rangle = \xi\langle\mu^{\text{recon.}}(Z,\rho_e)\rangle, \quad (7)$$

where $\langle\mu^{\text{calib.}}(Z,\rho_e)\rangle$ and $\langle\mu^{\text{recon.}}(Z,\rho_e)\rangle$ are the calibrated and reconstructed attenuation coefficients, respectively. To provide an analytic estimate of $\xi(\mu^{\text{recon.}})$, the measured values of the function were fitted using a fourth order polynomial via χ^2-minimisation.