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Review

Dual energy CT in radiotherapy: Current applications and future outlook

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A B S T R A C T

Dual energy CT (DECT) scanners are nowadays available in many radiology departments. For radiotherapy purposes, new strategies using DECT imaging are investigated to optimize radiation treatment for multiple steps in the radiotherapy chain. This review describes how DECT based methods can be used for electron density estimation, effective atomic number decomposition and contrast material quantification. Clinical radiotherapy related applications for improved dose calculation accuracy of brachytherapy and proton therapy, metal artifact reduction techniques and normal tissue characterization are also summarized together with future perspectives on the use of DECT for radiotherapy purposes.

Imaging is one of the cornerstones for diagnosis, radiation treatment planning and follow-up assessment of cancer patients. Computed tomography (CT) using X-rays is the most frequently used imaging modality for radiation therapy (RT), i.e. brachytherapy, external photon, electron and proton beam treatment [1]. Mainly the relatively easy calibration of Hounsfield Units from the CT scanner into electron density makes this modality perfectly suited for accurate dose calculation purposes in external photon beam RT. The use of other imaging modalities has increased over the past years such as magnetic resonance imaging (MRI) for regions where a high soft-tissue contrast is necessary, or functional imaging with dedicated radioactive tracers for positron emission tomography (PET). These modalities have for some treatment sites been integrated into the routine workflow for cancer patient imaging, but are nowadays typically still used in addition to CT imaging.

In recent years, the use of dual energy (DE) CT imaging has gained increased attention in radiology departments. Currently, multiple new strategies for using dual energy CT (DECT) systems for the entire chain of radiotherapy are also being investigated: e.g. improved dose calculation accuracy for brachytherapy and proton therapy, metal artifact reduction techniques and normal tissue characterization. This review article will describe the current technology available for DECT, review the possible applications and show future perspectives on the use of DECT for radiotherapy purposes.

Dual energy CT imaging: technology and physics

Imaging equipment for dual-energy CT imaging

The use of DE in CT scanners is not a recent idea. Already in the early days of CT imaging DE techniques were described [2,3]. Back then mainly technical limitations and computational power restricted the implementation of DE in routine practice. It was only a decade ago that DE imaging was introduced again, when the first clinical DECT scanner in 2005 became available [4,5]. This scanner used a dual-source technique by making use of two orthogonally-mounted X-ray tubes with corresponding detectors installed in the scanner, rotating around the patient independently operated with different kilovoltage settings. Since then, DECT has become a valuable tool in daily routine practice for different clinical mainly diagnostic applications e.g. differentiation of urinary stones, imaging of pulmonary embolism, neuro imaging or differentiation of pulmonary nodules [6–10].

Its easy use and radiation dose neutral application compared to standard routine protocols has made DECT an important new tool in daily routine practice. Furthermore, the use of DECT and further energy decomposition analysis offers a huge potential for improvement of image quality and further reduction of radiation dose [11].

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The most simple approach for acquiring DE images is the rotate–rotate DE, which acquires two different scans with two different kV settings sequentially, which inherently has the disadvantage of different time points of scanning, which might be influenced by patient movement, patient breathing or different contrast media timing. Nowadays different vendors offer different solutions for DE imaging with CT [12]. As mentioned, one conceptually simple approach uses two orthogonal X-ray tubes, which are operated with two different kV settings. The X-ray tubes rotate around the patient and therefore simultaneously acquire images of different kV. Typically combinations of 80/140 kV or 100/140 kV were used, while with the latest generation even spectra of e.g. 70/150 kV are possible with the advantage of further separating the energy spectra allowing better material decomposition. Another technology uses rapid kV switching (usually in less than 0.5 ms) of the X-ray tube. Alternatively, spectral information may be obtained by a dual layer detector, which allows the detection of high and low energy photons from the same imaging beam but on two different detector layers. A recent solution available is the so called twin beam technology which offers a DE option for single source scanners by automatically splitting the X-ray beam using dedicated filters into two energy spectra at the kV source, which then expose different geometric parts of the detectors [12]. The extension of DECT from two to multiple energies by the use of photon counting detectors with energy discrimination, termed spectral CT, has been explored by investigators [13]. These systems currently exist as prototypes for pre-clinical investigation [14] and it may be necessary to further improve detector technology to compete with dual source systems [15]. An overview of the current DECT scanners is given in Supplemental Table 1.

For routine diagnostic purposes there is a strong demand for low-dose imaging and major steps in dose reduction have been achieved over the past years. One needs to keep in mind that for RT purposes the dose acquired from all imaging procedures including image-guided RT using multiple cone beam CT imaging is typically only a fraction of the dose that is delivered by the actual therapy later on [16,17]. Furthermore, with DECT imaging the dose is actually divided over the two energies used for the acquisition leading to dose levels that are approximately equal to a single energy scan [18] and acquisition of DECT images do not necessarily increase the imaging dose to the patient. However, optimization of the balance between dose burden (ALARA principle) and imaging quality should be evaluated in the RT setting which opens possibilities to optimize the acquisition protocol for the specific RT goals maximizing a benefit-to-risk ratio [19,20].

Photon attenuation coefficient decomposition

X-ray CT images are three dimensional reconstructions of the effective photon linear attenuation coefficient $\bar{\mu}$. Because of the strong energy E dependence of the linear attenuation coefficient μ and due to the poly-energetic spectra employed in X-ray CT imaging, the effective $\bar{\mu}$ is usually calculated as the sum of the contributions from photo-electric effect, Compton scattering and Rayleigh scattering to the attenuation. Whereas the photo-electric component of μ has a strong dependence on the atomic number Z, Compton scattering is governed solely by the electron density ρ_e. The attenuation coefficient allows parameterization [21], that typically can be done as functions of an effective atomic number Z_{eff} and ρ_e [3,22], as initially proposed by Hounsfield in his seminal article [2]. An example of such parameterization for $Z_{\text{eff}} < 20$ is:

$$\mu(E) = \frac{c\rho_e Z_{\text{eff}}^a}{E^b} + \sigma_{\text{KN}}(E)\rho_e + \frac{e\rho_e Z_{\text{eff}}^d}{E^f} \quad (1)$$

where a–f are best fit material and energy independent parameters and σ_{KN} is the Klein–Nishina cross section for Compton scattering [23]. While Z_{eff} and ρ_e parameterizations have an intuitive physical interpretation, any two basis functions may be employed to decompose μ. Investigators proposed to use μ of two arbitrary materials as basis functions [24,25], for example water and calcium [25]. The energy dependence of μ for the basis materials water and iodine, which are used in many applications discussed further in this review article, is illustrated in Fig. 1 together with two special purpose DECT polychromatic spectra that are produced by the X-ray tubes.

Early on, DECT basis material decomposition algorithms were divided into two main categories: projection based [3] and image based [26]. While projection based methods intrinsically correct for beam hardening effects, they require data consistency (i.e. same ray paths) between the high and low energy projections [27]. This requirement severely limits the applicability of the method to sequential scanning (due to motion) or to dual-source helical scanning where two measurements of the exact same ray-paths may not be available. Thus the image-based formalism has been employed more frequently in modern DECT implementations and we will focus on these studies in this review.

Radiotherapy interest in DECT was first raised by Devic et al. for brachytherapy applications [28]. This initial investigation estimated the linear attenuation coefficient at low photon energies from DECT; that was followed by a thorough comparison of Z_{eff} and ρ_e based parameterization to the basis material method, where the latter was found preferable for brachytherapy in terms of photon cross section estimation accuracy [23]. Other pioneering work on decomposition was based on parameterization of tabulated μ/ρ values of Bazalova et al. [29–31], which is an extension of the mono-energetic synchrotron X-ray work [32–34] to polyenergetic X-ray spectra, but requires knowledge of the X-ray spectra employed [35].

Following these initial publications several algorithms to decompose DECT images into Z_{eff} and ρ_e have been proposed [36–40]. Of practical interest are calibration-based methods where a phantom with known Z_{eff} and ρ_e in the range of human tissues is scanned and fit parameters calculated which can be used to obtain

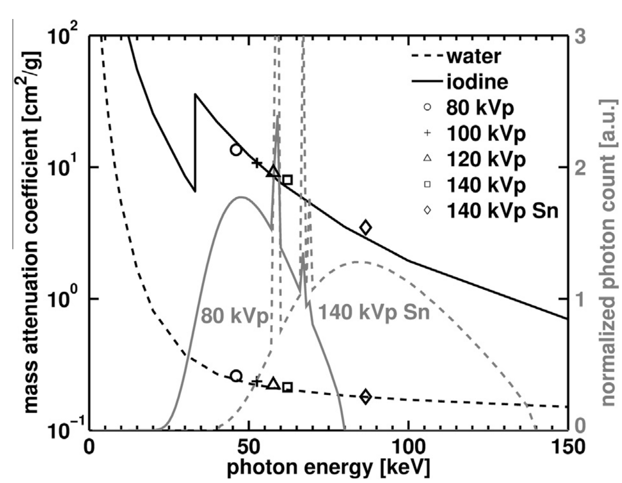

Fig. 1. Photon mass attenuation coefficients in the diagnostic imaging energy range from the NIST XCOM database [102] for the basis materials water and iodine indicated by the black dashed and solid lines, respectively. Symbols indicate the average values expected from typical X-ray spectra with various kVp generated with SpekCalc [103], plotted as a function of mean photon energy. 140 kVp Sn is filtered by tin for better spectral separation in DECT. Two typical X-ray spectra for 80 kVp and 140 kVp Sn are also shown as solid and dashed gray lines, respectively [85].

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