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Original paper

Objective assessment of low-contrast computed tomography images with iterative reconstruction

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ABSTRACT

Objective: This study aims to assess low-contrast image quality using a low-contrast object specific contrast-to-noise ratio (CNR_{LO}) analysis for iterative reconstruction (IR) computed tomography (CT) images.
Methods: A phantom composed of low-contrast rods placed in a uniform material was used in this study. Images were reconstructed using filtered back projection (FBP) and IR (Adaptive Iterative Dose Reduction 3D). Scans were performed at six dose levels: 1.0, 1.8, 3.1, 4.6, 7.1 and 13.3 mGy. Objective image quality was assessed by comparing CNR_{LO} with CNR using a human observer test.
Results: Compared with FBP, IR yielded increased CNR at the same dose levels. The results of CNR_{LO} and observer tests showed similarities or only marginal differences between FBP and IR at the same dose levels. The coefficient of determination for CNR_{LO} was significantly better ($R^2 = 0.86$) than that of CNR ($R^2 = 0.47$).
Conclusion: For IR, CNR_{LO} could potentially serve as an objective index reflective of a human observer assessment. The results of CNR_{LO} test indicated that the IR algorithm was not superior to FBP in terms of low-contrast detectability at the same radiation doses.

1. Introduction

The technical advances in multi-detector computed tomography (CT) radiation dose reduction, which have included the installation of various types of hardware and applications intended to reduce radiation doses during examinations on the latest CT scanners [1,2], have been remarkable. More recently, CT manufacturers have developed iterative reconstruction (IR) algorithms that enable radiation dose reduction while maintaining an image quality comparable to that achieved using conventional filtered back projection (FBP). IR has since attracted the attention of radiologists and physicists, and has been the subject of many clinical and physical studies [3–8].

As IR images are designed based on non-linear processes [9], the frequency properties of such images are known to differ from those of FBP images. IR indicates the adaptive changes of image properties, which depend on the radiation dose and object contrast. Therefore, in spatial resolution assessments of IR images, results obtained using conventional standard measurement methods do not necessarily reflect the image resolution characteristics. To solve this problem, several new modulation transfer function (MTF) measurement methods that account for the levels of noise and contrast in images have been proposed [4,6].

The contrast-to-noise ratio (CNR) has been widely used to quantify low-contrast detectability in CT images [3,10–12]. CNR is a simple and highly quantitative evaluation technique and has thus been used in several previous studies to conduct low-contrast detectability assessments of IR images. However, as the processes used to calculate CNR do not include the image frequency components, this parameter is invalid for evaluations of IR technology. Schindera et al. described low-contrast detectability in Adaptive Iterative Dose Reduction 3D (AIDR 3D; Toshiba Medical

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Systems, Tokyo, Japan) abdominal CT images obtained using a liver phantom [11,12]. These authors noted that, although AIDR 3D maintained or improved the quantitative image quality (image noise and CNR), the low-contrast detectability was not necessarily maintained. Their results indicated that CNR could not reflect the physical characteristics of the IR images.

In previous studies of CT image assessment, several mathematical model observer tests have been presented to investigate the influence of low-contrast detectability [13–15]. It has been recognized that mathematical model observer tests such as the non-pre-whitening model (NPW) are alternative methods by which it is possible to objectively assess the image quality. In the present study, we propose a quantitative low-contrast assessment metric that corresponds to the diameter of the target object and image noise property (low-contrast object specific CNR; $\mathrm{CNR_{LO}}$). $\mathrm{CNR_{LO}}$ assessment might reflect both image frequency characteristics and the frequency components corresponding to lesion size. $\mathrm{CNR_{LO}}$ is not only a useful objective image quality assessment method for IR images but can also yield results corresponding to human observer assessments, such as other NPW models. This study aims to assess the low-contrast image quality yielded by a $\mathrm{CNR_{LO}}$ analysis of CT images generated via the IR algorithm.

2. Materials and methods

2.1. Measurement theory of low-contrast object specific CNR

With a view to quantitatively evaluate the low contrast detectability of IR images, we propose a CNR analysis metric that corresponds to the diameters of the low contrast objects and image noise property ($\mathrm{CNR_{LO}}$). In a previous study, Loo et al. [16], reported an index value derived from the frequency characteristic of the signal and the spatial frequency component of the images, showing a comparable correlation with the visual detectability of the nylon beads in radiographs. In their study, the signal spectrum of a low-contrast object was derived with the film gradient at the reference density, the object Fourier spectrum, and MTF of the imaging system. Meanwhile, Diameter of the low-contrast tumour targeted in the clinical CT studies are not so small. For example, the hepatic cellular carcinoma minimum size that detected in clinical CT studies are 8-mm and more [17,18]. In addition, in the low dose lung CT screening, 5-mm thresholds for follow-up are indicated in several guidelines. Though the contrast of the ground-glass opacity is not low contrast (>100 HU), its appearance is similar to low-contrast object because the noise level is markedly high due to the very low doses [19]. Therefore, in the $\mathrm{CNR_{LO}}$ assessment, low-contrast objects and the noise level corresponding to the spatial frequency obtained with mean square root bandwidth $\bar{u}$ calculation for each object size were included. Since the images of the tumour-simulating rods were circular, their frequency components $S(u)$ can be expressed by the following equation, which uses the Bessel function of the first kind of the first order:

$$S(u) = \frac{J_1(\pi d u)}{2\pi d u}, \tag{1}$$

where J_1 ( ) is a first-order Bessel function of the first order and d is the target rod diameter (mm) and u denote the frequency. In our study, cross-sectional images of the simulated tumours are also circular, and the spatial frequency components of the signal can be calculated from Eq. (1).

To determine the spatial frequency components ($\bar{u}$) corresponding to the diameter of the target, the mean-square-root bandwidth, or $\bar{u}$, of a target rod $S(u)$ was defined by the following equation [20]:

$$\bar{u}^2 = \frac{\int_0^\infty u^2 |S(u)|^2 du}{\int_0^\infty |S(u)|^2 du}. \tag{2}$$

where $\bar{u}$ was defined as the most contributing spatial frequency for detectability corresponding to the diameter of the target lesion.

$\mathrm{CNR_{LO}}$ can be calculated from the following equation, which incorporates the noise power spectrum (NPS):

$$CNR_{LO}(\bar{u}) = \frac{ROI_M - ROI_B}{\sqrt{NPS(\bar{u})}}, \tag{3}$$

where ROI_M and ROI_B are CT values measured in the rod and background regions of interest (ROI), respectively, and $NPS(\bar{u})$ denotes the NPS at the spatial frequency ($\bar{u}$). $\mathrm{CNR_{LO}}$ is an index that reflects contrast of the lesion, image frequency characteristics and the frequency components corresponding to lesion size. $NPS(\bar{u})$ represents the amount of noise at the spatial frequency that involves detection of the target lesion.

2.2. Phantom

The low-contrast phantom was composed of cylindrical rods with diameters ranging from 3 to 10 mm implanted in a uniform material with an overall diameter of 200 mm (Fig. 1). The CT values of the background region and cylinder rods were 50 and 35–45 Hounsfield units (HU), respectively. Accordingly, the contrast between the background and rods was approximately 5–15 HU, equivalent to the contrast between a tumour and the liver parenchyma in a plain abdominal CT [21]. The phantom was placed at the isocentre with its cross section parallel to the in-plane. In the present study, we assessed two rods with different diameters and contrasts (a 5.0-mm-diameter rod with a 15-HU contrast and a 7.0-mm-diameter rod with a 10-HU contrast). We analysed the relationship between the human observer test and the objective quantitative image evaluations.

2.3. CT scanner and data acquisitions

All phantom data were acquired with an area detector CT scanner (Aquilion ONE ViSION Edition; Toshiba Medical Systems, Tokyo, Japan). Images were reconstructed using FBP and AIDR 3D, an iterative algorithm for image noise reduction. AIDR 3D uses an algorithm in both raw data and image data domains to reduce the image noise caused by the radiation dose reduction while preserving or even improving the spatial resolution and structural edges [22]. Moreover, AIDR 3D incorporates four noise reduction levels: weak, mild, standard and strong. Acquired image data were output in the Digital Imaging and Communication in Medicine

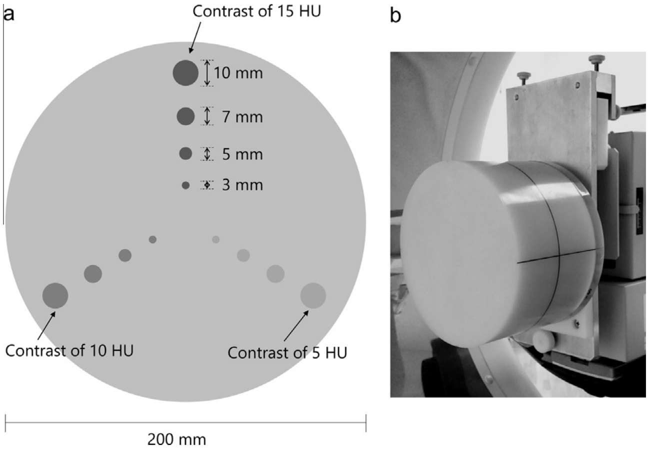

Fig. 1. Schema of the phantom used for analysis. (a) Cross section of the phantom indicating the diameters. (b) Photograph of the phantom.