

Feasibility of low-radiation-dose CT for abdominal examinations with hybrid iterative reconstruction algorithm: low-contrast phantom study

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Abstract Our purpose in this study was to evaluate the image quality of low-radiation-dose CT using hybrid iterative reconstruction (HIR), and to compare the results with those of filtered back projection (FBP) at routine doses. We measured the mean values and standard deviation of the CT numbers within and outside a 15-mm low-contrast object cylinder at 1.0 % contrast level. The noise reduction levels of the HIR were 1 (weak) to 7 (strong). Visual inspection of the low-contrast detectability was done by six radiologic technologists. The low-contrast detectability of the cylinder at the 1.0 % contrast level with HIR at all mAs levels was equal to that obtained with FBP, and thus the use of HIR did not result in any improvement of low-contrast detectability.

Keywords CT · Iterative reconstruction (IR) · Low-contrast detectability · Hybrid iterative reconstruction (HIR) · Filtered back projection (FBP)

1 Introduction

The ALARA (as-low-as-reasonably-achievable) principle is the primary challenge in modern CT (computed tomography) technology. Funama et al. [1] reported that low-contrast detectability was an important factor in abdominal CT, and especially for hepatic CT; however, reduction of the radiation dose in a CT examination causes a degradation of low-contrast detectability due to increased image noise compared to that obtained with a routine radiation dose. One of the proven solutions for noise reduction is the iterative reconstruction (IR) algorithm; however, the IR algorithm requires greater computer power and longer reconstruction times compared to the conventional filtered back projection (FBP) algorithm.

IR algorithms have been used in some CT systems with faster computers and hardware implementation [2–5]. A hybrid iterative reconstruction algorithm (HIR) reduces noise in both the raw data space and the image data space; therefore, the HIR allows low-radiation-dose CT examinations without degradation of low-contrast detectability [6–9]. Other groups have compared low-radiation-dose CT examinations using HIR with routine-dose CT examinations using FBP in clinical studies with a few dose reductions [10–12], but, to our knowledge, no investigators have published reports of low-contrast detectability measurement in low-radiation-dose CT using various dose reductions in a phantom study. Murazaki et al. [13] reported a decrease in the CT number with HIR compared to FBP in their coronary-calcification phantom study, but another investigator

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reported that the use of HIR did not result in such CT number changes compared to FBP [2]. Our purpose in the present study was to determine the feasibility of HIR for low-radiation-dose CT as compared with routine-dose abdominal CT examinations with FBP by evaluating the influence of CT number, contrast, noise, and low-contrast detectability in a phantom study.

2 Materials and methods

2.1 Hybrid iterative reconstruction algorithm

The HIR algorithm was available in the Philips CT system called the “iDose⁴” iterative reconstruction technique. The HIR algorithm is based on a two-step image-quality improvement process: the first step is optimizing the image quality and artifact prevention in the projection space, and the second step is model-based noise removal and resolution improvement in the image space [14]. There are seven noise reduction levels—from weak (level 1) to strong (level 7)—which are selectable by the users.

2.2 Phantom scanning and image reconstruction

A phantom (Catphan 600 Phantom; Phantom Laboratory, Salem, NY, USA) was scanned ten times for each current-time product (mAs) by 256-slice multidetector CT (Brilliance iCT; Philips Healthcare, Cleveland, OH, USA). The scanning parameters included a rotation time of 500 ms; 128×0.625 mm collimation; tube voltage, 120 kV; and 10, 20, 30, 40, 50, 60, 70, 80, 90, and 100 mAs; the corresponding calculated CT dose index was 0.6, 1.3, 1.9, 2.6, 3.2, 3.8, 4.5, 5.1, 5.8, and 6.4 mGy; small focal spot; pitch, 0.586. The images were reconstructed by FBP and by all seven levels of HIR (i.e., the iDose⁴) with the following reconstruction parameters: slice thickness, 5.0 mm; reconstruction kernel, standard (B); image matrix size, 512×512

pixels; display field of view, 20.0 cm. We used 100 mAs with FBP as the routine dose.

2.3 Measurement of CT numbers, contrast, and noise

A section on the phantom with varying sizes of cylinders in a circular configuration with a uniform-density background in the phantom was used for measurements of the CT number, contrast, and noise. The size of the region of interest (ROI) was 2.0 cm^2 . The attenuation of the signal was measured in a 15-mm diameter low-contrast cylinder at the 1.0 % contrast level, and the background was measured 18 mm outside of the signal ROI (Fig. 1a). The mean CT numbers were measured for both the signal and the background, and the contrast was calculated as the signal minus the background. For noise measurement, we used the same ROI as that used for the CT numbers. The SD (standard deviation) values of the background were measured as the noise.

2.4 Visual inspection of low-contrast detectability

The images reconstructed with FBP and all seven levels of HIR were evaluated by six radiologic technologists who agreed to participate in this study and who specialized in CT with an average of 10 years of experience (range 2–27 years). The evaluators inspected 800 image sets (10 times scanned with each 10–100 mAs value with FBP and all seven levels of HIR), with the CT window level set at 80 Hounsfield units (HU) and the width at 80 HU. The diameters of the low-contrast cylinder at the 1.0 % contrast level were 15, 9, 8, 7, 6, 5, 4, 3, and 2 mm (Fig. 1b). Each evaluator recorded the detectable minimum diameter of the low-contrast cylinder at the 1.0 % contrast level. The low-contrast detectability for each mAs value was the median of the diameters reported by the six evaluators. The color liquid crystal display monitor used was a Flex Scan S1910 (Eizo Co., Ishikawa, Japan) with a spatial resolution of $1,280 \times 1,024$ pixels.

Fig. 1 **a** ROIs' locations and **b** phantom featuring. ROI_S is the signal area, and ROI_{BG} is the background area. ROI region of interest

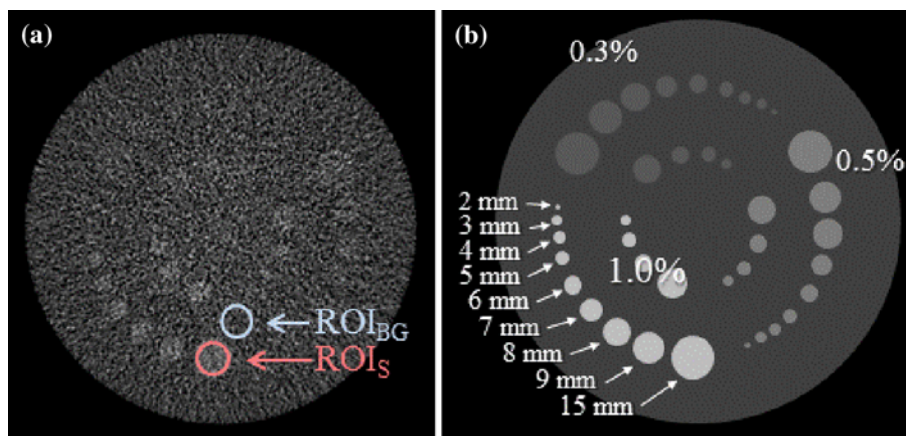


Fig. 2 Differences in **a** signal, **b** background, and **c** contrast with FBP and HIR at all seven noise reduction levels of the HIR. *FBP* filtered back projection, and *HIR* hybrid iterative reconstruction

2.5 Statistical analysis

The comparisons of the signals, the backgrounds, and the contrasts with HIR and 100 mAs with FBP were performed by use of Dunnett’s test. The statistical analyses were performed with commercially available software (JMP, version 8.0, SAS Institute, Cary, NC). Data are expressed as mean ± SD. For all statistical analyses, a *p* value of <0.05 was considered to indicate significance.

3 Results

3.1 Influence of dose reduction on CT number, contrast, and noise

3.1.1 CT number of signal

The CT number of the signal at 100 mAs with FBP was 69.8 ± 0.5 HU. Significantly increased CT numbers of the signal compared to 100 mAs with FBP were observed at 10 and 30 mAs with FBP (10 and 30 mAs: 71.3 ± 1.8 and 70.6 ± 0.8 HU; $p < 0.0001$ and $p = 0.0172$, respectively) and 10 mAs with HIR at level 1 (70.5 ± 1.5 HU; $p = 0.0247$; Fig. 2a). A significantly decreased CT number of the signal compared to 100 mAs with FBP was observed at 50 mAs at level 7 (68.8 ± 0.6 HU; $p = 0.0480$).

3.2 CT number of background

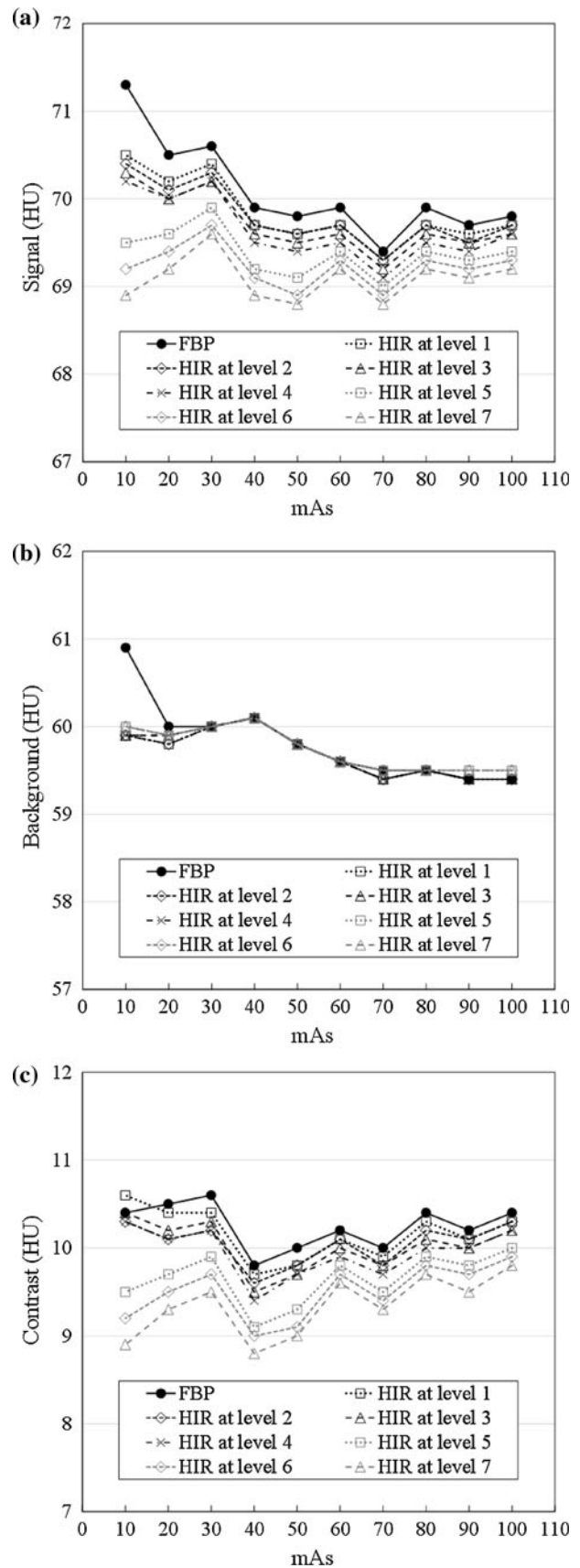
The CT number of the background at 100 mAs with FBP was 59.4 ± 0.3 HU. A significantly increased CT number of the background was seen at 10 mAs with FBP (60.9 ± 1.2 HU; $p < 0.0001$; Fig. 2b).

3.2.1 Contrast

The contrast at 100 mAs with FBP was 10.4 ± 0.3 HU. Significantly decreased contrast values compared to 100 mAs with FBP occurred at 50 mAs with HIR at level 6 (9.1 ± 0.7 HU; $p = 0.0453$) and at 20, 40, and 50 mAs with HIR at level 7 (20, 40, and 50 mAs: 9.3 ± 1.2 , 8.8 ± 0.7 , and 9.0 ± 0.7 HU; $p = 0.0434$, $p = 0.0224$, and $p < 0.0113$, respectively; Fig. 2c).

3.2.2 Noise

The noise at 100 mAs with FBP was 9.7 ± 1.0 HU. The noises with HIR were smaller than those obtained with



FBP (Fig. 3). The mAs levels with HIR that had the same noise compared with 100 mAs with FBP are listed in Table 1.

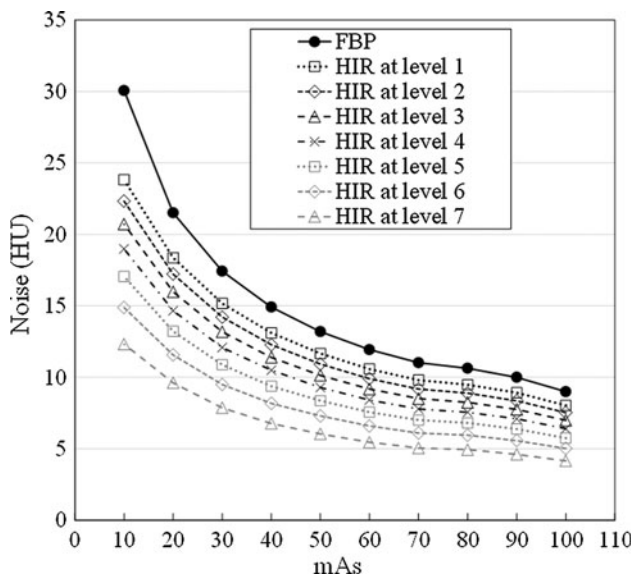


Fig. 3 Noise with FBP and HIR at all seven noise reduction levels of the HIR

Table 1 The mAs and the level of HIR that had the same noise compared with 100 mAs with FBP

mAs	HIR level
90	1
80	2
70	3
60	4
50	5
40	6
30	7

3.3 Low-contrast detectability

The detectable minimum diameters of the low-contrast cylinder at the 1.0 % contrast level were equivalent between FBP and HIR at all mAs (Table 2). The phantom images of 100 mAs with FBP and 60 mAs with FBP and all seven levels of HIR are shown in Fig. 4 with the CT window level set at 80 HU and the width at 80 HU.

4 Discussion

CT requires a higher radiation dose than do other X-ray examinations, and therefore dose-reduction techniques for CT have received much attention [15–17]; reduction of the CT radiation dose is an important goal.

The CT numbers of the signal were increased at lower than 30 mAs with FBP compared to 100 mAs with FBP. The cause of the increasing CT numbers of the signal was photon starvation, called “bias artifacts” by Scibelli [14]; therefore, the increase was the greatest at 10 mAs. An increase in the CT number of the background with FBP was observed at 10 mAs. Photon starvation occurs more easily at the signal of the cylinder because of the photon numbers detected compared to the background on the phantom. The use of HIR corrected the CT number increase at 10–30 mAs, but it simultaneously decreased the CT number at the mAs levels. It was therefore apparent that HIR did not contribute to the correction of the photon starvation.

The CT numbers of the signal with HIR were decreased depending on the level of HIR. We could not obtain effective information about the algorithm of HIR from the CT manufacturer, and we were therefore unable to determine the reason for the decrease in the CT numbers.

The decrease of the CT number of the signal reduced the contrast at mAs values lower than 50 with HIR at levels 5–7. Awai et al. [18] reported that there were some cases in

Table 2 The detectable minimum diameters of the low-contrast cylinder at the 1.0 % contrast level with FBP and HIR

Reconstruction type	Detectable minimum diameter (mm)									
	10 mAs	20 mAs	30 mAs	40 mAs	50 mAs	60 mAs	70 mAs	80 mAs	90 mAs	100 mAs
FBP	15	8	7	6	5	4	4	4	4	4
HIR at level 1	15	8	7	6	5	4	4	4	4	4
HIR at level 2	15	8	7	6	5	4	4	4	4	4
HIR at level 3	15	8	7	6	5	4	4	4	4	4
HIR at level 4	15	8	7	6	5	4	4	4	4	4
HIR at level 5	15	8	7	6	5	4	4	4	4	4
HIR at level 6	15	8	7	6	5	4	4	4	4	4
HIR at level 7	15	8	7	6	5	4	4	4	4	4

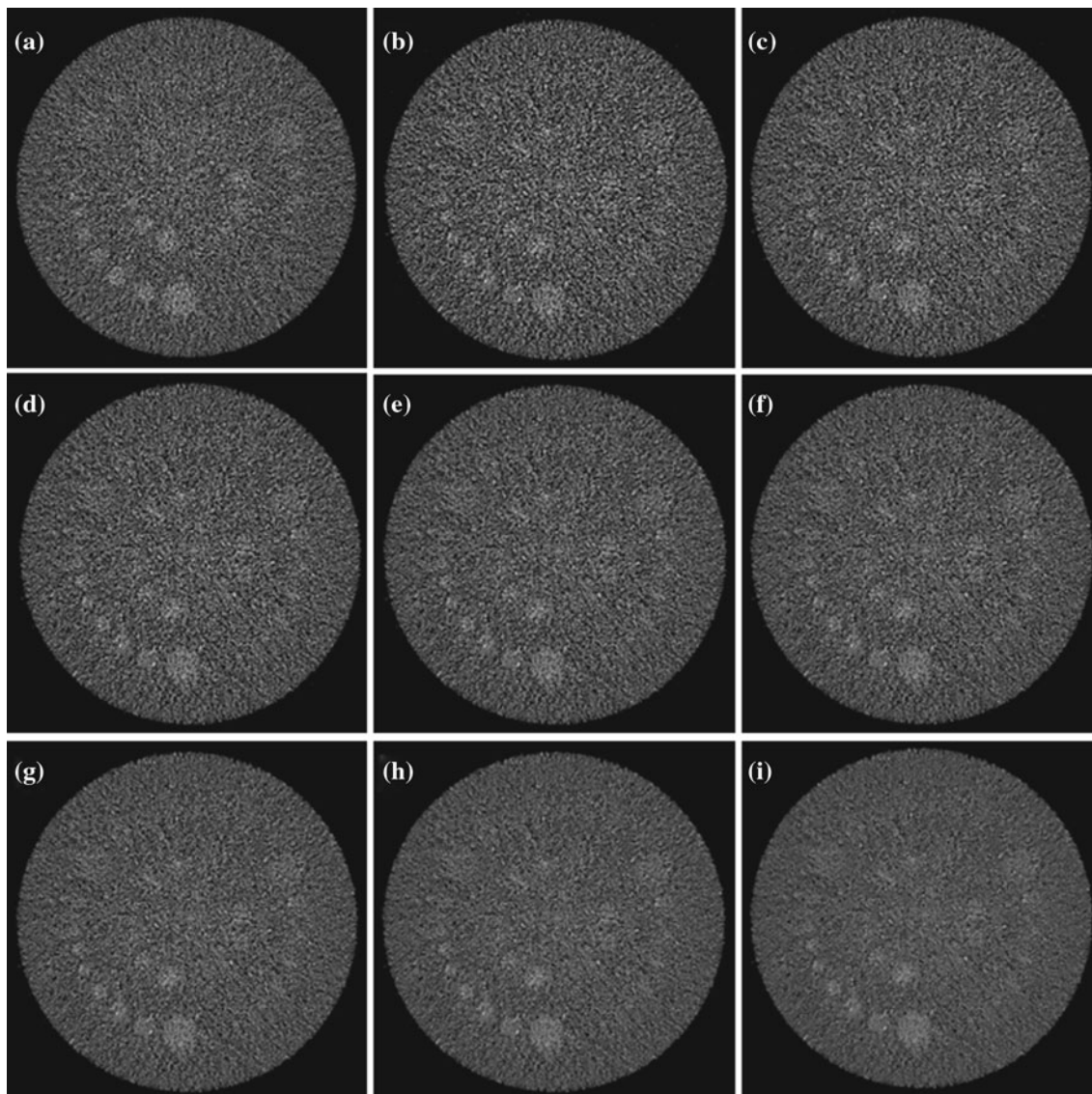


Fig. 4 Phantom images: **a** 100 mAs with FBP, and 60 mAs with **b** FBP; **c** HIR at level 1; **d** HIR at level 2; **e** HIR at level 3; **f** HIR at level 4; **g** HIR at level 5; **h** HIR at level 6; and **i** HIR at level 7

which the tumor-to-liver contrast was just a few HU in enhanced CT, which suggests that it would be appropriate to choose an HIR level lower than 4 to prevent the significant decrease in the CT numbers of the signal at under 50 mAs.

HIR decreases noise with maintained structural edges by iteration processes, but HIR does not correct a structure destroyed by noise. Consequently, the low-contrast detectability of HIR was no improvement compared to that of FBP. Kihara et al. [19] reported that the use of HIR provided an improvement in low-contrast detectability compared to FBP, but Takata et al. [20] concluded that other commercial HIR algorithms did not provide any significant improvement in low-contrast detectability. The reason(s) for these inconsistent findings may be resolved

by further studies of low-contrast detectability as examined here, or by more extensive low-contrast detectability studies.

Our phantom investigation had the following limitations. Because it was an investigation of low-contrast detectability in phantom images, spatial resolution was not involved, and clinical studies are required for the confirmation of our findings. In addition, because the phantom size was fixed, the influence of the beam-hardening effect when the object size changes was not considered. Only the standard reconstruction kernel for an abdominal examination was used; the findings may not apply to a head CT examination. The evaluators' knowledge of the cylinders' sizes and locations might have created a bias in the visual inspection [21]. However, Takata et al. [20] reported no

improvement of low-contrast detectability when other commercial HIR algorithms were used; therefore, such bias in our study may not have affected the results obtained.

We conclude that the use of HIR did not result in an improvement of low-contrast detectability with 1.0 % contrast at all mAs levels. However, HIR was effective for correcting the CT number increase caused by photon starvation at extremely low-dose levels, whereas the indicated CT number and contrast decreased at HIR levels 5–7.

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