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ORIGINAL PAPER

CT iterative reconstruction in image space: A phantom study

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Abstract Although iterative reconstruction is widely applied in SPECT/PET, its introduction in clinical CT is quite recent, in the past the demand for extensive computer power and long image reconstruction times have stopped the diffusion of this technique. Recently Iterative Reconstruction in Image Space (IRIS) has been introduced on Siemens top CT scanners. This recon method works on image data area, reducing the time-consuming loops on raw data and noise removal is obtained in subsequent iterative steps with a smoothing process. We evaluated image noise, low contrast resolution, CT number linearity and accuracy, transverse and z-axis spatial resolution using some dedicated phantoms in single, dual source and cardiac mode. We reconstructed images with a traditional filtered back-projection algorithm and with IRIS. The iterative procedure preserves spatial resolution, CT number accuracy and linearity moreover decreases image noise. These preliminary results support the idea that dose reduction with preserved image quality is possible with IRIS, even if studies on patients are necessary to confirm these data.

Introduction

The current reconstruction method available in CT application is filtered back projection (FBP), but recently new methods, based on iterative reconstruction algorithms, have been introduced [1,15].

In classic iterative reconstruction the CT scanner should be modelled mathematically [3], the correction loops directly on raw data and the reconstruction process is repeated many times, as a consequence iterative approach is much slower than analytical methods, but now increasing computation power and simplified models introduced in commercial software allow its use in clinical practice.

Since the second quarter of 2010 on Siemens CT Somatom Definition product family is available an iterative algorithm [2] named IRIS (Iterative Reconstruction in Image Space) that uses an initial FBP algorithm to reconstruct the images, in a second moment noise is removed in iterative steps: this process should preserve the spatial resolution obtained during the first traditional reconstruction cycle [4]. The

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idea is to decouple the spatial resolution, that is substantially determined by FPB, and noise. Spatial transverse/coronal resolution, CT number accuracy and linearity are supposed to be determined by a master image obtained during the first step with a conventional analytical method, the subsequent iterations in image domain have the unique aim to reduce noise by the application of a prior that smooths homogeneous regions inside the image [14].

This article explores, from a physical point of view, the main features of IRIS on phantoms images through the evaluation of classic CT image quality parameters as spatial and low contrast resolution, CT number accuracy and linearity, water CT number and noise.

The purpose of our measurements was to verify that spatial resolution, CT number accuracy and linearity are the same with IRIS and FBP while noise is reduced by iterative reconstruction.

Materials and methods

We performed our measurements on a Somatom Definition Flash (Siemens Healthcare), equipped with two independent x-ray tubes, each of them coupled with a 64 rows detectors array [5]. On this scanner IRIS algorithm is available in all protocols, including high pitch acquisitions, and can be selected during the reconstruction procedure: different types of IRIS convolution kernels are available in the same list of conventional filters. We acquired some phantoms normally used to investigate image quality in CT with single, dual source and cardiac protocols. All details on physical parameters investigated, scan and reconstruction data and phantom used are reported in Table 1.

We scanned different modules of a Catphan 500 phantom (The Phantom Laboratory, Salem, NY, USA) [6] to

Table 1 Details on physical parameters investigated, scan and reconstruction data and phantom used.

Physical parameter investigated	Scan parameters	Reconstruction parameters	Phantom used
Transverse spatial resolution (Fig. 2)	120 kV, 200 mAs, 1.0 s rotation time, 32 $\times$ 1.2 mm collimation, 5 mm slice thickness, pitch 1.0	FBP: B70s IRIS: I70s	Catphan 500, high resolution module
Low contrast resolution (Fig. 3)	120 kV, 340/170 mAs, 1.0 s rotation time, 128 $\times$ 0.6 mm collimation, 5 mm slice thickness, pitch 1.0	FBP: B31s IRIS: I31s	Catphan 500, low contrast resolution module
CT number linearity and CT number accuracy (Table 2)	120 kV, 200 mAs, 1.0 s rotation time, 32 $\times$ 1.2 mm collimation, 5 mm slice thickness, pitch 1.0	FBP:B40s IRIS: I40s	Catphan 500, sensitometry module
Coronal spatial resolution (Fig. 4)	120 kV, 500 mAs, 1.0 s rotation time, 128 $\times$ 0.6 mm collimation, pitch 1.0	FBP: B46f IRIS: I46f MPR: 1 mm recon thickness, 0.1 mm image increment	3D Spatial Resolution Phantom
Coronal spatial resolution in cardiac DS protocol (Fig. 5)	120 kV, 250 mA for each tube, 0.28 s rotation time, 128 $\times$ 0.6 mm collimation, artificial ECG-gated signal of 75 bpm, pitch 0.28	FBP: B46f IRIS: I46f MPR: 1 mm recon thickness, 0.1 mm image incrementsingle segment reconstruction 75 ms temporal resolution	3D Spatial Resolution Phantom
Water mean CT value and standard deviation, radial noise power spectrum (NPS) in Single Source protocol (Table 3, Fig. 6)	120 kV, 140/70 mAs, 1.0 s rotation time, 128 $\times$ 0.6 mm collimation, 0.6 mm slice thickness, pitch 1.0	FBP:B40s IRIS: I40s	30 cm diameter water-filled phantom
Water mean CT value and standard deviation in Dual Source protocol for obese patients (Table 3)	120 kV, 70 mAs for each tube, 1.0 s rotation time, 32 $\times$ 0.6 mm collimation, 0.6 mm slice thickness, pitch 1.0	FBP:B40s IRIS: I40s	30 cm diameter water-filled phantom
Water mean CT value and standard deviation in cardiac Dual Source protocol (Table 3)	120 kV, 70 mA for each tube, 0.28 s rotation time, 128 $\times$ 0.6 mm collimation, 1 mm slice thickness, artificial ECG-gated signal of 75 bpm, pitch 0.28	FBP: B40f IRIS: I40f single segment reconstruction 75 ms temporal resolution	30 cm diameter water-filled phantom

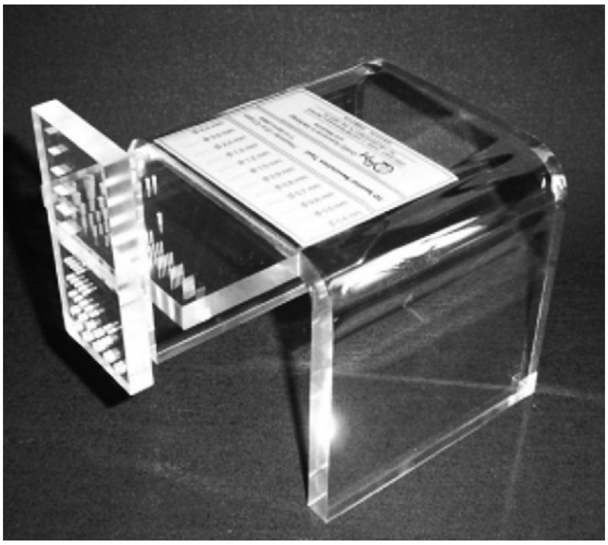

Figure 1 3D Spatial Resolution Phantom, QRM, Möhrendorf, Germany.

measure transverse spatial resolution, low contrast resolution, CT number linearity and CT number accuracy.

Transverse spatial resolution module is composed of bar patterns with different spatial frequencies that go from 1 to 21 lp/cm.

In low contrast resolution module we can find three areas with different nominal contras levels: 1%, 0.5% and 3%, in each contrast level there are targets with decreasing diameters (15, 9, 8, 7, 6, 5, 4, 3, 2 mm).

CT number linearity and CT number accuracy module has sensitometry targets made of teflon, delrin, acrylic, polystyrene, low density polyethylene (LDPE), polymethylpentene (PMP) and air in order to cover a wide range of electron densities.

We processed the images of each module with a conventional algorithm and with IRIS to outline possible differences in image quality.

We investigated the coronal spatial resolution using a dedicated phantom (3D Spatial Resolution Phantom, QRM, Möhrendorf, Germany) [7], with circular holes of varying diameter from 4.0 mm down to 0.4 mm aligned along the z-axis (Fig. 1). We scanned this phantom with a single source protocol and with a dual source protocol, commonly used in cardiac application to evaluate calcium scoring, with an artificial ECG trigger of 75 bpm. The images were reconstructed with FBP and with IRIS to compare the details phantom detection.

Image uniformity and noise were evaluated using a 30 cm diameter acrylic cylinder phantom, filled with water. We acquired the phantom in single source mode (Thorax Routine), in dual source mode with a protocol dedicated to obese patients (DS XXL Thorax) and in dual source mode with a cardiac protocol (DS CaScore) using again an artificial ECG signal. Also in this case images were processed using an FBP kernel and IRIS to investigate the noise behaviour under different scan and reconstruction conditions.

The images were evaluated with ImageJ 1.43u software (http://rsb.info.nih.gov/ij/).

Results

Fig. 2 (window width ww = 1 and window centre wc = 200) shows the high resolution module of Catphan 500 reconstructed with FBP and with IRIS, the maximum spatial frequency that can be visualized is 9 lp/cm for both images. The bars' visual appearance is not exactly the same, iterative reconstruction seems more blurry/blotchy: it could be explained looking at the difference in spatial frequency characteristics of image noise produced by different algorithms.

The results for the low contrast module are presented in Fig. 3 with ww = 80 and wc = 80: the fourth object with a nominal supra-slice contrast of 0.5% can be resolved in the FBP image obtained with 340 mAs, with IRIS reconstruction using 170 mAs is possible to detect the same detail, the image texture is quite similar between the two images even if FBP data are a bit less blurred and show sharpest edges.

The measured mean CT values in circular regions of interest (ROIs) positioned over the seven test objects of different materials in sensitometry module are reported in Table 2 for both reconstruction methods. These values confirm that in IRIS images CT numbers, for a big variety of materials, are the same obtained from FBP.

Linearity between CT number and nominal targets electron density is a basic requirement to use absolute CT numbers to separate tissue types, a linear fit between targets CT numbers and their electron densities has provided the same correlation coefficient (0.996) for FBP and IRIS data. Fig. 4 shows MPRs (ww = 1100 wc = 100) of the 3D Spatial Resolution Phantom acquired with a single source protocol and reconstructed with a 1 mm nominal slice thickness and a 0.1 mm image increment with FBP and with IRIS. For both reconstructions all the details down to 0.6 mm diameter can be resolved and geometrical

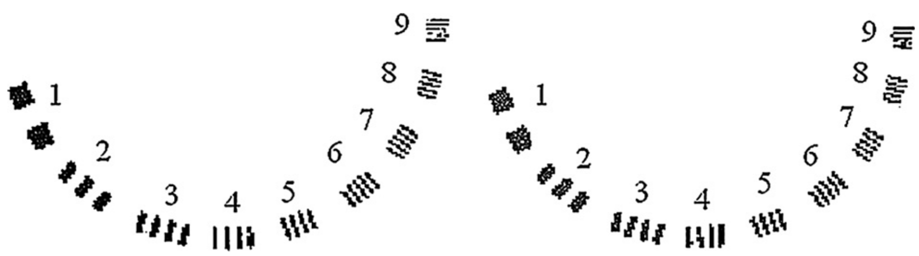

Figure 2 High resolution module of Catphan 500, left FBP reconstruction, right IRIS reconstruction.

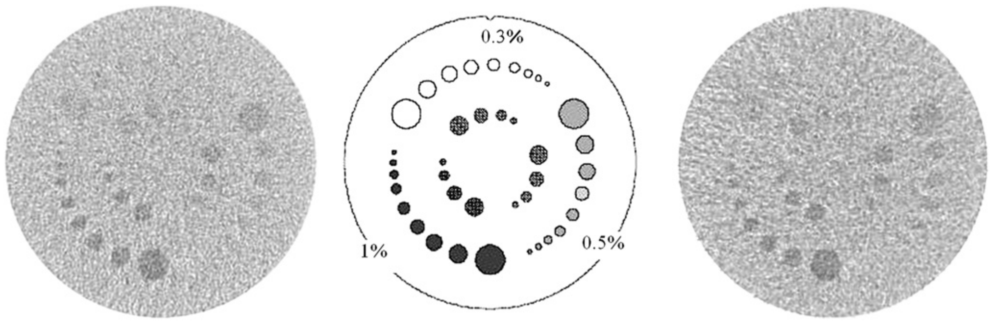

Figure 3 Low contrast module of Catphan 500, left FBP reconstruction (340 mAs), centre, phantom feature, right IRIS reconstruction at half dose (170 mAs).

Table 2 Data obtained with FBP and IRIS reconstruction in sensitometry module of Catphan 500.

	Air	PMP	LDPE	Polystyrene	Acrylic	Delrin	Teflon
Nominal CT number	−1000	−200	−100	−35	120	340	990
Measured CT number FBP (B40s)	−990.6	−179.5	−87.8	−33.3	123.3	342.2	929.7
Measured CT number IRIS(I40s)	−983.2	−177.4	−86.3	−32.4	122.3	342.1	925.1

distortions are not visible on images. We repeated the measurements with a dual source protocol used in cardiac applications with an artificial ECG-gated signal of 75 bpm (pitch 0.28). Fig. 5 (ww = 1100 wc = 100) shows the obtained MPRs images, *z*-axis spatial resolution is the same for both reconstruction methods.

All images collected during the water phantom acquisitions were analyzed evaluating image noise and water CT number in a central circular ROI and averaging data among 30 consecutive slices.

The results are presented in Table 3 for all the scans.

The mean CT number for water is constant in all modalities, but as expected the mean standard deviation of water CT number is always lower in IRIS reconstruction than in FBP one, even if with different percentage of reduction.

In single source mode the noise recovery obtained by iterative algorithm is more stressed: using 70 mAs and IRIS we can obtain the same noise value present in FBP images with 140 mAs. The extrapolation of these results in clinical images is not trivial due to the different nature of noise in phantoms and patients and should be verified in clinical practice [12,13].

To explore the frequency components of noise we processed the water-filled phantom images acquired in single source mode to obtain the radial noise power spectrum (NPS) in a central squared region of 20×20 cm^2 [8]. The results of this analysis are presented in Fig. 6 for both

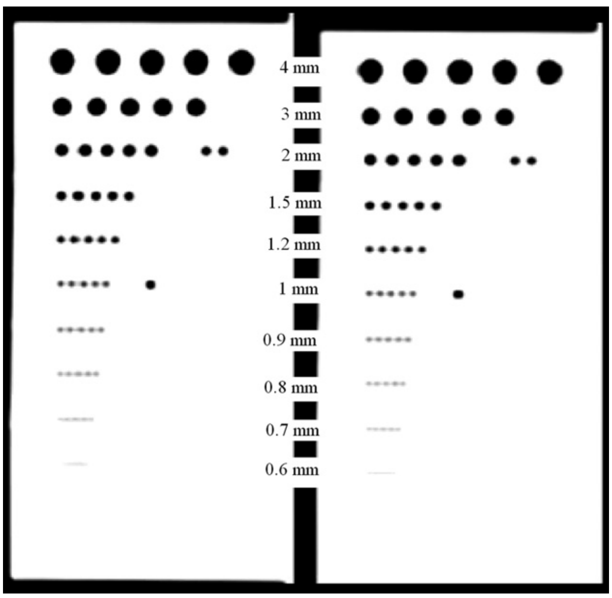

Figure 4 MPRs of the 3D Spatial Resolution Phantom in a single source protocol, left FBP reconstruction, right IRIS reconstruction.

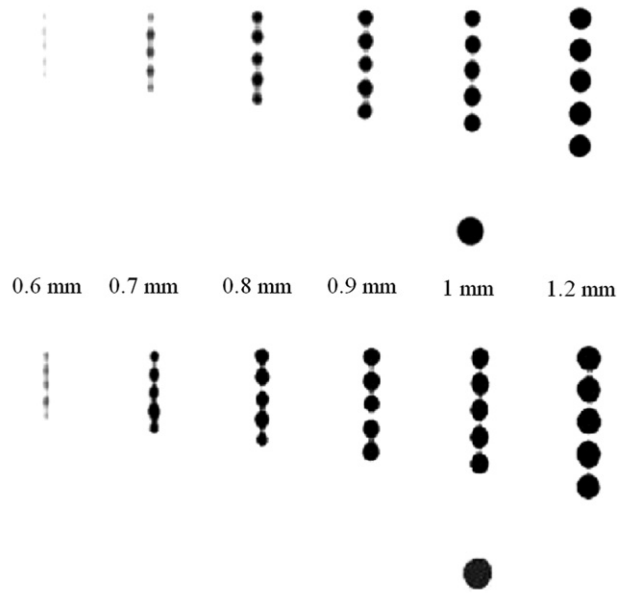

Figure 5 MPRs of the 3D Spatial Resolution Phantom in a dual source cardiac protocol, up FBP reconstruction, down IRIS reconstruction.

Table 3 Water CT number and noise obtained with FBP and IRIS reconstruction for different scan protocols in a 30 cm diameter water-filled phantom.

	Single Source			Dual Source XXL		Dual Source CaScore 75 bpm	
Recon Type	FBP 140 mAs	IRIS 140 mAs	IRIS 70 mAs	FBP	IRIS	FBP	IRIS
Mean CT number	0.95	1.23	1.77	−0.23	−0.20	−1.31	−0.21
Standard deviation CT number	64.7	39.9 (−38.4%)	58.9	49.1	33.2 (−32.3%)	108.1	90.1 (−16.7%)

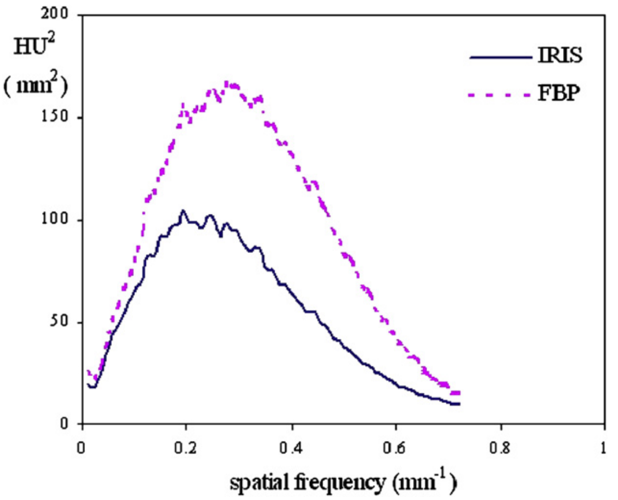

Figure 6 NPS curves for FBP and IRIS reconstruction for a single source acquisition protocol at 140 mAs.

reconstruction methods. The IRIS NPS data are always inferior to those obtained with FBP with an increase span at frequencies between 0.2 and 0.4 lp/mm. The general shape of NPS curves is the same, this confirm the visual impression of a similar image texture in both data sets.

Conclusions

Our phantom study on IRIS reconstruction method indicates that iterative algorithm preserves transverse and *z*-axis spatial resolution, CT number accuracy and linearity. IRIS decreases image noise in particular at frequencies between 0.2 and 0.4 lp/mm, so theoretically it allows to perform dose reductions without a significant loss in image quality. Dose control is a main topic in paediatric, follow-up and cardiac procedures and a lot of different approaches have been used in the effort to minimize radiation to patients [9—11], IRIS seems to be a promising tool to achieve this goal, even if further studies on samples of patients are essential to confirm these preliminary results [12—14].

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