Physica Medica 29 (2013) 684–694

Contents lists available at SciVerse ScienceDirect

Physica Medica

journal homepage: http://www.physicamedica.com

Original paper

Effects of computing parameters and measurement locations on the estimation of 3D NPS in non-stationary MDCT images

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A R T I C L E I N F O

Article history:
Received 19 December 2011
Received in revised form
25 June 2012
Accepted 1 July 2012
Available online 1 August 2012

Keywords:
Computed tomography
MDCT
CBCT
Noise stationarity
Noise power spectrum
Image quality
3D imaging
Measurement method

A B S T R A C T

The goal of this study was to investigate the impact of computing parameters and the location of volumes of interest (VOI) on the calculation of 3D noise power spectrum (NPS) in order to determine an optimal set of computing parameters and propose a robust method for evaluating the noise properties of imaging systems.

Noise stationarity in noise volumes acquired with a water phantom on a 128-MDCT and a 320-MDCT scanner were analyzed in the spatial domain in order to define locally stationary VOIs. The influence of the computing parameters in the 3D NPS measurement: the sampling distances $b_{x,y,z}$ and the VOI lengths $L_{x,y,z}$, the number of VOIs N_{VOI} and the structured noise were investigated to minimize measurement errors. The effect of the VOI locations on the NPS was also investigated.

Results showed that the noise (standard deviation) varies more in the *r*-direction (phantom radius) than *z*-direction plane. A $25 \times 25 \times 40$ mm^3 VOI associated with DFOV = 200 mm ($L_{x,y,z} = 64$, $b_{x,y} = 0.391$ mm with 512×512 matrix) and a first-order detrending method to reduce structured noise led to an accurate NPS estimation. NPS estimated from off centered small VOIs had a directional dependency contrary to NPS obtained from large VOIs located in the center of the volume or from small VOIs located on a concentric circle. This showed that the VOI size and location play a major role in the determination of NPS when images are not stationary.

This study emphasizes the need for consistent measurement methods to assess and compare image quality in CT.

Introduction

Fourier-based metrics have been established as the principal figures of merit to investigate and quantitatively evaluate the image quality performance of imaging systems [1]. Since the introduction of computed tomography (CT) scanners, the modulation transfer function (MTF) has been used to characterize the spatial resolution of CT systems [2–10] and the noise power spectrum (NPS) has been established as the gold standard to describe the noise frequency content in reconstructed data [4,7,8,11–13]. Although the pixel variance is still used to quantify the image noise in quality control programs [14], the NPS enables a more complete description of noise, taking into account the noise correlation induced by the reconstruction, filtering and processing steps. The NPS, associated with the MTF, provides the basis for an objective and quantitative evaluation of image quality and plays a central role in the estimation of task-based image quality through information transfer as detectability index. As recently shown in digital mammography and tomosynthesis [15–18], for well-defined conditions, such metrics can be used to validate cascaded systems analysis or to directly compute 2D/3D model observers based on a 2D/3D diagnostic task and predict the performance of CT systems. Although such tasks are more difficult to establish for multi-row detector CT (MDCT) examinations, the development of model observers based on 2D/3D Fourier metrics is an active area of ongoing clinical MDCT research.

To date, theoretical and experimental studies have been carried out to characterize and analyze the properties of 2D and 3D NPS for MDCT and prototype cone-beam (CB) installations, respectively

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http://dx.doi.org/10.1016/j.ejmp.2012.07.001

[4,7,19–27]. However, while several quantitative measurement techniques have been investigated and proposed for the standardized measurement of 2D and 3D MTF [4–8], very few studies have investigated the NPS measurement. Yang *et al.* measured the 2D and 3D NPS in CBCT for large images/volumes by dividing them into smaller overlapped regions of interest (ROIs) or volumes of interest (VOIs) [24]. On the other hand, Brunner *et al.* [8] estimated the noise from the entire image region divided into 32 × 32 non-overlapping ROIs that were separated by 8 pixels to ensure uncorrelated ROIs. Tward *et al.* showed that cone-beam systems were rarely stationary and that a local measurement should be performed in order to obtain accurate 3D NPS results [25–28]. Extraction using a small VOI, located along concentric circles to avoid regions with artifacts (i.e. at the center and edge of the noise volume) has been proposed [25]. However, Pelc *et al.* recently showed that the NPS extracted at different local regions in fan-beam and CBCT presented symmetric and asymmetric shapes as well as different directional dependencies that can produce different detectability [20–22].

Because the NPS is a fundamental metric, uncertainty in its measurement or inaccurate measurement methods may significantly affect dependent metrics, such as the noise equivalent quanta (NEQ), the detective quantum efficiency (DQE) and signal-to-noise-ratio (SNR), and could lead to inexact predictions of CT system performance as well as a biased diagnostic task-based detectability estimation [20–22]. The AAPM task group TG 169, created in 2007, has emphasized the importance of establishing a standardized method for the measurements of 3D metrics and especially NPS.

The goal of this study was first to investigate the impact of computing parameters on the calculation of the 3D NPS metric in order to determine a set of computing parameters that would minimize the bias and fluctuation errors. Because MDCT scanners are non-stationary systems, we first quantified the effect of the scan conditions on the noise stationarity. Then, the size of VOIs was determined in order to satisfy the local stationarity assumption criteria defined in this study, before calculating 3D NPS from these VOIs for various parameter combinations and quantified the errors on the 1D representation of the NPS. In the second part of this study, we computed local 3D NPS as a function of the location of the VOI in the scanned volume. The influences of the VOI location on the NPS were examined within the framework of fan-beam and CBCT system theory. Finally, our method was compared and discussed with respect to other published NPS methods.

Materials and methods

Phantom acquisitions and image reconstructions

A large number of experimental parameters can affect the degree of stationarity in CBCT such as geometry, scatter conditions (grid, air gap, and longitudinal FOV), detector type, pixel binning, reconstruction filters, voxel size, radiation dose level, use of a bowtie filter and phantom configuration [26,27].

In our study, acquisition and reconstruction conditions correspond to that of standard abdomen examinations. Scans were performed in sequential mode at 120 kVp on two commercial MDCT scanners: a 128-MDCT Brilliance iCT (Philips Healthcare, Cleveland, OH, USA) and a 320-MDCT Aquilion ONE (Toshiba Medical Systems, Nasu, Japan) with 128 × 0.625 mm and 320 × 0.5 mm detector collimations, respectively. A homemade cylindrical water phantom (28 cm long and 24 cm in diameter) with PMMA walls (5 mm thick) was placed at the isocenter of the CT gantry and scanned with a constant current, selected to give a CT dose index ($CTDI_{vol}$) equal to 13 mGy (Fig. 1). When necessary, identical acquisitions were performed without moving the CT table. Data sets were reconstructed with the standard reconstruction filters using a 512 × 512 matrix and various display fields of view (DFOV). It is worth noting that in this work all images were reconstructed with non-iterative methods. The acquisition and reconstruction parameters used for the analysis of the noise stationarity, the optimization of the NPS calculation and the measurement of the local NPS as a function of the location are summarized in Table 1. Then, CT data files (DICOM files) were transferred from the CT units to a standard desktop workstation and MATLAB R2011a (Mathworks, Natick, MA, USA) was used for analysis. In our study, a stack of reconstructed images (128 or 320 images) obtained from an axial acquisition was defined as “a volume of noise” and was utilized for the extraction of VOIs as illustrated in Fig. 2.

Noise stationary analysis in the spatial domain

To quantify the degree of noise stationarity, we simply measured the standard deviation (SD) along the radial or longitudinal direction (*r*- or *z*-direction) of a noise volume acquired from a single scan. The VOI size, equal to 10.2 × 10.2 × 10.0 mm^3, was small in comparison to the size of the noise volume (200 × 200 × 80 mm^3 and 200 × 200 × 160 mm^3 for 128-MDCT and 320-MDCT scanners, respectively) but large enough to have a sufficient number of voxels in order to minimize the statistical fluctuations of SD. With a 200 mm DFOV, a 512 × 512 matrix and a slice thickness of 0.625 mm or 0.5 mm, each VOI was formed of 26 × 26 × 16 or 26 × 26 × 20 voxels.

3D NPS calculation

3D NPS definition

Experimentally, the 3D NPS is computed from the 3D fast-Fourier transform (FFT) using the following equation:

$$\mathrm{NPS}\left(f_x,f_y,f_z\right) = \frac{b_x b_y b_z}{L_x L_y L_z}\cdot\frac{1}{N_{\mathrm{VOI}}}\cdot\sum_{i=1}^{N_{\mathrm{VOI}}}\left|\mathrm{FFT}\left\{\mathrm{VOI}_i(x,y,z) - \mathrm{VOI}_{\mathrm{background}\ i}(x,y,z)\right\}\right|^2\cdot\frac{1}{s} \tag{1}$$

where b_x, b_y and b_z are the sampling distances (in mm) in the *x*-, *y*-, and *z*-direction, respectively, and L_x, L_y and L_z are the VOI lengths in voxels in the *x*-, *y*-, and *z*-direction. N_{VOI} represents the number of

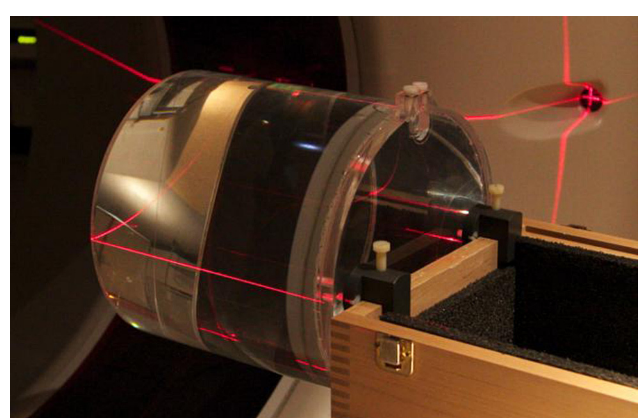

Figure 1. Cylindrical water phantom (28 cm long and 24 cm in diameter) used for this study. Acquisitions were performed at 120 kVp and 13 mGy on two different MDCT scanners (a 128- and a 320-MDCT scanner) using the axial acquisition mode.