A simple approach to measure computed tomography (CT) modulation transfer function (MTF) and noise-power spectrum (NPS) using the American College of Radiology (ACR) accreditation phantom

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Purpose: To develop an easily-implemented technique with free publicly-available analysis software to measure the modulation transfer function (MTF) and noise-power spectrum (NPS) of a clinical computed tomography (CT) system from images acquired using a widely-available and standardized American College of Radiology (ACR) CT accreditation phantom.

Methods: Images of the ACR phantom were acquired on a Siemens SOMATOM Definition Flash system using a standard adult head protocol: 120 kVp, 300 mAs, and reconstructed voxel size of $0.49 \text{ mm} \times 0.49 \text{ mm} \times 4.67 \text{ mm}$. The radial (axial) MTF was measured using an edge method where the boundary of the third module of the ACR phantom, originally designed to measure uniformity and noise, was used as a circular edge. The 3D NPS was measured using images from this same module and using a previously-described methodology that quantifies noise magnitude and 3D noise correlation.

Results: The axial MTF was radially symmetrical and had a value of 0.1 at 0.62 mm^{-1} . The 3D NPS shape was consistent with the filter-ramp function of filtered-backprojection reconstruction algorithms and previously reported values. The radial NPS peak value was $\sim 115 \text{ HU}^2 \text{ mm}^3$ at $\sim 0.25 \text{ mm}^{-1}$ and dropped to $0 \text{ HU}^2 \text{ mm}^3$ by 0.8 mm^{-1} .

Conclusions: The authors have developed an easily-implementable technique to measure the axial MTF and 3D NPS of clinical CT systems using an ACR phantom. The widespread availability of the phantom along with the free software the authors have provided will enable many different institutions to immediately measure MTF and NPS values for comparison of protocols and systems.
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Key words: computed tomography, CT, modulation transfer function, MTF, resolution, noise-power spectrum, NPS, noise, American College of Radiology phantom, ACR phantom

I. INTRODUCTION

The American College of Radiology (ACR) computed tomography (CT) accreditation program not only demonstrates commitment to image quality and patient safety but also benefits an institution's profile and reputation in the medical imaging community, and is required by some insurers, including Part B of Medicare. The ACR CT accreditation program is designed to be educational while evaluating personnel qualifications, scan parameters/techniques and performance, patient dose, and image quality. This process requires use of a specific phantom (model 464, Gammex-RMI, Middleton, WI), and image quality parameters assessed include high contrast spatial resolution (hereafter simply referred to as spatial resolution) and noise.¹ While the ACR program goals are commend-

able, spatial resolution measurements are restricted to simple qualitative evaluation of bar patterns, and noise measurements are limited to standard deviation measurements. The Fourier metrics of modulation transfer function (MTF) and noise-power spectrum (NPS), describing spatial resolution and noise, respectively, have become standardized for many x-ray based systems,²⁻⁴ replacing bar patterns and standard deviation measurements, and we hope to enable the same for CT.

The qualitative use of bar patterns to assess spatial resolution gained widespread use and remains common clinical practice⁵⁻⁷ due to the relatively easy and quick means of making the measurement. However, this technique is biased by observer subjectivity, and provides little information of system spatial resolution beyond a limiting value. The MTF is a metric describing resolution of linear and shift-invariant

FIG. 1. Two images with the same standard deviation but showing (a) high-frequency noise and (b) low-frequency noise content. The stark difference between the images illustrates the limitations of using standard deviation to fully assess image noise.

imaging systems in the Fourier domain, and the three-dimensional (3D) MTF, $\text{MTF}(u, v, w)$, is defined in terms of the 3D system point-spread function, $p(x, y, z)$,⁸

$$\text{MTF}(u, v, w) = |\mathcal{F}\{p(x, y, z)\}|, \quad (1)$$

where u , v , and w are spatial-frequency variables corresponding to spatial variables x , y , and z , respectively, $\mathcal{F}$ is the 3D Fourier transform operator, and $p(x, y, z)$ is normalized to unity volume. The MTFs of CT systems, calculated as early as 1976 by Judy,⁹ are used by many investigators^{10–13} to provide information across all spatial frequencies of interest.

The use of standard deviation to estimate CT image noise has gained similar widespread use^{5–7} due to the relatively easy and quick means of measurement. However, it provides a very limited description of noise magnitude, and two images with very different noise texture may have identical standard deviations, as illustrated in Fig. 1. The NPS describes noise in the Fourier domain for linear, shift-invariant systems with wide-sense stationary statistics. The 3D NPS, $\text{NPS}(u, v, w)$, is defined as the 3D Fourier transform of the system autocovariance function, $K(\tau_x, \tau_y, \tau_z)$,¹⁴

$$\text{NPS}(u, v, w) = \mathcal{F}\{K(\tau_x, \tau_y, \tau_z)\}, \quad (2)$$

where τ_x , τ_y , and τ_z represent the distance between voxels in x , y , and z directions, respectively. While calculation guidelines for CT NPS are still being formalized [American Association of Physicists in Medicine (AAPM) task group TG169], measurements were made as early as 1978 by Riederer *et al.*,¹⁵ and many investigators have described 3D CT NPS over recent years.^{12,16–18}

We demonstrate a novel and relatively straightforward method to make axial MTF and 3D NPS clinical CT measure-

ments using only two scans of a standard ACR phantom under similar parameters as already required for ACR accreditation. Widespread phantom availability and detailed parameter guidelines put forth for ACR accreditation would enable for easy CT system characterization, maintenance, and comparison between systems in different institutions based on MTF and NPS measurements.

II. METHODS

Data were collected on a Siemens CT system (SOMATOM Definition Flash) using the parameters outlined in Table I, corresponding to ACR values for an adult head scan. The system used 736 projection channels, 2304 projection views with angular flying spot, and 12 rows were acquired for each step-and-shot bed position of the axial scan, requiring 8 bed positions to cover the whole phantom. The ACR phantom (model 464, Gammex-RMI, Middleton, WI), shown in Fig. 2, was carefully centered in the scanner and properly aligned using the method described by McCollough *et al.* with aid of the accompanying stand.¹ This technique uses images of the wire pattern found in the first module of the phantom, obtained using a high resolution chest protocol with slice thickness < 2 cm, for alignment confirmation.¹ Only slices corresponding to the third (uniform) module of the phantom, intended for uniformity and noise measurements as well as distance accuracy and slice sensitivity profile (SSP), were used for all analyses. Slices containing partial volume effects with surrounding modules were excluded. This corresponded to a 3.27 cm region along the z direction. Two consecutive scans using each technique were acquired to enable data detrending for NPS calculation (see below).

TABLE I. Details of the computed tomography (CT) protocols used to image the American College of Radiology (ACR) phantom.

Protocol	Scan type	kVp	mAs	Scan FOV ^a (cm)	Reconstructed FOV ^a (cm)	Voxel size (mm ³)	Filter
Adult head	Axial	120	300	50	25	$0.49 \times 0.49 \times 4.67$	H30s

^aFOV = field of view

FIG. 2. An ACR CT accreditation phantom. Images of the third module, indicated by the * symbol, can be used to calculate MTF and NPS, describing resolution and noise, respectively.

All data were analyzed using custom-written software code under the MATLAB environment (Version R2010a, The MathWorks Inc., Natick, MA) and is freely available under a standard open-source software distribution license.

II.A. Modulation transfer function calculation

The MTF was calculated based on the edge method^{9,19} and adaptation of a more generalized 3D technique using a spherical edge phantom.^{10,12} The following steps were used in the technique:

1. The phantom slices selected for analysis were averaged together along the z direction on a voxel-by-voxel basis to create a single low-noise two-dimensional (2D) axial image. This also properly reduces the system point-spread function from 3D to 2D,

$$p(r, \theta) = \frac{1}{\Delta z} \int_{\Delta z} p(r, \theta, z) dz, \quad (3)$$

where $r = \sqrt{x^2 + y^2}$ represents the radial (axial) spatial variable, $\theta = \tan^{-1}(\frac{y}{x})$ is the corresponding polar coordinate angle, and Δz corresponds to the distance over which the slices are averaged and must be chosen such that it is greater than the correlation distance (i.e., correlation along the z direction). It is assumed that axial spatial resolution is isotropic along x and y directions [i.e., $p(r, \theta) = p(r)$], which is typically a good assumption.

2. Regions of unusable image data corresponding to ball bearings (BBs) in the phantom, Teflon rings from the phantom stand, the patient table, and nearby pixels,

were excluded from the analysis by creating a “mask” image of unusable pixels.

3. An open-field approach is used for MTF normalization,²⁰ where the 2D image data were mapped to the range of values [0,1], where 0 represented air and 1 represented the average pixel values of the uniform region of the phantom, on a pixel-by-pixel basis using the following formula:

$$x_{i,j}^{\text{norm}} = \frac{x_{i,j} - \bar{x}_{\text{air}}}{\bar{x}_{\text{phantom}} - \bar{x}_{\text{air}}}, \quad (4)$$

where $x_{i,j}$ represents the pixel at the (i,j) location, $\bar{x}_{\text{air}}$ represents the average pixel value corresponding to air, and $\bar{x}_{\text{phantom}}$ represents the average pixel value corresponding to the uniform phantom. Regions of interest (ROIs) of size 4.9 cm $\times$ 4.9 cm in the image corresponding to air and phantom were used to calculate $\bar{x}_{\text{air}}$ and $\bar{x}_{\text{phantom}}$, respectively.

4. Pixels corresponding to the cylindrical phantom were identified using thresholding,²¹ and the center determined by calculating the centroid of the circular region.
5. The oversampled edge-spread function (ESF) was calculated by mapping all points to their polar-coordinate equivalent and rebinning based on distance from the phantom’s center with a subvoxel bin size of 0.01 mm.¹⁹ This is equivalent to combining information from pixel values along isoradial contours of 0.01 mm in width, as illustrated in Fig. 3.
6. The ESF was differentiated to produce the corresponding line-spread function (LSF). Normalization of the LSF area to unity is not performed, as it has been previously shown to produce inflation and is not necessary for an open-field approach.²⁰
7. A Hann window was applied to the LSF prior to calculating the fast Fourier transform (FFT), and the radial (axial) MTF was calculated by taking the modulus of the FFT.

The resulting MTF was calculated *without* zero-frequency normalization to prevent inflation of all nonzero-frequency values.²⁰ Data from different angle ranges, including -6° to 6° , corresponding to the horizontal axis, 84° to 96° , corresponding to the vertical axis, and 0° to 360° , were used to calculate orientation-dependent MTFs and test the isotropic nature of the axial spatial resolution. As a second test of phantom alignment, the MTF was recalculated using only half of the image slices covering the third module, and compared to the other results.

II.B. Noise-power spectrum calculation

A 3D approach was taken to NPS calculation¹⁶ using the following steps:

1. Voxels corresponding to multiple 2D image slices of the third (uniform) module of the ACR phantom were identified and the center location of the phantom determined by identifying the phantom based on Hounsfield

FIG. 3. A schematic diagram of the third module of the cylindrical phantom, as seen in the axial image used for analysis that was obtained by averaging along the z direction. The transition between the phantom and air creates the edge used in the calculation, and the edge profile is obtained by looking at a single line from the center of the phantom outwards. The isotropic nature of the axial resolution enabled an oversampled radial edge-spread function to be formed by combining measurements along isoradial contours of the image, as indicated by the color and number coding between example points and radial contours.

- unit (HU) and using thresholding,²¹ and then calculating the centroid of this circular region, respectively.
2. The difference between the two acquired scans, both using the same given technique, was calculated on a voxel-by-voxel basis to produce a single detrended dataset in which the voxel values have a zero mean and only image noise is present. As this process also accounts for changes in voxels due to the BBs, data corresponding to the BBs were properly detrended and *not* excluded from the analysis. It should be noted that the same NPS results will be obtained regardless of which scan is subtracted from the other.

3. From the detrended dataset, ROIs were created, centered along a circle of radius 5 cm (half of the phantom radius), as illustrated in Fig. 4. Details of the ROI dimensions and arc length distance between ROIs are summarized in Table II.
4. The 3D FFT was taken of each detrended ROI *without* prior application of a window function (e.g., multiplication by the Hann function). The NPS was calculated as the ensemble average of the squared FFT values.
5. The calculated NPS values were divided by 2 to account for doubling of measured noise power caused by the subtraction operation of the two original scans.

FIG. 4. A schematic diagram of the third module of the cylindrical phantom as seen in a single axial image. The ROIs of size $\Delta x \times \Delta y \times \Delta z$ (see Table II for values) were centered on a circle with radius 5 cm. Only 10 3D ROIs are shown for clarity but 32 were used for calculations resulting in an overlap of $>50\%$ between ROIs (see Sec. IV). It is also important to note that the ROIs were taken from the dataset calculated from the *difference* between the two scans for each technique, and a schematic of the raw phantom images is shown simply to help visualize ROI locations.

TABLE II. ROI details used to calculate the NPS.

Voxel size (mm ³)	ROI size (cm ³)	ROI size (pixels ³)	Arc distance between ROIs (cm)	Arc distance between ROIs (voxels)	Total number
$0.49 \times 0.49 \times 4.67$	$6.27 \times 6.27 \times 3.27$	$128 \times 128 \times 7$	0.98	20	32

6. A radially-averaged axial NPS profile was generated using a similar rebinning technique as described above.

III. RESULTS

III.A. Modulation transfer function

Figure 5 shows the average axial image calculated from slices along the 3.27 cm length in the third module of the ACR phantom used to calculate the MTF. Regions excluded from the calculations are indicated by bounding boxes.

The corresponding MTFs as calculated using data from -6° to 6° , 84° to 96° , and 0° to 360° using the full third module 3.27 cm length, and 0° to 360° using only a 1.87 cm length of the module are shown in Fig. 6(a). Excellent agreement is found between all measured curves. The 10% value corresponds to 0.62 mm^{-1} and is an indication of the maximum detectable spatial frequency. This result agrees with the qualitative measurement obtained using the bar pattern in the second module of the ACR phantom, shown in Fig. 6(b), which indicates a maximum spatial resolution between 0.6 and 0.7 mm^{-1} .

FIG. 5. Averaged phantom image corresponding to the third module of the ACR phantom using a typical adult head protocol (see Table I). Regions enclosed inside the boxes in red correspond to ball-bearing locations, Teflon rings on the phantom stand, and the patient bed, and were not used in the MTF calculation. Image window and level have been adjusted to show noise within the cylinder.

III.B. Noise-power spectrum

The 3D NPS slices along $w = 0$, $v = 0$, and $u = 0$ are shown in Fig. 7. The corresponding radially averaged (axial) NPS along $w = 0$ is shown in Fig. 8(a) with the line profile along $u = 0$ and $v = 0$ shown in Fig. 8(b). It should be noted that the line profile uncertainty in the latter is expected to be

FIG. 6. (a) The axial modulation transfer function (MTF) as a function of spatial frequency using a typical adult head protocol (see Table I) and calculated using data from -6° to 6° (x -dir), 84° to 96° (y -dir), 0° to 360° (radial), and 0° to 360° but only using half the axial slices in the measurement (reduced ROI radial). The curve drops to a value of 0.1 at a frequency of 0.62 mm^{-1} . (b) ACR phantom bar pattern viewed using a window of 100 and optimized level of 1100. The patterns correspond to frequencies of (from top going clockwise): 1.2, 1.0, 0.9, 0.8, 0.7, 0.6, 0.5, and 0.4 mm^{-1} . An axial resolution (for example, the frequency at which the MTF falls to ~ 0.1) between 0.6 and 0.7 mm^{-1} is indicated.

FIG. 7. Three representative planes from the three-dimensional NPS as measured using images from the third module of an ACR phantom. Slices shown correspond to a typical adult head protocol (see Table I). The general shape of the NPS is consistent with that expected of filtered-backprojection reconstruction. Note that the resolution of the NPS in the z direction depends on the slice thickness and size of Δz in the ROIs used, and is expected to be worse than the radial (axial) direction due to the limited 4 cm width of the module.

larger as no radial averaging could be performed along that direction. Undersampling is also noted along the w axis due to the limited number of slices along z that could be included in the 3D ROIs used.

The axial NPS exhibits a filtered-ramp characteristic of filtered backprojection, increasing from ~ 40 to $\sim 115 \text{ HU}^2\text{mm}^3$ between 0 and $\sim 0.25 \text{ mm}^{-1}$ due to the ramp filter, and then decreasing down to $\sim 0 \text{ HU}^2\text{mm}^3$ by $\sim 0.8 \text{ mm}^{-1}$ due to

FIG. 8. One-dimensional profiles of the 3D NPS showing (a) a radial average along $w = 0$, and (b) a z -directional (w axis) profile of the NPS for a typical adult head protocol (see Table I). The number of points in the z -directional profile are limited by the 4 cm width of the phantom, and undersampling and aliasing are likely present.

low-pass smoothing filters. The NPS along the w axis decreases from a value of $\sim 35 \text{ HU}^2\text{mm}^3$ at 0 mm^{-3} to $\sim 25 \text{ HU}^2\text{mm}^3$ at 0.09 mm^{-3} .

IV. DISCUSSION

CT modulation transfer function and noise-power spectrum measurements can be made using images of a widely-distributed ACR phantom, a standard computer, and open-source MATLAB-based software, made freely available, with minimal effort and no additional equipment. The free, open-source nature of the code will enable easy portability to more robust programming languages, such as C, as well as community-based personalization and optimization, including a more friendly graphical user interface. Greater consistency is achieved between measurements if the same analysis code is used, and the ACR may wish to provide officially-sanctioned analysis software as part of their accreditation service if these techniques are adopted. Even if these metrics are deemed undesirable for ACR accreditation, the widespread availability, precise construction, and ACR phantom design make it ideal for more standardized CT measurement of radial MTF and NPS.

Use of the MTF and NPS require assumptions of shift invariance, linearity, and wide-sense stationary statistics. Rigorous satisfaction of these requirements is rarely achieved, and most systems either approximately satisfy the conditions, or do so when measured under specific conditions. Use of detector collimators (i.e., scatter rejection grid), bowtie filters, and even some aspects of image reconstruction may violate requirements of shift invariance within our system. Because of these violations, care must be taken when interpreting the results as they are likely only an approximation or an average of the true system properties. However, better MTF and NPS characteristics likely correspond to systems and/or techniques with better spatial resolution and noise properties. Thus measurements of MTF and 3D NPS made with the ACR phantom remain very useful for quality assurance, system comparison, and even system design. With greater system control, such as the ability to remove the bowtie filter, greater accuracy could be achieved between measured values and the system property being described. Recent advances in 3D image reconstruction methods (e.g., iterative and model-based) heighten this consideration further, and metrics of MTF and NPS deserve careful consideration with respect to linearity and stationarity of such new algorithms.

IV.A. Modulation transfer function

MTF determination of CT systems has been avoided by some investigators due to the difficulty of accurate edge alignment in three dimensions. Edge misalignment between slices can create an artificial blurring of the edge when calculating the low-noise averaged image, decreasing MTF measurement accuracy. Use of a sphere phantom has been suggested by some investigators,^{10,12} but requires a specialized new phantom. The described technique is an adapted 2D version of their 3D technique, enabling measurements along x and y

directions, and only requires the ACR phantom. The built-in features of the ACR phantom, including outer markings and wire patterns in the first module, enable quick and accurate alignment. This was tested on MTF results by varying the number of slices used when creating the averaged image. Agreement between all measurements made indicates phantom misalignment effects on MTF measurements were negligible over the distances tested.

MTF measurements using an ACR phantom and an edge technique have previously been made by Richard *et al.*¹³ They showed that the MTF is dose and contrast dependent when using adaptive statistical iterative and model-based iterative algorithms, but not filtered-backprojection algorithms, and thus introduce the concept of a task-based MTF. Their technique differs from the one presented in this paper in a few important ways. They use the first module of the phantom, relying on the disk images generated from the rod inserts, rather than the edge of the phantom in the third module. This necessarily limits the size of the region-of-interest that may be used for measurements. It has previously been shown that small ROIs can result in spectral leakage.²⁰ Their technique also relies on zero-frequency normalization, which has been shown to cause inflation of all nonzero-frequency values, especially when used in combination with small ROIs.²⁰ While the technique described in this paper would not show the dose and contrast dependence of the MTF, most clinical systems use a filtered-backprojection algorithm, which does not show this dependence. Those wishing to explore the MTF dependence could adapt our method and code, and make measurements of the first module as described by Richard *et al.*¹³

High-precision MTF determination was achieved by radial averaging and rebinning of image data, which relies on an assumption of radially isotropic spatial resolution. This was confirmed by calculating the MTF using data from varying angles and showing close agreement. Even if system spatial resolution is not isotropic, restricting measurements to angles close to the x and y axes would enable accurate and precise MTF measurements in those directions, as described elsewhere.¹⁰

Inaccurate calculation of the center of the phantom will result in a distance to the edge that changes with angle; if a point on the edge is closer by a small amount to the calculated center, then the corresponding point on the opposite side will be farther away by the same amount. This can create an artificial blurring of the edge, again decreasing MTF measurement accuracy. Close agreement between MTF calculations from varying angles indicates that any error caused by poor calculation of the phantom center is smaller than the uncertainty within the MTF measurements. Should accurate phantom center calculations not be possible, accurate MTF measurement is still possible. By restricting measurements to a small range of angles, all data used in the calculation will share approximately the same shift in edge position. It has previously been shown that MTF measurements are not affected by an error in edge position as long as the error is constant across all combined rows.²⁰ It should be noted that misalignment of the center of the phantom with the isocenter of the scanner is not expected to affect MTF accuracy, due

to the isotropic nature of the spatial resolution, and is actually required for measurement of the presampling MTF, as explained below.

It is the authors' professional assessment that the method described in this paper enables measurement of the presampling MTF. In order to prevent aliasing and obtain the presampling MTF, oversampling of the phantom, which is used to create our impulse response function, must be achieved at two important stages: in the projection data and the final reconstructed image. Placement of the phantom slightly offset from the system's isocenter, as will naturally occur even when aligning the phantom based on the technique described by McCollough *et al.*¹, ensures oversampling of the projection data. The slight offset results in the edge of the phantom being projected onto a different pixel location in each projection. This results in the familiar sine-wave pattern of the sinogram created from projection data and in this manner all "phases" of the impulse response function are obtained. In the final reconstructed image, oversampling is achieved using the edge method described here. Thus, what is measured is the presampling MTF of the entire imaging chain including system blurring processing and the reconstruction algorithm. However, a formal description with accompanying proof showing that aliasing is avoided at all image formation stages has yet to be given within the literature, and is an area of future research.

IV.B. Noise-power spectrum

It is known that the NPS shape can be ROI-location dependent,¹⁷ and radially symmetric (low-signal) streaking and ring artifacts are typically present in most images, indicating that the assumption of wide-sense stationary statistics is not completely satisfied. This was also observed in our measurements. To minimize this problem, the ROIs used to calculate the NPS were centered at a constant radius from the reconstruction isocenter and cover the full 360° of data where noise properties are approximately stationary.

An overlap between ROIs of <50% has traditionally been recommended when calculating the NPS.²⁻⁴ However, use of increased overlap is not expected to affect the accuracy of the calculated NPS, but may result in increased computation time if it results in the use of additional ROIs. The NPS was calculated using 4, 8, and 16 ROIs (not shown) in addition to the results presented using 32 ROIs, representing a range from minimal to very large ROI overlap. Good agreement was found between all curves and improved precision was noted when using 32 ROIs, thus these results were chosen for presentation.

Phantom alignment in the scanner is of limited importance for NPS measurements due to the uniform nature of the third module. Alignment in the z direction is only of importance to ensure that the selected 3D ROIs all remain within the region of the phantom. Alignment within the x,y plane is of importance due to the assumption that the phantom center corresponds to the reconstruction isocenter. While the results presented here indicate that the alignment technique described by McCollough *et al.*¹ is sufficient, should proper alignment

not be possible, determination of the isocenter should be calculated using an alternate method.

Inclusion of two BBs in the third module of the phantom causes a deviation from the phantom uniformity desired for the NPS measurement. While BB values are successfully removed during detrending, changes in noise properties in these voxels may produce artifacts in the NPS. However, these voxels only affect a small percentage of the ROIs, and even then only a small percentage of the voxels within these ROIs, and thus produce only a small effect. Attempts to exclude ROIs with voxels corresponding to the BBs may produce larger artifacts due to radial-symmetry loss of the CT data used.

The measured NPS curve shapes are consistent with those predicted for filtered-backprojection systems and previously reported clinical systems,^{15,17,18,22} showing the increasing values at lower frequencies consistent with a ramp filter, and the decrease of values at larger frequencies consistent with the apodization filter. It is known that NPS scaling is inversely proportional to the number of input x-ray quanta, which in turn is proportional to the radiation dose,¹⁷ emphasizing the need to measure the NPS under the clinical conditions of interest.

The nonzero z -direction NPS values at larger frequencies strongly suggest the presence of noise aliasing. Volumes acquired at different slice thickness, pitch, and levels of "z interpolation" in spiral (helical) acquisition modes are expected to exhibit different levels of correlation in the z direction. Thinner CT slices are expected to produce reduced aliasing in the NPS as well as better sampling along the z direction using the described technique. Thus, the conditions under which standardized NPS measurements are made should carefully be considered, and measurements using thinner slices are both likely desired and recommended by the authors. NPS dependencies, such as slice thickness and pitch, can be quantified fairly easily with the phantom and analysis software, and are areas of ongoing and future work.

IV.C. Three-dimensional approach

While it is tempting to treat CT data as a series of independent 2D axial images, such an approach will lead to inaccurate and even nonsensical results; CT data are fundamentally 3D volume datasets and thus need a proper 3D approach to MTF and NPS measurements, even if 2D axial results are only of interest. Thus, multiple slices of data *must* be averaged prior to calculation of axial MTFs. Similarly, 3D ROIs *must* be used for NPS calculations along with a 3D FFT. Attempts to analyze isolated 2D axial images (without appropriate normalization factor use) will result in incorrect measurements, since they ignore significant levels of out-of-plane correlation, and do not even yield the correct units.¹⁶ An analogy that aids in understanding is the erroneous method in 2D radiography in which one extracts 1D rows or columns from the 2D image to analyze the NPS; such an analysis ignores important correlations in the image and would not be expected to yield an accurate NPS.

The techniques described provide reproducible MTF and NPS measurements in the radial (axial) direction. However, no measurement of the z -direction (i.e., along the w axis) MTF is provided from the ESF approach described above. MTF calculation along the z direction would require a separate phantom, such as the spherical one previously mentioned.¹⁰ The fairly thick slices resulted in undersampling (aliasing) of the NPS in the z direction. Improved z -directional NPS sampling requires more image slices, which, in turn, would require the third module of the phantom to be longer or use of thinner slices.

The techniques described and the freely-available analysis code will help enable investigators and institutions to make more sophisticated and quantitative quality assurance measurements in day-to-day clinical physics practice. In addition, they will hopefully facilitate accurate and precise measurements of CT MTF and NPS that can be easily and readily used to assess CT system imaging characteristics within and between different institutions.

V. CONCLUSIONS

The spatial resolution and noise of CT images from a clinical Siemens SOMATOM Definition Flash system using a standard adult head protocol were measured based on described CT analysis techniques along with open-source analysis code to determine the radial MTF (providing a more comprehensive description of axial spatial resolution) and the 3D image NPS (providing a more comprehensive description of image noise) using the ACR accreditation phantom. The approach offers to transcend more simplistic and less informative metrics of bar patterns and standard deviation measurements.

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