International Journal of Progressive Sciences and Technologies (IJPSAT)
ISSN: 2509-0119.

https://ijpsat.org/
Vol. 55 No. 1 January 2026, pp. 197-205

Effect of Reconstruction Filter and Wire Diameter on Modulation Transfer Function Values in Various CT Image Fields

Ronel Arida Missinychrista*, Heri Sutanto*, Jatmiko Endro Suseno

Department of Physics, Faculty of Science and Mathematics, Diponegoro University, Jl Prof Jacob Rais, Tembalang, Semarang, 50275, Indonesia

*Corresponding Author: Ronel Arida Missinychrista, Heri Sutanto

E-mail: ronelaridamissinychrista@gmail.com, herisutanto@live.undip.ac.id

Abstract – This study analyses the effects of wire diameter and variations in reconstruction filters (bone and soft tissue) on the 3D Modulation Transfer Function (MTF) and the Oblique MTF. The reconstruction filter is a parameter that affects CT scans' ability to distinguish structural details, depending on the imaging purpose. The wire diameters used, namely 0,2 mm and 0,3 mm, provided differences to determine the effect of diameter on MTF values. The results showed that small-diameter wires yielded higher MTF 10% values (better resolution) in all planes when using bone filters. The results showed that the resulting MTF 10% values ranged from 0,61/mm to 1,22/mm and were above the tolerance limit for MTF 10% values, which is ≥ 0,5/mm. Furthermore, smaller-diameter wires produced higher MTF 10% values (better resolution) in all planes when using a bone filter. Conversely, with soft tissue filters, wire diameter affected only the sagittal and coronal planes. The slight differences in the sagittal and coronal planes are due to interpolation. The bone filter consistently improves image sharpness and MTF 10% values (including in the oblique direction), while the soft tissue filter reduces these values; however, overall, the MTF 10% values obtained meet the spatial resolution standards for clinical diagnosis.

Keywords: Bone, Modulation Transfer Function, Soft tissue, Wire.

I. INTRODUCTION

Currently, Computed Tomography (CT) is the primary modality in radiology installations used for diagnosis and interventional procedures such as biopsy or ablation [1]. CT utilises X-rays to produce images. However, CT is not immune to technical errors. Therefore, daily Quality Control (QC) is required to ensure that CT images are suitable for use [2]. The operator can use phantoms to perform daily QC to maintain CT quality. The operator must evaluate several parameters using phantoms, including noise, spatial resolution, artifacts, and CT linearity [2] .

Phantoms are made from specialised physical models and can serve as a QC medium [3]. Researchers have widely developed phantoms to evaluate CT images. One phantom used to evaluate CT image quality is the wire phantom, which serves to determine spatial resolution [4]. Spatial resolution is the ability of an imaging system to distinguish the smallest objects in two dimensions; the smaller the objects it can capture, the higher the resolution [4]. We describe spatial resolution by using the Modulation Transfer Function (MTF), which quantifies how system modulation changes with spatial frequency and defines the resolution limit at the 10% level [5]. MTF also describes the relationship between system modulation and spatial frequency. The tolerance limit for an MTF value of 10%, which is the limit of human vision, is ≥ 0.5/mm. We can determine the MTF in CT images using a wire phantom, and we must consider several important parameters, including phantom material, wire material, field of view (FOV), image reconstruction, slice thickness, focal spot, CT scan type, and image reconstruction algorithm. Then, the MTF curve can be generated from calculations using the Line Spread Function (LSF), Point Spread Function (PSF), and Edge Spread Function (ESF) [4].

Accurate MTF calculations require complex slice directions. In clinical practice, CT imaging is used to depict internal anatomical structures in 3D, with the x, y, and z axes. The problem occurs when the image slice direction is not parallel to the x, y, or z axis [6]. Additional geometric approaches can improve spatial direction coverage and spatial resolution estimation [7]. The addition of diagonal wires that intersect the phantom's centre (oblique) can represent more complex anatomical conditions that are not always aligned with the x, y, and z axes. The more complex the data representation of the MTF value, the more the resulting MTF value represents the actual anatomical condition [8].

Therefore, to test the MTF values produced in complex anatomical conditions that are not parallel to the x, y, and z axes, a wire phantom was created with additional diagonal wires that intersect the centre point of the phantom. It can provide a more complex representation of MTF values [8]. We used stainless steel wires with diameters of 0.2 mm and 0.3 mm to represent delicate anatomical structures in the human body, such as bone trabecula and thin cortical layers, and blood vessels, which are often the target objects during clinical CT examinations [9]. Then, it can determine and compare 10% MTF values under varying wire diameters and reconstruction filter types, namely, bone and soft tissue filters. The resulting MTF values will serve as a standard reference for CT images, used as a tool for clinical diagnosis and for viewing human anatomy at small diameters.

II. EXPERIMENTAL PROCEDURE

1. Materials

This study used several materials, namely acrylic as the primary material for phantoms equivalent to soft tissue, stainless steel wires with diameters of 0.2 mm and 0.3 mm as test objects, distilled water as the phantom filling fluid, and axial images from CT scans, which we used to calculate MTF values and analyse the effects of reconstruction filter variations and wire diameter. We performed the scans using a GE Revolution EVO 123-slice CT scanner.

2. Designs

In this study, we used four phantoms: the first and second contained 0.2 mm-diameter stainless steel wire, and the third and fourth contained 0.3 mm-diameter wire. We chose stainless steel because it provides accurate MTF measurements, and we often use it as a standard for image quality evaluation [10]. We used stainless steel wires with diameters of 0.2 mm and 0.3 mm to represent delicate anatomical structures in the human body, such as bone trabecula and thin cortical layers, and blood vessels, which are often the target objects during clinical CT examinations [9].

Figure 1a shows the first phantom design with a test object consisting of a 0,2 mm diameter stainless steel wire placed inside the phantom. This phantom was designed with three wires, positioned parallel to the principal axes (x, y, and z) to represent orthogonal directions. Then, in Figure 1b, four wires were used, with three wires positioned parallel to the x, y, and z axes, and one diagonal wire added to intersect the centre point of the phantom. This diagonal (oblique) wire represents anatomy that is not always parallel to the x, y, and z axes. Thus, a more complex and realistic representation of anatomical conditions will be obtained in the MTF calculation.

Figure 2b shows the second phantom design, with a test object consisting of 0,3-mm diameter stainless steel wires. This phantom has a similar design to the first phantom but differs in the diameter of the test-use wires, which are larger diameter. This difference in size was specifically designed to evaluate the effect of wire diameter variation on MTF measurements. By comparing the MTF values obtained from both phantoms, it is possible to analyse the extent to which object size can affect MTF values, both in the axial, sagittal, coronal, and oblique directions.

International Journal of Progressive Sciences and Technologies (IJPSAT)
ISSN: 2509-0119.
© 2026 Scholar AI LLC.
https://ijpsat.org/
Vol. 55 No. 1 January 2026, pp. 197-205

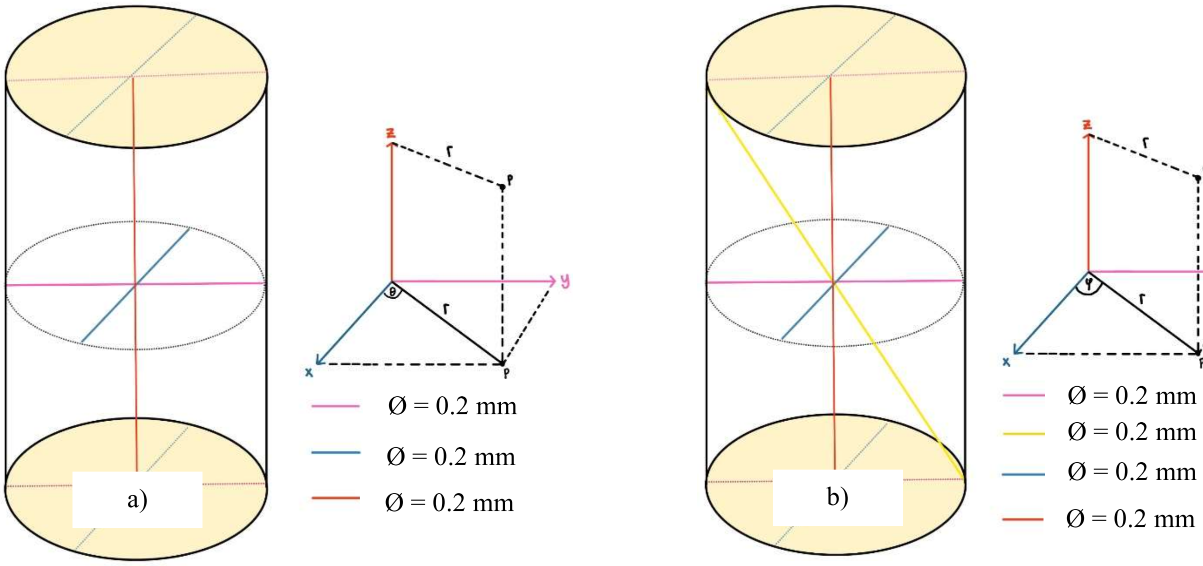

Fig. 1. Phantom Design with 0.2 mm diameter wire components on the (a) x, y, and z axes and on the (b) x, y, and z axes and the addition of diagonal wires

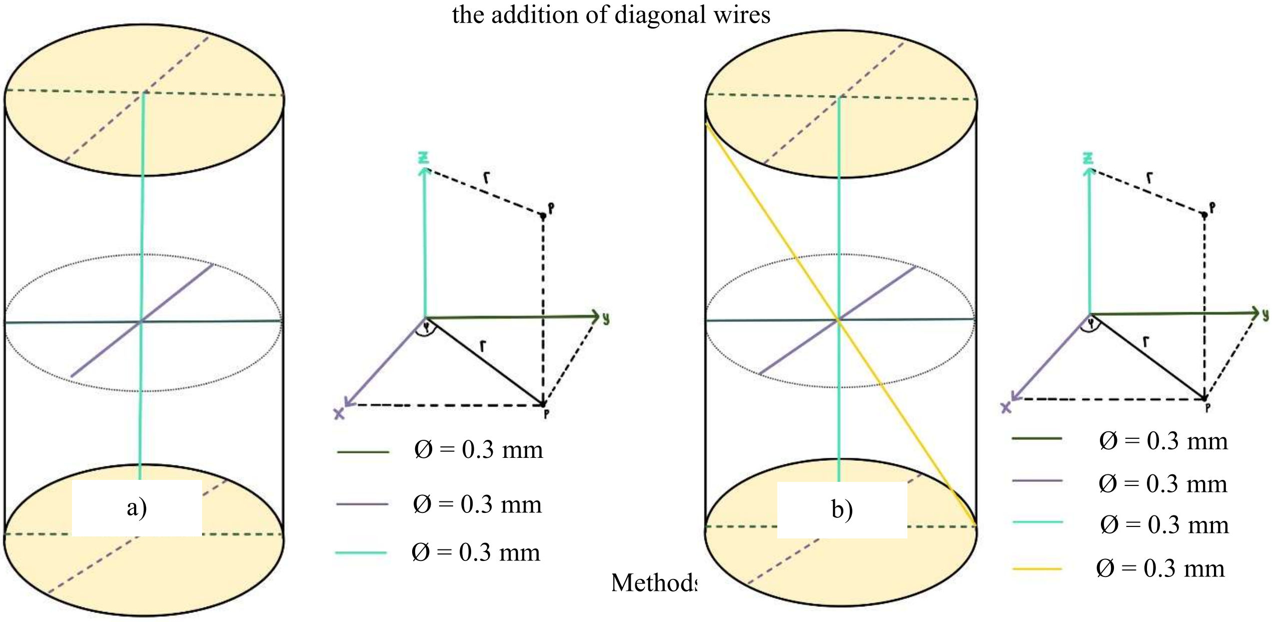

Fig. 2. Phantom Design with 0.3 mm diameter wire components on the (a) x, y, and z axes and on the (b) x, y, and z axes and the addition of diagonal wires

We performed phantom scanning with a 128-slice CT scanner, using settings specifically adjusted for image-quality analysis in MTF calculations. In this procedure, the only parameter changed was the reconstruction filter type, namely, soft and bone reconstruction filters. Conversely, other scanning parameters remained constant: the tube voltage, the tube current, the slice thickness, and the Field of View (FOV). All scans were performed in axial mode.

The axial image dataset was then reformatted using the IndoQCT software with the cubic interpolation method. We used axial slices in this process. We chose cubic interpolation because it produces smoother, more stable changes in pixel values, thereby reducing the potential for image distortion during rotation and reorientation [11]. 3D MTF measurements are automatically performed using the IndoQCT software. First, the slices used to measure each field are selected. The MTF values for the x and y axes are obtained from axial images, while the MTF values for the z axis are obtained from sagittal and coronal images.

We reformatted the axial images to produce sagittal and coronal images; we then used the two reformatted images to determine the MTF along the longitudinal axis [12]. To obtain oblique images, we performed reconstruction by first drawing lines perpendicular to the orientation of the diagonal wires on the coronal image plane. These lines serve as extraction paths for intensity profiles representing the oblique direction. Reconstruction along this perpendicular line produces an image (point-like image) that represents the system's response to the wire in the oblique direction. This point image is then used to form a spread function and., subsequently, to calculate the MTF oblique value.

III. RESULT AND DISCUSSION

Table. 1. MTF 10% Value of the Fantom Images

Phantom	Reconstruction filter	MTF 10% (mm^{-1})				
		Axial (x)	Axial (y)	Sagital (z)	Koronal (z)	Oblique
A. 0,2 mm Phantom without Diagonal Wire	Bone	1,17 ± 0,02	1,19 ± 0,03	0,78 ± 0,14	1,09 ± 0,03	-
	Soft tissue	0,62 ± 0,00	0,61 ± 0,01	0,65 ± 0,02	1,07 ± 0,01	-
B. 0,2 mm Phantom with Diagonal Wire	Bone	1,16 ± 0,02	1,19 ± 0,02	1,04 ± 0,05	1,06 ± 0,08	1,15 ± 0,00
	Soft tissue	0,61 ± 0,01	0,63 ± 0,01	0,98 ± 0,01	1,02 ± 0,01	0,62 ± 0,00
C. 0,3 mm Phantom without Diagonal Wire	Bone	1,09 ± 0,01	1,11 ± 0,01	0,77 ±0,08	1,01 ± 0,04	-
	Soft tissue	0,61 ± 0,00	0,61 ± 0,00	0,70 ± 0,01	1,02 ± 0,00	-
D. 0,3 mm Phantom with Diagonal Wire	Bone	1,10 ± 0,00	1,09 ± 0,01	0,90 ± 0,09	0,83 ± 0,08	1,06 ± 0,00
	Soft tissue	0,62 ± 0,02	0,61 ± 0,01	0,89 ± 0,09	0,89 ± 0,13	0,61 ± 0,00

Table 1 shows that overall, all phantoms produced MTF 10% values within the recommended standard range, which is ≥ 0,5/mm for each plane, including the axial (x), axial (y), sagittal (z), coronal (z), and oblique planes. The MTF 10% value is calculated to ensure that the CT system's spatial resolution meets the criteria for use in clinical assessment.

To ensure measurement accuracy, each MTF value presented in the table was calculated from five repeated measurements for each phantom configuration, reconstruction filter type, and image field. The values shown in the table are the mean ± standard deviation (SD) of these five measurements. These variations are due to sample noise, the interpolation process, and the reconstruction kernel.

The relatively small standard deviation values (ranging from 0,00 to 0,14) indicate that this method has good repeatability and low measurement variability. This confirms that MTF measurements in both orthogonal and oblique directions are stable. These results also support the conclusion that variations in MTF values are not caused by measurement instability, but rather by differences in wire diameter and the influence of reconstruction filters.

In the oblique direction, the difference in MTF 10% values is most apparent. With the bone filter, the 0,2 mm-diameter wire consistently provides higher MTF 10% values than the 0,3 mm wire across the entire field. Conversely, with the soft filter, the effect of wire diameter is only apparent in the sagittal and coronal planes. Evaluation of the oblique direction is critical because the orientation of anatomical structures is rarely parallel to the three principal axes. Hence, MTF performance in the oblique direction reflects a combination of lateral resolution in the axial plane and longitudinal resolution along the z-axis [13].

The effect of the reconstruction filter is most dominant in the axial (x-y) plane. The bone filter significantly increases the MTF 10% value by sharpening object edges, while the soft filter actually reduces the response at the high frequencies. However, when we view the image in the sagittal and coronal planes, the effect of the filter is almost imperceptible because these two planes are reformatted from the axial image and are more influenced by the detector size in the z-direction than by the reconstruction filter [14]. Figures 3 - 7 show graphs of MTF values in the axial (x), axial (y), sagittal (z), coronal (z), and oblique planes.

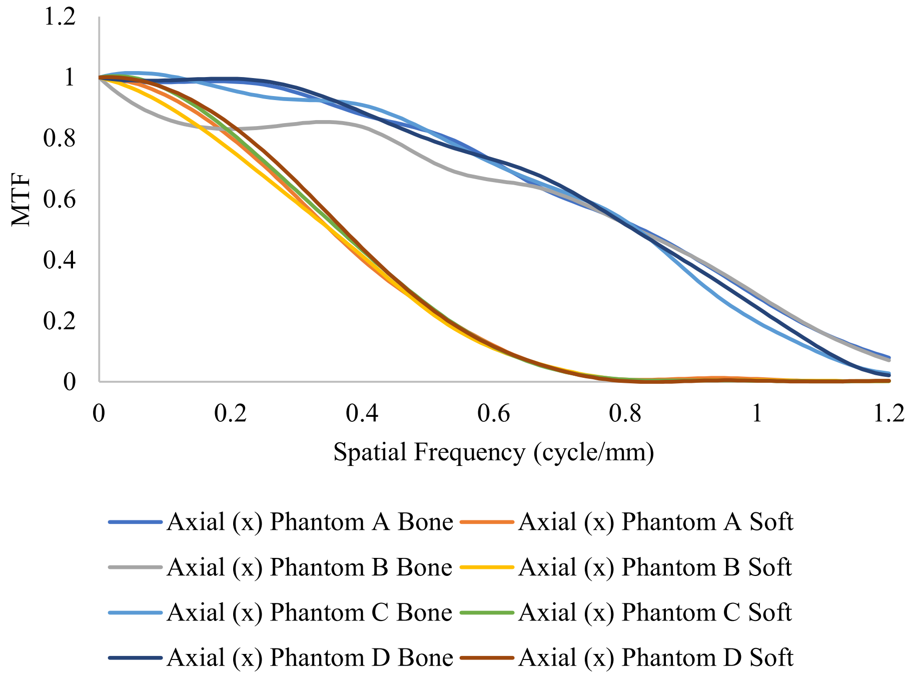

Fig. 3. MTF Value Graph on the Axial Plane (x)

International Journal of Progressive Sciences and Technologies (IJPSAT)
ISSN: 2509-0119.

https://ijpsat.org/
Vol. 55 No. 1 January 2026, pp. 197-205

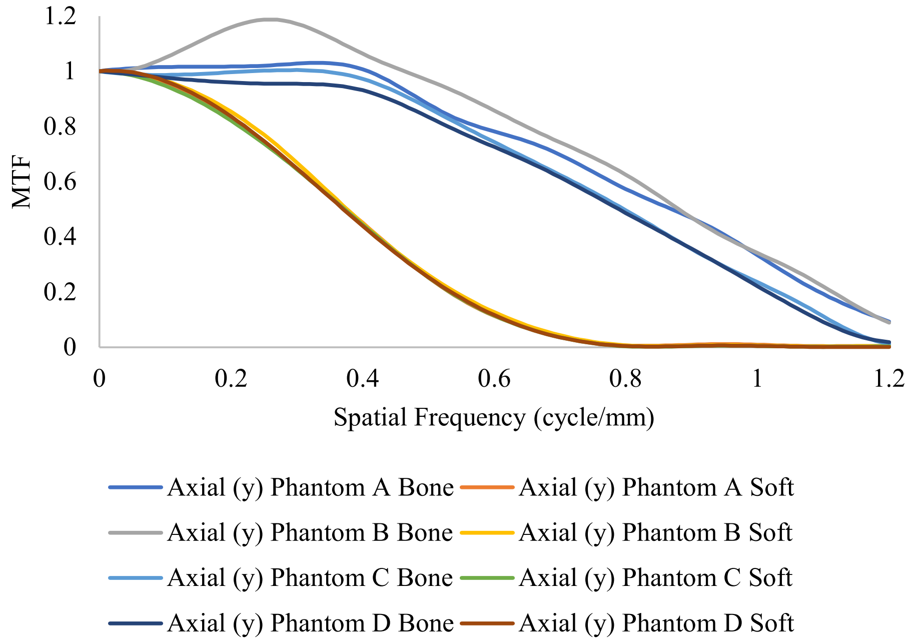

Fig. 4. MTF Value Graph on the Axial Plane (y)

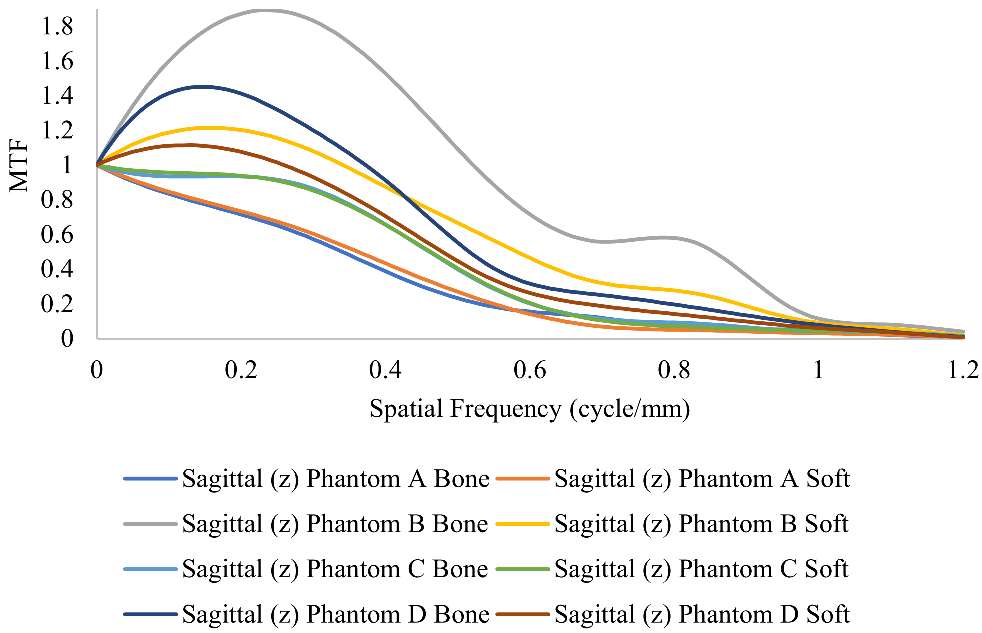

Fig. 5. MTF Value Graph on the Sagittal Plane (z)

International Journal of Progressive Sciences and Technologies (IJPSAT)
ISSN: 2509-0119.

https://ijpsat.org/
Vol. 55 No. 1 January 2026, pp. 197-205

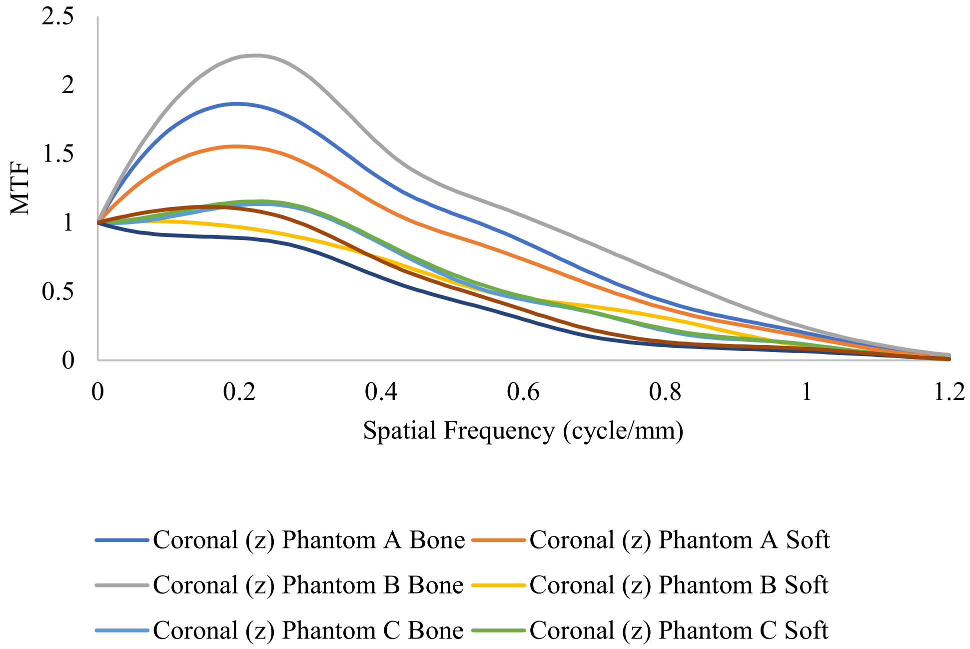

Fig. 6. MTF Value Graph on the Coronal Plane (z)

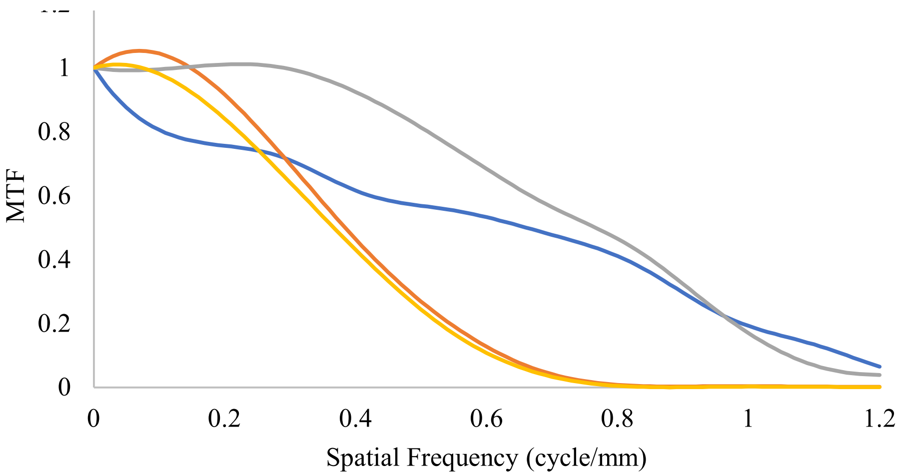

Fig. 7. MTF Value Graph on the Oblique Plane

10% values, especially with the bone filter [4] [15]. However, with a soft filter, the effect of wire diameter is not visible in all planes: only the sagittal and coronal planes show changes. In contrast, the axial and oblique planes are not affected by variations in wire size. When using soft filters, the effect of wire diameter is only visible in the sagittal and coronal planes because these two planes are more influenced by resolution along the z-axis (slice-thickness-dependent). This is in contrast to the axial and oblique planes, which are entirely dependent on the smoothing in-plane filter when using a soft filter. Thus, the effect of the wire diameter used is blurred.

The stability of the 3D MTF values is particularly evident when the wire diameter remains constant. In the axial plane, the MTF 10% value remains stable because the in-plane pixel size directly controls its resolution. In contrast, the sagittal and coronal planes show slight variations due to interpolation along the z-axis and the filter's smoothing characteristics, resulting in MTF values that are not fully isotropic.

IV. CONCLUSIONS

The evaluation results show that wire diameter strongly influences the 3D MTF, along with reconstruction filter type and analysis plane orientation. A smaller wire diameter yields a higher MTF 10% value, especially with a bone filter, whereas with a soft filter, this effect is only apparent in the sagittal and coronal planes. The reconstruction filter has a significant impact on the axial plane, where the bone filter consistently increases the MTF 10% value. In contrast, in the sagittal and coronal planes, the effect is minimal because both planes originate from axial image reformatting. A fairly noticeable difference in MTF values is observed in the oblique direction, reflecting a combination of lateral and longitudinal resolution. It is essential for assessing anatomical orientations that are not parallel to the central axis. Overall, all phantoms provided MTF 10% values within the recommended standard limits, indicating that the spatial resolution of CT images is feasible and reliable for clinical use.

This phantom is pratical for daily QC because the scanning process takes less than 1 minute, and MTF analysis can be complate in 5-10 minutes. Operators only neeed basic training in software operation, making this phantom suitable for routine QC implementation in radiology facilities.

REFERENCES

[1] S. A. M. Tunissen *et al.*, "Development, validation, and simplification of a scanner-specific CT simulator," *Med. Phys.*, vol. 51, no. 3, pp. 2081–2095, Mar. 2024, doi: 10.1002/mp.16679.

[2] RSD Gunung Jati Kota Cirebon, Jawa Barat, Indonesia, N. Asni, Moh. S. Nur Utami, and RSUD Cideres Kabupaten Majalengka, Jawa Barat, Indonesia, "QUALITY CONTROL CT SCAN (ANALISIS DAN EVALUASI KUALITAS CITRA)," *Pros. Semin. Si-INTAN*, vol. 3, no. 1, pp. 82–86, Oct. 2023, doi: 10.53862/SSI.v3.092023.014.

[3] U. Mahmood *et al.*, "Quality control of radiomic features using 3D-printed CT phantoms," *J. Med. Imaging*, vol. 8, no. 03, June 2021, doi: 10.1117/1.JMI.8.3.033505.

[4] C. Anam *et al.*, "Automated MTF measurement in CT images with a simple wire phantom," *Pol. J. Med. Phys. Eng.*, vol. 25, no. 3, pp. 179–187, Sept. 2019, doi: 10.2478/pjmpe-2019-0024.

[5] Nofrianto Nofrianto, Choirul Anam, Eko Hidayanto, and Ariij Naufal, "Comparison of MTFs Measured using IndoQCT and ImQuest Software on GE CT Phantom Images," *Int. J. Sci. Res. Sci. Technol.*, pp. 852–858, June 2023, doi: 10.32628/IJSRST523103156.

[6] E. Mejia, S. Sweeney, and J. E. Zablah, "Virtual 3D reconstruction of complex congenital cardiac anatomy from 3D rotational angiography," *3D Print. Med.*, vol. 11, no. 1, p. 4, Jan. 2025, doi: 10.1186/s41205-024-00247-6.

[7] M. A. Lodge, J. P. Leal, A. Rahmim, J. J. Sunderland, and E. C. Frey, "Measuring PET Spatial Resolution Using a Cylinder Phantom Positioned at an Oblique Angle," *J. Nucl. Med.*, vol. 59, no. 11, pp. 1768–1775, Nov. 2018, doi: 10.2967/jnumed.118.209593.

[8] C. Tominaga, H. Azumi, M. Goto, M. Taura, N. Homma, and I. Mori, "Tilted-wire method for measuring resolution properties of CT images under extremely low-contrast and high-noise conditions," *Radiol. Phys. Technol.*, vol. 11, no. 2, pp. 125–137, June 2018, doi: 10.1007/s12194-018-0443-8.

[9] A. Mondal, C. Nguyen, X. Ma, A. E. Elbanna, and J. M. Carlson, "Network models for characterization of trabecular bone," *Phys. Rev. E*, vol. 99, no. 4, p. 042406, Apr. 2019, doi: 10.1103/PhysRevE.99.042406.

[10] L. F. M. de Almeida, R. F. da Silva, E. R. Leite, P. L. Squair, and M. do S. Nogueira, "XXIV International Symposium on Solid State Dosimetry and 6th Symposium on Clinical Medical Physics," *XXIV Int. Symp. Solid State Dosim. 6th Symp. Clin. Med. Phys.*.

[11] E. Enjilela *et al.*, "Cubic-Spline Interpolation for Sparse-View CT Image Reconstruction With Filtered Backprojection in Dynamic Myocardial Perfusion Imaging," *Tomography*, vol. 5, no. 3, pp. 300–307, Sept. 2019, doi: 10.18383/j.tom.2019.00013.

[12] Choirul Anam, Betha S. Wulandari, Heri Sutanto, Riska Amilia, and Yuliana Lakapu, "The Evaluation of the Tube Current Impact on Axial, Sagittal, and Coronal MTFs on CT Images Using an In-House Phantom," *Int. J. Sci. Res. Sci. Technol.*, vol. 11, no. 6, pp. 349–354, Nov. 2024, doi: 10.32628/IJSRST24116185.

[13] A. M. Hernandez, P. Wu, M. Mahesh, J. H. Siewerdsen, and J. M. Boone, "Location and direction dependence in the 3D MTF for a high-resolution CT system," *Med. Phys.*, vol. 48, no. 6, pp. 2760–2771, June 2021, doi: 10.1002/mp.14789.

[14] Bushberg, J., Seibert, A., Leidholdt, E., and Boone, J., *The Essential Physics of Medical Imaging*.

[15] R. Zeng *et al.*, "Performance of a deep learning-based CT image denoising method: Generalizability over dose, reconstruction kernel, and slice thickness," *Med. Phys.*, vol. 49, no. 2, pp. 836–853, Feb. 2022, doi: 10.1002/mp.15430.