

Effect of Clinical Acquisition Parameters to Dose Distribution Adaptive Radiotherapy in Two Institutions

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Abstract

Introduction: This work aims to examine image quality and dose based on Cone Beam Computed Tomography using phantoms in the two institutions

Materials and methods: The Rando and Catphan Phantoms were scanned using identical image acquisition and reconstruction settings with Cone Beam Computed Tomography (CBCT). The calibration curve was obtained based on the Hounsfield Unit (HU) values on the Gammex 467 Phantom.. This study was performed at two institutions equipped with the same CBCT and CT Simulator. The acquisition with CBCT HU values that closely matched CT HU values employed a protocol of 120 kVp, Bowtie Filter (F1), and 80 mA for 10 ms within a Small Field of View (FOV). For the Medium FOV, the closest protocol also used 120 kVp, a Bowtie Filter, and 64 mA for 32 ms, while for the Large FOV, it was set at 120 kVp, F1, and 50 mA for 40 ms. The parameters chosen from each reconstructed FOV served as calibration curves for planning head and neck and pelvic treatments. Planning followed the Task Group 119 recommendations incorporating IMRT techniques.

Result: The best acquisition protocol for accurate dose calculations in head and neck cases is Field of View Small with 20 collimation using bowtie filter with 120 kVp and 80 mA for 10 ms/frame, while for Pelvic cases, it is FOVMedium with 20 collimation using bowtie filter with 120 kVp and 64 mA for 32 ms/frame. In image quality assessments, the lowest contrast visibility and uniformity scores were observed with FOV Small with 20 collimation using bowtie filter with 120 kVp and 80 mA for 10 ms/frame, using the reconstruction parameters Quality Image (QI) of 0.65% and 1%.

Conclusion: Selecting the appropriate CBCT acquisition protocol can lead to favorable outcomes in both dose calculation and image quality.

Keywords: Adaptive Radiotherapy; Cone Beam Computed Tomography (CBCT); CBCT Field of View (FOV)

Introduction

Image-guided radiation therapy (IGRT) is a radiotherapy technique to employ imaging techniques, such as computed tomography (CT) scan or magnetic resonance imaging (MRI), to ensure accurate radiation dose delivery to the tumor and the correct positioning of patients. IGRT uses Cone Beam Computed Tomography (CBCT), which produces 3D images similar to a CT scan. While CBCT allows for quick and accurate evaluation of dose distribution, it faces challenges related to image quality, consistency of Hounsfield unit (HU) values, and limited acquisition capabilities compared to CT. To perform dose calculations with the Treatment Planning System (TPS), converting HU values into relative electron densities (RED) is crucial.

The HU values in CBCT images vary due to several acquisition parameters, unlike those in CT images. The quality of CBCT images is influenced by its imaging technique and the amount of scattered radiation. For CBCT image acquisition, adjustments can be made to tube voltage, collimation, and filter type. Alterations in the acquisition parameters for kV-CBCT can impact both image quality and pixel value distribution¹. A widely used CBCT machine is the Elekta XVI (X-ray Volumetric Imaging) (Elekta Oncology Systems, UK), which operates within an energy range of 70 – 120 kVp and features three FOVs.

Previous work by Ubaidillah et al. indicates that the use of CBCT in adaptive radiotherapy is affected by various acquisition methods. This investigation was limited to the S20 FOV

mode for cases involving head and neck, showing that acquisition and reconstruction parameters influence the HU values relevant to the HU-RED calibration curve. The discrepancies across the full calibration curve remained below 1%, satisfying the gamma passing rate requirements. Additionally, the image quality assessment yielded scores of 1.2% for uniformity, 2.14% for low contrast visibility².

This study evaluated CBCT using three FOVs: Small (S), Medium (M), and Large (L), along with various acquisitions for each FOV. The parameters used for acquisition include a bowtie filter, differences in kV and mAs, and the scanning techniques employed. The Gammex phantom was utilized to assess the HU values of CBCT with a CT simulator. This study will analyze the impact of acquisition parameters on the HU values of CT and CBCT regarding dose reconstruction using a Treatment Planning System (TPS) and the Rando phantom as the subject. Additionally, this work examined image quality following the XVI Image Quality Test guidelines with the Catphan 530 phantom. To achieve a larger sample size, this research was conducted at two different institutions.

Materials and Methods

This study employs the CBCT XVI Synergy (Elekta, Stockholm, Sweden) alongside the GE Discovery RT 32-slice CT Simulator. The Elekta CBCT XVI is integrated into the Elekta Synergy gantry (Elekta, Stockholm, Sweden). It offers three Fields of View (FOV) tailored to various needs: Small (S) at 27 cm, Medium (M) at 41 cm, and Large (L) at 52 cm. Each FOV can accommodate additional collimators in cassette form, available in sizes 10, 15, and 20, which fit into the XVI outlet. Additionally, bowtie filters F0 and F1 are provided, where F0 indicates no filter and F1 indicates a filter is used.

The phantoms employed include the Gammex model 467 phantom (**Figure 1a**), the Rando phantom (The Phantom Laboratory, NY) and the Catphan 530 phantom (Phantom Laboratory, Salem, NY, USA). The Gammex 467 phantom diameter measures 33 cm and its length this 5 cm and comprises 4 solid water rods along with 12 tissue substitute rods. It is utilized to derive HU values for calibration curves, where each insert has

a diameter of 2.8 cm and a length of 7 cm. Rando phantom (**Figure 1b**) stands 175 cm tall and weighs 73.5 kg, containing 35 slices, each 2.5 cm thick. The Catphan 530 phantom (**Figure 1c**) is used to assess the image quality of CBCT, featuring a sensitometry sample for Teflon, Delrin, acrylic, polystyrene, Low Density Polyethylene (LDPE), Polymethylpentene PMP, air, and water, with a diameter of 15 cm and a thickness of 40 mm. This study's dose calculation was performed using the Treatment Planning System (TPS) Monaco version 6.1.2.0, which employs the Monte Carlo algorithm.

During this process, a Gammex 467 phantom was scanned using a CT simulator according to the radiation parameters listed in **Table 1**. Scanning was also conducted on the CBCT XVI Elekta, following the CBCT radiation parameters shown in **Table 2**. Image acquisition on the CBCT XVI involved three fields of view (FOVs): S20, M20, and L20, achieved by moving the gantry by 200° or half-scan. After obtaining the Gammex 467 phantom images from the CT simulator, an automatic calibration curve was generated for planning. The Gammex phantom was additionally scanned with the CBCT XVI Elekta using the specified acquisition settings.

After using the CBCT XVI Elekta for scanning, the data obtained by the detector were reconstructed to create a volumetric data set. This reconstruction relies on pre-set acquisition parameters such as voxel size, number of projections, field of view (FOV), and the specific software and hardware employed. It is essential that this reconstruction is completed quickly, ideally in under 3 minutes. During this process, four types of reconstructions were utilised: three standard image reconstructions from the CBCT XVI system and one that was modified by the researcher. The reconstruction parameters relevant to the study are illustrated in **Table 3**.

Table 1. CT Simulator Acquisition Irradiation Parameters

Irradiation Parameters	
Tube Voltage (kVp)	120
Effective mAs	560
Slice Thickness (mm)	5
FOV (mm)	312

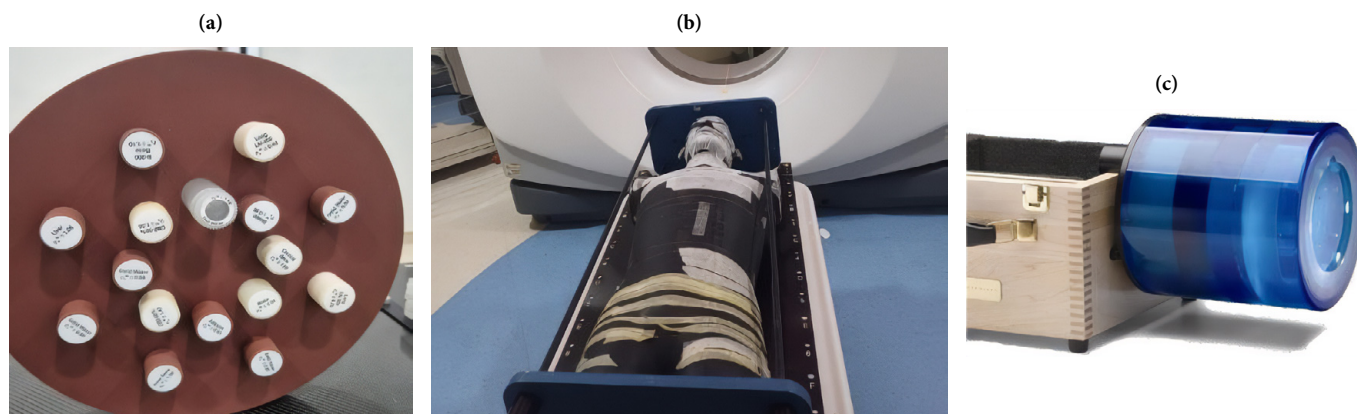


Figure 1. The phantoms used in this study : (a) Gammex 467 (b) Rando (c) Catphan 530 phantom

Table 2. CBCT XVI Elekta Acquisition Irradiation Parameters

Acquisition Parameters S20	
Collimator	S20
kVp	100;120
(mA, ms/frame)	(10,10); (50,10); (80,10)
Total mAs	(36,1); (180,5); (288,8)
Filter	F0; F1
Frame	361
Acquisition Parameters M20	
Collimator	M20
kVp	120
(mA, ms/frame)	(25,40); (40,40); (64,32)
Total mAs	(660); (1056); (1351,68)
Filter	F1
Frame	361
Acquisition Parameters L20	
Collimator	L20
kVp	120
(mA, ms/frame)	(40,32); (50,40); (64,40)
Total mAs	(650); (832); (1300); (1516)
Filter	F1
Frame	650

In the Medium and Large FOV, the “Reconstruction-DataType” parameter uses short, which means that XVI performs fast reconstruction and results in lower image quality compared to float. The “PreFilter” parameter in X-reconstruction has also been changed to “Median9”. This value will apply a median window filter to the projected image, where the window also includes one pixel and four nearest pixels. “Median9” produces higher contrast than “None”, but its resolution is lower than “None”. The last change is the “ProjectDownSizeFactor” parameter, which affects the speed of 3D reconstruction. A value of 1 means that 3D reconstruction is better but takes longer. All the changed parameters correlate with how the algorithm interpolates the information received by the detector.

The reconstructed CT and CBCT images were imported into the Monaco TPS for the purpose of three protocols were selected. This analysis utilized a circular ROI with a diameter of 2.5 cm volume ROI in Gammex 467 phantom (**Figure 2**). The measurement means HU for volume of one ROI from the CBCT images, collected through various acquisition and reconstruction methods, was compared to the mean HU from the CT images. This involved examining the differences between the two data sets. The chosen CBCT acquisition protocol was the one that exhibited the least difference in mean HU compared to the CT mean HU, ensuring that it did not surpass

Table 3. CBCT Reconstruction Protocols

Acquisition Parameters S20				
Acquisition Parameters	HR	MR	SR	IQ
Reconstruction VoxelSize	0.5	1	0.5	0.5
Reconstruction Dimension X	540	270	540	540
Reconstruction Dimension Y	520	264	520	520
Reconstruction Dimension Z	540	270	540	540
Reconstruction Offset X	0	0	0	0
Reconstruction Offset Y	0	0	0	0
Reconstruction Offset Z	0	0	0	0
Reconstruction Filter	Wiener	Wiener	Wiener	Wiener
NumberOfReconstruction	2	2	2	2
FilterParameters				
ReconstructionFilterParameter1	0.05	0.05	0.05	0.05
ReconstructionFilterParameter2	90	90	10	90
Interpolation	Partial2	Partial2	Bilinear	Partial2
ScatterCorrection	Uniform	Uniform	Uniform	Uniform
NumberOfScatterCorrectionParameters	1	1	1	1
SactterCorrectionParameter	0.24	0.24	0.24	0.24
ReconstructionDataType	Float	Float	Short	Short
PreFilter	None	None	Median5	None
ProjectionDownSizeFilter	1	2	1	1

Table 3 cont.

Acquisition Parameters M20				
Acquisition Parameters	HR	MR	XR	IQ
Reconstruction VoxelSize	0.75	1	0.5	0.75
Reconstruction Dimension X	550	410	728	550
Reconstruction Dimension Y	328	264	328	328
Reconstruction Dimension Z	540	410	728	540
Reconstruction Offset X	0	0	0	0
Reconstruction Offset Y	0	0	0	0
Reconstruction Offset Z	0	0	0	0
Reconstruction Filter	Wiener	Wiener	Wiener	Wiener
NumberOfReconstruction	2	2	2	2
FilterParameters				
ReconstructionFilterParameter1	0.05	0.05	0.05	0.05
ReconstructionFilterParameter2	90	90	90	90
Interpolation	Partial2	Partial2	Partial2	Partial2
ScatterCorrection	Uniform	Uniform	Uniform	Uniform
NumberOfScatterCorrectionParameters	1	1	1	1
SactterCorrectionParameter	0.24	0.24	0.24	0.24
ReconstructionDataType	Float	Float	Short	Short
PreFilter	None	None	Median5	None
ProjectionDownSizeFilter	2	2	1	2
Acquisition Parameters L20				
Acquisition Parameters	HR	MR	ZR	IQ
Reconstruction VoxelSize	0.75	1	1	0.75
Reconstruction Dimension X	670	500	670	670
Reconstruction Dimension Y	328	264	328	328
Reconstruction Dimension Z	670	500	670	670
Reconstruction Offset X	0	0	0	0
Reconstruction Offset Y	0	0	0	0
Reconstruction Offset Z	0	0	0	0
Reconstruction Filter	Wiener	Wiener	Wiener	Wiener
NumberOfReconstruction	2	2	2	2
FilterParameters				
ReconstructionFilterParameter1	0.05	0.05	0.05	0.05
ReconstructionFilterParameter2	90	90	90	90
Interpolation	Partial2	Partial2	Partial2	Partial2
ScatterCorrection	Uniform	Uniform	Uniform	Uniform
NumberOfScatterCorrectionParameters	1	1	1	1
SactterCorrectionParameter	0.24	0.24	0.24	0.24
ReconstructionDataType	Float	Float	Short	Short
PreFilter	None	None	Median5	None
ProjectionDownSizeFilter	2	2	1	2

*MR is Medium image reconstruction parameters (recommendation XVI), HR is High image reconstruction parameters (recommendation XVI), and SR, XR, ZR, IQ is researcher made image reconstruction parameters

100 HU from the CT value. Three acquisition protocols were chosen, including one that is recommended by the manufacturer's protocols included in the XVI system is Field of View Small with 20 collimations not used bowtie filter with 100 kVp and 10 mA for 10 ms/frame.

The acquisition of phantom images for Rando was conducted using nine selected acquisition protocols based on the mean HU values, with three protocols each from S20 FOV, M20 FOV, and L20 FOV. This process yielded in ten Rando phantom images, consisting of one CT image and nine CBCT images. All Rando phantom images were sent to the TPS Monaco for contouring in accordance with TG 119 for head and neck and pelvic cases. After sending the CT and CBCT images to

TPS Monaco and performing contouring on them, treatment planning was carried out following the guidelines of TG 119 (**Figure 3**)³. Parameter of treatment plan required grid spacing 3 mm, sequencing parameters 30 maximal control point, minimum segment width 0.5 cm, statistical uncertainty 1%, and medium fluence smoothing.

TG 119 specifies the dose objectives essential for IMRT, especially in head and neck cases, where careful consideration of specific structures, target volumes, and dose constraints is vital to assess the accuracy of both planning and delivery. The initial planning phase utilizes images from a CT simulator, identifying the left and right parotid glands and spinal cord as organs at risk (OAR) within the head and neck phantom.

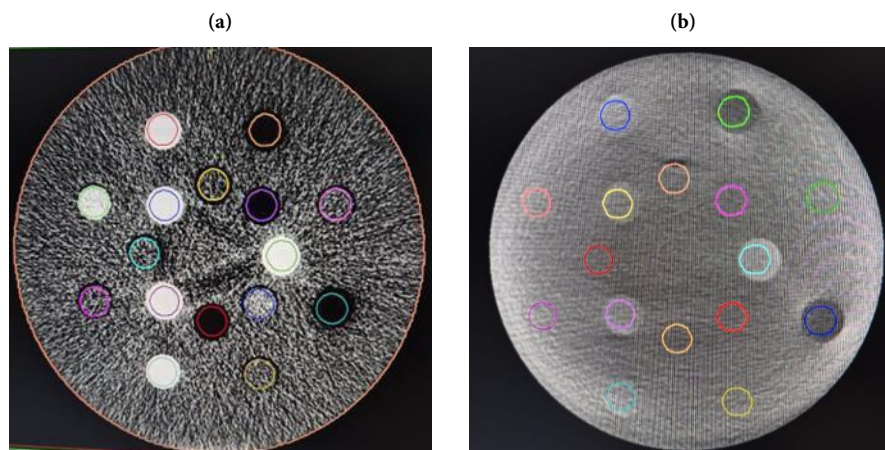


Figure 2. ROI 2.5 cm of TPS (a) CT Image (b) CBCT Image

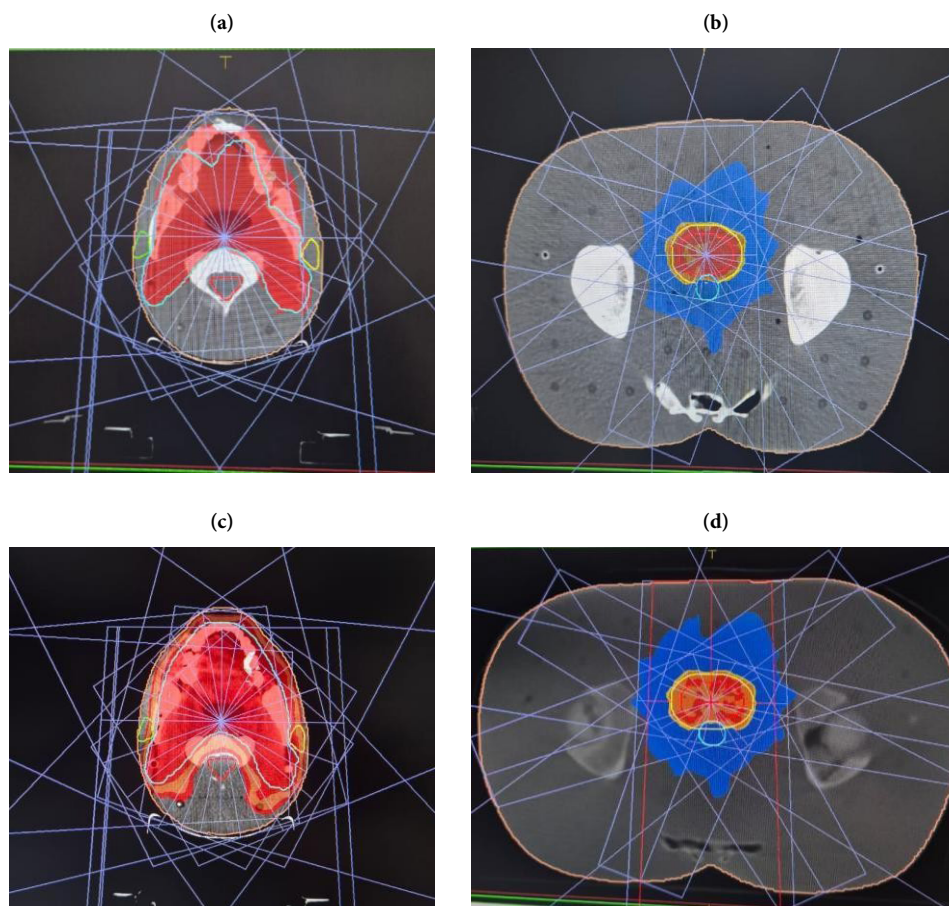


Figure 3. IMRT Planning Process (a) head and neck in CT (b) pelvic in CT (c) head and neck in CBCT (d) pelvic in CBCT

In pelvic cases, the bladder and rectum are recognized as the relevant OARs. Following this, dose distribution evaluation is performed by analyzing the dose volume histogram (DVH) for PTV at levels of 2% (near maximum), 50% (median dose) and 98% (near minimum), in addition to the OAR dose for each case. Another evaluation technique involves calculating the conformity index (CI) and homogeneity index (HI) using the equations specified in the equations section.

$$CI = \frac{PTV \times PIV}{PTV_{PIV}^2} \quad (1)$$

$$HI = \frac{D_{2\%} - D_{98\%}}{D_{mean}} \quad (2)$$

Where PTV is planning target volume; PIV is prescription iso-dose volume; PTV_{PIV}^2 is planning target volume receiving the prescribed dose or more; $D_{2\%}$ is minimum dose to 2% volume of the PTV; $D_{98\%}$ is minimum dose to 98% volume of PTV; D_{mean} is mean dose of PTV.

The phantom imaging of Catphan using the Elekta CBCT XVI was executed with nine selected ones. This image quality assessment aimed to identify the acquisition and reconstruction parameters that provide the optimal image quality. The evaluations performed comprised uniformity testing and low contrast visibility.

In the uniformity test or image consistency, the determination is based on the pixel values in the center of the CT slice in both horizontal and vertical directions. Information is gathered by taking the mean value from four points spaced 4.5 cm apart, as well as at the center point of the isocenter. Once the mean value is obtained, uniformity can be calculated using **Equation 3**. This test allows for a tolerance level of 3%.

$$Uniformity = \left(\frac{mean(high) - mean(low)}{mean(high)} \right) \times 100\% \quad (3)$$

Where, mean (high) is highest mean CT number among all ROIs (center and peripheries); mean (low) is lowest mean CT number measured among all ROIs.

The low contrast visibility test assesses the quality of CBCT images by checking the system's capability to detect subtle variations in tissue density. This evaluation requires the mean value and standard deviation of LDPE and Polystyrene fillings. An object is deemed visible if the mean CT value varies by more than the standard deviation, which serves as a measurement for noise. The test allows for a tolerance level of 3%. After obtaining the mean value and average standard deviation for both fillings, calculations can be performed using the specified equation.

$$LCV = \frac{\frac{CT_{polystyrene} - CT_{LDPE}}{mean_{polystyrene} - mean_{LDPE}}}{\frac{SD_{polystyrene} + SD_{LDPE}}{2}} \times 100\% \quad (4)$$

*In the documentation supplied with the catphan phantom, from the sensitiometry section, get the values of $CT_{polystyrene}$ is -35, and CT_{LDPE} is -100. $mean_{polystyrene}$ is the mean HU value of polystyrene; $mean_{LDPE}$ is the mean HU value of LDPE; $SD_{polystyrene}$ is standard deviation of HU in polystyrene; SD_{LDPE} is standard deviation of HU in LDPE.

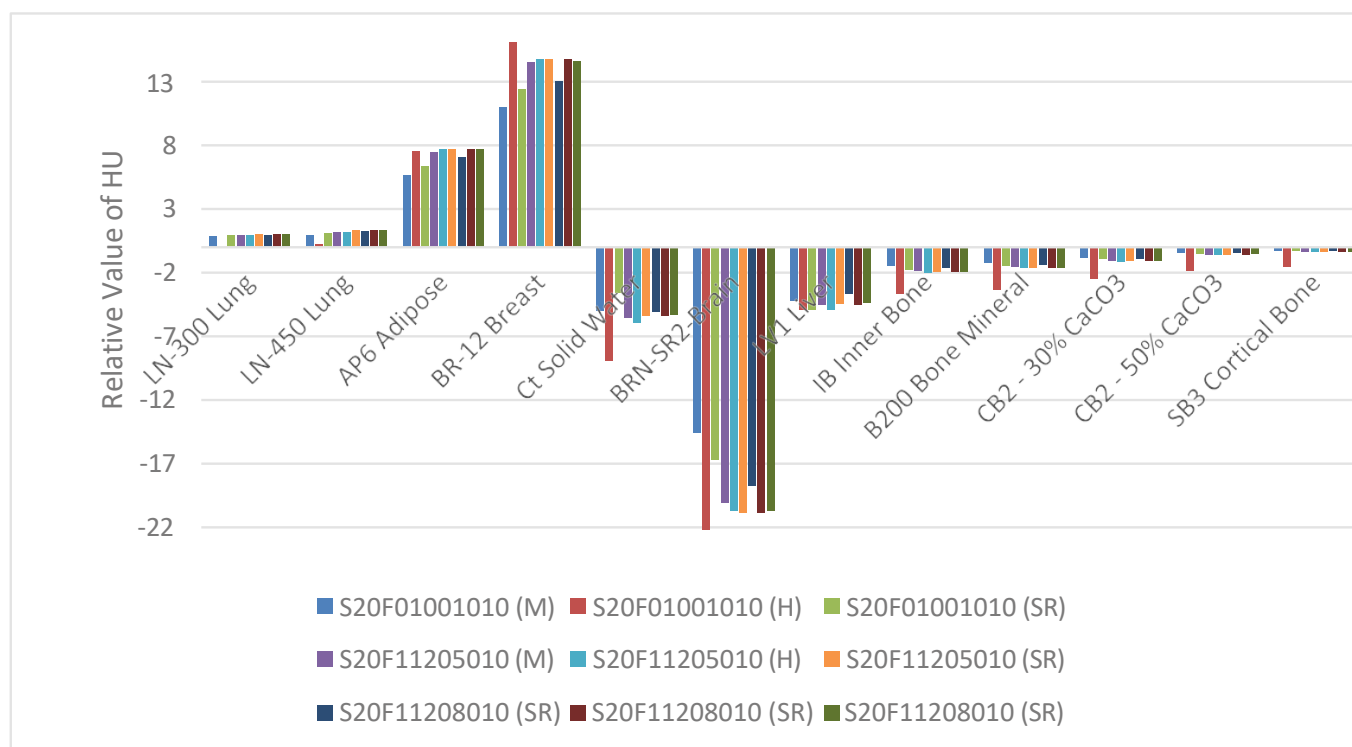
Results

Figure 4 presents the relative HU curve of CBCT in comparison to CT across several acquisition protocols. **Figures 4a and 4b** illustrate the curve results from two distinct institutions, revealing no significant differences between them. This similarity arises from the use of identical equipment, materials, and software. The image features various materials, including adipose, breast, solid water, and brain, which exhibit fairly different values compared to CT about -24.3 HU for the brain material obtained with the S20-F0-100-10-10 acquisition, attributed to its position within the inner phantom (**Figure 1**).

Figure 5 illustrates that in the head and neck cases, the D90 parotid peak is 1.030 in the S20-F0-100-10-10 acquisition protocol at the second institution. In contrast, the first institution reports a value closer to the OAR CT at about 0.98. The S20-F1-120-80-10-zr acquisition protocol yields relative OAR values close to 1. In the pelvic cases, the F1-120-40-32-m acquisition protocol shows the greatest deviation from 1 for the Rectum organ, registering at 0.979 at the second institution. The D90 bladder values hover around 1.002 across several acquisition protocols at both institutions. Notably, the M20-F1-120-64-32 acquisition protocol exhibits values approaching 1, indicating closeness to the OAR value on CT across all reconstructions.

Figure 6 shows the relative values of HI and CI for CBCT compared to CT. In the head and neck cases using a Small FOV, almost all relative values for CI are close to one. A relative value of 1.013 is consistently observed in the acquisition protocols F0-100-10-10-sr, F1-10-50 across all reconstructions, and F1-10-50-80-10-m at the first institution. In contrast, at the second institution, the relative CI values range from 0.985 to 1.015. For the pelvic cases, Medium and Large FOVs were used, where the first institution reported. Medium FOV and 0.815 to 1.130 for the Large FOV. At the second institution, the relative CI values ranged from 0.941 to 1.216 for the Medium FOV and 0.862 to 1.059 for the Large FOV. The most distant relative CI value compared to the head and neck cases is 0.985 in the acquisition F0-100-10-10. For the pelvic cases, it is 0.815 in the acquisition F1-120-50-40 with the Large FOV. The HI values at the first institution across all FOVs show values below one, with the furthest value being 0.79 from CT in the Medium FOV, while the second institution's furthest values are 1.317 for the head and neck cases and 1.18 for the pelvic cases in the Large FOV.

(a)



(b)

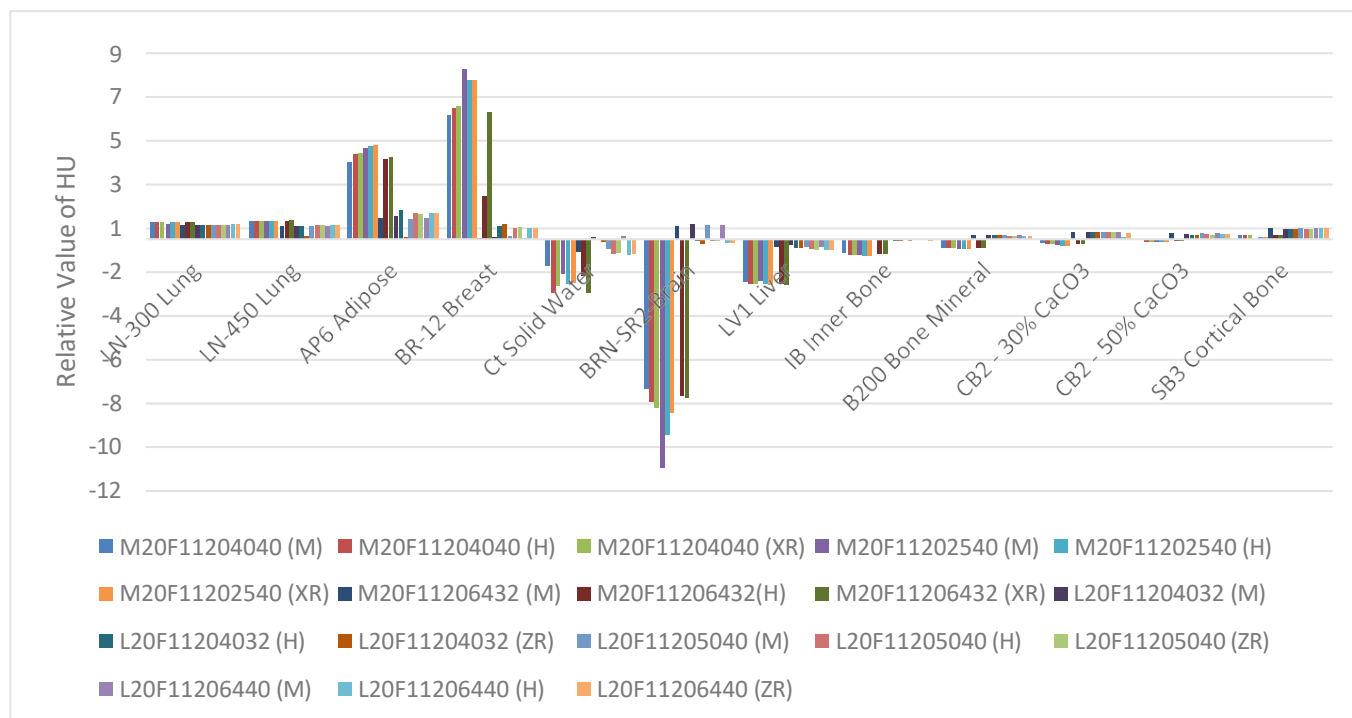
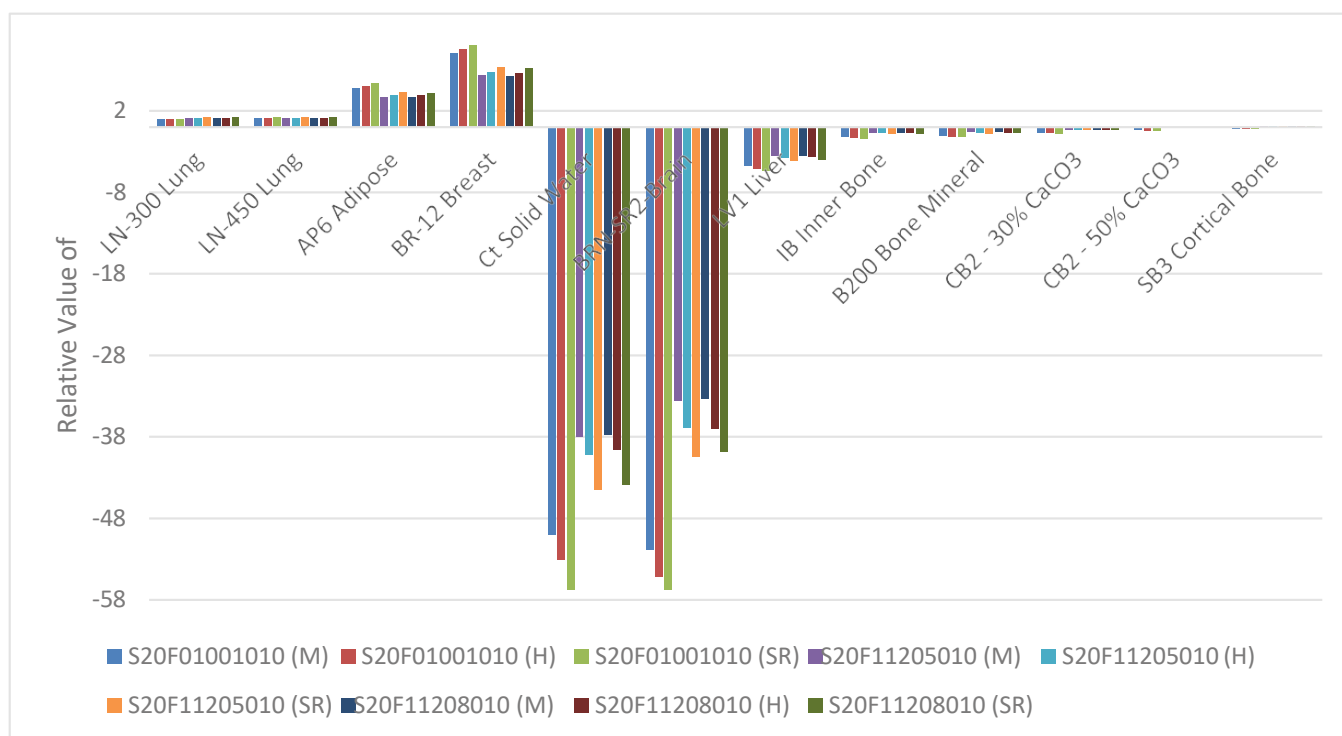


Figure 4ab. Relative HU values of CBCT to HU of CT (a) Small FOV first institution (b) Medium and Large first institution

(c)



(d)

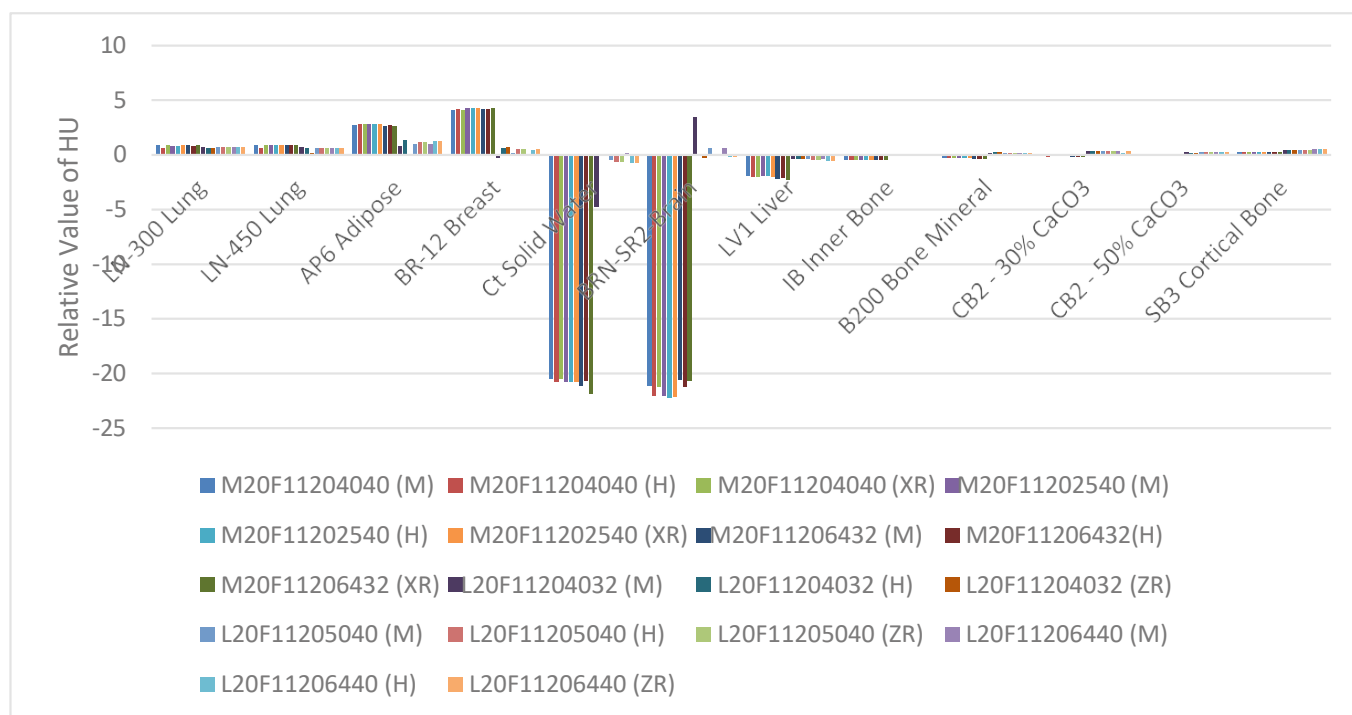


Figure 4cd. Relative HU values of CBCT to HU of CT (c) Small FOV second institution (d) Medium and Large second institution

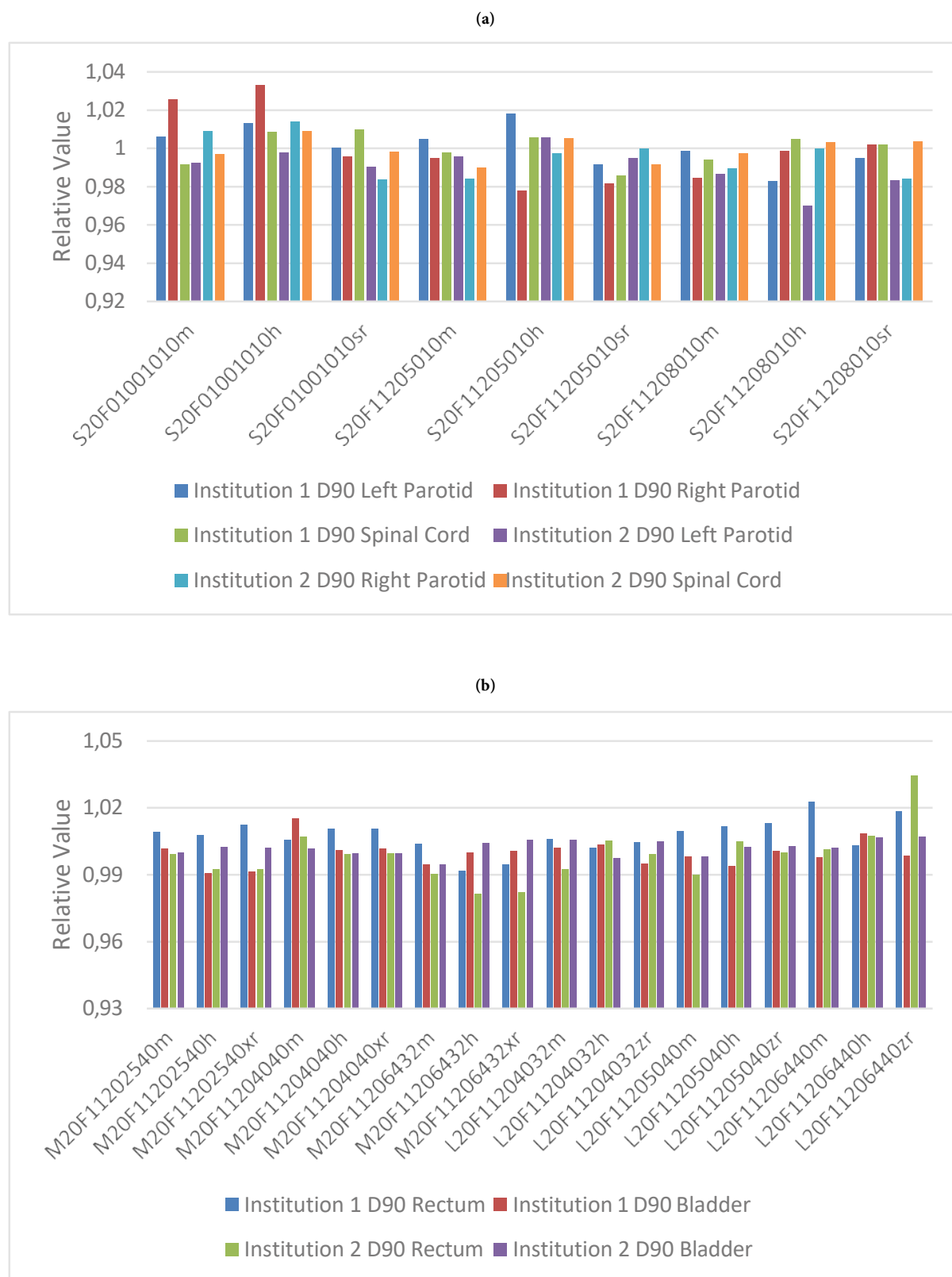


Figure 5. Relative Value of Organ at Risk CBCT values compared to Organ at Risk CT (a) head and neck (b) pelvic



Figure 6. Relative values of HI, CI, and CBCT compared to CT: (a) Small FOV; (b) Medium and High FOV.

In **Figure 7** all calculations result a difference of no more than 1.5%. The S20-F0-100-10-10-m protocol had the highest relative value at D98, at 1.0187 at the second institution for head and neck cases. Meanwhile, for pelvic cases, the highest relative value between PTV CBCT and CT was seen at D98, at

1.016, with the acquisition parameters L20-F1-120-64-40. This is consistent with the study by Souleyman and Dunlop, which showed that the dose difference was no more than 2.5% when using the S20-F0-100-10-10 protocol^{1,4}.

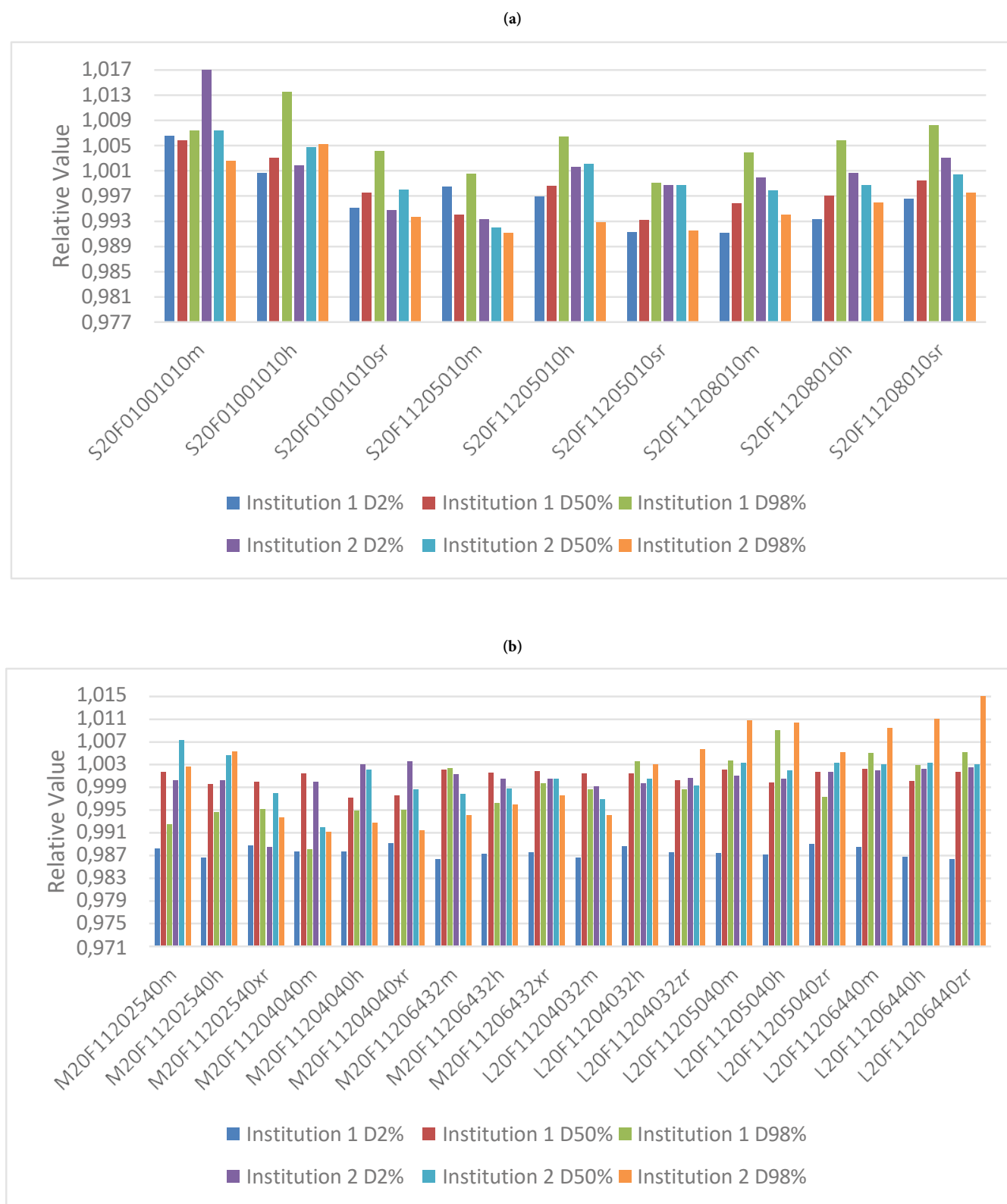


Figure 7. Relative values of Dose PTV compared to CT: (a) Institution 1; (b) Institution 2

In **Figure 8**, the S20-F0-100-10-10 acquisition parameter with medium reconstruction demonstrates uniformity test values and low contrast visibility (LCV) that are significantly above the tolerance limits in both institutions. At the first institution, the results were 4.2 % as a value of uniformity test and 8.90 % for the LCV test, while at the second institution, the results were 5.1 for the uniformity test and 8.87 for the LCV test.

The optimal values were achieved from the S20-F1-120-80-10 acquisition protocol with Image Quality (IQ) reconstruction developed by the author. At the first institution, the results were 1.12 for the uniformity test and 0.65 for the LCV test, while at the second institution, the results were 1 for the uniformity test and 0.78 for the LCV test.

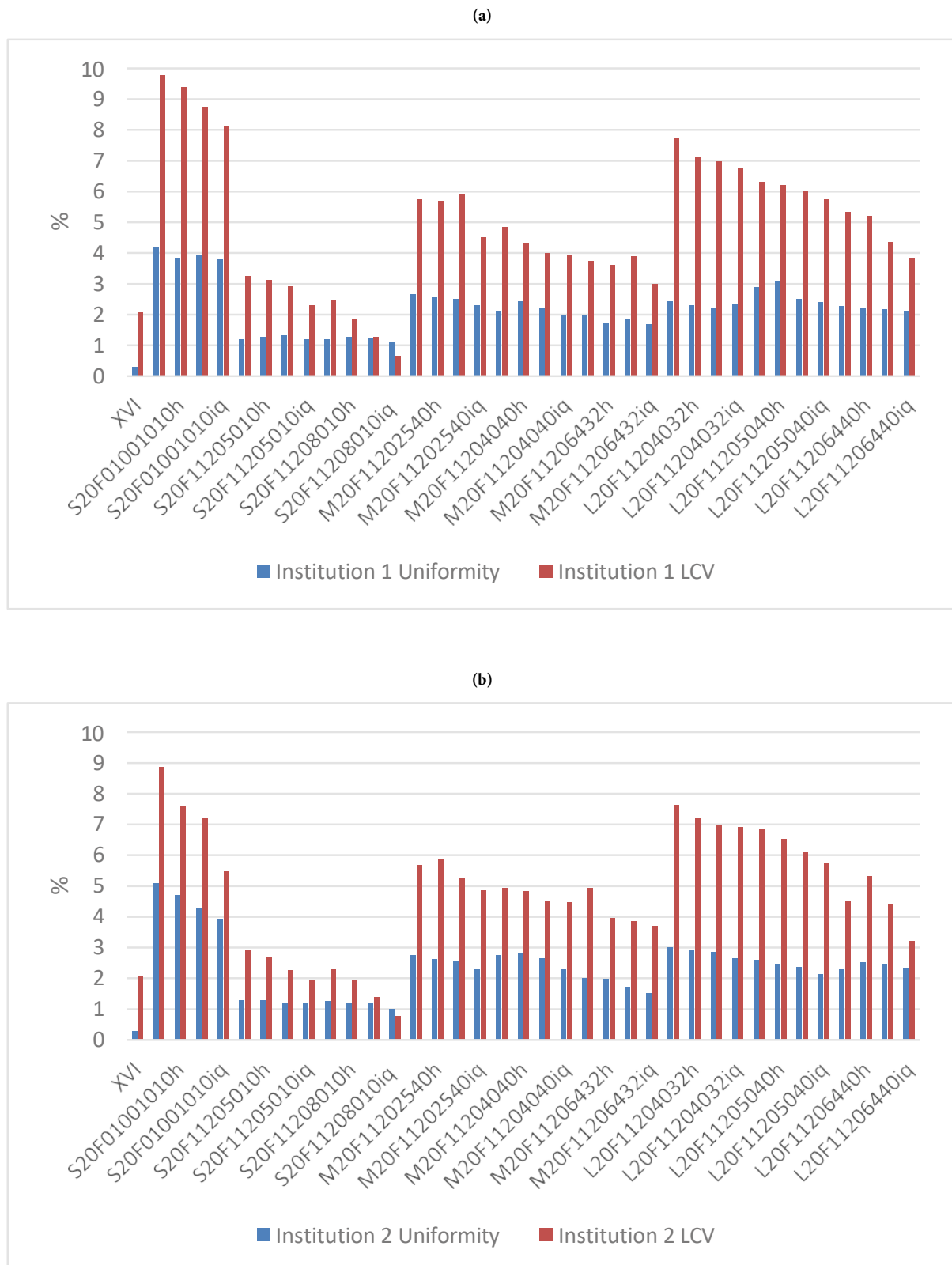


Figure 8. Image Quality Test Values (a) Institution 1 (b) Institution 2

Discussion

In the small FOV with specific acquisitions, the values closely align with CT values, with some materials showing a difference in HU values of under 100. In contrast, the Medium and Large FOVs exhibit a difference in HU values for nearly all materials that exceeds 100 HU⁵. This discrepancy relates to the outer diameter of the Gammex 467 phantom, which measures 22.4 cm. The small FOV, with a diameter of 27 cm, is the most comparable, while the larger FOVs induce a considerable amount of scatter, leading to inaccuracies in the CT to ED values. Acquisition protocols, including the use of filters, voltage, and mAs, significantly affect the HU values. As illustrated in **Figure 4**, the relative value approaches 1 when higher filters, kV, and mA are applied. The bowtie filter reduces the need for high current for the CBCT HU values to align closely with the CT HU values; this occurs because the filter adjusts the x-ray beam to create a more uniform exposure across the entire FOV. Moreover, the filter enhances image quality and the precision of HU measurements, particularly in regions adjacent to bone or dense tissue⁶. In addition to the use of filters, the voltage level also affects the difference in HU values. As the voltage increases, the energy of the X-ray beam will increase, leading to higher HU values. This is because photons interact more with the tissue, resulting in a stronger signal. Research by Giacometti et al. showed that changes in voltage have a more significant impact on HU values compared to changes in mAs (tube current)⁷. Higher mAs values will produce more photons, which can improve image quality by reducing noise. This also affects the accuracy of HU; as the mA used increases, the relative value of CBCT to CT approaches 1, meaning the difference between CBCT and CT values becomes smaller. The relationship between mAs and HU is most direct and linear when other parameters remain constant⁸.

The placement of material inserts in the inner and outer layers affects the presence of cupping artifacts. The HU values from CBCT for the inner material are higher in the Large FOV, except for bone material, while in the small FOV, the values are lower compared to the outer layer⁹. Materials with high density, such as IB Inner Bone, B200 Bone Mineral, CB2-30% CaCO₃, CB2-50% CaCO₃, and SB3 Cortical Bone, have relatively low values in succession. The largest difference in HU values is observed in the densest material insert (Cortical Bone SB3) across the entire FOV in two institutions. In the first institution, the highest relative HU CBCT value compared to CT for the Cortical Bone material was found in the S20-F0-100-10-10 acquisition at -2026 HU, HU value bones are negative because beam hardening and scatter distort attenuation of dense tissue like cortical bone, lowering their reconstructed gray levels. While the lowest was in the S20-F1-120-80-10 acquisition at 1023 HU. In the second institution, the highest relative HU CBCT value compared to CT was also for the Cortical Bone material at -1970 HU in the S20-F0-100-10-10 acquisition, and the lowest was in the S20-F1-120-80-10 acquisition at -1016 HU. These research

results are consistent with Souleyman et al.¹, which states that using the Gammex 467 phantom, the highest difference in HU values was obtained from Cortical Bone.

According to Srivivasan et al.¹⁰, the cupping artifact is characterized by a reduction in the isocenter attenuation area in the CBCT image, which becomes more pronounced in a larger FOV. This is because a larger FOV encompasses more material, resulting in beam hardening and greater scatter. A smaller FOV can minimize this effect, but it can create other issues such as image projection clipping. This effect is clearly visible in the isocenter area of the object, making it appear darker. This is consistent with what is presented in **Figure 2b**, where in the large FOV, the isocenter area is darker compared to the off-center area. The contribution of scatter and beam hardening effects is evident in this FOV. A study conducted by Sarria et al.¹¹ stated that the cupping artifact occurs in the large FOV due to a difference between the phantom size and the scanning mode, resulting in significant errors in the isocenter area of the large FOV. A Bowtie filter can be used to reduce beam hardening that causes the cupping artifact. Reconstruction algorithms can also be employed to reduce scatter radiation from the projection images.

The evaluation of dose distribution was carried out using the Treatment Planning System (TPS) Monaco, focusing on the CI, HI, and OAR to understand the impact of CBCT value errors on dose calculations. CT and CBCT images were uploaded to the TPS Monaco for treatment planning in head and neck and pelvic cases. The CI and HI values were derived through calculations based on **equations 1 and 2**. CI assesses the dose delivered accurately to the target volume, while HI measures the uniformity of dose distribution within that volume. A CI value nearing 1 signifies improved conformity, indicating that the dose to the organ volume is accurate. Conversely, an HI value approaching 0 denotes increasing homogeneity in dose distribution. As illustrated in **Figure 6**, the relative CI and HI values for CBCT versus CT were closest to 1 in the S20-F1-120-80-10 acquisition protocol for the head and neck case; for the pelvic case, the relative value closest to 1 was seen in the M20-F1-120-64-32 acquisition.

The indirect reconstruction algorithm greatly influences the dose distribution and image quality of CBCT. This relationship is clear in **Figures 5 and 6**, where the CI and HI values, despite using the same acquisition parameters, show varying outcomes based on different reconstruction methods. In the image quality assessment, the Uniformity and LCV values for the acquisition parameters, when reconstructed according to the author's modifications, fall below the tolerance thresholds advised by XVI system manufacturer protocols.

The reconstruction recommended by the author has better values for dose distribution and image quality testing. The reconstruction algorithm uses "Partial 2" interpolation, which interpolates by a factor of two and performs near-neighbor interpolation along the back projection, thus providing optimal image quality. This mode is used as the standard mode in the XVI reconstruction protocol. The use of the "Uniform"

parameter means that XVI employs an algorithm to estimate the amount of scattered radiation, which will later affect the projection data.

Uniformity and low contrast visibility tests were performed following the recommendations of XVI using the Catphan 530 Phantom. The uniformity test's tolerance is established at 1.5%, as stated in XVI's guidelines. **Figure 7** illustrates the uniformity test values across various FOVs along with their corresponding reconstructions. The highest value, exceeding the tolerance limit, appears in the S20-F0-100-10-10 acquisition protocol with medium reconstruction, while the lowest is recorded from S20-F1-120-80-10 with image quality reconstruction produced by the author.

FOV also influences image quality testing, as illustrated in **Figure 7**, showing that a larger FOV enhances the uniformity test value, moving it further from the tolerance limit. When we compare the S20-F1-120-80-10 acquisition to M20-F1-120-64-32 and L20-F1-120-60-40, all utilising the same reconstruction, the uniformity test values vary. The Small FOV yields the highest test value among the FOVs assessed. This is attributed to the fact that a larger FOV captures a greater volume of the object, increasing scattering and noise, which degrades image quality. Additionally, a larger FOV is more prone to artifacts, such as beam hardening. The supplementary reconstruction developed by the researchers (**Table 3**) enhances the test values, particularly with image quality (IQ) reconstruction, where the result reaches 0.65%, signifying a good uniformity level or fairly consistent HU values. In contrast, the S20-F0-100-10-10 acquisition parameter, even with the same reconstruction, yields results above the tolerance limit. Inappropriate image acquisition parameters cannot be rectified through reconstruction algorithms.

In the LCV test, the tolerance threshold is established at 3% by examining the differences between LDPE and Polystyrene materials. This illustrates that an effective reconstruction algorithm can enhance test outcomes. Additionally, the difference in Field of View (FOV) influences image quality; a smaller FOV tends to yield better results, whereas a larger FOV may reduce LCV due to scattering effects. Key acquisition

parameters such as voltage, mA, and filter usage play a crucial role in determining image quality. When comparing the S20-F0-100-10-10 acquisition parameter with a quality image (QI) to S20-F1-120-80-10 under identical reconstruction, the differences in results are noticeable. Increasing mA enhances image quality, which supports findings by Li et al.,¹² indicating that a setting of 100 mAs improves phantom image visibility, whereas 40 mAs results in diminished image quality, leaving the target partially obscured. Employing a bowtie filter can mitigate scattering and artifacts, while higher voltage and current can improve Uniformity and Low Contrast Visibility (LCV). Posiewnik et al. study also explored low contrast visibility tests across different FOVs, revealing that the Small FOV achieves a superior value of 0.7% compared to the Medium and Large FOVs¹³.

Conclusions

Using the right acquisition parameters can influence image quality and HU values, which are indirectly linked to dose distribution. The HU values from CBCT that most closely match CT HU are observed in the Small FOV, whereas the Medium and Large FOVs exhibit greater discrepancies. This is primarily attributed to the cupping artifacts present in Medium and Large scan modes, resulting from the mismatch in phantom size. For head and neck cases, the suggested acquisition parameters are S20-F1-120-80-10, whereas for pelvic cases, they are M20-F1-120-64-32. The values for uniformity and low contrast visibility are notably low at 1% and 0.78%, respectively, when using the same acquisition parameters along with the author's preferred reconstruction algorithm.

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References

1. Souleyman S, Maria KD, Cheikh T, Karima KK. Impact of Acquisition Protocols on Accuracy of Dose Calculation Based on XVI Cone Beam Computed Tomography. *Journal of Medical Physics*. 2021;46(2):94-104. doi:10.4103/jmp.jmp_128_20
2. Ubaidillah A, Fauzia AR, Purnomo AT, Nasution N, Wibowo WE, Pawiro SA. Effect of the Small Field of View and Imaging Parameters to Image Quality and Dose Calculation in Adaptive Radiotherapy. *Polish Journal of Medical Physics and Engineering*. 2023;29(2):130-142. doi:10.2478/pjmpe-2023-0014
3. AAPM. TG-119 IMRT Commissioning Tests Instructions for Planning, Measurement, and Analysis. 2009. https://www.aapm.org/pubs/tg119/TG119_Instructions_102109.pdf
4. Dunlop A, McQuaid D, Nill S, et al. Comparison of CT number calibration techniques for CBCT-based dose calculation. *Strahlenther Onkol*. 2015;191(12):970-978. doi:10.1007/s00066-015-0890-7
5. Richter A, Hu Q, Steglich D, et al. Investigation of the usability of conebeam CT data sets for dose calculation. *Radiat Oncol*. 2008;3(1). doi:10.1186/1748-717x-3-42
6. Rafic M, Ravindran P. Evaluation of on-board imager cone beam CT hounsfield units for treatment planning using rigid image registration. *Journal of Cancer Research and Therapeutics*. 2015;11(4):690-696. doi:10.4103/0973-1482.146087
7. Giacometti V, King RB, Agnew CE, et al. An evaluation of techniques for dose calculation on cone beam computed tomography. *The British Journal of Radiology*. 2019;92(1096). doi:10.1259/bjr.20180383

8. Holm AIS, Nyeng TB, S. Møller D, et al. Density calibrated cone beam CT as a tool for adaptive radiotherapy. *Acta Oncologica*. 2021;60(10):1275-1282. doi:10.1080/0284186x.2021.1945678
9. Taniguchi T, Hara T, Shimozato T, et al. Effect of computed tomography value error on dose calculation in adaptive radiotherapy with Elekta X-ray volume imaging cone beam computed tomography. *J Applied Clin Med Phys*. 2021;22(9):271-279. doi:10.1002/acm2.13384
10. Srivasan K, Mohammadi M, Shepherd J. Applications of linac-mounted kilovoltage Cone-beam Computed Tomography in modern radiation therapy: A review. *Pol J Radiol*. 2014;79:181-193. doi:10.12659/pjr.890745
11. Sarria GR, Schmitt H, Jahnke L, et al. Cone Beam CT-Based Daily Adaptive Planning or Defined-Filling Protocol for Neoadjuvant Gastric Cancer Radiation Therapy: A Comparison. *Advances in Radiation Oncology*. 2021;6(1):100593. doi:10.1016/j.adro.2020.09.026
12. Li J, Yao W, Xiao Y, Yu Y. Feasibility of improving cone-beam CT number consistency using a scatter correction algorithm. *J Applied Clin Med Phys*. 2013;14(6):167-176. doi:10.1120/jacmp.v14i6.4346
13. Posiewnik M, Piotrowski T. A review of cone-beam CT applications for adaptive radiotherapy of prostate cancer. *Physica Medica*. 2019;59:13-21. doi:10.1016/j.ejmp.2019.02.014