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Enhancement Image Quality of CT Using Single Slice Spiral Technique

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Abstract — Today, CT is one of the most important methods of radiological diagnosis. It delivers non-superimposed cross-sectional images of the body, which can show smaller contrast differences than conventional X-ray images.

Single slice spiral technique is widely used in the equipment of medical application, the first spiral CT scanner was a Siemens SOMATOM Plus system. Many techniques have emerged as possible alternative to exact analytical solution including Filtered Back-Projection (FBP) algorithms. The Measurement of image quality is of great importance in the field of medical image applications such as X-ray tomography. In this paper, comparison of reconstruction FBP (analytical) with respect to the reconstructed image quality is presented using different types of window. Projections calculated using single slice spiral technique in FBP algorithm and applying common filters and proposed filter to get high quality and fast implementation of the reconstructed image.

Index Terms —Filtered back projection algorithm (FBP), Computerized Tomography (CT), Fan beam, Single slice spiral.

I-INTRODUCTION

Reconstructing 3D objects from cone-beam projections is a fairly recent accomplishment, in conventional fan-beam CT, individual axial slices of the object are sequentially reconstructed using a well-known mathematic technique filtered back-projection and subsequently assembled to construct the volume. However, with 2D X-ray area detectors and cone-beam geometry, a 3D volume must be reconstructed from 2D projection data.

II- SINGLE SLICE SPIRAL CT

The first spiral CT scanner in routine operation is appeared in 1989. In the patient coordinate system the X-ray source and the single row detector move on a spiral. In a single row scanner there are always projection data at z-positions $z(\varphi)$ close to the image plane at z_{image} as shown in the figure 1.

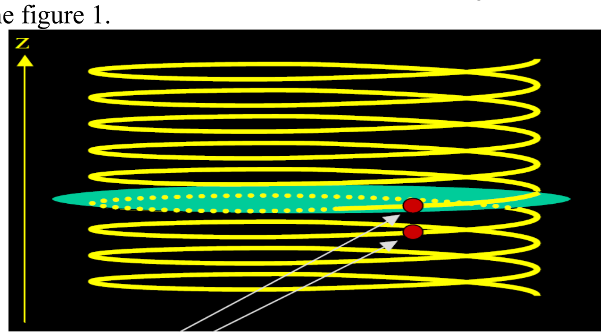


Fig.1. Single slices CT in spiral mode at z position

In practical helical beam X-ray CT machines, the source and array of detectors are mounted on a rotating gantry while the patient is moved axially at a uniform rate. Earlier X-ray CT scanners imaged one slice at a time by rotating source and one dimensional array of detectors while the patient remained static as shown in the figure 2. The helical scan method reduces the X-ray dose to the patient required for a given resolution while scanning more quickly. This is however at the cost of greater mathematical complexity in the reconstruction of the image from the measurements [1].

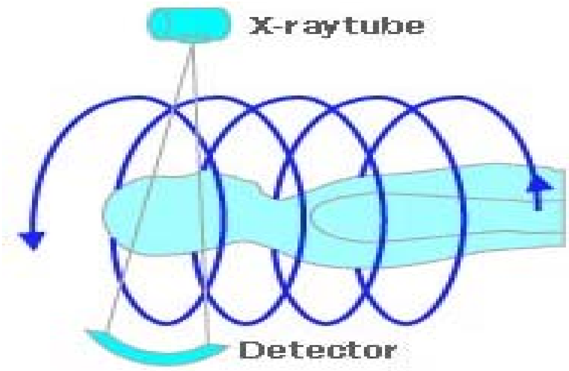


Fig.2. Illustration spiral around the body

The advantage of spiral CT in clinical use is complete coverage of organs in a single respiratory position, short scan times (resulting in fewer motion artifacts and a lower contrast medium requirement), additional diagnostic information due to improved resolution (thinner slices) and 3D visualization in routine operation, special cost-effective applications based on spiral CT, reduces the X-ray dose to the patient required for a given resolution, and overlapping reconstructions offer the advantage of better image quality due to lower noise and easier and more accurate diagnosis of small structures when the increment determines the distance between images reconstructed from a data volume. If an appropriate increment is used, overlapping images can be reconstructed. In sequential CT, overlapping images are obtained only if the table feed between two sequences is smaller than the collimated slice thickness. This, however, increases the patient dose. In spiral CT the increment is freely selectable as a reconstruction parameter, i.e. by selecting the increment the user can retrospectively and freely determine the degree of overlap without increasing the dose modern CT systems allow the reconstruction of slices with arbitrary increments. A clinical useful overlap is 30%–50% [2].

There is the clinical benefit of fast spiral scanning beyond the threshold of the breath hold time there is no huge clinical benefit foreseen for further reducing the scan time as shown in the Figure 3.

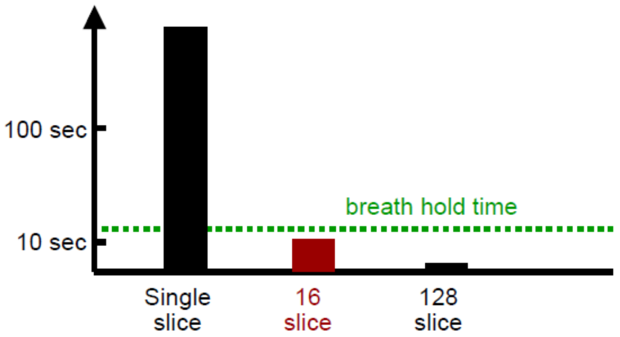


Fig.3. Spiral scanning beyond the threshold of the breath hold time

The single slice spiral beam reconstruction algorithm can be broken into three steps:

Step 1:

Applying of a smoothing (low pass) filter that can be used with Ramachandran-Lakshiminarayanan (Ram- Lak or R-K) filter, such as Gaussian, Kaiser, Cosine, Hamming, Hanning filters and proposed filter has the formula[3].

$$w[n]=\begin{cases}\cos^{5.3}(2n/M-1) & 0\le n\le M\\ 0 & otherwise\end{cases} \qquad \ldots(1)$$

Where M is window order

Filtering the projections uses Ramp filter in the frequency domain multiplied by window filter. The filtering process required for FBP algorithm amplifies the high spatial frequencies. To avoid the excessive noise amplification at these frequencies, a band-limited filter called Ram-Lak filter is often used in image reconstruction. Ram Lak filter is band limited ramp filter which belongs to its vendors Ramanathan and lakshminarayan. Usually, frequencies slightly below the cutoff frequency are unreliable and for reducing the high spatial frequency characteristics of image results sharp edges and sensitive to noise, smoothing window is used to reduce these artifacts. Many smoothing windows and functions are available for this application [3].

Step 2:

Get the projections for the first slice and pre-weigh the projection data according to

$$q'_{\beta}(\gamma)=q_{\beta}(\gamma)D\cos\gamma \qquad \ldots(2)$$

Where: β is the angle that the X-ray source makes with the reference axis, γ is the angle gives the location of a ray within a fan, and *D* is the distance of the X-ray source from the origin.

Step3:

Calculation of 1D Fourier transforms for each Projection $q'_{\beta}(\gamma)$ to arrive $Q''(w,\beta)$.

Step 4:

Multiplying each projection with the filter *H(w)* in Frequency domain

Step 5:

Finally, each weighted projection is back-projection by taking the inverse Fourier transform of the above result.

Then get the next slice for the object.

The final reconstruction formula for single slice spiral scanning becomes:

In this equation, *D* is the distance of the X-ray source to the iso-center, *L* is the distance from the X-ray source to the point of reconstruction (*x*, y), γ is the detector angle, β is the view angle, β_0 is the starting angle of the helical dataset, *π* is

$$f(x,y)=\int_{\beta_0}^{\beta_0+\Pi} L^{-2}d\beta \int_{-\gamma_m}^{\gamma_m} w(\gamma,\beta)q(\gamma,\beta)h''(\gamma'-\gamma)D\cos\gamma d\gamma \qquad \ldots(3)$$

the angular span of the dataset, *w*(γ,β) is the weighting function for back projection, *q*(γ,β) is the fan-beam projection , and *h*″(γ) is the filter function. The advantage of this approach is its computational efficiency and ability to minimize spatial resolution impact [4].

III. Multi Slice CT

Cone beam CT uses a 2D detectors (the increased number of detector rows can be used for) to obtain a 2D array of projection image, as show in Figure 4. Cone beam X-rays penetrate the whole target, forming a complete 2D projection imaging. All slices of projection data in one angle is acquired after just one shooting. For the parallel beam CT and fan beam CT, one slice of projection image is taken after a 360° rotation, and then the generator and detector (or patient) is moved to get the next slice of projection image. For cone beam CT, usually only one 360° gantry rotation is needed to acquire whole slices of projection images.

Obvious advantages of such a system, which provides a shorter examination time, include the reduction of image un-sharpness caused by the translation of the patient, reduce image distortion due to internal patient movements, and increased X-ray tube efficiency. However, its main disadvantage, especially with larger field of view FOVs, is a limitation in image quality related to noise and contrast resolution because of the detection of large amounts of scattered radiation, and the high cost with increasing the number of detectors [5].

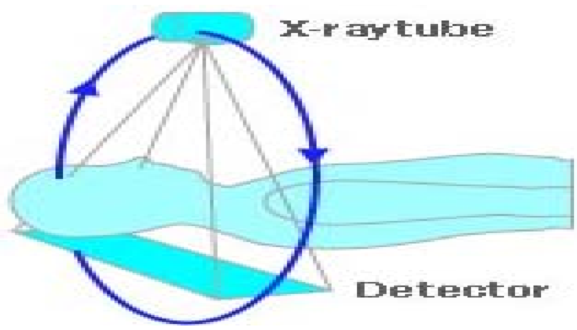


Fig.4. Cone beam geometry multi slice

IV. Result And Discussion

The computer implementation of the filtered back-projection algorithms using single slice CT in spiral geometry, as illustrated in the steps before. The 3D phantom (test image) used for generating the projections for different slices of the reconstruction image shown in figure 5, image size (128 x128x27).

The input parameter D-distance between source and the center of object, $\Delta\gamma$-the angular spacing of the beams is parameter with value 0.25 degree. D parameter must be large enough to ensure that the cone beam is outside of the image at all rotation angles and 360 projections with uniform increment of unity are used. The moving X-ray source and detectors to the next slice to obtain the projections at the next z position to another slice.
We used multi-detector CT (MDCT) scanners to acquire up to 27 slices, considerably reducing the scanning time compared with single-slice systems and allowing generation of 3D images at substantially lower doses of radiation than single detector fan-beam CT arrays.
It provides clear slices images of highly contrasted structures and is extremely useful for evaluating bone.

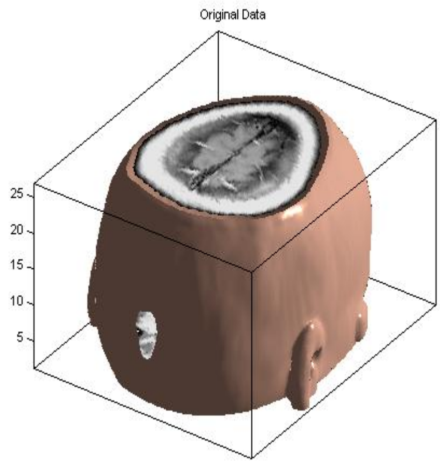
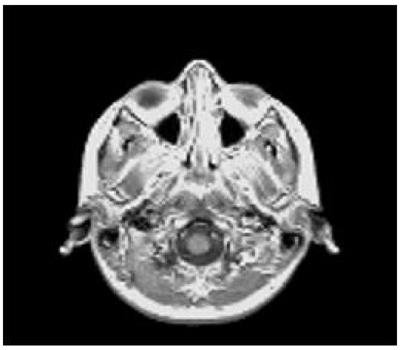
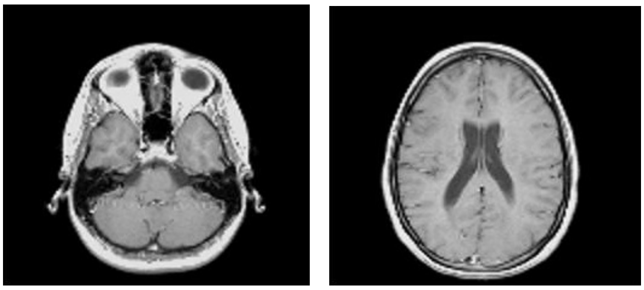


Fig.5. Three dimension head image

We using a MATLAB program to obtained a 2D slice from 3D test image and now choosing three slices as shown in the figure 6 and applying common filters and proposed filter on each slice to show the best filter can be using that gives excellent reconstructed image and fast implementation by FBP single slice CT in spiral mode cone beam algorithm. Figure 7 shown the sinogram of each slice we used,
We select the second, six, seventeen, slices and applying the Hann, Hamming, Kaiser, Cosine, Ram-Lak, Gaussian, and proposed filters to the filter back-projection algorithm as shown in the figure 6.



Z=2



Z=6 Z=17

Fig.6. different slices chosen along the Z-orientation of the original brain data

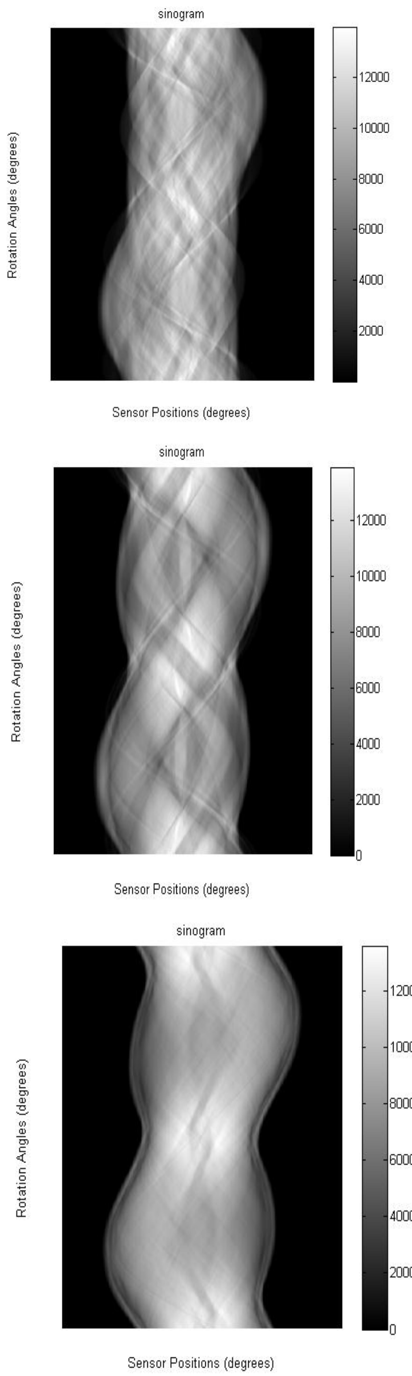


Fig.7. Sinogram of the slices z=2, 6, 17 along the Z-orientation

TABLE 1 Quality measurements of image reconstruction for second slice

Filter Type	SNR	RMSE	NAE	MD	Time (sec)
Ramp-Lak	12.9555	28.6902	0.1311	56.4062	46.596
Cosine	16.8622	18.297	0.1537	54.713	63.278
Gaussian	15.7667	19.876	0.15223	54.6332	69.767
Hamming	17.8658	16.3012	0.1262	48.2622	59.413
Hann	18.2589	15.587	0.1261	47.554	56.61
Kaiser	18.2923	15.5197	0.1262	47.0363	68.767
Proposed	18.1975	15.8722	0.1262	47.4966	50.668

TABLE 2 Quality measurements of image reconstruction for sixth slice.

Filter Type	RMSE	SNR	NAE	MD	Time (sec)
Ramp-Lak	28.4983	13.0138	0.1619	57.1775	46.577
Cosine	18.0337	16.9885	0.1537	54.713	62.053
Gaussian	19.0112	15.5564	0.1533	54.776	69.019
Hamming	16.0243	18.0146	0.151	53.7647	56.584
Hann	15.2954	18.419	0.1506	53.4674	59.389
Kaiser	15.3267	18.4524	0.1504	53.4136	66.840
Proposed	15.5923	18.252	0.1506	53.5818	52.365

TABLE 3 Quality measurements of image reconstruction for seventeenth slice.

Filter Type	SNR	RMSE	NAE	MD	Time (sec)
Ramp-Lak	13.0228	28.4688	0.1456	56.2411	47.691
Cosine	17.0092	17.8908	0.1393	51.5297	63.952
Gaussian	16.785	20.339	0.1391	51.1128	68.118
Hamming	18.044	15.9703	0.1375	50.2059	53.934
Hann	18.452	15.2374	0.1372	49.855	56.648
Kaiser	18.4867	15.1766	0.1371	49.6369	63.952
Proposed	18.2842	15.5346	0.1371	49.8562	53.588

Different quality parameters such as mean square error (MSE), signal to noise ratio (SNR), maximum difference (MD) and normalized absolute error (NAE) used to show the accuracy of reconstructed image with different filters
Maximum error it's the variation of the method of paired comparisons.

$$MD = \max\left(\left|\hat{f}(x,y) - f(x,y)\right|\right) \quad \ldots(4)$$

Signal to noise ratio is used to measure the difference between two images, It is defined by [6]:

$$SNR_{ms} = \frac{\sum_{x=0}^{M-1}\sum_{y=0}^{N-1} \hat{f}(x,y)^2}{\sum_{x=0}^{M-1}\sum_{y=0}^{N-1}\left[\hat{f}(x,y) - f(x,y)\right]^2} \quad \ldots(5)$$

If $\hat{f}(x,y)$ is considered to be the reconstructed of the original image *f(x, y)* and an error or "noise" signal *e(x, y)*.
Normalized absolute error NAE it's a numerical difference between the original and reconstructed image.

$$NAE = \sum_{x=o}^{M-1}\sum_{y=0}^{N-1}\left|\hat{f}(x,y) - f(x,y)\right| \Big/ \sum_{x=0}^{M-1}\sum_{y=0}^{N-1}\left|f(x,y)\right| \quad (6)$$

….

Where the image are of the size M x N, The root-mean-square error, e_{rms} between *f(x, y)* and $\hat{f}(x,y)$ is then the squared error averaged over the M x N array, or

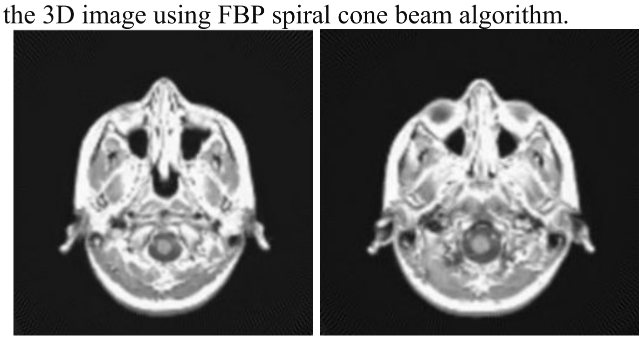
$$e_{rms} = \left[\frac{1}{MN}\sum_{x=0}^{M-1}\sum_{y=0}^{N-1}\left[\hat{f}(x,y) - f(x,y)\right]^2\right]^{1/2} \quad \ldots(7)$$

As shown in the table (1) for z=2 slice the value of SNR is 18.419 dB for Hann filter and SNR is 18.252 dB for proposed filter the value of this parameter gives better reconstructed image. The root mean square error, the difference between the original image slice and reconstructed one, as shown in table (1), It may be noted that as the quality of the reconstructed image improves and produce least error in the part of proposed, Hann, and Kaiser filters with values 15.2954, 15.3267 and 15.5923 respectively these values are very close together and the effect is not so clear on image reconstruction by comparing with other parameters. Maximum difference for z=2 is minimum using FBP with Kaiser, proposed and Hann with values 53.4136, 53.5818 and 53.4674 respectively. Normalized Absolute error given the minimum values when using Kaiser, proposed and Hann filters with values 0.1504, 0.1506 and 0.1506 respectively As shown in the table (1). The performance measurement gives the minimum reconstruction time by applying proposed filter with 52.365819 sec but Kaiser (66.840975 sec) and Hann (59.389967 sec).
Different quality parameters used to show the accuracy of reconstructed image with different filters as shown in the table (2) for z=6 slice the value of SNR is 18.452dB for Hann filter and SNR is 18.252dB for proposed filter the value of this parameter gives better reconstructed image.

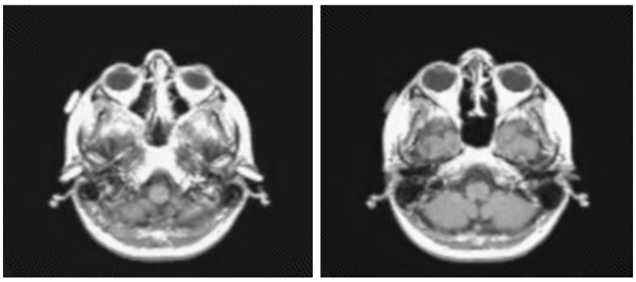
The root mean square error, as shown in table (2), It may be noted that as the quality of the reconstructed image improves and produce least error in the part of proposed, Hann, and Kaiser filters with values 15.5346, 15.2374and 15.1766 respectively these values are very close together and the effect is not so clear on image reconstruction by comparing with other parameters. Maximum difference for z=6 is minimum using FBP with Kaiser, proposed and Hann with values 49.6369, 49.8562and 49.855 respectively. Normalized absolute error given the minimum values when using Kaiser, proposed and Hann filters with values 0.1371, 0.1371, and 0.1372 respectively as shown in the table (2). The performance measurement gives the minimum reconstruction time by applying proposed filter with 53.58864 sec but Kaiser 63.95227 sec. and Hann 56.64804 sec.

For z=17 slice the quality parameter with different filters as shown in the table(3) the value of SNR is 18.2589dB for Hann filter and SNR is 18.1975dB for proposed filter the value of this parameter gives better reconstructed image. The root mean square error, as shown in table (3), It may be noted that as the quality of the reconstructed image improves and produce least error in the part of proposed, Hann, and Kaiser filters with values 15.8722, 15.587 and 15.5197 respectively. Maximum difference for z=17 is minimum using FBP with Kaiser, proposed and Hann with values 47.0363, 47.4966 and 47.554 respectively. NAE given the minimum values when using Kaiser, proposed and Hann filters with values 0.1262, 0.1262 and 0.1261 respectively, as shown in the table (3). The performance measurement gives the minimum reconstruction time by applying proposed filter with 50.66875sec but Kaiser 68.76743 sec and Hann 56.6168 sec.

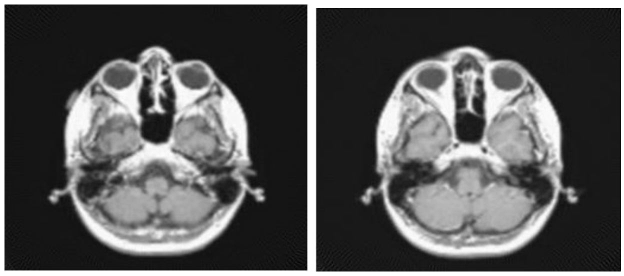
Result relating different quality-measures are listed in the tables for varies slices of the 3D head phantom in the table (1), table (2), and table (3) respectively, the value of SNR for proposed gives excellent reconstruction images as shown in the tables. MD is minimum using FBP with proposed filter. Normalized absolute error given the minimum values when using proposed filter. RMSE follow the same trend Keeping in the view the elapsed time recorded for each image reconstructed for the filters used and the quality of the image reconstructed the proposed filter appeared to give best output, time reconstruction by used this filter is less than using other filter that give high performance for reconstructed image filter and excellent quality. And as shown in the figure 8 used the proposed filter to reconstructed 27 slices along the Z-orientation of the 3D image using FBP spiral cone beam algorithm.



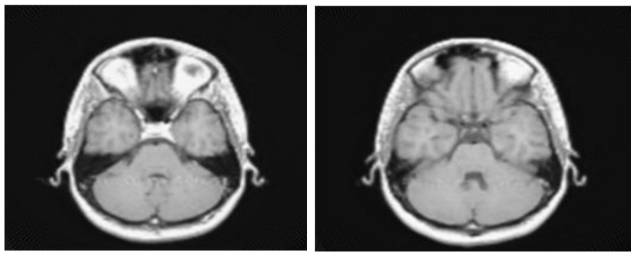
Z=1 z=2



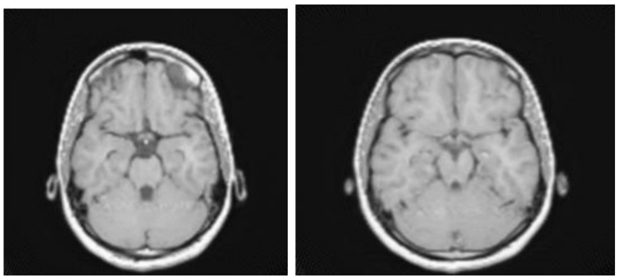
Z=3 z=4



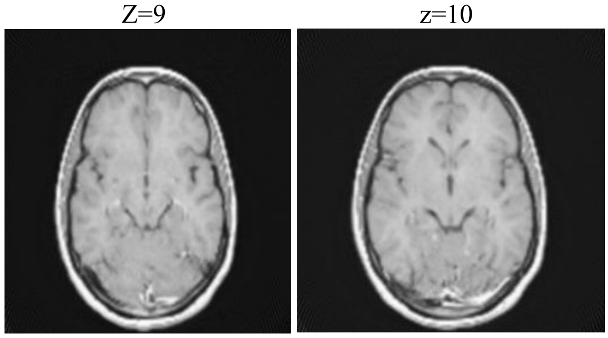
Z=5 z=6



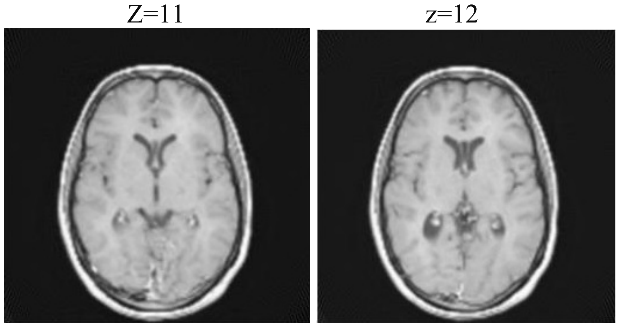
Z=7 z=8



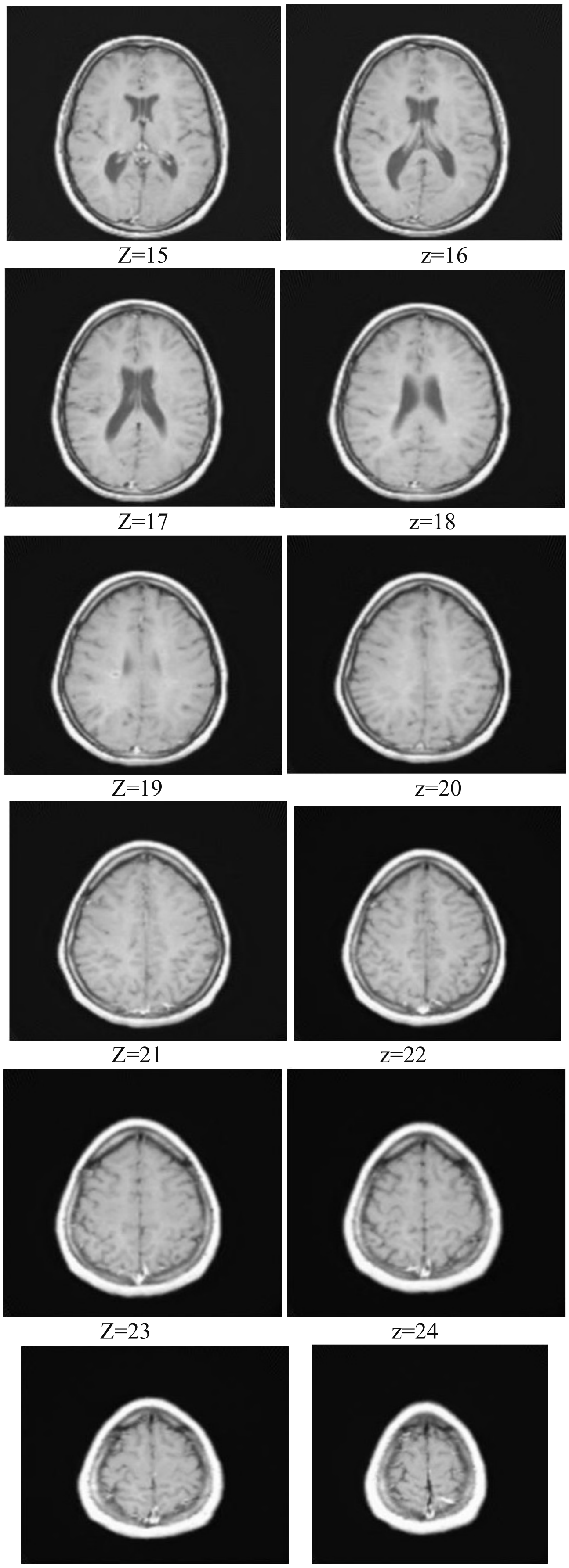
Z=9 z=10



Z=11 z=12



Z=13 z=14



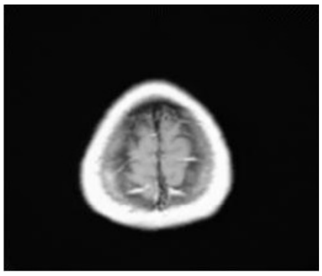

Z=27

Fig.8. Slices along the Z-orientation of the reconstruction image using FBP spiral cone beam algorithm with proposed filter

VI- CONCLUSIONS

The method that implemented can be used for image reconstruction in CT is cone beam single slice spiral geometry. In this work, objectively measurement by mean square error, signal to noise ratio, maximum difference, normalized absolute error and estimated time led to ability to compare the quality and performance for the tomographic reconstructed image. The main conclusion that can be deduced from the results obtained from FBP algorithm implemented of the cone beam single slice spiral reconstruction method that FBP algorithm with common filters (Hann, Hamming, Kaiser, Cosine, Ram-Lak, Gaussian, and proposed) applying by using cone beam single slice spiral geometry in three dimension head 3D phantom image size (128 x 128 x 27) and get the elapsed time recorded for each image reconstructed for the filters used and the quality of the image reconstructed. The proposed filter appeared to give an best output, time reconstruction by used this filter is less than using other filter that give high performance for reconstructed image filter and excellent quality.

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