Modelling the Siemens SOMATOM Sensation 64 Multi-Slice CT (MSCT) Scanner

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Abstract. Reconstructing large volumetric 3D images with minimal radiation dosage exposure with reduced scanning time has been one of the main objectives in the advancement of CT development. One of its advancement is the introduction of multi-slice arc detector geometry from a cone-beam source in third generation scanners. In solving this complex geometry, apart from the known vast computations in CT image reconstruction due to large CT images, iterative reconstruction methods are preferred compared to analytic methods due to its flexibility in image reconstruction. A scanner of interest that has this type of geometry is the *Siemens SOMATOM Sensation 64 Multi-Slice CT (MSCT) Scanner*, which has a total of 32 slices with 672 detector elements on each slice. In this paper, the scanner projection is modelled via the intersecting lengths between each ray (exhibited from the source to the detector elements) with the scanned image voxels, which are evaluated using the classical *Siddon's* algorithm to generate the system matrix, H. This is a prerequisite to perform various iterative reconstruction methods, which involves solving the inverse problem arising from the linear equation: $\mathbf{S} = H{\cdot}\mathbf{I}$; where $\mathbf{S}$ is the projections produced from the image, $\mathbf{I}$. Due to the 'cone-beam geometry' along the z-axis, the effective field-of-view (FOV) with voxel dimensions $(0.4{\times}0.4{\times}0.4)$ mm^3 is $512{\times}512{\times}32$ voxels. The scanner model is demonstrated by reconstructing an image from simulated projections using the analytic *Feldkamp-Davis-Kress* (FDK) method against basic iterative image reconstruction methods.

1. Introduction

Third generation CT scanners are currently widely being used clinically [1]. The manufacturers use a simple acquisition principle [2] with an x-ray focal spot source that rotates on a circular path around the isocentre of the gantry, and a cylindrical detector located on the opposite side with its centre axis positioned on the focal spot. This set up known as the 'rotate/rotate' geometry [3] allows the rotating gantry to be mounted with both the x-ray tube and detector revolving around the patient.

Obtaining volumetric 3D images has been one of the objectives in the advancement in CT imaging. From the utilisation of subsequent axial scans in 'step-and-shoot'/sequential mode, to spiral/helical imaging, the technology has evolved with the introduction of multi-slice detector modules that allows volumetric acquisition even during axial scanning [4]. This is useful, especially in dynamic volume imaging for volume perfusion studies [3].

Introduced in 2004, the Siemens SOMATOM Sensation 64 Multi-Slice CT (MSCT) Scanner is a scanner model that allows an acquisition of 64-slice simultaneously with an adaptive 32-row array detector design with a double z-sampling technique [3] using a periodic motion of the focal spot in the

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z-direction [5], which is also known as the z-flying focal spot (z-ffs). Nevertheless, its potential can still be enhanced. This paper provides the development of modelling the scanner as a prerequisite to perform various iterative reconstruction methods. Preliminary image reconstruction results are also presented to validate the model, using standard reconstruction techniques.

2. Methodology

The methodology adopted to model the MSCT scanner is based on a previous study in modelling a PET/CT scanner [6], divided into four aspects. Firstly, the geometry of the scanner gantry is modelled to simulate the rays transmitted from the source to the multi-slice 2D detector arrays. Secondly, the geometry of the phantom image is modelled to establish the field-of-view (FOV) across the axial and sagittal planes to accommodate the cone-beam geometry of the scanner. Thirdly, the intersection lengths between the rays and each of the image voxels are traced to generate the system matrix that models the scanner. The last aspect of the methodology is performing image reconstruction based on the model to validate it. This part involves reconstruction using the analytic Feldkamp-Davis-Kress (FDK) [8] method to compare with the iterative Algebraic Reconstruction Technique (ART) [7] method. The model of the scanner, and image reconstruction were performed in MATLAB.

2.1. Scanner geometry

The two main components of the Siemens SOMATOM Sensation 64 MSCT scanner are the source and detector. The distance between the source and detector is 1040 mm, and the distance between the source and isocentre is 570 mm.

The source uses the Siemens STRATON X-ray tube technology [5] that allows a user to choose between acquiring 32 or 64-slice during scanning. This is based on the availability of the z-flying focal spot (z-ffs) option that utilizes a periodic motion of the focal spot in the longitudinal z-direction. Nevertheless, the scope of this paper is limited to modelling 32-slice with the z-ffs option switched off [2] for simplicity of the model.

The detector [3] is made of scintillator ceramics known as Ultra-Fast Ceramic (UFC), and is arranged in 2D modules. Each module consists of 40 × 16 detector pixels along the longitudinal and azimuthal directions (i.e. z-axis and x-y plane). Along the longitudinal direction, the detectors are arranged in an adaptive array with 32 central rows and 4 outer rows on both sides to accommodate for different collimated slice widths. In this model, only the central rows with a 0.6 mm slice width measured at the isocentre is considered. Thus, the total length at the isocentre is 32 × 0.6 mm = 19.2 mm. Taking the ratio between the distances of the source to the isocentre and detector locations, the actual detector width is evaluated as 35.03 mm, with each slice width approximately 1.09 mm. This produces a cone angle, $\Gamma \approx 1.93°$ along the z-axis direction.

Across the x-y plane, 42 detector modules are arranged in an arc with its centre axis positioned at the location of the focal spot source. Each module is separated by an equal-angle, $\Delta\varphi \approx 1.24°$ at the source with the length of each module in the azimuthal direction is 22.51 mm, given that the fan angle, $\Phi \approx 52°$ [2]. Figure 1 depicts an illustration of the fan-beam and cone-beam geometries.

The basis of CT is the detection of attenuated x-rays transmitted from the source by each of the scintillator detector elements. The capability of a scanner depends on the number of x-rays that the detector can capture, thus is reflected by the number of detector elements available. For each longitudinal row, the number of detector elements is 42 × 16 = 672 elements across the azimuthal direction. Therefore, for this MSCT scanner, the total number of detector elements is 672 × 32 = 21,504 elements.

2.2. Image geometry

A realistic 3D head phantom with dimensions: 512×512×512 voxels is used, as shown in figure 2. Assuming each voxel length is 0.4 mm, the total volumetric dimension lengths would be 204.8 mm each. This is acceptable since the MSCT scanner geometry [5] has an aperture length of 700 mm, and a range of reconstruction field-of-view (FOV) between 50 – 500 mm. With these dimensions, the head

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phantom would entirely be covered by the fan-beam geometry with a FOV of 204.8×204.8 mm^2 on the axial plane

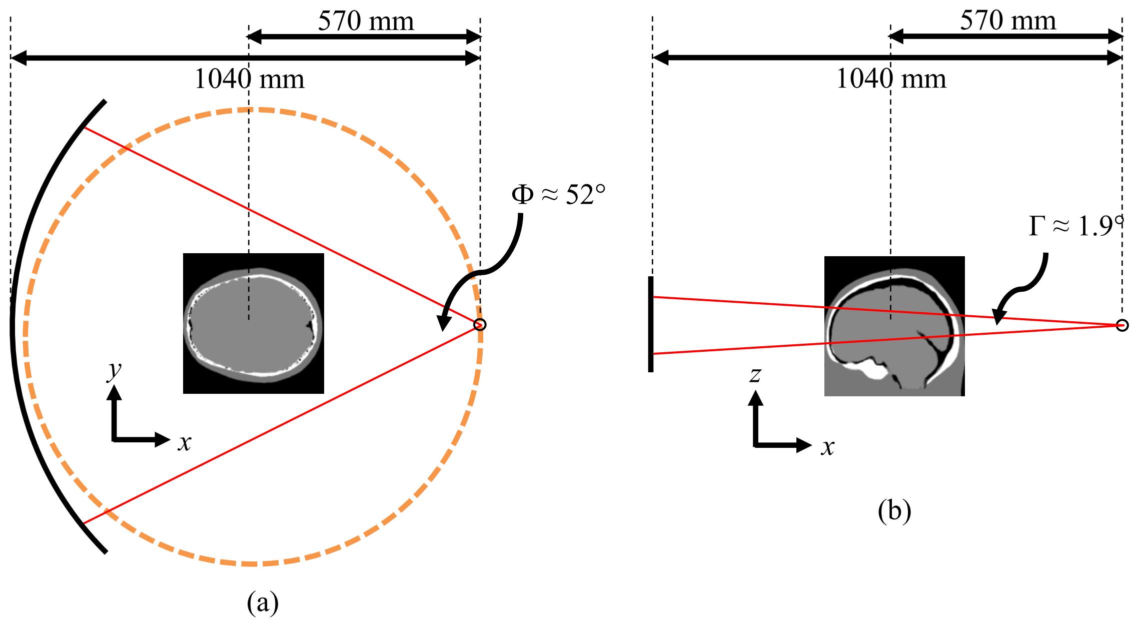

Figure 1. The source-detector configuration at view angle θ = 270° depicting the: (a) fan-beam, Φ ≈ 52°, and (b) cone-beam, Γ ≈ 1.9° geometries from axial and sagittal planes respectively.

Multi-slice detectors have proved to be advantageous due to its simplicity and robustness especially in reducing the radiation dose during scanning [9]. However, the cone beam angle from this geometry has its limitations such as image artefacts arising due to several factors, such as 'axial truncation' that limits the FOV, since the object is not entirely covered in the longitudinal direction. Discarding the slices that are not entirely covered by the cone angle, only 32 pixels along the z-axis are considered. This leads to a FOV of 204.8×12.30 mm^2 on the sagittal plane. Therefore, the head phantom's 3D FOV is 512×512×32 voxels that correspond to 204.8×204.8×12.30 mm^3.

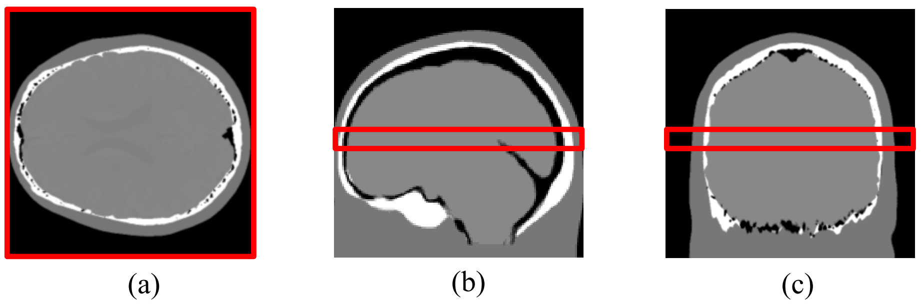

Figure 2. Realistic head phantom images used from different plane of views: (a) axial, (b) sagittal, and (c) coronal planes. The field-of-view (FOV) of the scanner is evaluated as 512×512×32 voxels as highlighted.

2.3. Ray tracing

Transmitted x-rays that penetrate a scanned object are attenuated to a degree depending on the type of tissue as well as its thickness located within the path of each ray. In this model, the radiological path of a ray from the source through the image voxel arrays onto the detector elements, are evaluated by tracing the intersection lengths of each ray with the voxels using Siddon's algorithm [10]. Assigning each intersection lengths parametrically with a designated coordinate based on the ray and voxel number will eventually generate the system matrix, H that models the scanner for one view.

Each view will produce a projection that contains all of the generated data [11] by the transmitted x-ray from that source location. The collection of projections from all views can also be referred to as the sonogram, **S**. Minimizing the total number of projections, i.e. number of views used in a CT scan is important [12] as it will affect the radiation dose. The number of required projections is highly dependent on the scan protocols that varies for different organs [13]. A total of 360 scan views with a 1° increment around the head phantom in a circular trajectory are used in this model, with the system matrix for each view is generated and stored individually to enable different number of views to be simulated repeatedly.

The total size of the system matrix depends on the total number of rays × total number of image voxels. For 360 views, the total number of rays, $P = 360 \times 21{,}504$ rays = 7,741,440 rays, and the total number of image voxels = 512×512×32 = 8,388,608 voxels. Therefore, the total size of the system matrix for this model, $H = [7{,}741{,}440 \times 8{,}388{,}608]$.

2.4. *Image reconstruction*

Two methods of reconstruction are considered here which are: analytic and iterative reconstruction methods. Each of the methods is briefly described below.

2.4.1. Analytic reconstruction. It is known that 3D cone-beam reconstruction is a challenging problem to be solved analytically. A popular method in solving this geometry is the Feldkamp-Davis-Kress (FDK) method [8]. This method re-writes the Radon transform used in the popular Filtered Back Projection (FBP) approach in solving 2D fan-beam reconstruction into a convolution and backprojection form that is extended to the 3D cone-beam case [14].

Nevertheless, analytic reconstruction methods are well-known to be customized for idealized models that assume the data samplings are of continuous in measurements when in reality, the projections are actually discretely sampled onto the detector module elements. In practical CT applications, the projections are generated from a finite set of measurements depending on the number of detected rays, which causes the reconstruction to be ill posed.

2.4.2. Iterative reconstruction. Unlike the analytic approach, iterative reconstruction methods treat the image reconstruction as a discrete problem from the start [11]. It focuses on solving the linear system equation representation of the problem: $\mathbf{S} = H \cdot \mathbf{I}$, with **S** being the vector representation of projections generated from the image vector representation, **I**, and H is the system matrix of the scanner. The Algebraic Reconstruction Technique (ART) is used [14] to iteratively reconstruct the image (**I**) using a uniform image ($\mathbf{I}_0$) as an initial guess and recursively computing a sequence of estimates ($\mathbf{I}_k$) subsequently until a maximum number of iterations, k, is satisfied. Based on Kaczmarz's algorithm, the inductive step in ART for determining the next image estimate ($\mathbf{I}_k$) is shown in equation (1) below:

$$\mathbf{I}_{k+1} = \mathbf{I}_k + \frac{S_n - \mathbf{h}_n^{\mathrm{T}} \cdot \mathbf{I}_k}{\left\|\mathbf{h}_n\right\|^2} \cdot \mathbf{h}_n \tag{1}$$

where $\mathbf{h}_n$ refers to the n-th row of the system matrix, H and $n = k \bmod P + 1$.

3. Results and discussion

The evaluation results of the MSCT scanner model are presented based on the results from the head phantom image reconstruction using the two methods described in the previous section. Each of the reconstructed images is quantitatively assessed using the normalized root mean squared difference (NRMSD) value. Although the reconstruction involves a 3D head phantom with dimensions 512×512×32 voxels as shown in figure 2, only the 2D central slice of the axial plane is presented here.

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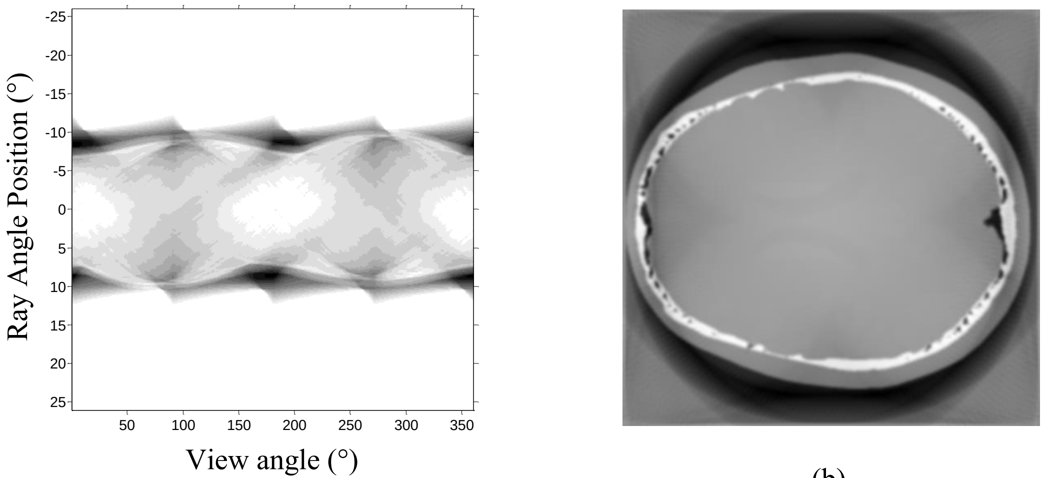
3.1. Analytic reconstruction

Figure 3(a) shows the synthetic projection of the central slice, generated using the system matrix, *H*, with 360 views as described in section 2.3. Figure 3(b) shows the reconstructed image using the analytic FDK method with the NRSMD = 19.1%. Compared to the original head phantom image shown in figure 2(a), it can be seen that there are numerous artefacts and distortions, due to the limited number of views.

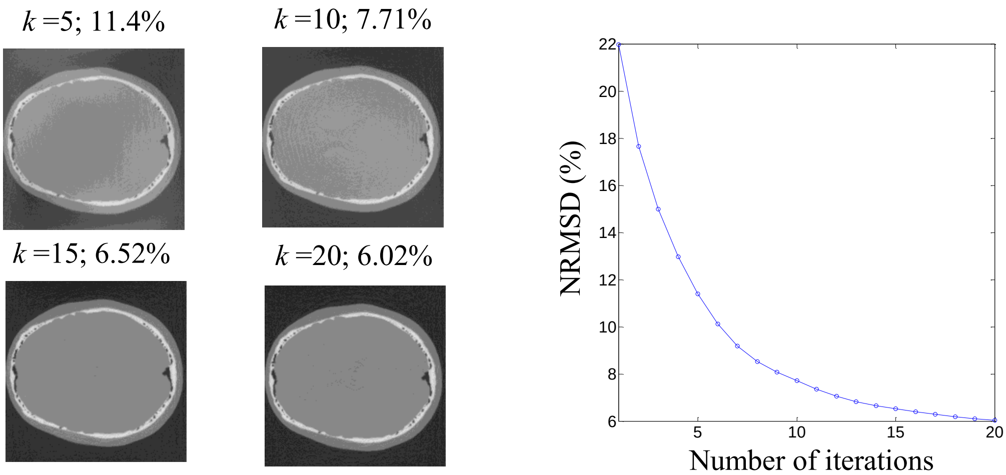


(a) (b)

Figure 3. (a) A synthetic 360-view projection of the central slice, generated using the obtained system matrix, *H*, from the head phantom image used in figure 2, and (b) an axial slice of the reconstructed image using the analytic FDK method, with NRMSD = 19.1%.

3.2. Iterative reconstruction

Using the iterative ART method based on Kaczmarz's algorithm, a maximum number of 20 iterations were used to reconstruct the same projection shown in figure 3(a), along with the intermediary iterations at iteration 5, 10 and 15 as presented in figure 4(a). The NRMSD values for all of the reconstructed images using this method are recorded and compiled as shown in figure 4(b).



(a) (b)

Figure 4. (a) Reconstructed images using the iterative ART method, at different values of iterations with their respective NRMSD values. (b) Decrement of the NRMSD value as the number of iterations is increased.

It can be seen that as the iterations progress, the percentage NRMSD between the reconstructed and actual images using the iterative ART method reduces greatly, with a higher accuracy compared to the analytic FDK method. This agrees with the fact that for a sparse number of views, iterative reconstruction methods are able to produce better quality images, indirectly minimising the need to high radiation dose during imaging.

4. Conclusion

The Siemens SOMATOM Sensation 64 Multi-Slice CT (MSCT) scanner was successfully modelled, with its system matrix for 360 views, H generated.

This will encourage more medical imaging related researches to be carried out with clinical images available from the scanner. An improvement on the model would be to incorporate the FFS option into this model, allowing more image slice acquisition for a single view position. Apart from that is the incorporation of noise into this model.

5. References

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