Projection Data and Scanograms in X-ray Computed Tomography

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Abstract

Adaptive sampling of data in x-ray computed tomography may be used to reduce patient dose and scan time. A medical Hitachi W1000 CT system is to be used for adaptive sampling studies, where the source of projection data will be both raw data and that obtained from scanograms (essentially digital radiographs). The geometrical distortion of the scanogram has been characterised with both unusual and non-medical applications in mind, and the conventional radiograph is compared with the scanogram. Data obtained from the scanogram requires correction prior to use as projection data for image reconstruction and the method of correction is described. Images of several types of rotationally symmetric phantoms have been successfully reconstructed using both types of data. The reconstructed images are very sensitive to scanner asymmetry in projection data.

1. Introduction

Adaptive sampling is a technique commonly used to reduce computation time where large amounts of data have to be processed, for example, image compression (Liu and Hayes, [4]), image synthesis (Kirk and Arvo, [3]) and image scanning (Lo Presti et al, [5]). Adaptive sampling of data in x-ray computed tomography (CT) is of interest as some x-ray measurements routinely taken in a normal scan, contain little additional information and therefore may have limited effect on the image. If such data could be omitted from the data set then it is possible that both patient dose and scan time could be reduced. Kachelriess et al [2] have applied non-linear adaptive sampling to helical and conventional CT scanning for this purpose. Apart from normal slice scans there is also interest in reducing the x-ray dose to patients undergoing scanograms (Perisinakis et al, [6]), particularly in cases where frequent and regular imaging is required (Romanowski et al, [8]). The main problem is that it is difficult to decide what constitutes information and which data to omit. It is necessary, therefore, to find suitable criteria for assessing which ray-sums (or data points) should be omitted. Alternatively, if there is some limitation or restriction on the set of projection views that it is possible to take, then it is necessary to decide which object orientations will be best imaged by the given set of views. It is our objective to adapt a commercial CT system for these studies.

One way these questions may be investigated is to selectively remove certain classes of x-ray measurements from the input data prior to reconstructing the image and then to assess which classes of data are important and which do not contain essential information for the CT image reconstruction. For this purpose, full control over the acquisition of the data is required.

For the Hitachi W1000 scanner, in a normal slice scan, the fan-shaped x-ray beam is measured by 768 detector elements. These elements are read at each of 952 angular positions (for the shortest, 1.2 s, scan) of the x-ray source and the resulting 731,136 data points are used to reconstruct the slice, or cross-sectional CT image. Each data set, consisting of all the detector measurements across the fan beam, for one angular position of the source, is termed a projection. It is not possible to modify the data acquisition from the data that constitutes each projection during a normal scan obtained from commercial CT systems, therefore, alternative methods of obtaining controlled projections must be found. Two possible alternative sources of projection data are:

- scanograms, normally used for positioning patients prior to taking slice scans and,
- raw data from the x-ray detectors.

Data of the types described have been used in the present investigation to form images of several types of phantom (described below).

X-ray CT scanners are designed specifically for forming slice images of the internal organs of human patients, but they can also be used for industrial non-destructive inspection or they may be used in unconventional medical applications. Burch [1] has used x-ray CT techniques for the measurement of density variations in powder compacts but points out that beam hardening and diffuse scattering complications have to be taken into account. The scanogram in medical CT imaging is normally used to position the patient relative to the scanner prior to taking slice scans and to plan the positions and number of the slice images to be taken, but there are other applications of the scanogram. Sivachenko et al [9] have used scanograms to assess the weight of the liver and other organs as an aid to diagnosis of disease. Wade [10] has used lateral or anterior-posterior (AP) scanograms for the measurement of distances within the pelvis and has discussed the accuracy of such measurements. Zonneveld [12] has discussed the general technique, geometrical and timing considerations and medical applications of the scanogram including radiotherapy planning and as a diagnostic tool. Because of the potential use of scanograms for non-medical or unusual purposes, it has become important to re-examine some of the characteristics of this procedure to assess whether it can be used effectively for other applications.

The purpose of this paper is, then:

- to demonstrate that CT slice images can be successfully reconstructed from projections generated using scanogram and raw data for future adaptive sampling studies,
- to define the geometrical characteristics of the scanogram for potential use in unusual or non-medical applications, and
- to investigate the properties of images of rotationally symmetric phantoms reconstructed using replicated single projections in preparation for adaptive sampling studies.

2. The Scanogram

The scanogram is similar to a conventional planar radiograph such as a chest x-ray but with some major differences. The scanogram is obtained by moving the patient smoothly through the x-ray beam, by traversing the patient table, such that the organs of interest pass through the beam and the image appears on the scanner screen in real time. An additional difference between a radiograph and a scanogram is that the x-ray beam in a radiograph diverges axially as well as laterally whereas in a scanogram, axial beam divergence is limited to only a few degrees by a collimator, and since the axial position of a point on the image is determined directly by the position of the patient table at the time at which that point is imaged, there is negligible stretching or distortion of the image in a scanogram in the axial direction.

The scanogram and the radiographic image of an object are superficially similar. However, the different types of distortion introduced by the two processes are quite marked, as demonstrated by Figures 1 and 2 for the same object in the same position. The object used was a calibrator designed for calibrating image intensifiers. It consisted of a flat perspex sheet 3 mm thick with 2 mm diameter steel ball bearings embedded to form a square array of 25 x 28 rows with adjacent balls being 15 mm between centres. In both figures, the calibrator was mounted at an angle of 18° to the vertical. It is evident that the differences between the two processes of acquiring a scanogram and a radiograph produce different geometric distortions of the image relative to a parallel beam view. If the scanogram is to be used as a source of projection data, or if it is to be used in unconventional applications, it is necessary to compensate for these distortions and other variations in the data due to the signal processing chain.

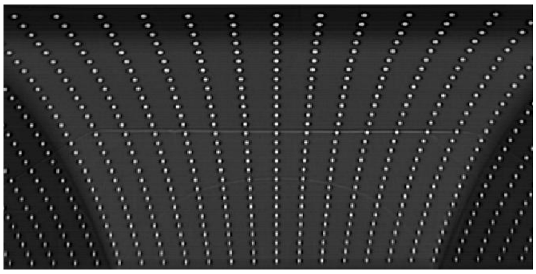

Figure 1. A scanogram of the ball calibrator.

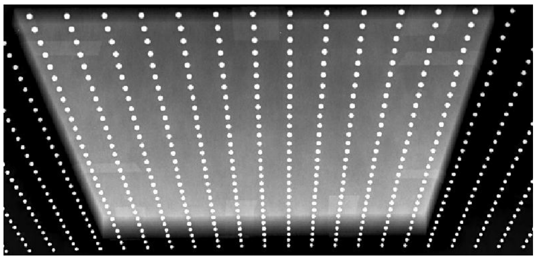

Figure 2. A conventional radiograph of the calibrator shown in Figure 1 standing at the same angle.

The CT processor applies filtration to the scanogram data to enhance the image contrast of bones and other organs to assist in the accurate positioning of the patient. For the purposes of the present investigation, however, the effect of this filtration was minimised by choosing the least active filter.

3. Raw Data

The second method of obtaining CT projection data is to acquire the data directly from the x-ray detectors. This can be done be accessing the so-called raw data. The scanner measures the x-ray intensity using a high-pressure xenon gas ionisation chamber partitioned into 768 independent detector elements across the fan beam. The elements are arranged on a circular arc centred on the x-ray focal spot and each element subtends an angle of approximately $0.06\,^\circ$ at the source, representing a width near the object centre of about 0.64 mm at mid-plane in the scanner aperture. The processor measures the x-ray flux incident on each element, or channel, in a given time, as an electric current, but several instrumental corrections are applied to this before the signal may be accessed as raw data. Measuring the raw data after the x-rays pass through a test object, it is found that the data values increase as x-ray attenuation increases and, typically the values range from zero for air (or no object) to more than 20,000 for several mm of lead.

4. Experimental

All of the experiments carried out using the Hitachi W1000 scanner in this series were obtained using an axial beam width (slice thickness) of 1.5 mm, a scan time of 1.2 s (except for scanograms where the scan time depends on the length scanned), a tube current of 50 mA and a tube voltage of 100 kV. The CT images reconstructed from projections generated from a scanogram or raw data used a third generation (rotate/rotate) calculation based on the summation convolution back-projection method (Wells et al 1994) with the Ram-Lak filter (Ramachandran and Lakshiminarayanan 1971).

Conventional radiographs were taken using a Watson Victor radiographic system with a tube current of 50 mA, a tube voltage of 70 kV, and an exposure time of 0.1 s. Agfa ADCC MD10 image plates were used for radiographic exposures and processed using an Agfa Musica Solo imaging system.

5. Characterisation

5.1. Raw Data

If the raw data is to be used for imaging purposes, then it must be fully characterised in terms of both geometry and magnitude. Stationary scans were used to generate projections when using raw data. During a normal rotating scan, the x-ray detectors are read by the processor at a series of times (or pulses) while the source and detector rotate synchronously around the object. The data from each of the 768 detector elements at any given pulse form a projection. In a stationary scan, the source and detector do not rotate. The same number of projections will be obtained but all of the angular projections will be identical. Accordingly, pulse 200 was arbitrarily selected and used throughout for all raw data measurements. Given the height of a horizontal plane relative to the centre of the scanner aperture, the lateral position of any point on this plane may be calculated given the channel number used to image it. From this it is possible to accurately determine the lateral position of any point in the object as long as its height relative to the centre of the aperture is known, and as long as the channel number imaging the point can be identified. Using the image intensifier calibrator, this procedure was used to determine the position of a series of balls and it was possible to locate them to better than ± 0.8 mm.

As regards the raw data magnitude, the raw data value obtained by reading each channel after a scan is a large integer for strongly attenuating objects and approximately zero for air (or no object). This is proportional to $-\ln(I/I_0) = \int \mu_e dt$, the ray-sum for the ray-path in question, where μ_e is the effective x-ray linear attenuation coefficient for the material (the radiation is not monochromatic) and dt is the elemental thickness. For a homogeneous object, the integral becomes $\mu_e t$ where t is the total thickness of the object. Using six detector channels, three on either side of centre, the raw data value for various ray-paths through a series of phantoms was measured and the average value per channel was found to be directly proportional to the ray-path length t, confirming that the measured data value is directly proportional to $\int \mu_e dt$ as expected.

5.2. Scanogram Data

Similarly, both the geometry and magnitude of the scanogram data must be characterised. The CT software has an on-screen distance measurement function. This function is actually intended for use in measuring distances and angles in a normal slice scan and care must be exercised in using it in scanogram mode. The ball calibrator was positioned at mid-plane with a ball centred as accurately as possible. The slice thickness selected was 1.5 mm so that each ball, being 2 mm in diameter, would virtually span the full beam thickness to ensure that the presence of a ball would show up strongly in the raw data. Taking a series of stationary scans, the calibrator was moved until the centre ball was accurately aligned with the centre pair of channels to within 0.3 mm. This ensured correct alignment rather than relying upon the projector beams.

A series of calibration curves was produced comparing the lateral distance of each ball from centre, measured using the distance function, with the true distance as determined using the calibrator. These curves were measured as a function of table height relative to the centre of the aperture and it was found that the dimensional error increases, in the worst case, to approximately 45 mm for a ball near the edge of the image at the lowest table position. The error is nearly zero at mid-plane. Further, the dimensional error is positive below the mid-plane and negative above. The table position at which the error will be near zero was determined by measuring 20 ball spacings at random positions on the image for each of several table heights and plotting the mean apparent ball spacing against height (see Figure 3). As can be seen from Table 1, the true ball spacing of 15 mm occurred near mid-plane.

Table 1. Apparent ball spacing vs table height relative to centre. (True ball spacing 15 mm.)

Table Height Relative to Centre (mm)	Measured Ball Spacing (mm)
+175	11.8 ± 0.6
+80	13.1 ± 0.7
0	15.0 ± 1.1
-80	17.4 ± 0.8
-175	21.5 ± 1.2

It is clear that, in scanogram mode, the processor images every point within an object as if it is located on this zero plane and its lateral position given by the distance function will be that of a point on the object at mid-plane being imaged by that particular channel.

The equation for the lateral position, l, of a point at height, h, above (positive) or below (negative) mid-plane using the imaging channel number, n, is given by

$$l = (R - h)\tan\left(\frac{n - n_m}{N} \times 23.3\,^\circ\right)$$

where R is the vertical distance from the focal spot to the mid-plane, 606.3 mm, n_m is the channel number corresponding to centre (in our case 384.5), and N is the range of channels from edge to centre of the fan (in our case 380.5). The angle $23.3\,^\circ$ is the half-angle of the x-ray fan beam.

Scanogram data has not yet been characterised in magnitude except for an examination of the effects of applied filters, the effects of which cannot be entirely eliminated. Scanogram data always has one of five filters applied to it in the W1000 scanner. These filters are called: head low contrast, head spatial, body low contrast, body spatial and bone edge enhancement. Scanograms of a 5 cm diameter uniform polyethylene cylinder were taken with each of these filters in turn and the scanogram profiled. It was found that all except filter 2, a low pass filter, produce a large negative excursion in magnitude at the edges of the object, the purpose of which is to make the image boundaries of bones and other organs more distinct. Filter 2 has a profile closest to that obtained using the raw data, therefore, all scanogram images were prepared using this filter.

5.3. Rotationally Symmetric Phantoms

To compare the two sources of projection data, namely scanogram and raw data, for subsequent image reconstruction, it was necessary to use rotationally symmetric phantoms as, having only one projection, it is necessary to duplicate it many times along the projection axis of the sinogram to generate an image.

Phantoms used were 230 mm diameter discs of:

- medium density fibreboard (MDF), and
- polyethylene.

Figure 3. Castellated MDF phantom; normal slice image.

The polyethylene phantom was a standard cylinder, 40 mm thick, used for routine calibration of the scanner. The wooden phantom was a cylinder, 18 mm thick, with five rectangular section circumferential channels machined into one side to form a castellated surface. The channels had centre radii ranging from 15 to 95 mm and each was 10 mm wide and 9 mm deep. This permitted either the castellated side or the plain cylindrical side to be used for imaging since the x-ray fan beam was only 1.5 mm thick. The purpose of the circumferential channels or castellations was to provide some edge features that could be used to compare the properties of the two different sources of projection data. Figure 3 shows a slice image of the castellated side of the MDF phantom, obtained using a 1.2 s scan and the scanner's normal reconstruction process. It gives a good representation of the physical appearance of the phantom.

5.4. Reconstructed Images

Image profiles using scanogram and raw data were obtained for each of the phantoms listed. Data was collected at 185 angular positions across each phantom in the case of the raw data, and 207 positions in the case of the scanogram data, to form the single projection data set. Since the raw data comes from detector elements which are equally spaced in angle across the fan beam, the raw data may be used directly as input to the fan beam reconstruction program. However, scanogram data points are not equally

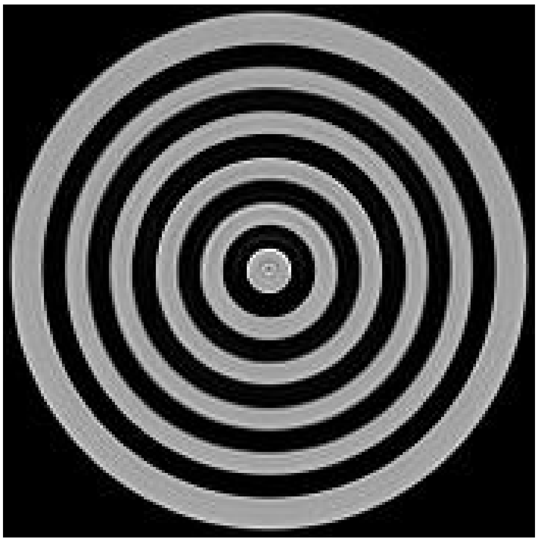

Figure 4. Castellated MDF phantom; projection generated from scanogram data.

spaced in angle across the fan beam, therefore, this data must be redistributed to make it suitable for input to the reconstruction calculation. The redistribution was done by taking a ray-path in the fan beam, calculating the corresponding lateral position at mid-plane in the object, then finding the magnitude of the profile line at that point by linear interpolation between adjacent measured points. This process was repeated for ray-paths at equal angular steps across the fan beam. Each angular step was taken to be the equivalent of approximately two detector elements which is the same angular interval used in obtaining the raw data. As a result of this procedure, the number of data points actually used in the reconstruction calculation for scanogram data was actually 183 compared with the 185 points used for raw data. The number of (replicated) projections in all cases was 360, representing an angular step between projections of $0.5\,^\circ$.

Figures 4 and 5 show images of the castellated MDF phantom obtained using a single projection generated using raw data and using scanogram data respectively, the scanogram data having been redistributed to make it equiangular across the fan beam as described previously. The most distinctive difference between the normal slice images from the CT scanner and those reconstructed from a single replicated projection is that the latter are smeared in the circumferential direction, reminiscent of lathe-turned wood, as might be expected because each projection used to generate the images is identical. Another noticeable difference is the

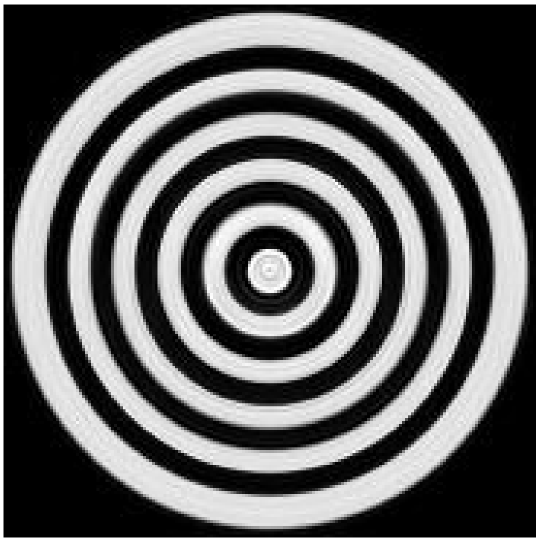

Figure 5. Castellated MDF phantom; projection generated from raw data.

conical depression at the centre of the raw data generated images which is not present at all in the normal slice images nor on the phantom itself. In addition, the image of the plain phantom (not shown) exhibits fine vertical streaking. This streaking is not a result of the printing process but is present on the image. The conical feature and the streaks are artefacts of the reconstruction process when using a single replicated projection. Imaging in this manner, is extremely sensitive to asymmetry as will be explained.

Figure 6 shows an image of a uniform MDF phantom generated from scanogram data. A close examination of this figure shows that, although near perfect rotational symmetry of the image might be expected from the manner in which it was generated, semicircular bands at about mid-radius differ, within the image, between the upper and lower half planes. The reason for the conical depression, the vertical streaks and the semicircular bands are due to asymmetry in the setting of the phantom relative to the fan beam. Any small asymmetry in projection data will manifest itself as an inconsistency when the projection angle is turned through $180\,^\circ$ to represent a view of the same phantom from

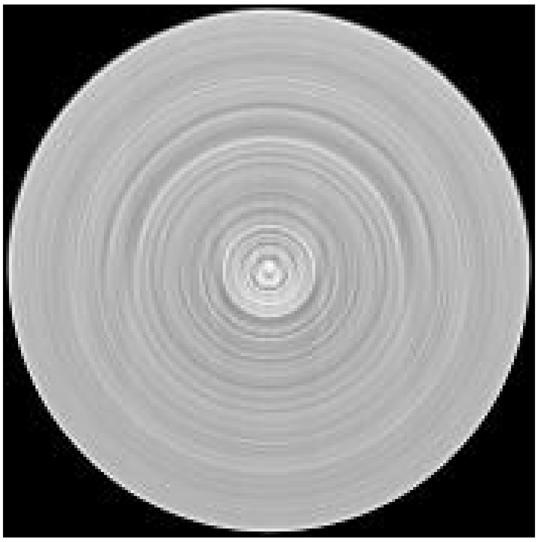

Figure 6. Plain MDF phantom; projection generated from scanogram data.

the opposite direction.

It was demonstrated that the semi-circular bands, the depression in the centre and the feint vertical streaks were due to mis-setting errors or natural asymmetry in the data by imaging a perfectly symmetrical data set. This data set was artificially generated by reflecting the data on one side of one of the projection datasets about the centre point to produce the data on the other side. The three features disappeared, producing a perfectly symmetrical image.

6. Conclusion

It has been demonstrated that CT slice images can be successfully reconstructed using projections generated from both raw and scanogram data. The object must be accurately centred in the aperture of the scanner. Any deviation from symmetry produces strong streak artifacts in the reconstructed image. It is necessary to correct for the non-linear distribution of scanogram data across the fan beam in the manner described in Section 6.2. Raw data does not require this correction.

The differences between scanogram and conventional radiographic images have been described. The scanogram image is produced by the processor in such a way that every point within the object is imaged as if it were located at the mid-plane of the object circle on the ray-path corresponding to the detector element imaging that point. Because of this, distances between two points measured by the distance function provided with the scanner software will only be accurate if the two points are located in the mid-plane of the image circle. Above and below this plane the lateral distances are given by the equation in Section 5.2 (for a posterior-anterior scanogram) but the height of the points above or below the mid-plane must be known.

It is proposed, next, to examine the relationship between the magnitude of raw data and scanogram data using a stepped phantom. Also, imaging of asymmetrical objects will proceed using limited raw or scanogram data sets. Scanogram data will be taken at different source angles and raw data from different pulses.

7. Acknowledgements

The authors acknowledge the generous donation of the Agfa Musica Solo imaging system by Agfa. One of the authors (RBH) gratefully acknowledges funding by the J L William Postgraduate Research Scholarship.

References

[1] S. F. Birch. *Insight-non-Destructive Testing and Condition Monitoring*, 43:29–31, 2001.

[2] M. Kachelriess, O. Watzke, and W. A. Kalender. *Medical Physics*, 28:475–90, 2001.

[3] D. Kirk and J. Arvo. *Computer Graphics (Acm)*, 25:153–6, 1991.

[4] S. Liu and M. Hayes. *IEEE Proceedings of SOUTHEAST-CON '91 (Cat. No. 91CH2998-3)*, 2:1096–100, 1991.

[5] L. Lo Presti, G. D'Amato, and L. Sambuelli. *IEEE Transactions on Geoscience and Remote Sensing*, 38:2608–16, 2000.

[6] K. Perisinakis, J. Damilakis, and A. Voloudaki. *Radiation Protection Dosimetry*, 93:173–8, 2001.

[7] G. N. Ramachandran and A. V. Lakshiminarayanan. *Proc. Nat. Acad. Sci. USA*, 68:2236, 1971.

[8] C. A. J. Romanowski, A. C. Underwood, and A. Sprigg. *British Journal of Radiology*, 67:1103–7, 1994.

[9] T. P. Sivachenko, A. K. Belous, and V. A. Romanenko. *Meditsinskaya Radiologiya*, 28:68–70, 1983.

[10] J. P. Wade. *British Journal of Radiology*, 65:261–3, 1992.

[11] P. Wells, J. Davis, and M. Morgan. *Materials Forum*, 18:111–33, 1994.

[12] F. W. Zonneveld. *Medicamundi*, 25:25–8, 1980.