A methodical approach for comparison of CT image quality relative to dose

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How often have you heard these...?

- The image quality on this scanner is better than on our other model.
- The scanner in my private clinic is worse than the one I use here.
- Are we giving more or less dose on this new scanner ?
- Which scanner gives me the best image quality for the least dose ?
- I want to buy a low dose scanner.

The aim of this exhibit is to give a step by step approach for comparing the image quality and dose in CT using standard accepted image quality and dose parameters. There is potential for numerical values to be extracted, however the results are best showed graphically.

Figure 1. Dilemmas and decisions.

Is it the protocol or is it the scanner?

Image quality and dose depend on how the scanner is constructed, and on how it is used. We can adapt how the scanner is used in order to arrive at the true comparison. Some of the factors that influence image quality and dose are indicated in table 1.

Scanner design
detectors (material, configuration, numbers, rows)
data sample rates
software corrections (eg. beam hardening)
x-ray tube, filtration, focal spot
geometry (eg focus-axis, detector distances)
Scan protocols
clinical application
tube current, voltage, focal spot
scan time
reconstruction algorithm
collimation width
image slice thickness
helical pitch, interpolation algorithm

Table 1. Scanner design and scanning factors

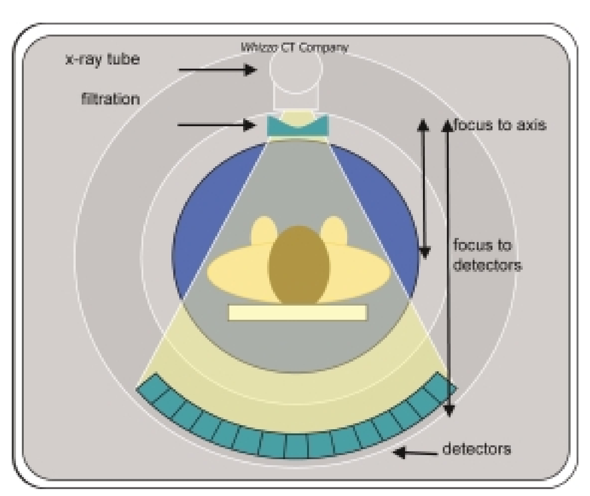

Figure 2. Scanner design

What do we mean by image quality?

Perception or numbers?

Patient diagnosis is based on perception of image quality, however numerical approaches are objective, and represent perception to a greater or lesser extent depending on the parameter.

Perception of image quality is affected by psycho-physical processes involved in vision and pattern recognition. However numerical quantities such as the noise power spectrum, the full curve of the modulation transfer function (MTF) for the scan plane resolution, and the shape of the imaged slice profile for z-axis resolution can closely describe the perceived image quality.

Numerical image quality is also commonly quoted using single figures for these imaging performance parameters, such as standard deviation for image noise, and specific values from the MTF and from the imaged slice profile[1]. In this poster we are using these recognised simple numerical indices of image quality, as well as the computed tomography dose index (CTDI)[2].

The measurement, and standard quotation, of these parameters are described in this poster. Common approaches can often be made for both sequential and helical scanning, on single and multi-slice scanners.

However, any particular comments on helical scanning are made in green, and particular comments for multi-slice scanning are made in red.

Image quality parameters

Image noise

Image noise is measured using a water filled phantom of the appropriate size for head or body.

A region of interest (ROI) is placed at the center of the image, and the standard deviation used as a measure of image noise (Figure 3).

In order for the noise value to be repeatable in subsequent slices, the diameter of the ROI needs to be large enough to contain a sufficient number of pixels without incorporating any non-uniformity effects across the phantom. As a guide, a ROI diameter which is about 40% of the phantom diameter can be used.

For multi-slice sequential scans the outer slices may contain about 4-5 % higher noise, so the average of all slices is taken.

For images reconstructed from a helical acquisition, single or multi-slice, noise is measured using the ROI in the normal way.

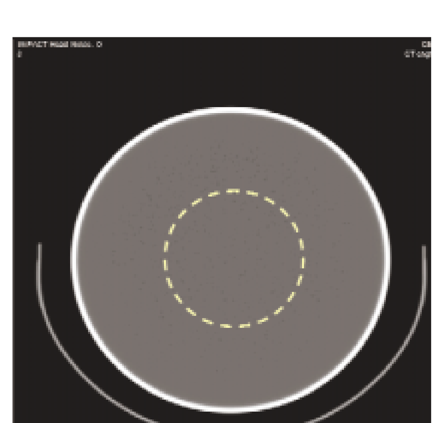

Figure 3. Image of water filled phantom, and superimposed ROI.

Scan plane spatial resolution

Spatial resolution is measured using a high contrast edge, and a modulation transfer function (MTF) analysis is carried out (Figure 4). The frequencies at which the curve falls to 50% and 10% of its original value are quoted. An average of these two numbers can be used as a simple, single figure representation.

The MTF is a measure of how the spatial frequencies, of the function that describes the edge, are transferred into the image. Remember: high spatial frequency = high spatial resolution.

This method can be used for all images from multi-slice sequential scans.

This method can also be used for images from helical acquisitions. Scan plane spatial resolution is not generally dependent on helical scan parameters, although the values may be slightly different from those taken with sequential scans due to small differences in acquisition and reconstruction techniques.

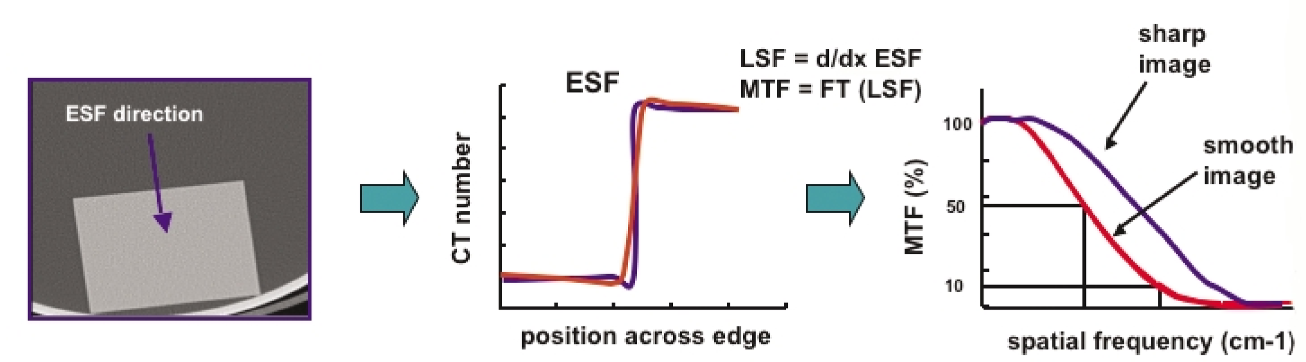

Figure 4. High contrast edge image. Graphs show edge spread functions (ESF) and corresponding MTFs for smooth and sharp algorithms.

Z-sensitivity (z-axis spatial resolution)

For axial scans, angled high contrast aluminium ramps in water are used. The extent of the ramp in the image represents the width of the slice.

This can be measured by finding the full width at half maximum of the profile of CT numbers. A correction is then made for the angle of the ramp to the image plane (Fig 5).

Artefacts appear if the ramp test tool is used for helical scanning, and a different method needs to be used.

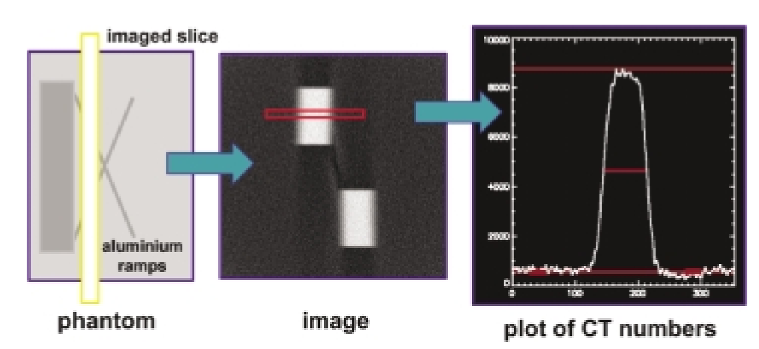

Figure 5. Technique for measuring the axial imaged width.

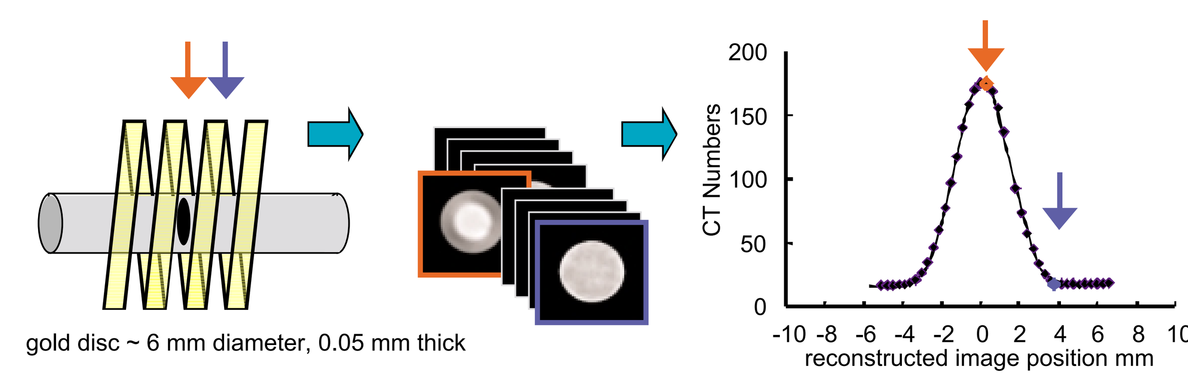

Figure 6. Measuring the image width from a helical scan.

The helical test tool consists of a gold disc, 0.05 mm thick, held in a polymethylmethacrylate (PMMA, ie Perspex, Lucite) rod. This is scanned, and images reconstructed at intervals of 1/10th the image width. A representation of the image profile is obtained by plotting CT numbers from the centre of the image of the rod against the z-axis position of the image (Figure 6).

Dosimetry parameters

The computed tomography dose index ($CTDI_{100}$) is measured using a 100 mm pencil ion chamber, at two positions, centre (c) and periphery (p), within standard head and body PMMA (Perspex or Lucite) phantoms[2] (Figure 7).

The weighted mean of the measurements gives the $CTDI_w$:

$CTDI_w = 1/3.CTDI_c + 2/3.CTDI_p$

The CTDI is dependent on the beam width. In multi-slice sequential scanning the reconstructed image width is not generally a consideration.

For helical scanning, the $CTDI_w$ is divided by the pitch to give a mean dose, $CTDI_{vol}$, along the z-axis: $CTDI_{vol} = CTDI_w$ / pitch

Figure 7. Standard CTDI phantom.

How do you compare image quality and dose?

What scan protocols should you use? How do you take into consideration different image widths, beam widths, reconstruction algorithms, and the effect of pitch and algorithm in helical?

How do you deal with results being different (Table 2)?

	Noise	Resolution^	Image width	Dose
Scanner A	7.6 HU	5.4 c/cm	4.5 mm	23 mGy
Scanner B	5.4 HU	5.9 c/cm	5.0 mm	40 mGy

^ resolution = average of 50% and 10% MTF values

Table 2. All measured parameters different

The answer is KIS: Keep it Simple

Don't make it more complicated than it is already. You can use a three stage process:

(1) Minimise scan protocol variation
(2) Make corrections for known trends and dependencies
(3) Identify trends in other variables

Sequential scanning represents the basic capability of the scanner. Helical scanning takes into consideration how the scanner is used. Multi-slice scanning (sequential and helical) is more complicated, but with careful consideration a valid comparison of the resultant image quality and dose can be made.

Minimise scan protocol variation

Clinical protocol

When making comparisons it is best to use protocols that are similar, such as a standard abdomen or head protocol. This allows for any anatomy specific algorithms that the manufacturer may use, such as beam hardening or adaptive algorithms.

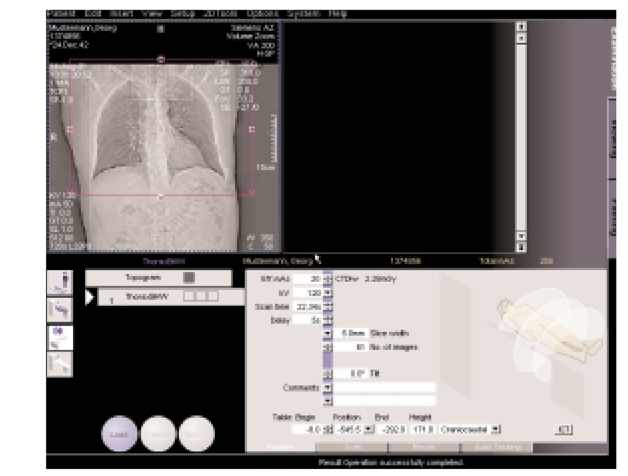

Figure 8. Protocol choices

kV and focal spot

Noise, dose, and to some extent image contrast, all vary with kV. Because of these variations, it is preferable to use the same kV where possible.

Most scanners have a choice of large or small focal spot. This can have an influence on image noise, dose and spatial resolution. It is often automatically selected by the scanner depending on algorithm, imaged slice width, collimation or tube current. It is preferable to use the same focal spot (large or small) in comparison tests.

Beam width

On single slice scanners the CTDI is independent of beam width except for the very narrow slices.

On multi-slice scanners the CTDI varies with beam width since the penumbra is not used to create the images. The proportion of penumbra relative to the nominal beam width increases for small beam widths.

So it is always best to use the same nominal beam width eg 20 mm.

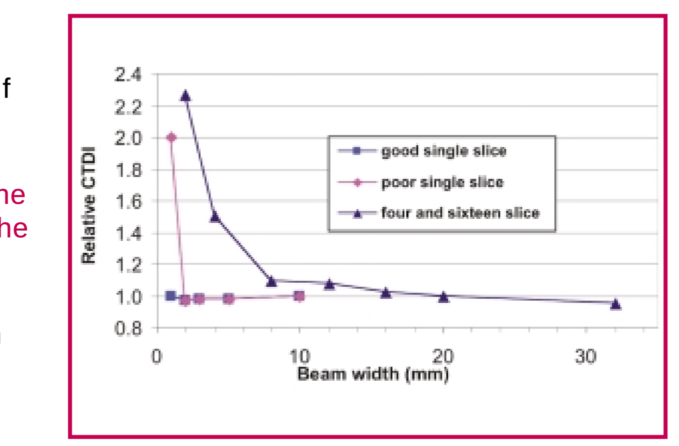

Figure 9. CTDI and beam width

Make corrections for known trends

Image width, tube current and scan time

Image noise is dependent on the number of photons that are used to create the image, and as such is inversely proportional to the square root of both the image width and the tube current-time product.

$$noise \propto \frac{1}{\sqrt{slice\ thickness}} \qquad noise \propto \frac{1}{\sqrt{mAs}}$$

For scanning it is best to select the same nominal image width eg. 5 mm, and to select mAs values which give similar dose. Small differences in the data can then easily be corrected using the above relationship (Table 3).

	Noise	Resolution	Image width	Dose
Scanner A	6.0 HU	5.4 c/cm	5.0 mm	40 mGy
Scanner B	5.4 HU	5.9 c/cm	5.0 mm	40 mGy

Table 3. Same image width and dose

Identify trends in other variables

Pitch and helical reconstruction methods

In both single and multi-slice helical scanning use the same pitch where possible.

In single slice helical, pitch does not affect image noise, but a larger pitch results in a greater image width, a less rectangular image profile, and the average dose ($CTDI_{vol}$) is lower.

In single slice helical, the interpolation algorithm affects both the image width and the noise, and where there is a choice, the same type of interpolation algorithm should be used where possible.

In multi-slice helical, the relationship of noise and pitch is dependent on the individual scanner, and this relationship should be identified. Some scanners adjust the mA with pitch, resulting in constant noise and dose ($CTDI_{vol}$).

Multi-slice scanners can have a selection of helical interpolation and cone beam correction algorithms, which are complex and scanner dependent. The relationship of noise and image width in these circumstances is assessed, but the algorithm normally used by the scanner is selected for comparison purposes.

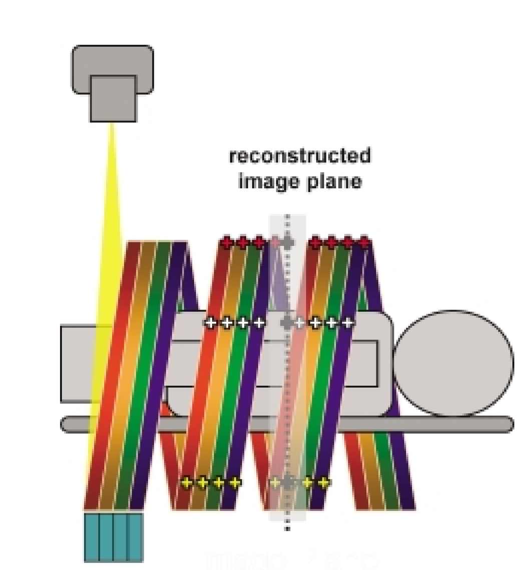

Figure 10. Helical pitch and reconstruction.

Convolution filter

The convolution filter (convolution kernel or reconstruction filter), can be a factor that is difficult to account for.

The image noise and spatial resolution are measured for the range of convolution filters that are available on each scanner.

The noise, adjusted for differences in dose and image width, is plotted against spatial resolution (Figure 11).

From this graph we can now visually compare image noise, at a constant dose and image width, for any spatial resolution. Or we can use interpolated values to standardise the final values in our table of image quality and dose (Table 4).

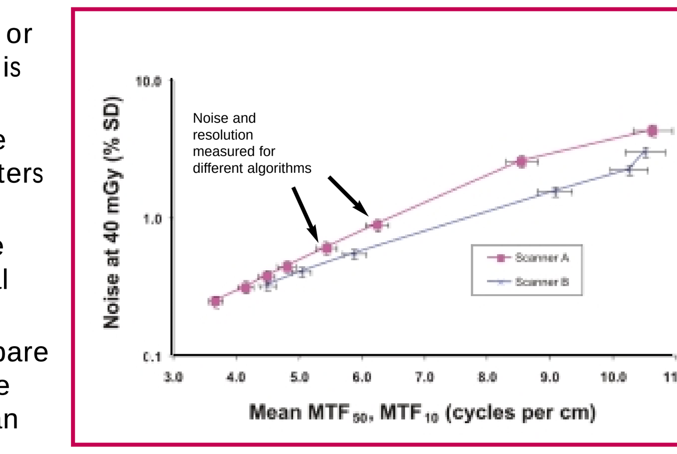

Figure 11. Noise and spatial resolution, at standard dose and image width.

	Noise	Resolution	Image width	Dose
Scanner A	7.8 HU	5.9 c/cm	5.0 mm	40 mGy
Scanner B	5.4 HU	5.9 c/cm	5.0 mm	40 mGy

Table 4. Noise values for same resolution, image width and dose

Can we tie all this into a single number?

If we fit power curves to the noise and resolution data given in figure 12 (four slice scanners, head protocols[3]), we can see that three of the scanners have high correlation co-efficients, with noise as a function of resolution to a power between 2.3 and 2.8. We can combine all the parameters to give a general equation:

$$noise \propto \frac{resolution^{a}}{\sqrt{dose \times image\ width}}$$

A single number for each scanner can then be calculated using a generalised relationship. However the fourth scanner is harder to characterise as there are a number of algorithms that give similar spatial resolution but different noise. This demonstrates a limitation of this approach.

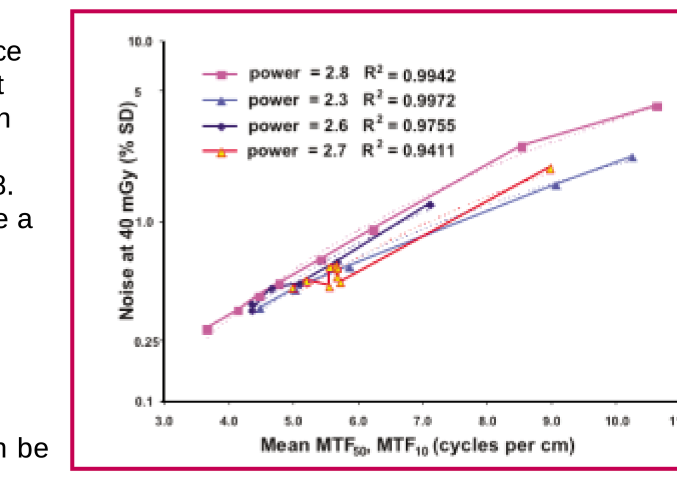

Figure 12. Curve fits to data

Sixteen slice scanners

By using the approach outlined in this poster, data are given for sixteen slice scanners[4]. The standard abdomen protocol was chosen as a baseline. A similar set of power relationships can be established for this set of data.

Scanner	Algorithm
GE Lightspeed 16	soft, standard, lung, detail, bone, bone+, edge
Philips Mx8000 IDT	A, EC ,B ,C , D
Siemens Sensation 16	B10, 20................80
Toshiba Aquilion 16	FC 10, 11, 12, 13, 14, 30, 31, 80

Table 5. Algorithms used for Figure 13.

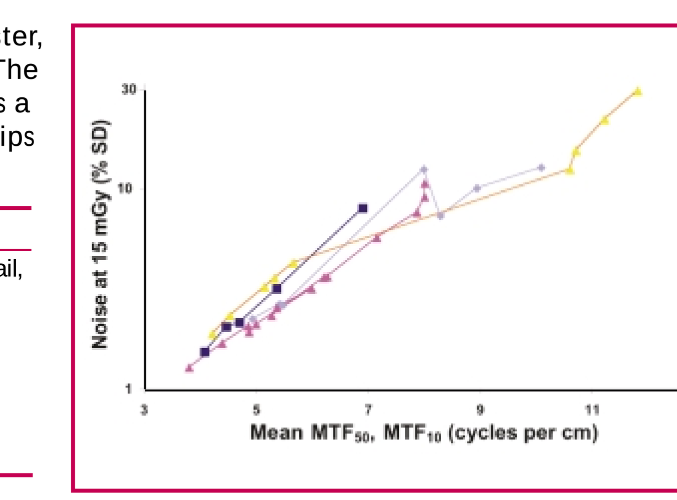

Figure 13. Abdomen protocol, 16 slice scanners.

Summary

The approach presented here is a step by step guide to enable a clear comparison of scanner numerical image quality. Variations in values, due to different scan settings and image quality dependent variables, which generally cloud the final picture, are eliminated.

Graphs are used as a visual demonstration for the noise and resolution relationship. A single number can be obtained from the graph, where a generalised relationship is found. A single comparative number is attractive and can give a general overview, but does not easily take into account particular exceptions that certain algorithms or scanners raise, and also does not include the full visual impact of the data.

This approach is a baseline for clarifying the issues in comparative image quality and dose studies. This baseline is essential before tackling the ever increasing complexity of dose and image quality issues of modern scanners, such as patient related mA modulation, and anatomy specific image reconstruction algorithms.

References

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4. Sixteen Slice Comparison Report Version 9 (in press), ImPACT Report MHRA03121, 2003, HMSO, Norwich, UK, 2003

ImPACT is an evaluation centre of the MHRA (The Medicines and Healthcare products Regulatory Agency, UK Department of Health). ImPACT web site: www.impactscan.org