The Relationship between Noise, Dose and Pitch in Cardiac Multi-Detector Row CT (MDCT)

Andrew Primak, Ph.D., Cynthia H. McCollough, Ph.D., Michael R. Bruesewitz, R.T.(R), Jie Zhang, Ph.D., J. G. Fletcher, M.D.
CT Clinical Innovation Center, Department of Radiology, Mayo Clinic College of Medicine, Rochester, MN

Abstract

In spiral CT, dose is always inversely proportional to pitch. However, the relationship between noise and pitch (and hence noise and dose) varies dependent upon scanner type (single slice (SS) vs. MDCT), and MDCT reconstruction mode (cardiac vs. non-cardiac). In single-slice spiral CT, image noise is independent of pitch. In MDCT however, noise is dependent on pitch, thus CT manufacturers adapt mAs based on pitch to assist the user in obtaining constant image noise when pitch is varied. This results in comparable noise when the ratio of mAs to pitch is held constant. In cardiac spiral CT, the behavior of noise vs. pitch returns to the SS paradigm, where noise is independent of pitch. Thus, for faster rotation times, where pitch must be lowered, patient dose is increased without a commensurate decrease in image noise. In this exhibit, we review these principles and demonstrate the relationships between image noise, dose and pitch for cardiac vs. non-cardiac MDCT.

Radiation Dose

Radiation dose is related to the amount of energy that x-ray photons deliver during an examination. It depends on the total number of photons and their individual energies. The energy distribution of these photons depends on the operating voltage (kV) of the x-ray tube and the spectral (bowtie) filter. The total number of photons is proportional to the tube current (mA) and, the "x-ray on" time (s) during a single gantry rotation (and hence their product, mAs).

Additionally, in sequential mode, the radiation dose is proportional to the detector width / table increment per rotation. Thus for the most common scenario of contiguous scans (where table increment = total detector width), dose is proportional only to mAs. When the table increment ≠ total detector width, the general relationship that dose is proportional to mAs/pitch must be used, where pitch = table increment per rotation/total detector width. For scans with no table increment, dose is proportional to the mAs x the number of rotations during the scan.

In spiral mode, the radiation dose is similarly proportional to mAs x total detector width / table increment per rotation. As contiguous scanning can not be assumed in spiral SS or MDCT, the degree of x-ray beam overlap must be communicated. Thus, in spiral mode, the radiation dose is proportional to mAs/pitch (which on some systems is called effective mAs, or mAs per slice).

Noise

Quantum image noise σ directly relates to the number of x-ray photons N, which have contributed to the data used to reconstruct the image. Since x-ray photon statistics obeys the Poisson distribution, σ is proportional to √N and the relative noise (σ/N) decreases with increasing N. The number of photons contributing to a reconstructed image is proportional to the tube current (mA) and the amount of time necessary to acquire all the projection data needed for the reconstruction. In sequential mode, this time equals to the "x-ray on" time per rotation, so N is proportional to mAs, and relative noise is proportional to 1/√mAs.

In spiral mode, however, the relationship between noise and pitch varies with scanner type (SS vs. MDCT), and MDCT reconstruction mode (cardiac vs. non-cardiac).

Single-slice spiral CT

In SS spiral CT, a gantry always has to rotate through a certain angle (dependent on the reconstruction algorithm) in order to acquire all the projection data needed for an axial image. Thus, N depends only on the gantry rotation time and not on the table speed or pitch. Pitch has an effect on image quality due to the interpolation necessary to generate planar data from the measured spiral data, but it has no effect on noise.

Multi-detector row spiral CT

Non-cardiac mode

In non-cardiac spiral MDCT, noise depends on pitch through the elaborate interpolation algorithm used to generate a set of planar projection data (Fig. 1). When pitch<1, the measured spiral data partially overlap in the z-direction (perpendicular to the gantry plane) as shown in Fig. 1b, so some segments of the planar projection data for the same axial image can be generated more than once (from the spiral data acquired by different detector rows). This redundant data results in more x-ray photons contributing to the reconstruction of the axial images compared to pitch=1, reducing the relative noise and making noise pitch-dependent.

Example: A scan with pitch=0.5 needs twice the number of gantry rotations to cover the same distance compared to pitch=1. This results in fully overlapped spiral data, which can generate two completely redundant sets of planar projection data for every axial image (Fig. 1c). Combining these two redundant data sets into one allows reconstruction of axial images where noise is reduced by a factor of √2.

When pitch>1, the measured spiral data have gaps in the z-direction, so some segments of the planar projection data cannot be generated at all (Fig. 1d). This missing data results in less x-ray photons contributing to the reconstruction of each axial image compared to pitch=1, increasing the relative noise and making noise pitch-dependent.

Generally, to offset the increase in noise as pitch is increased, the mAs needs to be increased approximately proportional to the increase in pitch. Thus, as long as "effective mAs" (defined as mAs/pitch) is held constant, both dose and noise remain fixed. This is demonstrated in Fig. 2 (a, b) where noise and dose are shown as a function of mAs, and as a function of effective mAs.

Cardiac mode

In cardiac spiral MDCT, the best possible temporal resolution is required to minimize artifacts resulting from cardiac motion. This goal is achieved by minimizing the amount of time necessary to acquire the data needed for image reconstruction. Using redundant (hence pitch-dependent) data is not acceptable because it degrades the temporal resolution.

Cardiac algorithms use partial reconstruction techniques, which needs only 180° + fan angle of the projection data to reconstruct an image. Thus, the number of photons N contributing to the cardiac reconstruction depends only on mA and the time it takes for the gantry to rotate through 180° + fan angle. Since this time is proportional to the rotation time, N (and hence noise) is dependent only on mAs and is not affected by pitch. Fig. 2 (c, d) demonstrates the independence of noise on pitch for cardiac reconstructions, even though dose remains dependent on pitch. Hence, a constant effective mAs no longer guarantees equivalent noise, but rather only equivalent dose.

Figure 1

Graphical description of the interpolation algorithm used to generate planar data from the measured spiral data. The dotted lines show the center of every detector row, while the brown solid lines indicate the detector boundaries. All spiral data within a pre-defined spiral interpolation window (shown in blue) are used to generate planar data for the image plane (shown in red). Pitch<1 results in overlapped spiral data (shaded areas), which generate segments of redundant planar data, while pitch>1 results in spiral data with gaps, which cause missing segments of planar data.

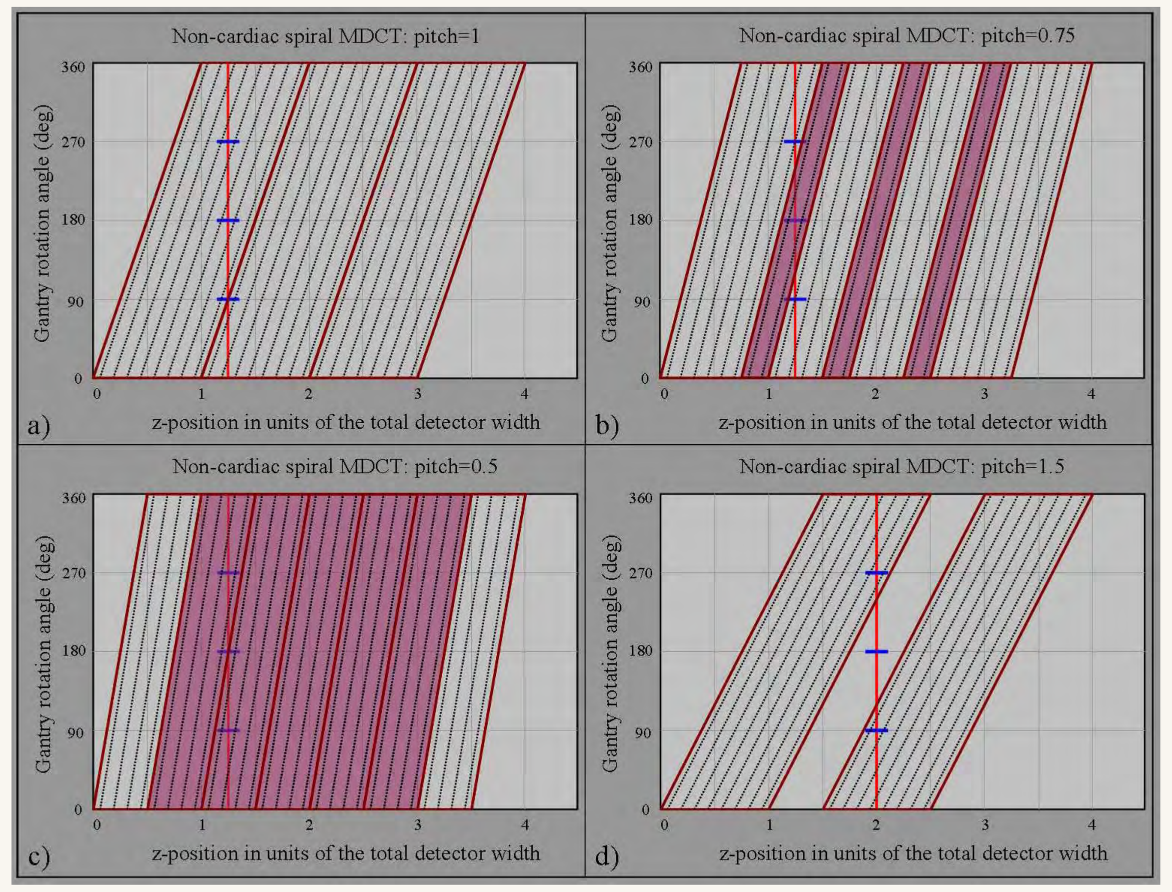
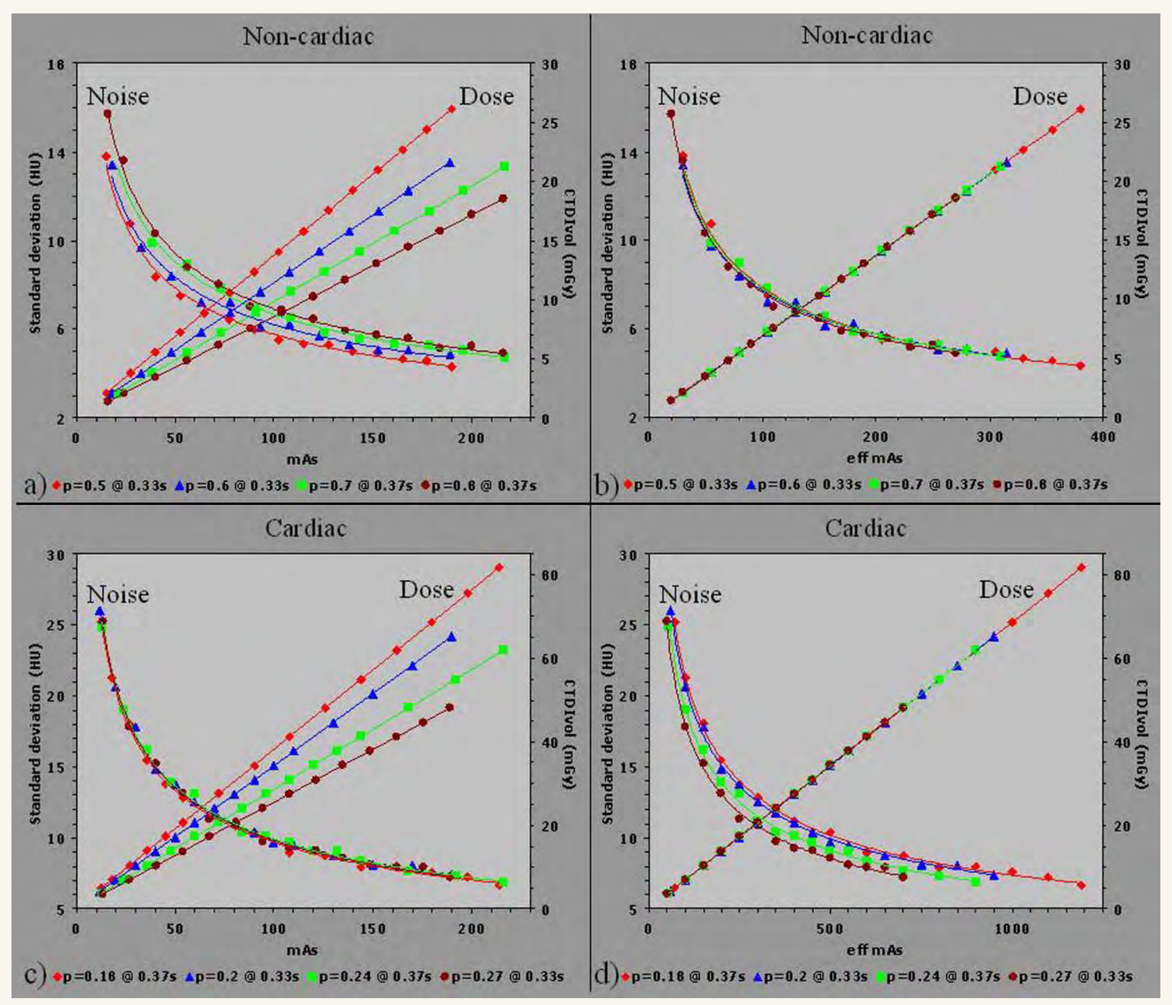


Figure 2

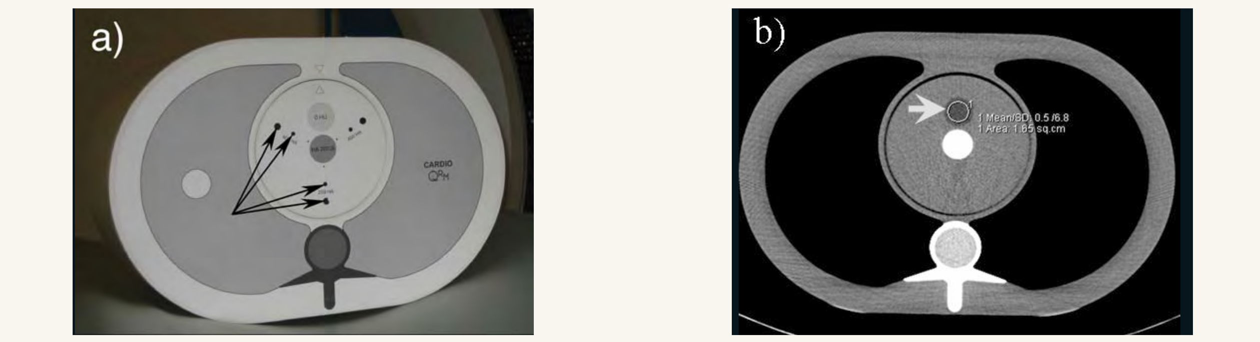
Noise and dose vs. mAs and effective mAs for (a, b) non-cardiac and (c, d) cardiac spiral modes at 4 different pitch values (p=pitch) and 2 differenct gantry rotation time (0.33s and 0.37s). Left y-axis corresponds to noise curves, right y-axis corresponds to dose curves.



Noise vs. mAs Measurements

Noise vs. mAs data were obtained on a Sensation 64 scanner (Siemens Medical Solutions, Forchheim, Germany) using an anthropomorphic cardiac CT phantom (QRM, Möhrendorf, Germany, Fig. 3a). The phantom was scanned using both cardiac and non-cardiac spiral modes with four different pitch values (0.18, 0.2, 0.24, and 0.27 for cardiac, and 0.5, 0.6, 0.7, and 0.8 for non-cardiac) and two rotation times (0.33 and 0.37 s). The noise was measured within the water equivalent insert embedded in the central portion of the cardiac phantom (Fig. 3b). Dose was assessed using CTDIvol, per IEC 60601-2-44.

Figure 3



(a) Anthropomorphic cardiac phantom with small calcified cylinders used for calcium quantitation (black arrows)

(b) Large water-equivalent cylinder used for noise vs mAs measurements (white arrow head).

Pitch in Cardiac MDCT

Temporal resolution is of fundamental importance to cardiac MDCT and was a driving force behind making gantries rotate faster and faster. However, faster gantry rotation requires a slower pitch in cardiac mode to avoid volume gaps in heart coverage between images reconstructed in consecutive cardiac cycles (Fig. 4). For example, on the 4-slice scanner, Siemens Volume Zoom, for a rotation time of 0.5 s, the necessary pitch was 0.375. For the Siemens Sensation 64, with a rotation time of 0.33 s, the necessary pitch is 0.2.

To achieve the same noise at faster rotation times (recall that noise is independent of pitch in cardiac mode), one has to use the same mAs and smaller pitch values and, hence, higher radiation doses result (dose is always inversely proportional to pitch). Thus, better temporal resolution for cardiac spiral CT comes with the price of higher dose.

We can derive a good approximation of the relationship between pitch, heart rate and the rotation time by taking into account that the table should not move more than the total detector width W within one heart cycle (RR-interval time T_{RR}). Hence, the table speed V should be less or equal to W/T_{RR}. Keeping in mind that $V=(\text{pitch} \times W)/T_{rot}$, where T_{rot} is the rotation time, we obtain

$$V \le \frac{W}{T_{RR}} \Rightarrow \frac{VT_{rot}}{W} \le \frac{T_{rot}}{T_{RR}} \Rightarrow \text{pitch} \le \frac{T_{rot}}{T_{RR}}$$

For the exact relationship, we have to consider a particular cardiac reconstruction algorithm. For example, the multi-slice cardiac volume reconstruction algorithm (MSCV) uses two interpolation partners for linear interpolation. The condition that there be no volume gap for the MSCV algorithm (Fig. 5) requires that the Nth slice position z_N at the start of the 1st cardiac cycle recon (t=0) is not less than the 1st slice position z_1 at the end of the 2nd cardiac cycle recon ($t=T_{RR}+T_Q$), where T_Q is the recon time window. If $z_1(t=0)=0$, then the n_{th} slice position at time t is given by $z_n(t)=Vt+(n-1)d$, where d is the collimation width ($N \times d=W$). Applying the no-gap condition we obtain

$$z_1(T_{RR}+T_Q) \le z_N(0) \Rightarrow V(T_{RR}+T_Q) \le (N-1)\,d \Rightarrow \frac{VT_{rot}}{Nd} \le \left(\frac{N-1}{N}\right)\left(\frac{T_{rot}}{T_{RR}+T_Q}\right) \Rightarrow \text{pitch} \le \left(\frac{N-1}{N}\right)\left(\frac{T_{rot}}{T_{RR}+T_Q}\right)$$

Figure 4

Graphical description of an ECG-gated spiral scan with pitch that is too high for the heart rate. Continuous coverage of the heart is not possible because of the volume gaps between the images reconstructed in the consecutive cardiac cycles.

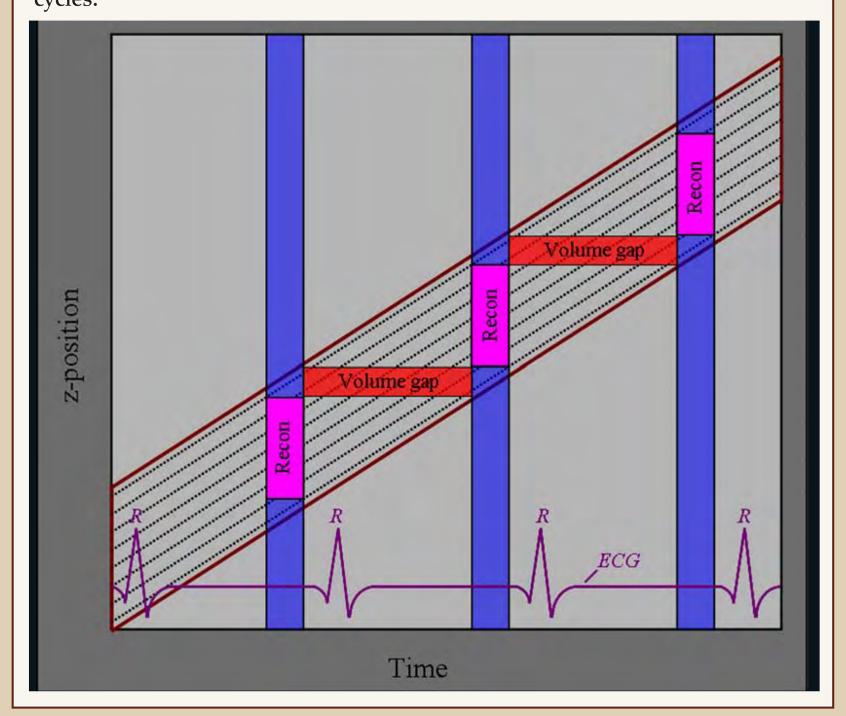
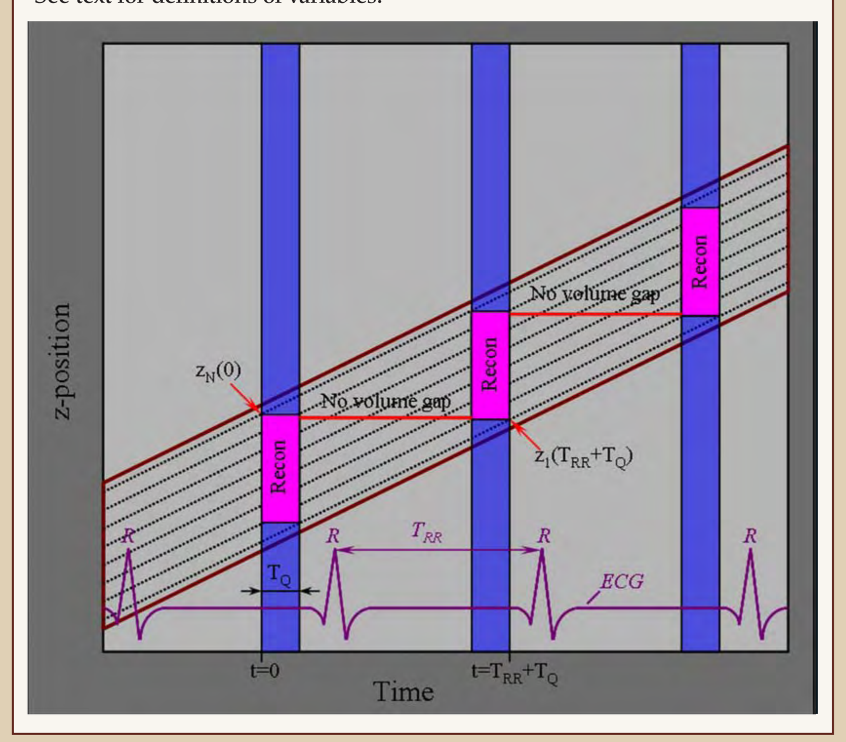


Figure 5

Graphical description of the relationship between pitch, heart rate and the rotation time for cardiac spiral scanning with the MSCV reconstruction algorithm. See text for definitions of variables.



Clinical Implications

Since faster rotation times in cardiac spiral MDCT require higher radiation dose (compared to slower rotation times) to achieve equivalent image noise, its use should be justified by clinical benefits for a particular application. We have examined two specific applications for cardiac CT: quantitative coronary artery angiography and quantitation of coronary artery calcium.

Example 1: The accuracy of percent area stenosis measurements in a moving stenotic vessel phantom improved significantly as the rotation time decreased from 0.5 to 0.33 s (Fig. 6). Hence the benefit of faster rotation time appears to outweigh the disadvantage of the increased dose (Scientific Poster LPB10-06-p).

Example 2: The accuracy of coronary artery calcium measurements in a rotating insert of the anthropomorphic cardiac phantom (Fig. 3) was not statistically different for calcifications larger than 3 mm and densities greater than 400 mg/cm3 (Fig. 7) amongst the 4 rotation times tested (0.33, 0.37, 0.42 and 0.5 s). We conclude that for screening purposes, slower rotation times should be used as they appear equally accurate yet deliver lower radiation dose to the patient.

Figure 6

Percent area stenosis measured in a moving stenotic vessel phantom scanned with Siemens Sensation 16 and 64. Data were acquired at rest (0.33 s rotation time only) and while the phantom was moving at physiological coronary artery velocities for multiple gantry rotation times (0.33, 0.42 and 0.5 s). The stenoses were measured using a semi-automated tool (VesselView, Siemens) for slice widths of 0.6 (Sensation 64 only), 0.75 and 1.0 mm.

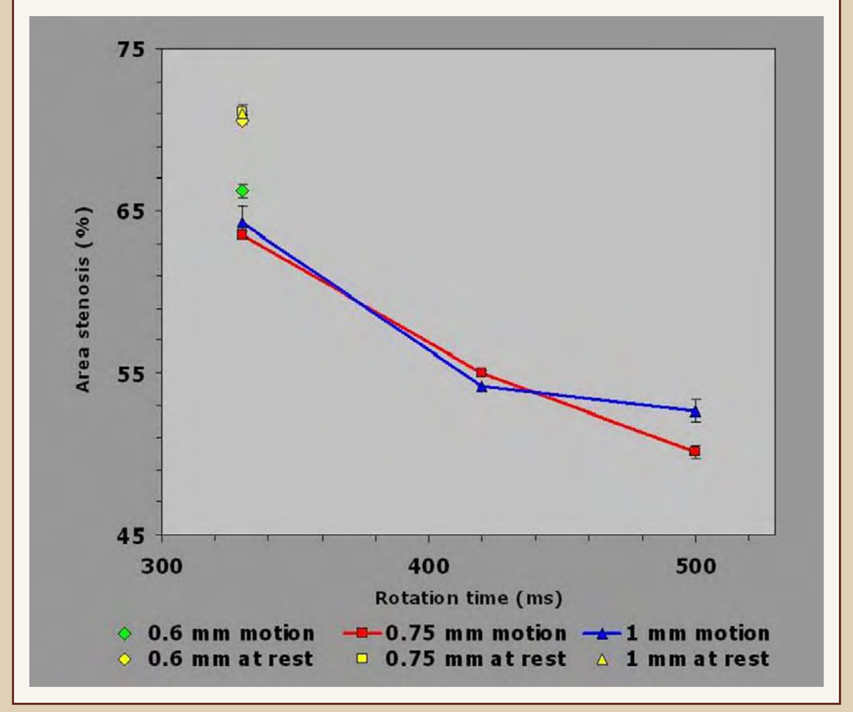


Figure 7

Calcium mass score measured in a rotating insert of the anthropomorphic cardiac phantom scanned with Siemens Sensation 16 and 64 using multiple gantry rotation times (0.33, 0.37, 0.42 and 0.5 s). Calcium scores were obtained using an automated tool (CaScoring, Siemens) for 4 hydroxyapatite (HA) cylinders of known sizes and densities (400 and 800 mg/cm³, 3 and 5 mm diameter) embedded into the phantom.