Published in final edited form as:

J Comput Assist Tomogr. 2009 ; 33(3): 338–347. doi:10.1097/RCT.0b013e318181fa7a.

Cardiac Computed Tomography Radiation Dose Reduction Using Interior Reconstruction Algorithm With the Aorta and Vertebra as Known Information

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Abstract

High x-ray radiation dose is a major public concern with the increasing use of multidetector computed tomography (CT) for diagnosis of cardiovascular diseases. This issue must be effectively addressed by dose-reduction techniques. Recently, our group proved that an internal region of interest (ROI) can be exactly reconstructed solely from localized projections if a small subregion within the ROI is known. In this article, we propose to use attenuation values of the blood in aorta and vertebral bone to serve as the known information for localized cardiac CT. First, we describe a novel interior tomography approach that backprojects differential fan-beam or parallel-beam projections to obtain the Hilbert transform and then reconstructs the original image in an ROI using the iterative projection onto convex sets algorithm. Then, we develop a numerical phantom based on clinical cardiac CT images for simulations. Our results demonstrate that it is feasible to use practical prior information and exactly reconstruct cardiovascular structures only from projection data along x-ray paths through the ROI.

Keywords

computed tomography; cardiac CT; local reconstruction; fan-beam reconstruction; parallel-beam reconstruction; radiation dose reduction

Cardiovascular disease (CVD) is the leading cause of death and a common chronic illness in the United States and worldwide. Every year since 1900, except for 1918, it has accounted for

more deaths than any single cause of death in the United States.¹ Noninvasive imaging techniques including computed tomography (CT) play an integral role in CVD diagnosis, which is important for early prevention and effective therapy. There have been tremendous developments in cardiac CT since the introduction of single-slice helical/spiral or slip-ring CT in 1998 and multidetector CT (MDCT) in 1999.² The subsecond rotation combined with the improved volume coverage can provide thinner slices that have improved both temporal and spatial resolution.³ This has led to an increased use of cardiac CT. In fact, the total number of cardiac CT scans doubled between 2002 and 2003 to 485,000 cases.^{4,5}

Cardiac CT continues evolving with the innovative technologies such as controlled^{6,7} and dynamic volumetric cardiac CT.^{8,9} The 320-slice MDCTs are now commercially available. Innovative scanning trajectories such as the saddle curve¹⁰ and compositing circles¹¹ have been proposed. Hence, dramatic progress of CT technology, especially MDCT, is foreseeable in the near future.¹² As a result, its use is likely to improve the diagnosis of CVD, including coronary artery disease, risk factor assessment, and other cardiovascular disease such as congenital heart disease in infant and children.

Computed tomography is the largest source of radiation dose in diagnostic imaging. A high dose of ionizing radiation associated with x-rays in CT can damage living cells, whereas the effect of low-level exposure (<100 mSv) remains unclear. It has also been estimated that the cancer risk caused by CT may be up to 1.5% to 2.0%.¹³ Current recommendations are to use radiation dose as low as possible while satisfying the diagnosis requirement. Therefore, there is an urgent need for effective dose reduction.

Several strategies were proposed to reduce the radiation dose, most of which focused on the modulation of scan parameters, modification of scanner geometry, and use of prospective gating, and so on.^{14,15} Although these methods can reduce the x-ray radiation significantly, further reduction can be made by developing new methods to reduce the amount of raw data. The first landmark work along this direction is the well-known fan-beam half-scan formula.¹⁶ In 2002, Noo et al¹⁷ reported a new imaging condition that makes exact and stable reconstruction possible when the object is not truncated in any required fan-beam projections, and all the lines through the region of interest (ROI) intersect with the source trajectory nontangentially. The recent milestone is the 2-step Hilbert transform method,¹⁸ which performs the reconstruction of a small region from a supershort scan possible. In 2006, Defrise et al¹⁹ enhanced this condition such that a line segment which also represents field of view (FOV) can be reconstructed if it has at least one intersection with the object convex compact support. However, these conditions do not cover the case of interior reconstruction where the FOV is just wide enough to cover an internal organ or ROI to be imaged. This new finding may help reduce dose significantly for interior reconstruction applications such as dental and cardiac imaging. In 2007, our group²⁰ proposed an improved condition along this direction. As illustrated in Figure 1, we proved that the image on an interior line segment can be exactly reconstructed as long as it is within the FOV and a portion of the segment is known. The line segment can now be completely within the object support, and the projection data necessary for reconstruction is fully truncated. Figure 2 compares the difference between the data acquisition configurations of conventional CT and interior CT. The reconstruction from truncated projection data can be effectively addressed in a backprojection filtration framework. The backprojection of the differential projection data is first computed in the FOV that corresponds to the scaled Hilbert transform of the image to be reconstructed. The original object image is then obtained by inverting the Hilbert transform.

For practical applications, the most important question is what regions can be assumed to be known. Because our interior tomographic approach is for interior imaging and uses localized projection data, the external markers on the human body cannot serve as known regions for

reconstruction. It is also necessary to choose the known regions that provide adequate signal intensity in contrast to other cardiovascular structures. More importantly, the ROI should have little variability in the signal intensities from slice to slice in MDCT but may change from subject to subject. It is preferable that these requirements are satisfied with contrast injection for future clinical applications in which contrast material can be used (eg, CT angiography). In MDCT, it is also necessary to know these regions in all the slices. The cross section of descending aorta in the mid thorax is perhaps the best known subregion within the ROI when the contrast dose is not used or already standardized, whereas the spine or vertebral bone is also a good candidate that has good signal intensity and little variation regardless of contrast enhancement. In this article, we assume that the attenuation coefficients of the blood in aorta and bone in vertebra are available. Blood attenuation can be obtained using an individualized previous scan of the blood sample. The required attenuation values may also be obtained from previous scans of the subject. In this way, the intersubject variability may be minimized.

The main objectives of this article are to: (1) study the feasibility of using the blood in aorta and vertebral bone as the known regions for interior ROI reconstruction and (2) make our simulations realistic by developing a new phantom resembling clinical cardiac CT images and using realistic noise levels. In the following sections, we will first describe the methods and then report numerical results. Finally, we discuss relevant issues to conclude the article.

METHODS

Backprojection

This is the first step in our reconstruction process used to obtain Hilbert transform of the image scaled by a factor of -2π . In a fan-beam acquisition geometry (Figure 3), the formula proposed by Noo et al¹⁸ for backprojection at a point $\mathbf{x} = (x, y)$ can be written as:

$$b_{\theta}(\mathbf{x}) = \frac{1}{2} \int_0^{2\pi} \frac{D^2}{v^2} \operatorname{sgn} \left(\sin \left(\phi - \arctan \left(\frac{\bar{u}}{D} \right) - \theta \right) \right) \frac{d}{du} \left\{ \frac{D}{\sqrt{D^2 + u^2}} p(\phi, u) \right\} \bigg|_{u=\bar{u}} d\phi + \frac{p(\phi_1, \bar{u}_1)}{\|\mathbf{x} - \mathbf{x}_1\|^2} - \frac{p(\phi_2, \bar{u}_2)}{\|\mathbf{x} - \mathbf{x}_2\|^2} \quad (1)$$

where $p(\phi, u)$ is the acquired fan-beam projection for source rotation angle ϕ and detector position u ,

$$\bar{u} = \frac{D}{v} (-x \sin(\phi) + y \cos(\phi)), \quad (2)$$

$$\bar{v} = R - x \cos(\phi) - y \sin(\phi), \quad (3)$$

θ is the desired filtering direction ($\theta = 0$ is the horizontal direction and $\theta = \pi/2$ is the vertical direction), the sgn term in the backprojection formula is used to reverse the sign at the critical angle, $\mathbf{x}_1$ and $\mathbf{x}_2$ are intersections of the line at an angle θ passing through point $\mathbf{x}$ with source locus,

$$\phi_1 = \theta - \arcsin((x \sin(\theta) - y \cos(\theta))/D), \quad (4)$$

$$\phi_2 = 2\theta - \phi_1 + \pi, \quad (5)$$

ϕ_1 and ϕ_2 are the angular positions of the intersection points $\mathbf{x}_1$ and $\mathbf{x}_2$, $\bar{u}_1$ and $\bar{u}_2$ are the values of u at $\phi = \phi_1$ and $\phi = \phi_2$, respectively.

We also carry out the backprojection in the parallel-beam geometry. In the following equations, a subscript p is used to distinguish parallel-beam parameters from the fan-beam counterparts. Because the original acquisition is based on fan-beam geometry before parallel-beam backprojection, we must rebin the fan-beam projection data by linear interpolation according to the following equations:

$$\phi_p = \phi - \arctan\left(\frac{u}{D}\right), \quad (6)$$

$$u_p = D \sin\left(\arctan\left(\frac{u}{D}\right)\right). \quad (7)$$

The parallel-beam backprojection formula presented by Noo et al¹⁸ can be written as:

$$\begin{aligned} b_\theta(\mathbf{x}) &= \frac{1}{2} \int_0^{2\pi} \text{sgn}(\sin(\phi_p - \theta)) \frac{d}{du_p} P(\phi_p, u_p) |_{u_p} \\ &= -x \sin(\phi_p) + y \cos(\phi_p) \Big|_p^{\phi_p}. \end{aligned} \quad (8)$$

Projection onto Convex Sets

Secondary to the truncation of the data, directly inverting the Hilbert transform image to reconstruct the image is challenging because of the ill conditioning of the Hilbert transform kernel. Here, we use an iterative projection onto convex sets (POCSs) method,²¹ which has been previously used by our group¹⁹ and others²⁰ to carry out the image reconstruction. Let us assume that an original image f belongs to an intersection C_0 of well-defined closed convex sets $C_1, \dots, C_m, \dots, C_M$, that is, $f \in C_0 = \bigcap_{m=1}^M C_m$. Then, let the projection operators onto these M convex sets be denoted as $P_1, \dots, P_m, \dots, P_M$, and the composite operator as $T = P_1 P_2 \dots P_M$. The POCS procedure can then be defined as $f_n = T^n f$, where n indicates the iteration number. The convex sets used here to invert the truncated Hilbert transform are presented in the Results.

ROI Reconstruction from a Small Known Subregion

We propose to carry out image reconstruction first in the horizontal or vertical direction using a small known subregion and then use the reconstructed image in addition to any other known subregions as prior information for inversion in its orthogonal direction. In both the horizontal and vertical direction, the reconstruction is carried out 1 line at a time. Because the reconstruction of an image requires inversion in 2 orthogonal directions, the backprojection also needs to be performed on the lines in these 2 directions.

Design of a Realistic 2-Dimensional Numerical Phantom

Based on a real human CT image shown in Figure 4A, the shape and location of the heart, aorta, thorax, lungs, sternum, ribs, and vertebra were modified in a central slice of the popular thorax phantom²² to obtain a new 2-dimensional (2-D) cardiac phantom shown in Figure 4B.

Whereas the original thorax phantom is defined in terms of density, the new phantom is defined in terms of Hounsfield units (HUs) as in the commercial CT scanners. The equation used to convert the linear attenuation coefficient from per centimeter to HU was as follows:

$$\mu_{\text{HU}} = 1000 \frac{\mu - \mu_w}{\mu_w}, \quad (9)$$

where μ is the linear attenuation coefficient of a structure in per centimeter at a particular KVp and μ_w is the linear attenuation coefficient of water at the same KVp.

To obtain the linear attenuation coefficients for the new phantom, we first calculated the HU values using 4 slices from a 3-D image for each structure in the phantom. The regions used in the slices are shown in Figure 5. Figure 5A was extracted from a noncontrast single-phase image acquired in a cine-mode acquisition at 120 KVp and 400 mA with a slice thickness of 2.5 mm. Figure 5B was obtained from Figure 5A by retaining the voxels corresponding to bones with HU above 200. Similar regions were then identified in 3 other consecutive slices. Means of the HU values within the regions shown in Figure 5A in all the 4 slices were used to estimate the HU values of the blood in aorta and heart, lungs, and tissue. The HU values for the bones were obtained from the mean HU values of the voxels corresponding to the bones in all the 4 slices. As seen from Figure 5B, the mean HU value of the ribs was derived from several rib regions. The noise was calculated as the mean of SDs within the tissue regions in the 4 slices. The linear attenuation coefficients of various structures in HU can be calculated using equation 9. The linear attenuation coefficient of water at 120 KVp was obtained via linear interpolation from its values at 100 KVp and 150 KVp,²³ yielding:

$$\mu_w(120 \text{ KVp}) = 0.1624 \text{ cm}^{-1}.$$

Table 1 summarizes the calculated linear attenuation coefficients. Table 2 provides a description of the new phantom geometry and associated linear attenuation coefficient for the vertebra. Table 3 lists the same for all the remaining structures. The regions of vertebra 1 and vertebra 2 shown in Figure 5B, respectively, corresponded to structure 1 and structures 3 to 8 in Table 3.

RESULTS

We carried out the numerical simulations using our modified new phantom. A linear 84-cm virtual detector consisting of 1050 equispaced elements was assumed. A circular trajectory was used and 1200 equiangular nontruncated projections were acquired over 360 degrees in the fan-beam geometry shown in Figure 3. This nontruncated acquisition mode gave us the freedom to simulate various truncated scenarios by retaining only the projections through an ROI. Poisson noise was also added to the projection data using the method described by Yu et al.,²⁴ with $N_0 = 4 \times 10^6$ as the original number of photons per ray. The standard derivation of noise in the reconstructed image from nontruncated projections was consistent with that of clinical images, which is approximately 19.0 HU in this case. Projection data in the parallel-beam geometry were obtained by rebinning the fan-beam data. Images were reconstructed in both fan-beam and parallel-beam geometries. The POCS method was used to reconstruct the final images. We used the same convex sets as in our previous article.²⁰ Assuming $f(x) \in L^2(\mathcal{R})$ as the 1-D image to be reconstructed, these sets can be defined in reference to Figure 1A by the following equations:

$$\begin{aligned}
C_1 &= \{f \in L^2(\mathcal{R}) \mid (Hf)(x) = g(x), x \in (p, r)\}, \\
C_2 &= \{f \in L^2(\mathcal{R}) \mid f(x) = f_o(x), x \in (p, q)\}, \\
C_3 &= \left\{f \in L^2(\mathcal{R}) \mid \int_{-1}^1 f(x) dx = C_f\right\}, \\
C_4 &= \{f \in L^2(\mathcal{R}) \mid f(x) \geq 0, x \in [-1, 1]\}, \\
C_5 &= \{f \in L^2(\mathcal{R}) \mid f(x) \leq f_{\max}, x \in [-1, 1]\}.
\end{aligned}$$

The image f corresponds to an image of linear attenuation coefficients μ along a given π -segment. $f_o(x)$ is the known portion. C_f is the projection along the x-ray line containing the π -segment. $f_{\max}$ is the upper bound of $f(x)$. In our simulation, we used $f_{\max} = 0.5$. Our fourth convex set holds because we do not expect any tissue/organ with an attenuation coefficient less than zero. The maximum linear attenuation coefficient $f_{\max}$ in the fifth convex set may be determined from the maximum HU present in a previous scan using equation 9 and the knowledge of linear attenuation coefficient of water at the corresponding KVp. Alternatively, because bone has the higher linear attenuation coefficient among the body structures, its highest physiological value may be used for the same. We performed 400 iterations in our simulations.

Figure 6 shows the 3 different ROIs and known subregions that we used for our simulations. The small rectangular strip (horizontal or vertical) within the rectangular ROI was first reconstructed. This strip then served as the known information for the reconstruction in the orthogonal direction. The known information was the attenuation values of blood in aorta and/or bone in vertebra. In Figure 6A, only the aorta was the known subregion, and horizontal inversion was first carried out. In Figure 6B, only the part of vertebra within the ROI was the known subregion, and vertical inversion was first carried out. In Figure 6C, the information about both aorta and vertebra within the ROI was known, and horizontal inversion was first carried out. The small rectangular region within aorta was used as the known subregion in the cases shown in Figures 6A and C. Images of reconstructed ROIs for simulation setup of Figures 6A through C with profiles are respectively presented in Figures 7 to 9.

In all the images, we observed some drop in intensities in the vertical strip between vertebra and sternum. This may be attributed to the vertebra and sternum, which are high-attenuation structures as explained by Defrise et al.¹⁹ It can also be noted that the quality of images in Figure 7 is slightly better than the corresponding images in Figure 8, and the quality of images in Figure 9 are better than the corresponding images in Figures 7 and 8. This phenomenon is more evident in the profiles. The poor image quality is observed in Figure 8 as compared with Figure 7 because the vertebra within the ROI was the only known subregion in the former case, and vertical inversion was carried out first. Serving as a known subregion, the poor approximation image of vertical strip image leads to an overall inferior image quality. Better image quality of Figure 9 can be attributed to 2 reasons: (1) horizontal inversion was carried out first, which provided a horizontal strip image as a better approximation to the known subregion for vertical inversion, and (2) some parts of vertebrae were additionally known that helped in the process of vertical inversion.

DISCUSSION

Because MDCT is now increasingly used for CVDs and other body systems, the radiation dose has become a growing concern. We believe that our exact local reconstruction approach²⁰ has a potential to reduce the radiation dose significantly in cardiac CT, improve the image quality, and be applied to other static body systems as well.

In this study, we designed a new 2-D phantom that is more realistic than the popular thorax phantom.²² Our interior reconstruction approach²⁰ was tested by reconstructing ROIs using various combinations of aorta and vertebra as known subregions. The method was tested using noisy and noise-free projection data, respectively. Our results indicate that it is feasible to use the attenuation coefficients of blood in aorta and/or vertebral body as prior information to carry out exact interior reconstruction. Some artifacts exist because of the limitations of POCS method and the presence of high-attenuation structures such as bones.

Future research directions include improving the image quality, reducing the computation time, identifying known-subregions, and testing with real projection data. Significant research efforts are necessary to improve the quality of reconstructed local images such that they are comparable to conventional reconstruction from nontruncated projection. Along this direction, it is necessary to try different methods like expectation maximization, and other optimization methods that might be able to provide better image quality. Being an iterative method, the POCS is computationally expensive and time consuming. Increased reconstruction speed may be achieved by implementing the iterative methods in a parallel fashion. Undoubtedly, developing an analytic method to inverse the Hilbert transform is the fast and best way. Which subregions within the ROI can be assumed to be known is an important issue too. Besides the blood and bone, it will be worthwhile to find other subregions that could be assumed to be known or develop strategies to introduce known regions when possible. As an example, we might introduce a calibration device in a specific position, such as axillary line (left) to provide an additional region of known information.

Currently, we are limited in our ability to apply our methods to real data because of the confidential nature of the raw projection data. We hope to overcome this obstacle in the near future. Applying these methods to real data will provide further insights into effectiveness and limitations of our approach. We are actively working to improve CT by reducing radiation dose and improving image quality to address challenging problems such as cardiac CT.

In conclusion, we have proposed a novel method to reconstruct the ROI from truncated projection data using the attenuation coefficients of blood in aorta and/or bone in vertebra as known information. The potential for dose reduction in cardiac CT has been demonstrated in numerical simulations with our new numerical cardiac phantom. In addition, a few research directions seem exciting to pursue.

Acknowledgments

The authors thank Debra Fuller, Josh Tan, and Carrie Smith of Wake Forest University School of Medicine for their assistance in obtaining and interpreting the clinical CT images.

This work was partially supported by the National Institutes of Health/National Institute of Biomedical Imaging and BioEngineering (grants EB002667, EB004287, and EB007288A).

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FIGURE 1.

Exact interior reconstruction settings. One known region adjacent to the reconstructible region (A) and 2 known regions adjacent to the reconstructible region (B), where $f(x)$ is the image and $g(x)$ is the Hilbert transform.

FIGURE 2.

Difference between conventional CT and interior CT. Conventional CT with the x-ray beam containing the object support (A) and interior CT with the x-ray beam only covering the ROI (B), where 1 is for an object support, 2 for a ROI, and 3 for a known subregion within the ROI.

FIGURE 3.
Fan-beam projection data acquisition geometry.

FIGURE 4.

Design of a new cardiac CT phantom. A clinical CT image with HU values mapped to gray-level values 0 to 255 (A) and the corresponding numerical phantom without contrast using -1000, 1000 as the display window (B).

FIGURE 5.

Slices showing the regions used to calculate the HU values for the new phantom. The slice used for estimation of representative tissue HU values (A) and the slice used for various bone HU values (B).

FIGURE 6.

Simulation setup with the outermost rectangles for the ROI and the narrow rectangular strips within the ROI for the first-pass reconstruction. The rectangle within aorta as the known region (A), the part of vertebra within the ROI as the known subregion (B), and both the rectangle within aorta and the part of vertebra within the ROI as known subregions (C). The display window is $-400, 400$ HU.

FIGURE 7.

Reconstructed ROI images for the simulation setup of Figure 6A with profiles shown for the dotted lines. A, Fan-beam backprojection with noise-free projection data. B, Fan-beam backprojection with noisy projection data. C, Parallel-beam backprojection with noise-free projection data. D, Parallel-beam backprojection with noisy projection data. Display window remains $-400, 400$ HU.

FIGURE 8.

Reconstructed ROI images for the simulation setup of Figure 6B with profiles shown for the dotted lines. A, Fan-beam backprojection with noise-free projection data. B, Fan-beam backprojection with noisy projection data. C, Parallel-beam backprojection with noise-free projection data. D, Parallel-beam backprojection with noisy projection data. Display window remains $-400, 400$ HU.

FIGURE 9.

Reconstructed ROI images for the simulation setup of Figure 6C with profiles shown for the dotted lines. A, Fan-beam backprojection with noise-free projection data. B, Fan-beam backprojection with noisy projection data. C, Parallel-beam backprojection with noise-free projection data. D, Parallel-beam backprojection with noisy projection data. Display window remains $-400, 400$ HU.

TABLE 1

Estimated Linear Attenuation Coefficients

Structures	Linear Attenuation Coefficient	
	cm ⁻¹	HU
Heart and aorta blood	0.1694	43
Tissue	0.1435	-116
Lungs	0.0254	-844
Ribs	0.259	595
Sternum	0.2183	344
Vertebra 1	0.2153	326
Vertebra 2	0.2538	563

TABLE 2

Description of the Vertebra in the New Phantom

No.	Objects	Center, cm	Axes Lengths Length, Breadth	Clip Lines (a, b, d)	Linear Attenuation Coefficient	
					cm ⁻¹	HU
1	Ellipse	0, -7	1.7, 1.85	0, 1, -8.3	0.2153	326
2	Ellipse	0, -7	1.0, 0.8	0, 1, -8.3	0.1435	-116
3	Rectangle	0, -8.4	1.1715, 0.1	—	0.2538	563
4	Rectangle	-2, -8.975	2, 0.475	-2, 32, -294.4; 0.95, -2.8285, 22.929	0.2538	563
5	Rectangle	2, -8.975	2, 0.475	2, 32, -294.4; -0.95, -2.8285, 22.929	0.2538	563
6	Rectangle	0, -9.35	0.8, 0.15	-2, 4, -38.7; 2, 4, -32, 294.4; -2, -32, 294.4	0.2538	563
7	Rectangle	0, -10.25	0.4, 0.75	—	0.2538	563
8	Ellipse	0, -11	0.4, 0.4	0, -1, 11	0.2538	563

The equation of clipping lines is of the form $ax + by > d$. These lines were derived from planes in Thorax Phantom [21]

TABLE 3

Description of Different Structures in the New Phantom

Structures	Center (x, y), cm	Rotation About z Axis, Degrees	Half-axes Lengths (x, y), cm	Clip lines (a, b, d)	Linear Attenuation Coefficient	
					cm ⁻¹	HU
Heart blood	-2, 3.5	-30	8, 6	—	0.1694	43
Heart tissue	-2.5, 3.5	-10	5.5, 3.5	—	0.1435	-116
Thorax tissue	-4, -4	—	17, 8	-1, 0, 0; 0, -1, 4	0.1435	-116
	4, -4	—	17, 8	1, 0, 0; 0, -1, 4	0.1435	-116
	-5, -4	—	16, 18	-1, 0, 0; 0, 1, -4	0.1435	-116
	5, -4	—	16, 18	1, 0, 0; 0, 1, -4	0.1435	-116
Lungs	-8, 1.25	-35	9, 11.5	0, -1, 5	0.0252	-844
	-8, 1.25	-35	9, 11.5	0, 1, -5; 0, -1, 3; -1, 0, 1	0.0252	-844
	-8, 1.25	-35	9, 11.5	0, 1, -3; -1, 0, 0; -0.3420, -0.9397, -8.664	0.0252	-844
	8, 1.25	35	9, 11.5	0, -1, 5	0.0252	-844
	8, 1.25	35	9, 11.5	0, 1, -5; 0, -1, 3; 1, 0, 1	0.0252	-844
Sternum	8, 1.25	35	9, 11.5	0, 1, -3; 1, 0, 0	0.0252	-844
	0, 11	-70	0.55, 1.9	1, 0, 0	0.2183	344
	0, 11	-70	0.1, 0.3	1, 0, 0	0.1435	-116
	0, 11	70	0.55, 1.9	-1, 0, 0	0.2183	344
Ribs	0, 11	70	0.1, 0.3	-1, 0, 0	0.1435	-116
	9, -10	85	0.5, 1.9	—	0.259	595
	9, -10	85	0.07, 0.75	—	0.1435	-116
	17.7, -5.5	-25	0.6, 1.5	—	0.259	595
	17.7, -5.5	-25	0.07, 0.75	—	0.1435	-116
	-18.5, -3.0	10	0.6, 1.55	—	0.259	595
	-18.5, -3.0	10	0.1, 0.75	—	0.1435	-116
	14.5, -9	-60	0.55, 1.7	—	0.259	595
	14.5, -9	-60	0.07, 0.75	—	0.1435	-116
	-17, -7	35	0.55, 1.55	—	0.259	595
	-17, -7	35	0.07, 0.75	—	0.1435	-116
	18.4, 0	10	0.5, 1.5	—	0.259	595

Structures	Center (x, y), cm	Rotation About z Axis, Degrees	Half-axes Lengths (x, y), cm	Clip lines (a, b, d)	Linear Attenuation Coefficient	
					cm ⁻¹	HU
	18.4, 0	10	0.07, 0.75	—	0.1435	-116
	-18.2, 2	-14	0.6, 1.5	—	0.259	595
	-18.2, 2	-14	0.07, 0.75	—	0.1435	-116
	17.2, 4.5	25	0.55, 1.55	—	0.259	595
	17.2, 4.5	25	0.07, 0.75	—	0.1435	-116
	-15.5, 7.5	-35	0.55, 1.6	—	0.259	595
	-15.5, 7.5	-35	0.07, 0.75	—	0.1435	-116
	11.5, 11	60	0.55, 1.7	—	0.259	595
	11.5, 11	60	0.07, 0.75	—	0.1435	-116
	-11.5, -10.2	-100	0.5, 1.8	—	0.259	595
	-11.5, -10.2	-100	0.07, 0.75	—	0.1435	-116
Aorta blood	-2.5, -1.5	—	1, 1	—	0.1694	43

The equation of clipping lines is of the form $ax + by > d$. Objects and clip lines were introduced to obtain the desired shape for the structures.