Determination of left ventricular mass in dogs with rapid-acquisition cardiac computed tomographic scanning

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ABSTRACT The development of left ventricular hypertrophy in patients with heart disease often has far-reaching clinical implications with respect to overall morbidity and mortality. Approaches used to assess left ventricular mass include electrocardiography, echocardiography, contrast ventriculography, single photon–emission tomography, and conventional computed tomography. However, all of these modalities suffer from some major draw back that precludes widespread application to all patients. In this study we assessed the accuracy of determinations of left ventricular mass in 22 dogs by rapid-acquisition (50 msec) computed axial tomography (RACAT), an ultrafast computed tomographic (CT) instrument. Electrocardiographically triggered, end-diastolic, short-axis cardiac scans were obtained from apex to base during administration of intravenous iodinated contrast. Myocardial edges were determined for each tomographic scan by two methods: the regional half-contour method (the CT density half way between that of the left ventricular myocardium and adjacent ventricular cavities or lung) and “interactive plateau thresholding” of the cardiac borders. Left ventricular mass by RACAT was calculated as the sum of the mass of each individual scan from apex to base (modified Simpson’s rule). Postmortem left ventricular mass ranged from 58 to 160 g. The correlation between true left ventricular mass and tomographically determined mass was excellent (r = .99), with the slope and y intercept not statistically different from 1 and 0, respectively. The standard error of the estimate was 4.1 g. Interobserver and intraobserver variability for determining left ventricular mass demonstrated excellent agreement (r = .99 and r = .99, respectively). We conclude that quantitative assessment of left ventricular mass can be accurately and reproducibly performed in dogs by rapid acquisition CT scanning. It is likely that this technique will be readily transferable to the clinical settings and prove to be an important method for quantifying left ventricular mass in patients.

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DEVELOPMENT of left ventricular hypertrophy is often a harbinger of adverse clinical consequences. It is known that the incidence of sudden death associated with acute myocardial infarction is increased threefold among patients with concomitant left ventricular hypertrophy and hypertension. Regardless of the cause, left ventricular hypertrophy has recently been implicated as a cause for diminished coronary artery reserve. Furthermore, in patients with left ventricular hypertrophy the functional integrity of the ventricle tends to deteriorate over time and may eventually result in overt heart failure.

Determination of left ventricular mass in vivo has been attempted by various techniques, including echocardiography, biplane left ventricular angiography, M mode and two-dimensional echocardiography, and 201Tl imaging in conjunction with single photon–emission tomography and gated and dynamic computed tomography, however, none are ideal.

A rapid-acquisition computed axial tomographic (RACAT) system is now available that is capable of obtaining real-time, stop-action tomographic images of the left ventricle in conjunction with excellent spatial...
tial resolution. The intent of this study was twofold: (1) to determine whether the theoretical advantages of RACAT scanning could be used to accurately determine left ventricular mass in dogs under conditions that would be potentially clinically applicable, and (2) to evaluate a myocardial identification algorithm that would facilitate determination of left ventricular mass.

Methods

RACAT scanner. The computed tomographic (CT) scanner used in this study was the Imatron C-100 (Cine CT). A prototype scanner was used for the first nine dog studies. The remaining 13 dogs were scanned on a production model. The scanner uses four semicircular x-ray-generating tungsten target rings that may be sequentially activated by focusing and sweeping an electrocardiographically triggered electron beam along each target ring. Each target ring may be activated in 50 msec, producing two contiguous, 8 mm thick tomographic slices. Each of the four targets is separated from its neighbor by 4 mm (figure 1). An individual target may be activated up to 17 times per second, thus effectively producing a tomographic cine composed of 50 msec tomographic frame images.

Animal preparation. Twenty-two dogs (weight 15 to 30 kg) were used in this study. Each was anesthetized with a combination of droperidol and fentanyl (Innovar-Vet, 1.0 ml/10 kg iv) and sodium pentobarbital (10 mg/kg iv). The dogs were mechanically ventilated via a cuffed endotracheal tube (tidal volume 12 ml/kg, rate 10/minute). Anesthesia was maintained by supplemental doses of sodium pentobarbital as needed. Electrocardiographic electrodes were attached to each dog to provide continuous monitoring of the heart rate as well as providing a trigger signal to the scanner. An angiographic catheter (No. 6F) was placed in the inferior or superior vena cava via a jugular or femoral vein. This catheter was to be used for contrast media injection. An 18-gauge catheter was placed in the femoral artery for continuous monitoring of arterial pressure.

Data acquisition

Experimental animal studies. Each dog was oriented in the scanner such that short-axis tomographic scans could be obtained from cardiac apex to base. This was accomplished by placing the dog right-side down with a shallow degree (15 to 25 degrees) of caudocranial slew (the dog’s head pointing approximately to 11:15–11:30 with respect to the scanner gantry in the y plane) of the imaging table with respect to the body’s long axis and scanner gantry. One or two localization scans were obtained without contrast to determine the scan level immediately below the cardiac apex. Once this level was identified, an intravenous injection of iodinated contrast (meglumine sodium diatrizoate) was administered over 15 sec. The dogs’ respirations were suspended at end inspiration and maintained for approximately 3 sec while scanning was performed. Electrocardiographically triggered scanning was begun 2 sec before the completion of the injection of the contrast media, thus assuring that adequate contrast enhancement would be present on both sides of the interventricular septum during data acquisition. The electrocardiogram triggered scans were initiated at 80% of the RR interval, thus allowing for acquisition of data during mid- to end-diastole. Sequential activation of the lowest and next highest target rings produced two pairs of scans (four contiguous slices) within 108 msec in a single cardiac cycle.

Two scanning methods of determining left ventricular mass were evaluated in this study. In the first method 15 dogs were evaluated by use of contrast injection 0.75 ml/kg of meglumine sodium diatrizoate for every four tomographic slices obtained. In this group only the central two target rings (figure 1, B and C) were sequentially activated during a 108 msec interval of a single cardiac diastole. With this method the completion of an apex-to-base mass study necessitated two injections. In the second method seven dogs were studied after a single contrast injection of 0.6 ml/kg was used to radiographically interrogate 8 cm of myocardium (four pairs of slices). This was accomplished by sequentially activating target rings D and C during the first cardiac diastole and then rings B and A on the second diastole. The rate administration of contrast was similar to that in the first set, although myocardial scanning was performed 3 to 4 sec before the completion of the bolus injection.

Postmortem measurement of myocardial mass. After the completion of the study, the animals were killed with an intravenous injection of potassium chloride. Their hearts were removed, the left ventricle was dissected free from the adjacent right ventricular free wall, and the atrium and right ventricular free wall were dissected from the anatomic left ventricle. Free-hanging right ventricular trabeculae were trimmed to the septal wall. Valvular tissue, atrioventricular groove fat, and epicardial fat surrounding the coronary arteries were carefully excised. The unfixed left ventricle was blotted dry and immediately weighed to the nearest 0.1 g.

Analysis of data. During the acquisition of each scan the projection data are supplied continuously to the scanner acquisition system and subsequently to the computer. Image reconstruction was performed by the filtered back projection technique. The data were processed to provide images composed of 256 × 256 picture elements (pixels). For the animal studies a reconstruction circle of 25 cm was used, so that each pixel represented a cross-sectional area of 1.02 × 1.02 mm². Each scan “thickness” was 8 mm. The resulting tomographic image were displayed on a monochrome video monitor with the brightness of each pixel linearly proportional to the x-ray attenuation coefficient of the tissues examined. The data are expressed in a quantitative manner in terms of Hounsfield numbers (Hn) derived from the equation

$$Hn = 1000 \left( A - A_w \right) / A_w$$

where A is the calculated x-ray attenuation coefficient of the tissue and $A_w$ is the attenuation coefficient of water. Thus water is assigned a Hounsfield number of 0. All tissue attenuation coefficients with and without contrast were compressed into a scale that ranged from –1000 (air) to +1000 (bone). Computer programs allowed selection for regularly or irregularly shaped regions of interest within a displayed image. The absolute Hounsfield numbers of these regions could be determined re-
Border identification. There were four myocardial borders in a tomographic image of the left ventricle that required separate identification before the left ventricular mass could be accurately determined. These four regions were the interface between the lung and epicardium, left ventricular cavity and adjacent endocardium, right ventricular cavity and right endocardial edge of the septum, and anterior chest wall and myocardium (figure 2). Individual determination of these four interfaces were necessary because the Hounsfield numbers of tissues adjacent to the left ventricular myocardial edges are highly variable between individual scan levels in the same animal as well as between individual dogs. Consequently, specific criteria were developed for identifying each of these edges. Two algorithms for myocardial border identification were evaluated. The first method was a regional half-height contour algorithm that has previously been described. The second method of border detection we term interactive plateau thresholding. With this technique the border in question was defined by evaluating the rate of change in distance vs CT density across a border. With each of these computer-assisted border identification methods the endocardial and epicardial borders of the left ventricle were traced superimposed on the displayed image with a trackball controlled cursor. The areas enclosed by the traced borders were automatically planimetered. The border of the papillary muscles were included in calculation in vivo of the left ventricular mass.

Image analysis

Regional half-contour border detection. At each corresponding tomographic scan level from cardiac apex to base 13 distinct regions of interest were defined: three in the right ventricular cavity, three in the left ventricular septum, one in the left ventricular cavity, three myocardial areas corresponding to the anterior lateral and posterior aspects of the heart, and three areas in the lung (figure 2). From these regions of interest average Hounsfield numbers were calculated for the lung, myocardium, septum, and right ventricular cavity, and appropriate window and level settings were determined.

Left ventricular cavity/endocardial edge. Without the aid of contrast administration it is virtually impossible to identify the interface between the ventricular cavity and endocardium because the CT tissue densities of myocardium and blood are nearly identical. However, with intravenous injection of iodinated contrast the ventricular cavities are clearly delineated. Six regions of interest, each approximately 5 mm2, were positioned midway between the epicardial and endocardial borders along the anterior, free wall, and posterior myocardium and left ventricular septum. In addition, a region of interest (7 × 7 mm2) was placed in the center of the left ventricular cavity (figure 2). Definition of the endocardial cavity edge was determined by setting the Hounsfield window level display setting half way between the average Hounsfield numbers of the four myocardial regions and that of the opacified left ventricular cavity. These window level settings varied for each image and between individual dogs, but ranged from approximately 50 to +150 Hounsfield units. After setting the window level the window width was set at 1, thus converting the image to a black and white binary image. With the aid of a track ball the black-white interface was traced. The outline of the papillary muscles was traced from the same binary image. The heterogeneity of CT numbers within the left ventricular cavity was less than 10% or ± 20 units in a given image at the center of the left ventricular cavity.

Right ventricular cavity/myocardial edge. In contradiction to the left ventricular cavity, which demonstrated a fairly homogeneous Hounsfield number distribution after contrast injection, the right ventricular cavity was less uniform. This difference may be related to the irregular shape of the right ventricle, heavy trabeculations in the right ventricle, and inhomogeneous admixture of contrast agent and blood following a right-sided contrast injection. Three separate regions of interest were placed in the interventricular septum. The average of these three was obtained. An additional three regions of interest were then placed in the right ventricular cavity close to the interventricular septum and the average Hounsfield number of these three was determined. Analogous to the determination of the left ventricular cavity endocardial border the septal border of the right ventricle was defined as the half-counter Hounsfield number between the average Hounsfield number of the right ventricular cavity and the average number for the interventricular septum. In a short-axis scan this edge defines approximately 33% of the left ventricular myocardium outer circumference.

Lung/epicardial edge. The placement of the lung/epicardial edge was facilitated by the large differences between the Hounsfield numbers of the myocardium (usually between +50 to +150) and the adjacent lung (less than −400). Unlike the ventricular cavity/endocardial edge the visual placement of the lung/epicardial edge is highly dependent on the Hounsfield window level chosen for display. The left ventricular--lung edge was defined with the use of the half-contour edge detection algorithm by analogy to the edge-detection method described.
above. The resulting values of the Hounsfield numbers ranged from −250 to −400, within the “plateau” range of Hounsfield numbers as described above. This edge defines approximately 56% of the left ventricular epicardial surface in the short axis.

Anterior chest wall/myocardial edge. The identification of the anterior chest wall/myocardial edge was facilitated by the observation that in the closed-chest dog there is a distinct visual plane separating these areas. A simple mathematic approach to defining this interface could not be developed because the Hounsfield numbers of the anterior mediastinum and adjacent myocardium are roughly the same. Consequently, the placement of the anterior chest wall/myocardial edge was made directly from the image display. This interface usually involved less than 10% of the total circumference and thus minor errors in this edge placement would not result in significant errors in estimation of left ventricular mass.

Interactive “plateau” thresholding method of edge detection. The computer software permits the operator to highlight, by means of the “blink” or “identify” mode, all those pixels within a range of CT densities. This highlighting is accomplished by alternation of the displayed intensity (grey level) of the pixel between bright white and the original grey-level values. For example, a window level of 100 and a window width of 1 will identify only those pixels with a CT density of 100. At a window level of 100 and a window width of 10 the identify mode will highlight all those pixels from 95 to 105 CT units. When the window width is set equivalent to the inherent background noise level of this instrument (approximately 15 CT units), as the window level blink is progressively increased from negative to positive numbers (lung to myocardium), the myocardial borders can be determined by observing that at various times the CT numbers in the blink mode align themselves in a uniform curvilinear pattern identifying first the epicardial and then the endocardial edges. The CT numbers at which this alignment occurred represented the values of the border of interest and corresponded closely to the CT density defined by the half-contour method previously described. Initially, with the use of the blink mode with a window level set at the mean lung density (−600 to −900), a random distribution of highlighted CT pixels was produced within the thorax. As the window level was progressively increased, the ragged and disorderly appearance of the highlighted pixels was suddenly outlined by a smooth curvilinear pattern in the epicardial edge. This pattern circumscribing the perimeter of the epicardium persisted without significant alteration as the threshold was increased over a range of CT numbers of approximately −250 to −350 CT units (the plateau) (figure 3). With further reduction in the window level, the highlighted pixels of the edge rapidly dispersed into the left ventricular myocardium and remained in a disorganized pattern for the next 150 to 300 CT units.

Further increases of the window level eventually produced a clear and well-organized curvilinear outline of the left ventricular endocardial border. Once again the position of the highlighted endocardial border remained constant and uniform for a plateau range of approximately 50 to 75 CT units. Further increase in the window level resulted in an abrupt and chaotic disorganization of the highlighted pixels that rapidly dispersed into the ventricular cavity. The right ventricular cavity and endocardial edge were determined in a similar manner. The limits of the threshold plateau for the endocardial, epicardial, and right ventricular endocardial interfaces were determined. After determining the CT limits of each border “plateau,” the window blink level was set at one-half this number. With a track ball the pixels highlighted by the blink CT numbers outlining each myocardial interface was traced.

Calculation of in vivo left ventricular mass. The mass of each short-axis left ventricular slice was then calculated as a product of the planimetered myocardial area, scan slice thickness (0.8), and specific gravity of the myocardium (1.05 g/cm³). The data for the 4 mm gap of unscanned myocardium was estimated by linear interpolation between the slice above and below the gap. The cardiac apex was defined as occurring at the lowest chest level scan that contained definable myocardium. The cardiac base was identified as occurring at the scan level immediately below the scan level at which the aorta appeared as a distinct structure (i.e., a circle). The left ventricular myocardial mass was calculated as a sum of the masses of the individual scanned sections (modified Simpson’s rule).

Statistical analysis. Statistical analysis of the 22 data points comparing the left ventricular mass derived by RACAT scan-
ning in vivo and the postmortem left ventricular mass were analyzed by linear regression. The first seven dogs comprised a training set in which the two edge-detection algorithms were validated. The remaining 15 dogs comprised the test set of experimental animals. The linear correlation coefficient, slope, and y intercept were calculated. Standard error of the estimate was determined in the usual fashion. Interobserver and intraobserver variability were calculated for representative animals in which left ventricular masses spanned the full range of those in all animals studied. The reproducibility of determinations of left ventricular mass was examined in six animals scanned for left ventricular mass between two and five times each.

Results

Data on left ventricular mass are presented for 22 dogs. In these dogs, heart rate varied from 60 to 90 beats/min and mean aortic blood pressure varied from 70 to 110 mm Hg.

A representative series of short-axis scans from cardiac apex to base in one experimental animal is shown in figure 4. The right ventricular tree wall is not readily apparent because of technical limitations in photo-

**FIGURE 4.** Typical short-axis scans in one dog from cardiac base to apex (right to left, top to bottom). The graphic at the upper left of each panel demonstrates the relative cardiac level at which each tomographic scan was obtained. A. The three most basal slices; B the midventricular through apical slices. All scans were obtained near end-diastole. Note the excellent spatial resolution of the left ventricular cavity, anterior and posterior papillary muscles, right ventricular cavity, and anterior, lateral, posterior, and septal myocardial surfaces. (The right ventricular free wall is not well visualized as a consequence of limitations in photographic reproduction.)
graphic reproduction. The data acquisition time for two pairs of 8 mm thick contiguous scans was 108 msec (4 cm of myocardium). All scans were obtained at mid- to end-diastole by electrocardiographic triggering at 80% of the RR interval. This method of scan acquisition provides excellent delineation of the left and right ventricular cavities and structure and clearly demonstrates the anterior, lateral, and posterior myocardial walls and interventricular septum.

Between six and 10 tomographic slices were required to scan the entire left ventricle from apex to base. In four animals in which the entire ventricle could not be scanned with the initial eight tomographic sections the table was incremented by 8 cm and the remainder of the ventricle scanned. Figure 5 shows the results of RACAT determination of left ventricular mass as compared with left ventricular mass at autopsy in all 22 dogs. Left ventricular mass ranged from 58 to 160 g, with a mean of 98.7 ± 33 g. The correlation coefficient and linear regression equation for the entire animal population (n = 22) by the interactive plateau thresholding technique were \( r = .99, y = 0.97x + 1.98 \), SEE = 4.1 g (figure 5). The slope and y intercept were not statistically different than 1 and 0, respectively. By the interactive plateau thresholding edge-detection method, the respective correlation coefficient, linear regression equation, and standard error of the estimate for the initial training set (n = 7) and testing set (n = 15) were \( r = .97, y = 1.02x - 1.09 \), SEE = 3.5 g and \( r = .99 y = 0.97x + 1.88 \), SEE = 4.3 g. There was no statistical difference between the slope and intercept for either the testing or training data set.

Observer variability for left ventricular mass determined by RACAT scanning. With the interactive plateau thresholding technique the intraobserver and interobserver variability for analysis of 42 ventricular scans from six dogs with myocardial masses of between 58 and 150 g was excellent (intraobserver variability \( r = .99, y = 0.96x + 3.5 \) [figure 6, A]; interobserver variability \( r = .99, y = 0.96x + 4.8 \) [figure 6, B]).

Variability between edge-detection algorithms. Fifteen random animals were evaluated with both the half-contour and interactive plateau thresholding edge-detection algorithms. The correlation coefficients, linear regression equations, and standard errors of the estimate for the two edge-detection methods were virtually identical. The autocorrelation between left ventricular mass determined by the half-contour and interactive plateau thresholding techniques was \( r = .99, y = 1.02x - 1.83 \), SEE = 3.25. The range of error for determination of the left ventricular mass with the half-contour method was from 0 to 7 g with an average of 3.3 g vs a range of 0 to 8 g with an average of 3.7 g for the plateau threshold technique.

Reproducibility of determinations of left ventricular mass. Studies of the reproducibility of scanning and determination of left ventricular mass was performed in six animals. Figure 7 demonstrates that determination of left ventricular mass in the same animal after repositioning and rescanning was highly reproducible with a small range of error (0 to 5 g).

Effect of positioning on determination of left ventricular mass. The relationship between position of the animal and accuracy of mass determination was evaluated in five dogs (range of mass 85 to 133 g). The dogs were
positioned in the scanner right-side down, with the table slewed clockwise 20 degrees with respect to the center of the gantry (figure 7, second data point in lines A thru E). The range of error for determination of left ventricular mass for the half-contour edge-detection method was 0 to 5 g and the range of error for plateau thresholding was 0 to 4 g.

Discussion

This study represents the first attempt to determine left ventricular mass in vivo with a commercially available RACAT instrument. The data demonstrate that by use of the RACAT technique and the method of cardiac edge definition described, left ventricular mass in the dog can consistently and accurately be determined.

The discussion section will be divided into three components: (1) comparison of RACAT determination of left ventricular mass with other methods, (2) myocardial edge detection and partial volume effect, and (3) potential problems, risks, and benefits when this technique is applied to patients.

Comparison of methods of determination of left ventricular mass in vivo. There are four clinical methods, in addition to computerized tomography, that have been used to determine left ventricular mass in vivo. These are (1) electrocardiography, (2) contrast ventriculography, (3) single photo–emission computed tomography, and (4) echocardiography. Determination of left ventricular mass from the surface electrogram is unreliable.4,5 Biplane angiographic estimates of left ventricular mass have been shown to correlate reasonably well with postmortem left ventricular mass despite its inherent dependency on fixed geometric assumptions of left ventricular shape.6,7 Unfortunately, the invasive nature of this procedure limits its practical application, especially for serial determination of left ventricular mass. The experimental accuracy of single photon–emission tomography with 201Tl is comparable to that of biplane angiography.13 However, since the distribution of 201Tl parallels myocardial perfusion, determination of left ventricular mass in patients would be severely limited by prior myocardial infarction or severe coronary artery disease. Compared with the earlier estimates of left ventricular mass with M mode echocardiography,8 two-dimensional echocardiographic imaging has substantially improved the accuracy of determination of left ventricular mass.9–11 Unfortunately, high-quality two-dimensional echocardiograms from standard transducer positions may not be obtained in a substantial number of adult patients. This limitation has hindered the widespread ap-

![Graphical representation of data](image-url)
plication of echocardiographic estimation of left ventricular mass to the majority of patients with cardiac disease.

Studies by Skioldebrand et al.\textsuperscript{14} and Peck et al.\textsuperscript{15} have demonstrated the feasibility of the use of computerized tomography to determine left ventricular mass in dogs. The results of these studies demonstrated that CT estimates of left ventricular mass were highly correlated with postmortem mass (r = .94, .95; SEE 8.4, 8.1 g, respectively). However, because of the excessively long scanning time (2 to 4 sec per tomographic image), these studies were performed under less than ideal physiologic conditions that included prolonged breath holding or respiratory paralysis. In addition, because only one scan could be acquired at a time the study necessitated multiple couch increments as well as continuous infusion of contrast medium. Despite these shortcomings, the potential applicability of computed tomography to the determination of myocardial mass was established.

Iwasaki et al.,\textsuperscript{19} using the dynamic spatial reconstructor (DSR) (an ultrafast CT research scanner capable of obtaining millimeter-thin multiple-level tomographic scans in under 20 msec), have recently reported highly accurate estimates of left ventricular mass (r = .99, SEE = 4.1 g) in eight dogs. In that study left ventricular mass was determined after three-dimensional reconstruction of a left ventricle from 1.8 mm slices and reformating the section to obtain 3.6 mm short-axis scans from apex to base. This is in contrast to our method whereby short-axis scans were directly obtained through the majority of the left ventricle. The tomographic thickness in the study by Iwasaki et al. was approximately one-half that of the present study (3.6 vs 8 mm). Theoretically, thinner tomographic slices should result in more accurate estimates of ventricular mass.\textsuperscript{20} However, despite significant differences in slice thickness, the results of the present study are virtually identical to those of Iwasaki’s. These findings demonstrate that subsecond tomographic techniques such as the DSR and RACAT methods provide the most accurate experimental estimates of left ventricular mass in vivo.

Accuracy of border identification. In this study we have used two algorithms for border detection. Interactive plateau thresholding in the blink mode allows for a rapid and accurate method of edge detection. Regional edge detection by the half-contour algorithm is time-consuming, but is equally as accurate. Figure 3 demonstrates a plot of CT numbers vs pixel location in a profile traversing the lung, myocardium, left ventricular cavity, septum, and right ventricular cavity in the right ventricular free wall. A clearly defined plateau (or steepest descent) corresponding to the myocardial border is observed. Plateau A defines the edge between the lung and the partial volume corona of the epicardial surface. Note that this plateau defines the area of least rate of change whereas change in CT numbers from \(-650\) to \(-200\) is accompanied by only a minimal (less than 1.5 mm) change in distance. Likewise, the endocardial edge is found along plateau B, whereas the change in CT numbers from \(+100\) to \(+230\) is accompanied by a minimal change in distance. In a sense the half-contour method and interactive plateau thresholding techniques are quite similar. If the mean CT numbers of the myocardium (x = 100) and the lung (y = \(-700\)) are used to compute the half-contour CT number (\(-400\)), then it is observed that this number will always fall along the lung-epicardial plateau.\textsuperscript{2} Consequently, it is likely that the regional half-contour algorithm for border detection works because it fortuitously falls within this plateau.

Myocardial edge detection and the partial volume effect. In this experiment each display pixel represented a voxel of 1.02 mm \(\times\) 1.02 mm \(\times\) 8 mm. The attenuation number (Houndsfield units) computed for each voxel is dependent on the relative proportion of the various tissue constituents that it contains. The partial volume effect results from inclusion of two dissimilar tissues (i.e., lung/myocardial or contrast-enhanced cavity/myocardial interface) in a voxel that is to be investigated. For example, in a voxel that contains 3 mm of myocardium and 5 mm of lung tissue (i.e., an apical slice) the Hounsfield number for the myocardium will be a negative number, although it is known that tissue density structures are all above zero. Consequently, if the myocardial borders were drawn with the use of set display parameters for all scans, centering only about positive Hounsfield numbers, a serious underestimation of myocardial mass would occur. Moreover, partial volume problems are more pronounced at the apex and base of the heart. The presently used algorithms for border detection are validated by the highly corellative relationships demonstrated in this study.

Potential problems, risks, and benefits of determination of left ventricular mass in patients by RACAT. Determination of left ventricular mass in patients would be facilitated by serial short-axis tomographic acquisition from apex to base. Although previous reports of patient studies employing cardiac computed tomography have not used short-axis projections, evaluation of the left ventricle in the short axis view is important for a number of reasons. Theoretically, myocardial edge detec-
tion is facilitated and the compounding influences of partial volume effect are minimized in this view. Second, Simpson’s approximation for determining left ventricular mass and volumes should be facilitated by use of scans perpendicular to the left ventricular long axis. Finally, the short-axis view is one of the most clinically desirable projections since it allows the inspection of the entire circumference of the heart at any level, with the exception of the apex. This is in contrast to the usual transaxial view obtained with conventional computed tomography, which obscures the posterior and inferior walls. Preliminary studies in our laboratory have demonstrated that by the use of a combination of cranial tilt and clockwise slew of the imaging table multiple short-axis views can be acquired in patients. One potential problem with this mode of acquisition is that a portion of the ventricular circumference of the lowest apical tomograms lies adjacent to the liver and thus edge detection in this area may be imprecise. However, the absolute error of border identification would be small compared with the total left ventricular mass.

The presence of irregular cardiac rhythms of frequent premature beats may cause inaccuracies in determination of left ventricular mass since this method involves nonsimultaneous acquisition of tomographic changes. Consequently, for the present time it would seem prudent to avoid determination of ventricular mass in subjects with grossly irregular rhythms.

The risks of RACAT determination of left ventricular mass in man relate to the cumulative exposure to radiation as well as the known risk of intravenous injection of iodinated contrast material. The dose of radiation to the heart is approximately 250 mrad.s. If only left ventricular mass is to be determined each section of the heart is scanned only once. Since the x-ray beam is well collimated, the cumulative dose to the whole organ per mass study is 250 mrad.s. In comparative terms the total dose of radiation for a RACAT mass study is about 2 to 2½ times greater than that for a routine posterior-anterior and lateral chest x-ray (100 mrad.s) or equivalent to 2 to 3 sec of fluoroscopy (exposure 80 to 160 mrad/sec).

Intravenous administration of iodinated contrast agents has been associated with a wide variety of minor side effects, although significant adverse reactions rarely occur. With the emergence of the newer isosmotic, nonionic contrast agents, the potential hazards associated with intravenous contrast agents should be substantially reduced.

In conclusion, we have measured left ventricular mass in 22 dogs with a high degree of accuracy using a RACAT scanner. This report presents a highly accurate method of estimation of left ventricular mass in vivo. The extremely favorable interobserver and intraobserver concordance for analysis of ventricular slices lend support to the algorithms for myocardial border identification used. Rapid-acquisition computed tomography represents a precise and minimally invasive approach for determination of left ventricular mass that should readily translate to the clinical arena.

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