CT of the Heart

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CT OF THE HEART

PRINCIPLES AND APPLICATIONS

EDITED BY

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Foreword

Radiologic technology has made dramatic advances in the last 25 years, and none have been more impressive than those in computed tomography (CT). The progress in the speed of obtaining images, computing, postprocessing, and spatial resolution has been incredible. The result is that CT has moved from displaying purely morphologic information to providing valuable physiologic data as well. Whether with electron beam or multidetector-row CT, advances are impressive and nowhere have the applications been more useful and dramatic than in the heart.

This multiauthored book, *CT of the Heart*, edited by U. Joseph Schoepf, MD, is a splendid rendition of the state-of-the-art in CT imaging of the heart; however, where appropriate, it also features comparisons with other technical approaches, such as magnetic resonance and ultrasound. The contributors are leading radiologists, cardiologists, physicists, engineers, and basic and clinical scientists from Europe, the United States, Israel, and Japan.

The entire contents are meticulous and comprehensive, from the introduction about the past, present, and future of CT of the heart, through the technical underpinning of the method and the various clinical, physiologic, and pathologic applications of CT in studying the heart.

This book fills an immense need, particularly at a time when cardiac screening with CT, whether one agrees with this practice or not, is a reality. Furthermore, with the rapid increase of aging populations in the industrialized world, noninvasive diagnostic approaches are increasingly needed. As technology continues to advance and applications of CT to heart studies expand, it is my hope that the editor will bring this book up to date with a new edition.

Alexander R. Margulis, ***MD, DSc (HON)***
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Preface

Through the ages of exploration and enlightenment the heart has kept its fascination as the metaphor of life. Its firm entrenchment in human emotion and consciousness and the overwhelming socioeconomic importance of its diseases make noninvasive visualization and diagnosis of the heart and its diseases the coveted "holy grail" of medical imaging.

Imaging of the heart has always been technically challenging, because of the heart's continuous motion. The introduction and ongoing technical improvement of fast ECG-synchronized computed tomography (CT) scanning of the heart has enabled imaging of the elusive but cardinal cardiac anatomy and pathology with a combination of speed and spatial resolution that is hitherto unparalleled by other noninvasive imaging modalities. Accordingly, considerable interest has been directed in recent years at the beneficial utilization of CT for noninvasive interrogation of the coronary arteries and for imaging studies of anatomic and functional sequelae of ischemic heart disease, such as cardiac perfusion, motion, and viability. The current and potential future roles of CT for these and other applications are the subject matter of this book.

CT of the Heart, however, does not claim to have all of the answers. CT of the heart is a nascent but rapidly evolving field, and the act of condensing expert knowledge and experience in the format of a book can only result in a snapshot of the status quo at a certain point in time. Novel iterations of existing technology and profoundly new concepts of medical imaging are already on the horizon. The benefits and indications of integrating CT into the diagnostic algorithm of heart disease is intensely researched and discussed: to date, the diagnostic value of CT coronary calcium measurements and the exact role of this marker for cardiac risk stratification remain unclear and controversial. We are only beginning to understand the usefulness and potential clinical application of CT angiography for noninvasive detection of coronary artery stenosis. Cross-sectional assessment of the coronary artery wall for noninvasive identification, characterization, and quantification of atherosclerotic lesions and disease burden is a promising and exciting but yet untested concept.

Within *CT of the Heart*, the reader will be exposed to a variety of expert opinions on the respective topics. Some authors will assume a more optimistic or a more conservative perspective on cardiac CT applications. Because of the lack of large-scale clinical studies, only future experience will show who may be right. It is a declared goal of this book to showcase the full scope of current developments, research, and scientific controversy regarding the principles and applications of CT of the heart. Truth is most likely to be found in the equilibrium of opinions. The publisher and I have striven to maintain this equilibrium by providing a platform for differing opinions and for different technical approaches to CT of the heart. To mitigate commercial overtones and bias, which so often accompany the first steps of a potentially important new technology, contributions of users and/or developers of all cardiac CT manufacturers were included. Scientists representing different companies graciously disregarded commercial divisions and agreed to co-author chapters in order to provide the reader with a truly balanced view on cardiac CT technology.

Accordingly, *CT of the Heart* is the work of many. I am indebted to Dr. Christopher Cannon, the series editor, and to Paul Dolgert of Humana Press for entrusting me with the role of editor. I am grateful to my chairman, Dr. Philip Costello, for his unfailing guidance, support, and friendship. I feel very, very honored by all the kindness that my many friends in the cardiac imaging community have shown me by volunteering their time, their knowledge, their experience, and the vision that went into their respective contributions. All authors are highly respected experts in their fields and this book would never have come to pass without their incredible support, for which I am so grateful. Finally I would like to thank Tracy Catanese and Craig Adams of Humana Press for so efficiently and expertly steering the production of *CT of the Heart*.

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INTRODUCTION AND HISTORICAL BACKGROUND

I

1 CT of the Heart

Past, Present, and Future

WILLIAM STANFORD, MD

INTRODUCTION

Imaging of the heart and great vessels has previously been done with plain film, cardiac catheterization, nuclear medicine, and echocardiography as the primary imaging modalities. The recent newer advances in the computed tomography (CT) and magnetic resonance imaging (MRI) technologies, however, have dramatically changed our approach to imaging cardiac disease. CT and MRI, supplemented by CT angiography and MRI angiography, are increasingly replacing the chest film, as well as nuclear and—to some extent—echo imaging as the primary modalities in evaluating heart disease.

CT images, which initially took over 4 min to generate, can now be obtained in 50–100 ms with electron beam imaging and in 125–500 ms with helical imaging. Importantly, these advances in temporal resolution are rapid enough to essentially stop cardiac motion, and thus visualization of extremely small structures such as calcium deposits within the walls of the coronary arteries is now possible. Along with this increased temporal resolution has come a concomitant increase in spatial resolution, and isotropic voxels as small as 0.5 mm^3 are now identifiable. These advances, along with the increasing use of 3D reconstruction techniques, have revolutionized cardiac imaging and have moved CT from only anatomic visualization into the arena of functional and perfusion imaging.

HISTORICAL

CT was first introduced by Sir Godfrey Hounsfield in the 1970s—a short 30 yr ago *(1,2)* (Fig. 1). Hounsfield was an electrical engineer working for EMI, an electronics firm in England. While there, he conceived the idea of taking cross-sectional X-ray data and reformatting these data into images. For this, he and Alan M. Cormack, a Tufts professor of mathematics working independently, received the Nobel Prize in medicine in 1979.

Hounsfield's first-generation scanner used a translate/rotate technology. In this methodology, an X-ray source moving laterally activated a series of single detectors before moving to another position and repeating the process. This translate/rotate process was repeated until the entire circumference of the patient was scanned (Fig. 2A). Scan times of 4.5 min per image were required.

The technology was then advanced with the addition of fan beam architecture to the translate/rotate process. With this refinement, each radiation beam activated multiple detectors rather than a single detector. Thus increased numbers of images were possible from each tube activation. With this upgrade, imaging times were reduced to approx 2.5 min per image (Fig. 2B).

The next advance was the introduction of a continuously rotating X-ray tube coupled with a continuously rotating detector array. This further decreased scan times to approx 18 s (Fig. 2C).

Around 1978, the rotating detector array concept was changed to that of a fixed detector array, and this further decreased scan times to approx 2 s per image, and this is the configuration present in many of our conventional CT scanners in operation today (Fig. 2D). As a consequence of these advances, CT was able to become a major workhorse for whole-body imaging, which included many cardiac applications. However, although the excellent resolution and absence of overlying structures allowed visualization of the pericardium and many of the relatively static abnormalities such as intracardiac filling defects from thrombi and tumors, the contracting heart was not well visualized.

An additional problem with this conventional technology was that the scanner cabling restricted tube movement and allowed for only a single tube rotation to a fixed point before the tube had to be returned to its original position. Thus a continuous tube movement was not possible and scan times were relatively fixed. Because of this, cardiac applications were restricted to a 2-s-per-image acquisition time. Yet, the ability to visualize not only the vessel lumina and cardiac chamber endocardium but also the vessel wall and the surrounding myocardium were important applications not heretofore possible. Thus CT provided a more comprehensive view of cardiovascular pathology.

In the early 1980s, an important advance moved CT into the realm of cardiac imaging. This was the introduction of the electron beam technology concept by Dr. Douglas Boyd of the University of California, San Francisco *(3)*. The electron beam scanner, while having an appearance similar to a conventional CT scanner, did not have an X-ray tube rotating around the patient. Instead, electrons were generated by a source and then

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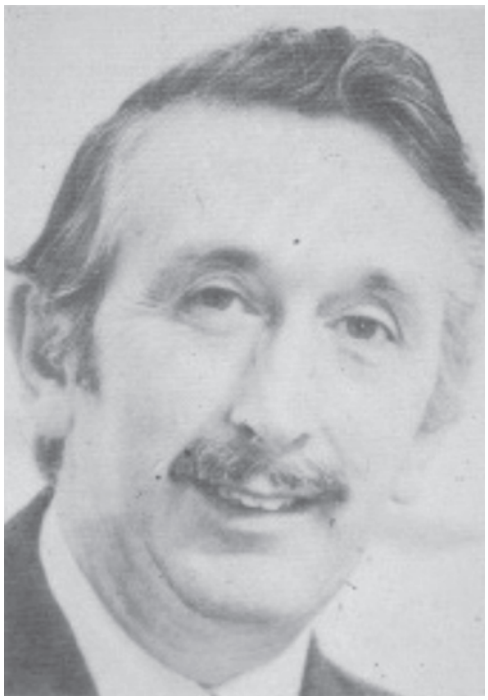
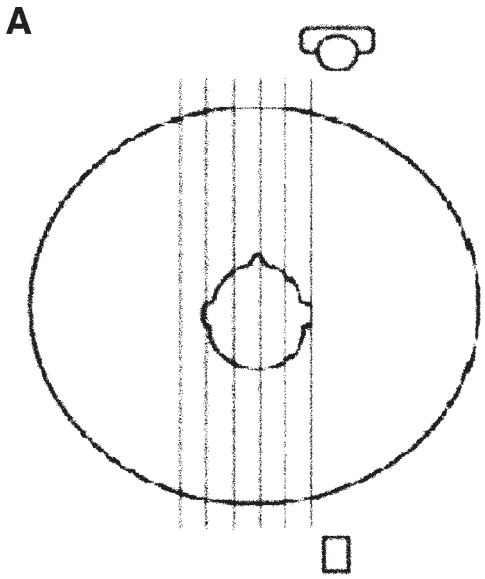
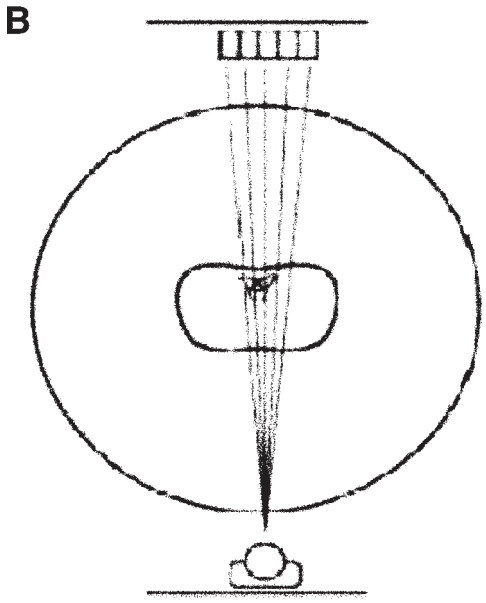
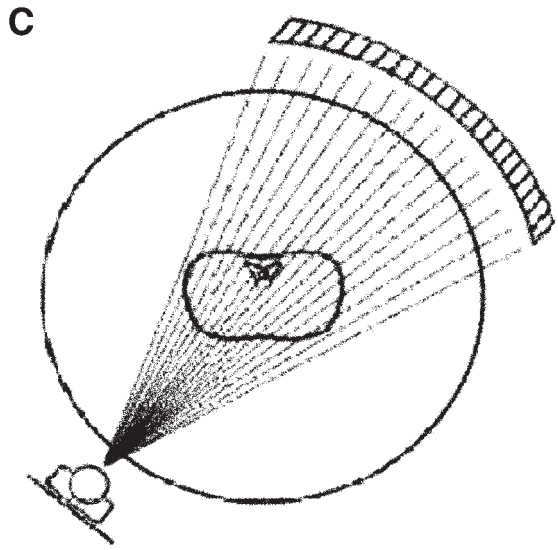


Fig. 1. Godfrey N. Hounsfield, the father of computed tomography. For their achievements, Hounsfield and Alan M. Cormack were awarded the Nobel Prize in medicine in 1979.







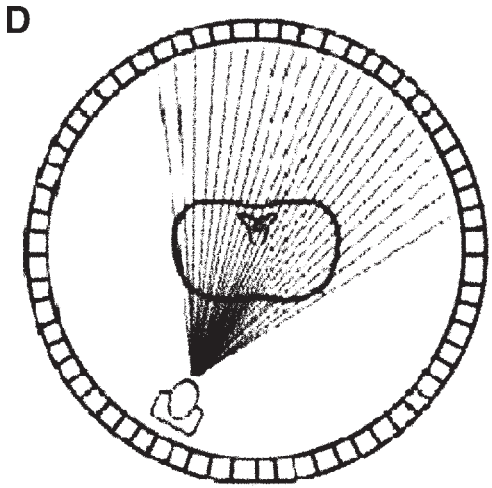


Fig. 2. **(A)** Diagram of the first-generation translate/rotate CT scanner. To create the image, a series of exposures were taken as the X-ray source moved a short distance laterally. The tube then rotated to a different position and the sequence was repeated. A single image required 4.5 min exposure time. **(B)** A later modification of the translate/rotate sequence introduced fan beam architecture to activate multiple detectors before moving to a new position. With this modification, scan times were reduced to 2.5 min/image. **(C)** The next generation of scanners used a rotating X-ray source coupled to rotating detectors. Scan times were reduced to 18 s/image. **(D)** Subsequent modifications used a rotating X-ray source with a fixed detector sarray. Scan times were 2 s/image.

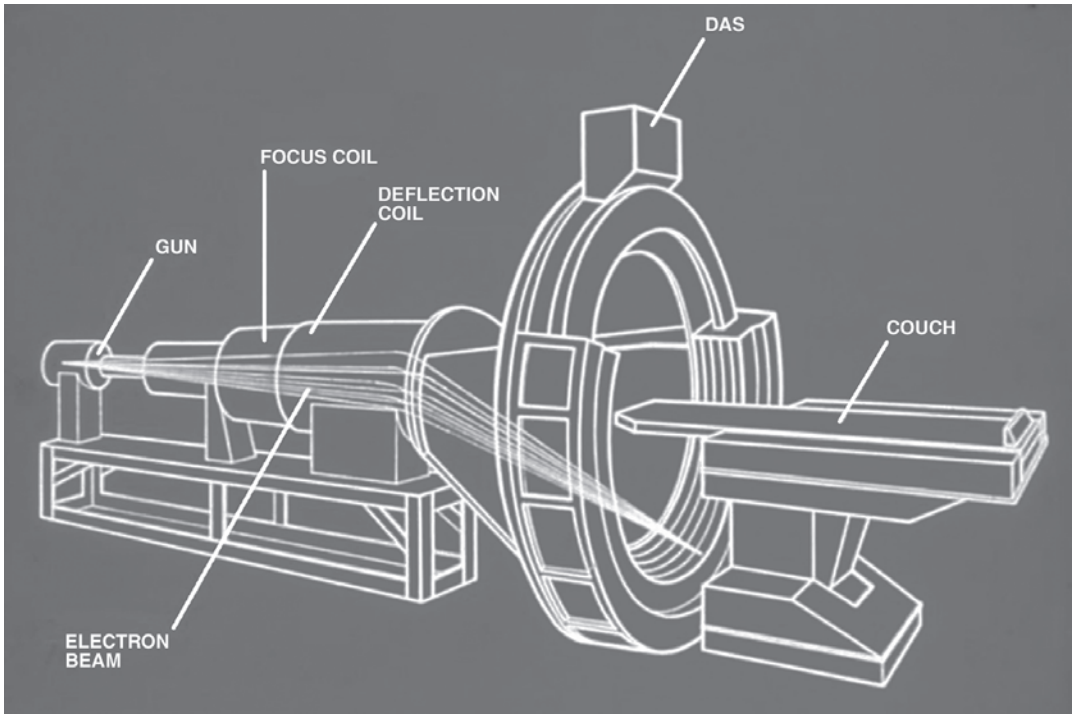


Fig. 3. Cross-sectional diagram of the GE-Imatron C150 XP electron beam scanner. Electron sweeps of tungsten target rings generate X-rays that traverse the patient to activate detectors in the gantry over the patient. With this development, imaging times were reduced to 50–100 ms. (Courtesy GE-Imatron Inc.)

bent electromagnetically to sweep tungsten target rings located in the gantry beneath the patient. The X-rays produced by the electron sweep traversed the patient and were collected by fixed solid-state detectors located in the gantry above the patient (Fig. 3). This technology decreased scan times to 50–100 ms, which essentially froze cardiac motion and thus dramatically changed our ability to image the beating heart. For the first time it was possible to view cardiac contractions and to visualize small structures such as calcium deposits within the walls of the coronary artery.

An additional major advance in CT imaging came in the early 1990s with the introduction of helical/spiral CT imaging and its slip-ring technology *(4)* (Fig. 4). With these advances, the X-ray beam was able to continuously rotate around the patient as the patient moved through the scanner gantry. These innovations decreased scan times to approx 500–1000 ms and ultimately produced data sets with spatial resolutions as small as 0.5 mm^3. The initial platform for helical CT scanners consisted of a single X-ray source and a single detector ring; however, subsequent developments in helical CT technology have enlarged imaging platforms from a single detector to 4 rows and now 16 rows, and in the future 32-, 64-, and 256-row detectors are expected. Along with the increase in numbers of detector rows were advances in the temporal resolution, and now imaging times now as fast as 420 ms are possible. These and other innovations in multislice helical CT have allowed the entire heart to be scanned within a 20–30 s breath-hold, and do so with excellent spatial resolution *(5)*.

THE PRESENT

CONVENTIONAL CT

The design of our current conventional CT scanners is that of a rotating X-ray source activating fixed detectors in the gantry surrounding the patient. The tube movement is limited by cables so that once the tube rotates around the patient, it has to stop and rewind. This constrained tube travel is a major limitation of conventional CT; however, the cross-sectional image detail is excellent and spatial resolutions of 9–12 line pairs (lp)/cm are possible.

ELECTRON BEAM CT

The electron beam CT (EBCT) scanner (GE-Imatron) works on a different principle than does conventional CT. With EBCT, electrons sweep tungsten target rings to produce X-rays that enter from beneath the patient, and are attenuated and collected by solid-state detectors in the gantry above the patient. The technology, because of the absence of moving parts, dramatically decreases scan times to 50–100 ms, which is rapid enough to essentially freeze cardiac motion.

There are several different EBCT scanner models in operation. The older C150 and C300 scanners operate at 130 kV, 625 mA, 83 kW to produce 100-ms, 1.5-, 3-, and 6-mm slice thickness images at resolutions of 9.5 lp/cm. The technology also makes it possible to generate flow and movie mode 50-ms, 8-mm dual slice thickness images from four target rings. The latter ability is important in functional imaging; its spatial resolution is limited at 4.5 lp/cm.

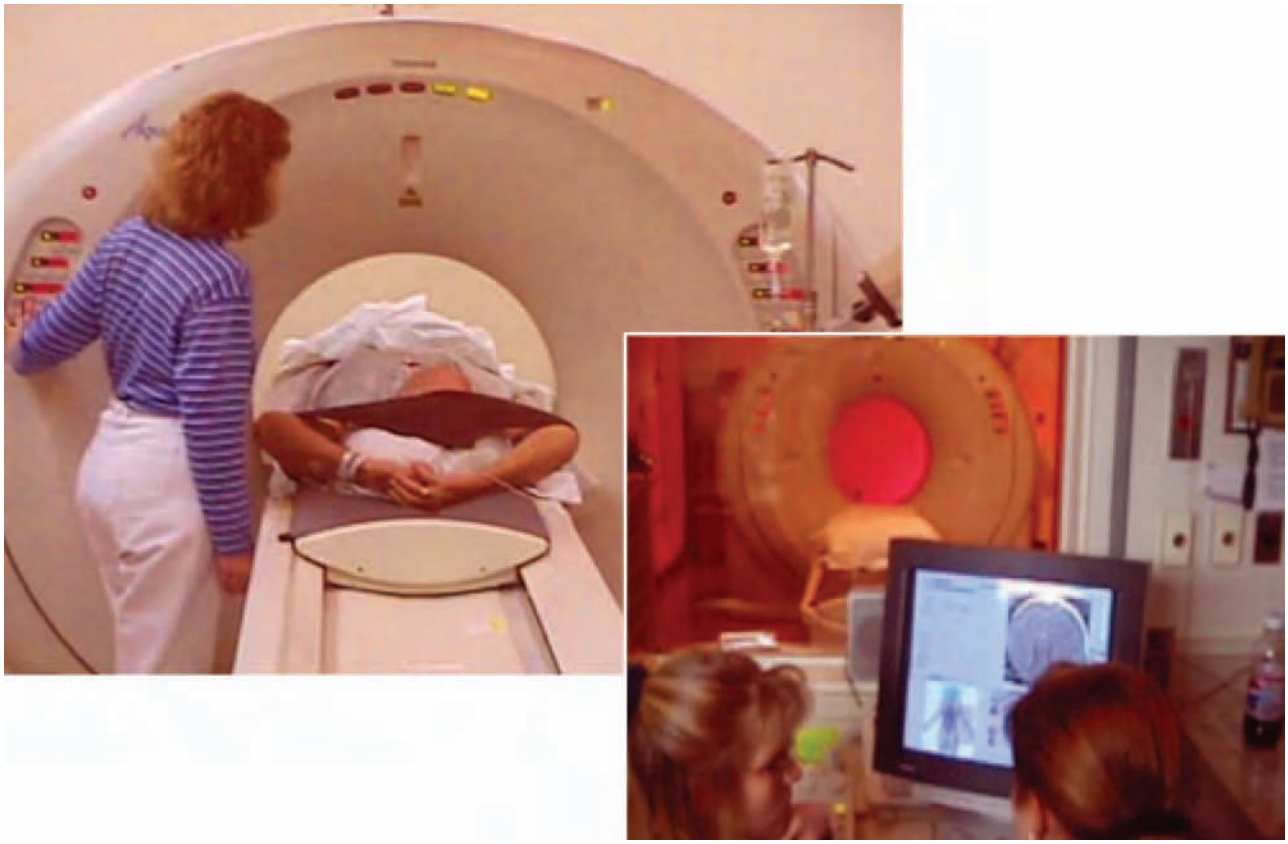


Fig. 4. A modern helical CT imaging suite.

The newer EBCT e-Speed™ version operates at 140 kV, 1000 mA, 140 kW. It can generate dual 1.5-, 3-, and 8-mm slices, at 50 ms temporal resolution. The spatial resolution is 10 lp/cm. This version also has a 100-ms mode for higher spatial resolution and a 33-ms mode for higher temporal resolution. Both are significant advances in EBCT technology.

Flow Mode

EBCT uses several sequences to image the heart. In the flow mode sequence, the C150 and C300 scanners generate eight 8-mm images in 224 ms. Then, following an 8-ms delay to reset the beam, an additional eight 8-mm slices can be generated in another 224 ms. The advantage of this configuration is that a stack of multilevel images can be obtained almost simultaneously, thus allowing visualization of a contrast bolus as it enters, peaks, and washes out of a region of interest (Fig. 5). This sequence is similar to the first-pass studies used in nuclear medicine. Flow mode sequences are useful in evaluating coronary artery bypass graft patency, intracardiac shunts, and arteriovenous malformations, as well as identifying bolus arrival times in aberrant vessels.

Movie Mode

A more commonly used EBCT sequence is the movie (cine) sequence. In this configuration, the C150 and C300 scanners can generate images of the contracting heart every 58 ms (17 images per second). These images can be acquired during systole and diastole, which allows visualization during a cardiac contraction (Fig. 6). This sequence, usually consisting of 10–12 same-level images, is triggered off the R wave of the ECG. Once completed, it then takes 8 ms for the scanner to reset, and another 10–12 image data set at the same or usually a different level can be obtained. The spatial resolution in the flow and movie modes is moderate at 4.5 lp/cm; however, with the new e-Speed model, resolutions of 10 lp/cm are possible. Movie sequences are important in quantifying cardiac function and in evaluating abnormal contraction patterns, such as those seen post myocardial infarction. Also, by completely opacifying the cardiac chambers, filling defects such as thrombi or tumors are readily identified.

Step Volume and Continuous Volume Scans

Lastly, in the step volume scan (SVS) and continuous volume scan (CVS) modes, 1.5-, 3-, or 6-mm single-slice thicknesses are obtainable. In the SVS sequence, the temporal resolution of the C150 and C300 platforms is 100 ms, which is 9 images per second, with the pixel sizes varying from 0.06 to 1 mm^2. With the e-Speed, 50-ms volume scan sequences at 17 or 34 images per second are possible, depending upon whether one or two detectors are used. The SVS images can be triggered at preset intervals from the ECG signal. The maximal resolution is 10 lp/cm, depending upon the field of view and the reconstructive algorithm selected. In the SVS mode, images are acquired in a manner similar to that of conventional CT scanners—i.e., a single target ring is swept to produce an image and then the table moves and a second image is generated. Multiple sweeps of the same target can be taken and volume averaged if

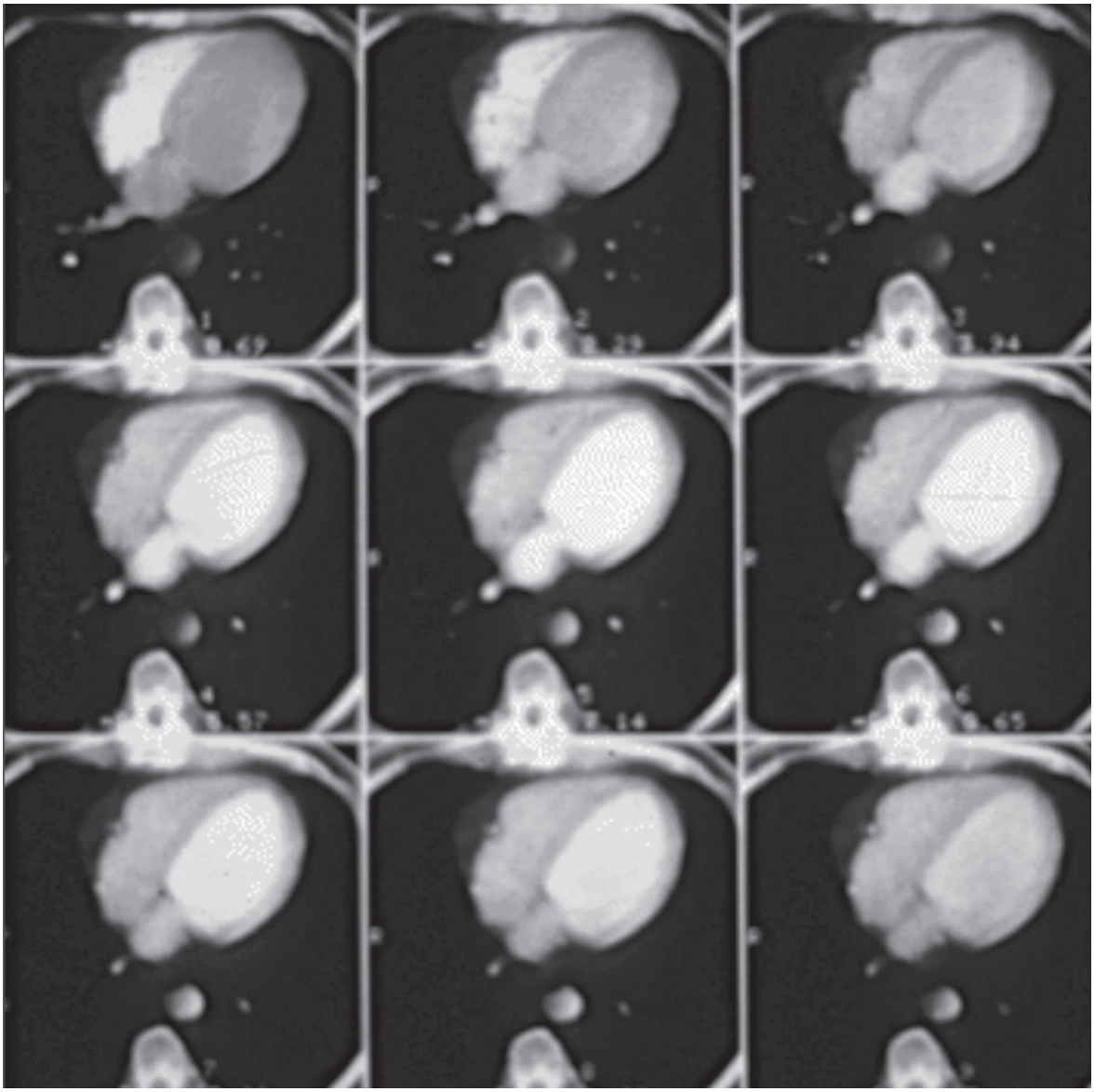


Fig. 5. An electron beam CT flow mode sequence showing a contrast bolus opacifying the right atrium and ventricle, and then washing out and later appearing in the left ventricle. This sequence is useful in evaluating bypass graft patency, intracardiac shunts, and arteriovenous malformations, as well as identifying contrast arrival in aberrant vessels.

additional resolution is required; however, this comes at the price of increased radiation exposure.

The continuous volume mode (CVS) is similar to that of spiral or helical CT methodologies, with up to 140 images on the C150, up to 280 images on the C300, and 400 images on the e$^-$Speed scanners possible during an acquisition period of about 33 s. The minimum exposure time is 100 ms (33 ms on e$^-$Speed) and the images can be contiguous or overlapped. On the C150 or C300 scanners, this mode allows a complete 140-slice, 3-mm thick data set to be acquired during a 17-second breath-hold. Scan widths of 1.5–10 mm are possible.

HELICAL CT

Current-generation helical/spiral CT scanners have multirow detectors capable of generating 1 to 16 slices of varying thickness with each gantry rotation. Technological advances have reduced gantry rotation times from 1 s to 420 ms, and with segmented reconstruction, image times approximating 100 ms are possible. An additional advantage of the helical technology is that the detector information is generated as a volumetric data set, and this permits later reformations of different slice thickness that allows 3D reconstructions. Helical CT images are commonly ECG gated, and this further decreases motion artifacts by being able to restrict image acquisition to the quiet phase of the cardiac cycle. The latter is particularly important in coronary calcification screening.

Helical/Spiral CT Sequence

In cardiac imaging, helical CT scanners are operated in two basic modes, both gated from the patient's ECG. The first mode, designated *prospective gated scanning*, activates the X-ray tube for only the time needed to acquire a partial image. This is

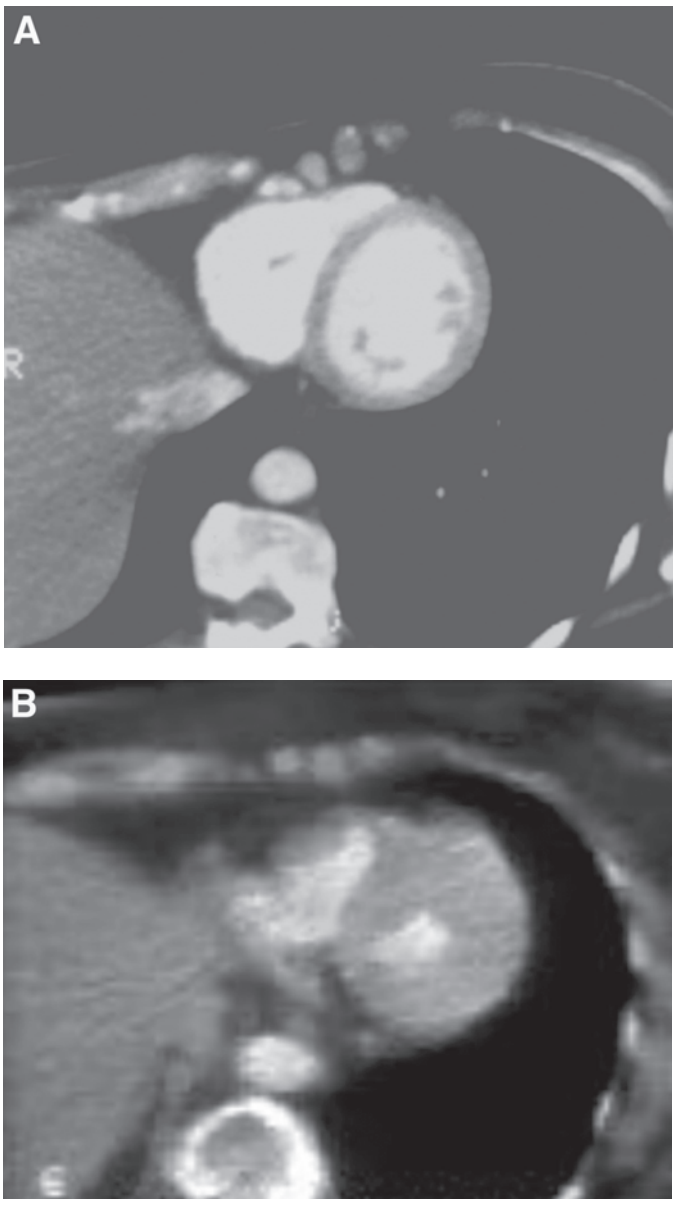


Fig. 6. Electron-beam short axis movie mode images at the mid-left-ventricular level in diastole (**A**) and systole (**B**). This sequence is useful in quantifying cardiac function and evaluating abnormalities of wall motion.

roughly half the gantry rotation time, or as short as 200–250 ms. The time of data collection is measured from the R wave, and the operator selects this delay. This mode is frequently used for coronary artery calcium imaging, because the radiation dose to the patient is kept to a minimum. However, rapid or irregular patient heart rates can affect image quality and the reproducibility of the study. Data collection periods that extend beyond diastole may exhibit motion artifacts that can reduce accuracy. A 4-detector ring scanner rotating at 0.5 s and programmed to provide 3-mm slices can acquire a data set in about 20 s, a reasonable breath-hold for most patients. Sixteen-row detector scanners in the same study reduced scanning times to roughly 8–12 s.

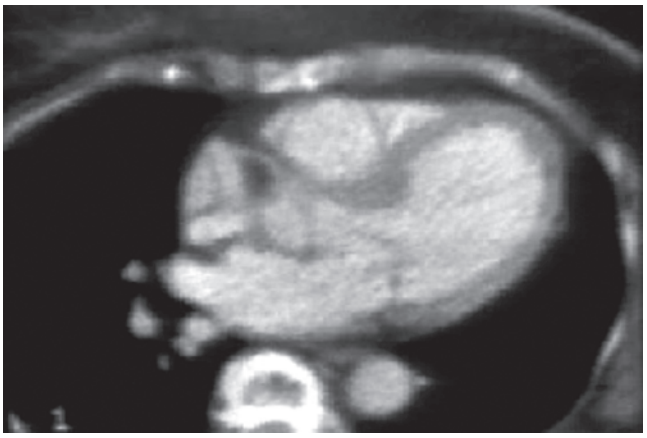


Fig. 7. Electron beam CT long axis (four-chamber) view image showing the left and right atria, left and right ventricles, and left-ventricular outflow tract.

Retrospective gated scanning is the other operating mode of helical scanners. This sequence involves the helical scanning of the entire heart while at the same time recording the patient's ECG. After the scan, images are reconstructed at a preselected phase of the cardiac cycle. This mode is commonly used for coronary artery angiography and for cardiac functional analysis. In order to avoid anatomical gaps in the data set, the helical pitch is set lower than that used for general body imaging. Thus the dose to the patient is higher. Nevertheless, the ability to reconstruct images in multiple phases from the same high-resolution data set can provide important information. Submillimeter slices, as small as 0.5 mm, can be used to achieve a high spatial resolution, and cardiac data collection can be accomplished in about 30 s. However, the patient's heart rate is again a major factor in maintaining high image quality. If the heart rate exceeds 70 beats per minute (bpm), segmented image reconstruction can be used. This algorithm uses data from two, three, or four consecutive cardiac cycles to reconstruct a single image, thus improving the temporal resolution significantly and extending the range of heart rates that can be scanned with high-quality images.

ADDITIONAL CONSIDERATIONS

IMPORTANCE OF TEMPORAL RESOLUTION

Although conventional, helical, and EBCT are able to identify cardiac anatomy, motion artifacts can still remain a problem. With contrast enhancement, conventional CT can image cardiac chambers and great vessel anatomy extremely well. However, helical/spiral CT and EBCT, because of their faster scan times, are suited better for imaging moving structures. In a typical cardiac sequence, the superior and inferior vena cavae, pulmonary arteries, and aorta are routinely visualized, as are the right and left atria and ventricles, and their outflow tracts (Fig. 7). With CT, the interfaces between the contrast-enhanced cardiac chambers and myocardium are usually well defined, especially if the image is acquired during diastole; in systole this interface may be partially degraded by motion. Overall, both helical/spiral CT and EBCT, because of their faster scan times, generally better define struc-

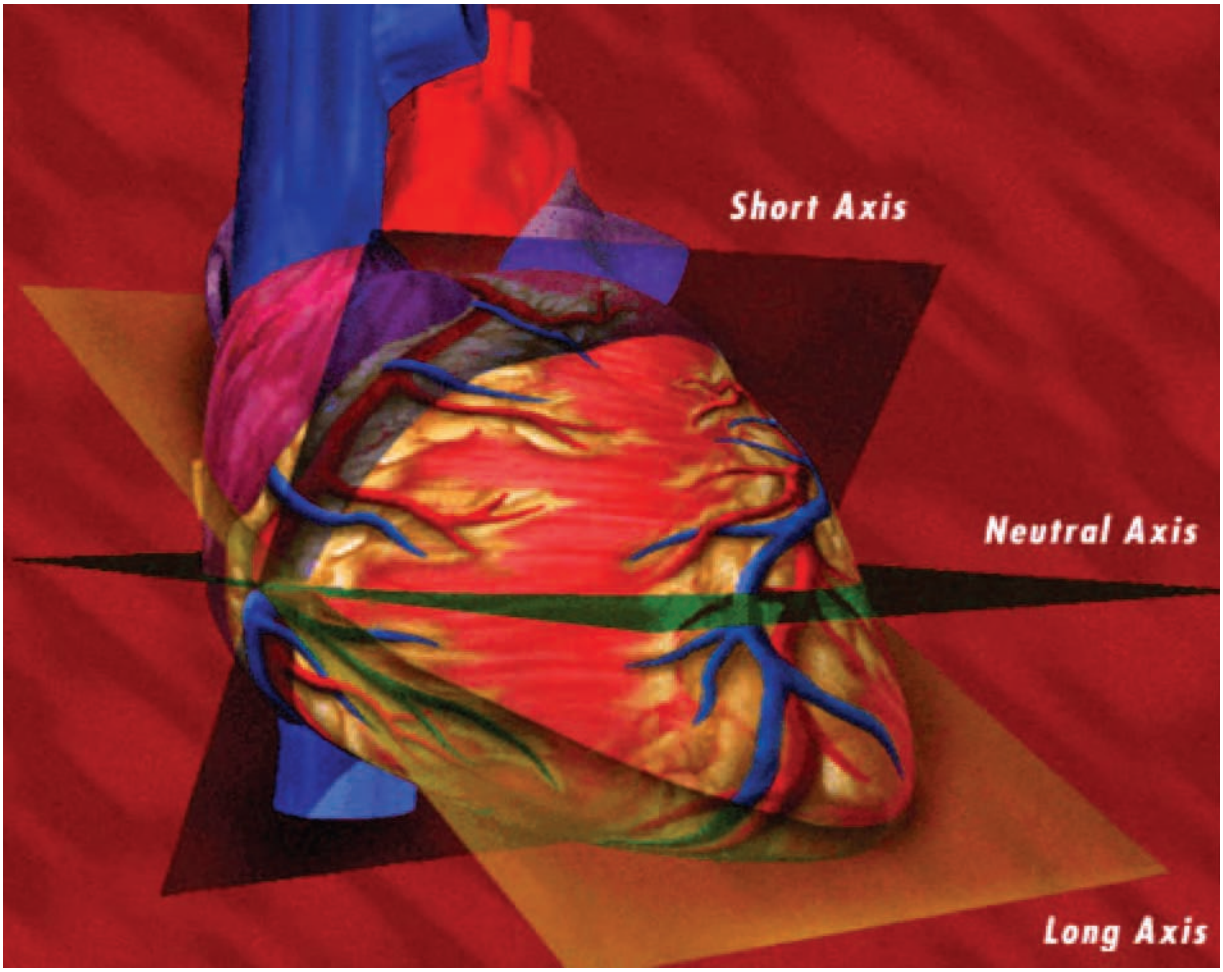


Fig. 8. Heart model showing the cross-sectional planes used in cardiac imaging.

tural detail *(5)*. Although it is important that the patient remain quiet and suspend respiration, often it is still possible with the EBCT and helical/spiral CT scanners to obtain satisfactory images, even if the individual is unable to hold his or her breath.

CONTRAST CONSIDERATIONS

In cardiac imaging, contrast administration is often necessary. The contrast material is usually administered as a bolus or as a continuous infusion, but because of the speed of the scanners, contrast arrival times become critical. Commonly, volumes of 80–150 mL of contrast material, varying from 240 to 370 mg of iodine per mL, are administered. Generally, the contrast material is infused at a rate of 1–4 mL/s using a power injector.

For contrast administration, timing methodologies become important to ensure contrast optimization. One method is to administer a test bolus of 10 mL of contrast material and perform repeated imaging over the area of interest to determine contrast bolus arrival time. Another is to perform a circulation time using either cardiogreen dye or a solution of 0.5% magnesium sulfate. Both are injected at approx 4 mL/s. With magnesium sulfate, approx 10 mL of 0.5% solution is commonly administered, with bolus arrival being manifested by a warm sensation in the back of the tongue or throat. With cardiogreen dye, an earlobe densitometer records the arrival of the bolus. Other techniques are the "sure start" technologies, where the scanner monitors the rise in contrast attenuation over the area of interest. When the attenuation approaches the preselected threshold, the scanner is triggered automatically. Alternatively, a fixed delay of 15–30 s may be used.

POSITIONAL CONSIDERATIONS

Cardiac CT scanning, especially in evaluating cardiac function, requires imaging in planes less familiar to radiologists. The oblique position of the heart requires modifications to our traditional scanning planes. The two configurations commonly used in cardiac imaging are the short axis view, wherein the ventricle is "bread-loafed" (Fig. 8), and the "four-chamber view," which slices the heart longitudinally from top to bottom and produces images that are similar to those familiar to our echocardiology colleagues (Fig. 7).

RADIATION DOSAGE

The effective dose is the overall radiation exposure to the patient. This is designated in mSv, and is frequently equated to months of background exposure. For an EBCT calcium study, the effective dose would be 1.0 mSv in males and 1.3 mSv in women (4.0 and 5.2 mo of background radiation, respectively) *(6)*.

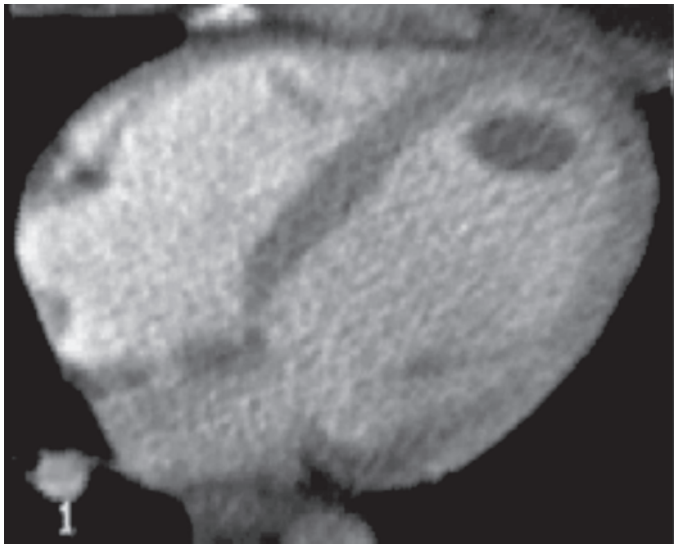


Fig. 9. Electron beam CT image showing an apical thrombus in the tip of an akinetic left ventricle in a postmyocardial-infarction patient.

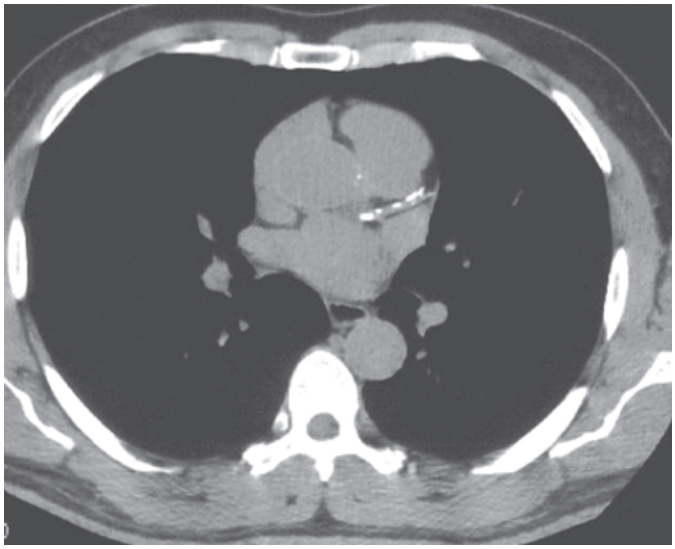


Fig. 10. Helical CT image of calcifications in the left anterior descending coronary artery.

For a multislice helical/spiral CT prospectively triggered coronary calcification study, the effective dose would be 1.5 mSv in males and 1.8 mSv in women (6 and 7.2 mo of background radiation). If a multislice helical/spiral CT coronary calcium study were retrospectively triggered, this would increase the radiation exposure to 3.0 mSv in males and 3.6 mSv in women, which would amount to 12 and 14.4 mo of background radiation *(6)*.

For a 50-ms EBCT cine sequence, the exposure would be 0.58 mSv per image; for a 10-image sequence, this would give an exposure of 5.8 mSv (23.2 mo of background exposure). One additional advantage of the electron beam technology is that the radiation beam enters from beneath the patient, which reduces exposure to breast and thyroid tissue.

If a CT angiographic study were done using retrospective gating, the radiation dose could reach as high as 10.9 mSv in men and 13.0 mSv in women (43.6 and 52 mo) *(6)*. However, tube current modulation techniques have now been introduced that can ramp down the power during noncritical times, and this can decrease radiation exposure by up to 80% *(5)*.

CARDIAC APPLICATIONS

Both EBCT and helical/spiral CT are useful in defining cardiac and great vessel anatomy. The excellent spatial and temporal resolutions allow identification of intracardiac filling defects such as thrombi and tumors *(7)* (Fig. 9), and the identification of defects of the pericardium *(8)*. Functional imaging can quantify stroke volumes and ejection fractions, as well as evaluate abnormalities of wall motion and perfusion, especially in post-infarct patients *(9,10,11)*. More recently, imaging of coronary calcification *(12,13,14)* (Fig. 10) and CT angiography for identifying coronary stenoses *(15,16,17)* (Fig. 11) and soft, noncalcified coronary atherosclerotic plaque *(18)* (Fig. 12) have become extremely important applications.

ADVANTAGES OF CT

The main advantage of CT imaging is that cross-sectional images without overlying structures are generated. The images have excellent temporal and spatial resolution; however, one must recognize that because some organs such as the heart do not lie in conventional imaging planes, true short and longitudinal axis anatomical images may not be possible without reformatting. To partially overcome this problem, some scanner couches and/or gantries can be tilted to position the heart into a more appropriate scanning plane. Generally, cardiac images are acquired in a short axis plane similar to "bread loafing," or in the longitudinal axis projection similar to the "four-chamber view" of echocardiography (Fig. 8). Because of the excellent resolution of CT, imaging of structures as small as 0.5 mm is possible. Additionally, tissues with different CT attenuations, such as fat, myocardium, and calcium, can be readily identified. An example is in imaging of the pericardium, where the low-attenuation epicardial and mediastinal fat lie adjacent to the higher-attenuation, fibrous pericardium. Commonly, the addition of iodinated contrast material allows further definition of the myocardium, and functional imaging of the heart is possible.

DISADVANTAGES OF CT

The disadvantages of even the faster CT technologies are the still somewhat long scan times which, unfortunately, do not completely eliminate cardiac motion. Other disadvantages inherent in all scanners are that they are stationary and the patient has to be transported to the scanner and be subjected to radiation. Also, contrast administration is often necessary, and weight restrictions may prevent some patients being scanned.

THE FUTURE

The ideal X-ray imaging system would be to combine the attributes of conventional radiography to produce images with spatial resolution in the order of 5 1p/mm and acquisition times of tens to hundredths ms that would be sufficient to freeze physiologic motion. In addition, the ideal system would respond linearly over a wide range, exclude scatter, and provide 3D

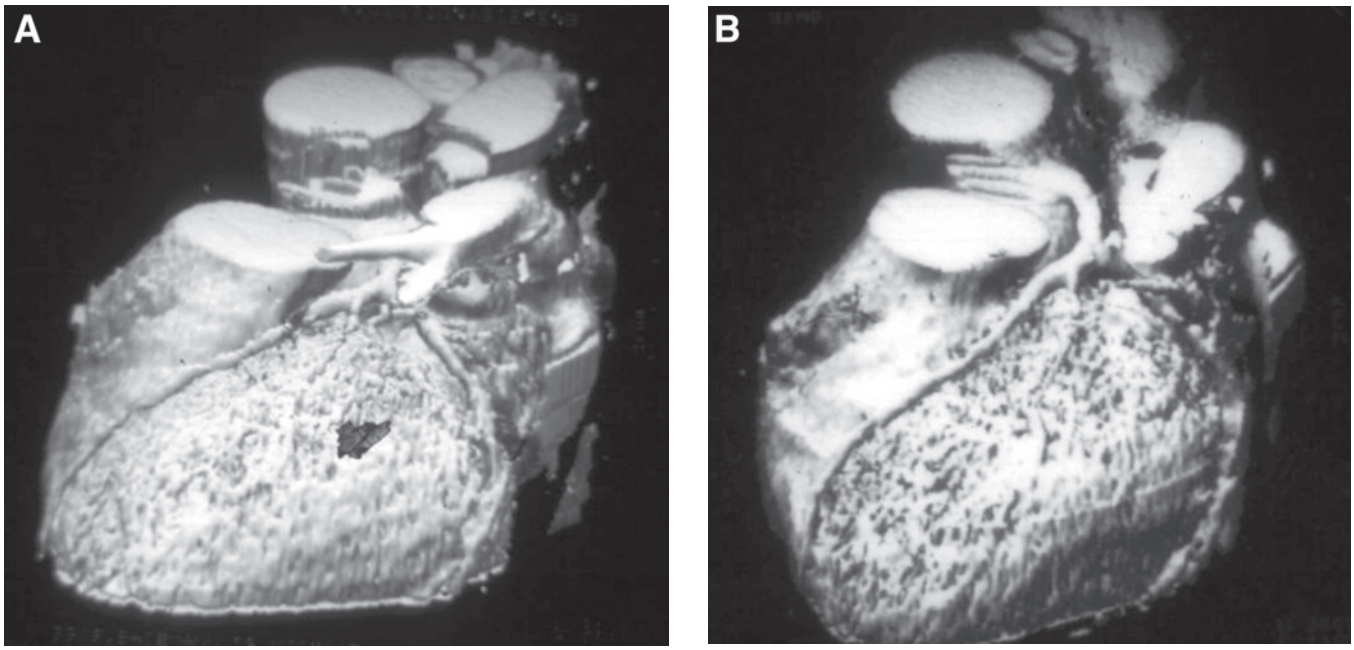


Fig. 11. Electron beam CT shaded surface display image showing a stenosis in the left anterior descending coronary artery pre- (**A**) and post- (**B**) balloon angioplasty. (Courtesy Moshage et al. *[15]*.)

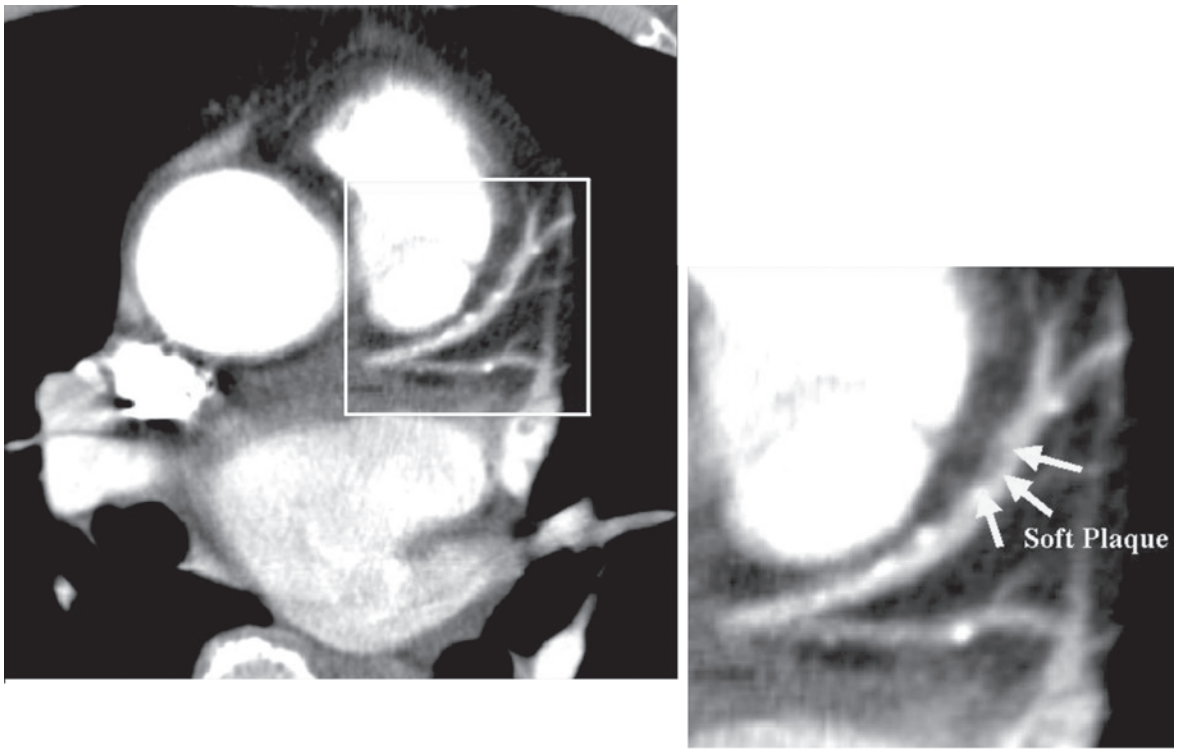


Fig. 12. Helical CT angiography image demonstrating areas of noncalcified plaque in the wall of an left anterior descending coronary artery.

information free of superimposing tissues with isotropic resolution throughout *(19)*. Also, it would provide the ability to image in transverse sections and thus avoid the problem of superimposed tissues. CT still has a long way to go, but continued improvements in temporal and spatial resolution are expected. Already, prototype platforms of 256 detector rows are being developed and tested. Isotropic voxel resolutions of $0.5 \times 0.5 \times 0.5$ mm are now possible, and from these 3D data sets it is possible to replicate true anatomy. Sector reconstruction can decrease scan times to below 100 ms, and EBCT scan-

ners that may reduce scan times to under 33 ms are under development. Techniques to decrease scatter and noise are being evaluated, and these should further improve image quality and decrease motion blur. These and other refinements have made possible functional and perfusion imaging.

Thus, the future of CT imaging of the heart is very exciting, and expectations are that CT will play an ever-increasing role in cardiac imaging.

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TECHNICAL BACKGROUND II

2 Electron Beam CT of the Heart

DAVID G. HILL, PhD

INTRODUCTION

Electron beam tomography (EBT)* was developed by Douglas Boyd, PhD, and his associates at Imatron, starting in 1977; the first clinical installation was at University of California at San Francisco in 1984. The challenge was to design a CT scanner capable of imaging the heart without motion artifacts; this required data acquisition times of 100 ms or less. In order to study perfusion, covering most of the heart without moving the patient table was also necessary. Acquisition times of 100 ms or less required a design with no moving parts except for the patient table. The result is EBT, where an electron beam generates the X-ray views needed for CT by sweeping across fixed targets.

Since 1983, through several models, EBT has been continuously improved in spatial resolution, temporal resolution, and software capability, and has been distributed worldwide. In addition to the original anatomy, wall motion, and perfusion applications, applications such as measurement of coronary calcium, CT coronary angiography, lung studies, and standard CT radiological examinations have been developed and introduced.

Unless otherwise stated, this article quotes specifications for the e$^-$Speed™ model.

TECHNOLOGY

EBT eliminates moving parts in the X-ray generation system by employing an electron beam that sweeps around a fixed tungsten-coated target to generate a fan of X-rays. The detector is mounted in a fixed position above the target ring, providing a fourth generation (fixed detector, moving beam) CT design. Sufficient data to reconstruct an image can then be acquired in one sweep of the beam around the target, which in the current model of the scanner (e$^-$Speed) can be as short as 33 ms. The speed and position of the beam and data acquisition system are under computer control, allowing flexibility in the choice of temporal resolution, spatial resolution, and X-ray signal.

The schematic shown in Fig. 1 is representative of an EBT system. The electron source accelerates electrons at potentials of up to 140 kV, producing a 1000-mA beam. Under high vacuum, the beam expands down a beam pipe until it reaches a group of magnets used to control and shape the beam. Two dipoles deflect the beam onto one of four fixed targets and sweep it along that target. A solenoid and two quadrupoles focus the beam and ensure a proper elliptical shape aligned so that the minor diameter of the ellipse is in the scan direction. The tungsten targets form a 210° arc and are backed by a water-cooling system, allowing at least 20 s of continuous scanning at full power without reaching thermal limits. X-rays are generated as the electrons strike the tungsten and are collimated into a thin fan that irradiates the patient. The entire beam system, from electron source to target ring, is under vacuum and can be considered a very large, water-cooled, stationary X-ray tube.

Reconstruction of a cross-sectional image from CT data requires that at least one ray in every direction pass through each pixel; a ray connects a source point and a detector (180° plus the X-ray fan angle). Current EBT geometry meets this requirement with a 216° detector and a 210° target, providing a 47.5-cm central region for imaging.

X-rays from any of the four targets can reach the dual slice detector. If the targets are used in series, 76 mm of a patient can be seen without moving the patient. Brass collimation rings minimize the number of X-rays that could pass through the patient but not strike a detector. Collimation and detector size give dual 7-mm slices with any of the targets. For narrower slices, one target is used, and an additional collimator is employed to limit the X-ray beam to dual 1.5-, single 3-, or dual 3-mm slices as measured at the isocenter. This additional collimation is also prepatient, to give maximum dose utilization. During image reconstruction, the data for dual 1.5-mm slices can be combined to make a single 3-mm slice, and the data for dual 3-mm slices can be combined to make a single 6-mm slice.

As in any CT scanner, X-rays are detected and digitized by solid-state detectors. Special EBT challenges are the high data rates and short integration times necessary to acquire sufficient data for an image in 33 ms or less.

Reconstruction of these images requires that the data be corrected for beam hardening, beam and detector position, scattered X-rays, and incoming flux. The result is a cross-sectional CT image with very high temporal resolution and high spatial resolution.

A single sweep is the passage of the electron beam along a target from beginning to end (210°), producing sufficient data to reconstruct an image. Most sweeps used for imaging move at constant angular velocity. A reset lasting 4–6 ms, during which no X-rays are generated, brings the beam back to the

*Also referred to as EBCT in the published literature.

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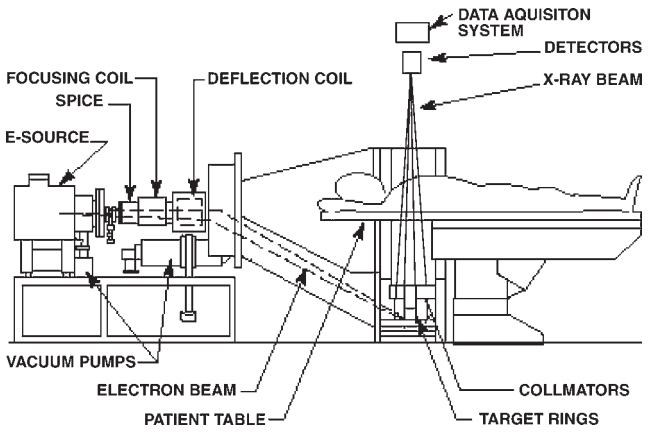

Fig. 1. Electron beam CT schematic.

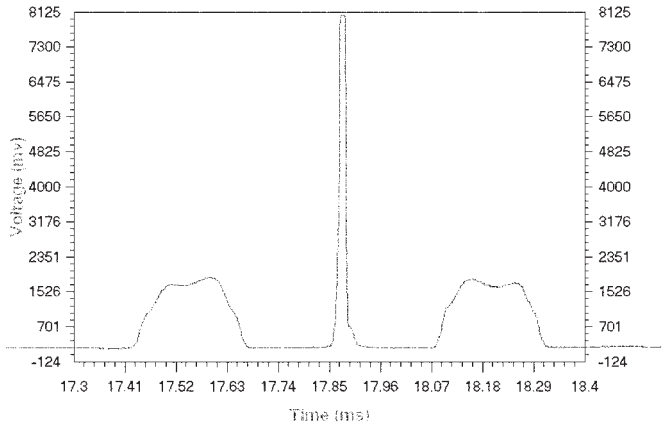

Fig. 2. Typical W wire signal.

starting position for the next sweep. If the next sweep does not immediately follow the preceding sweep, the beam is either turned off or deflected to a beam stop downstream of the deflection coils, where it continues to move until the next sweep is commanded. Because of its high power, the electron beam can never be allowed to stop on the target without damaging the target. Protections against such damage are built into the hardware and software controlling the scanner.

The geometry of the electron source requires the perveance, defined as $V^{1.5}/I$, to be a constant; there is only one mA value for a given kV setting. Thus, an e$^-$Speed with nominally 1000 mA at 140 kV has 895 mA at 130 kV.

EBT calibration and data correction involve scanning of air to get incoming flux, measurement of water to set the CT number of water to 0, and correction of beam hardening and scattering to ensure uniformity of CT numbers in imaging of an organ or object with constant density.

An additional EBT calibration is tuning, which is used to determine with high precision both the exact location of the beam with time, and the size, shape, and orientation of the beam spot. The beam condition at any given time is controlled by a sequence of coil currents vs time given to the magnets that steer and shape the beam.

An EBT scanner has a fifth target, located beyond those used for patient studies; a series of three-wire bundles in a W shape are located in front of this target. As the beam sweeps over this target and crosses one of the W wires, the location of the peak on the central wire gives the time of the beam with respect to data acquisition; the width of the peak on the central wire gives the beam spot width; the shape and similarity of the pulses from the angled side wires give the orientation of the elliptical beam spot and its length; the distance between the side peaks is related to the beam radius from isocenter. An example of a W wire signal is shown in Fig. 2.

The first part of tuning consists of adjusting the coil currents in time to give the desired parameters. Using equations that characterize the coil currents as a function of deflection angle, the tuned values are transferred to the sweep descriptions that will be used for scanning on the X-ray targets. The second part of tuning is a scan, with the transformed coil current values, of a phantom containing a set of pins. Analysis of the raw data from the pin scans gives the reconstruction system the location of the beam corresponding to the data as acquired, enabling interpolation of the raw data to the positions required by the reconstruction algorithm.

Control of the scanner requires numerous computers networked together, dealing with control of the beam, collimation, table, power supply, data acquisition, reconstruction, and operator interaction.

SCAN MODES

Standard sweep speeds with the e$^-$Speed are 33 ms, 50 ms, and 100 ms. Scan modes consist of different combinations of a series of sweeps on one or more targets, with patient-table motion or with no patient-table motion, and/or with an electrocardiogram (ECG) or timed trigger. If the patient table is not programmed to move while a sweep is being executed, the result is an axial scan mode. Axial scans, triggered from an ECG signal, are typical for cardiac studies. If the patient table moves while the X-rays are on, the result is a continuous volume scan, analogous to a spiral or helical scan with a mechanical CT. Continuous volume modes are typically used for CT angiographic studies of the great vessels, for example.

An external ECG monitor generates triggers at the R wave. Sweeps begin at a user-specified time after the R wave, which can be set either as a time interval in seconds or as a percentage of the RR interval. If triggering is based on percentages, the time is estimated from the previous seven heart RR intervals and thus adapts to changes in the heart rate. Extensive clinical studies *(1)* show, for an EBT scanner with its high temporal resolution, that the point in the RR interval with minimum motion is about 300 ms after the R wave, near the end of the T wave. Since the length of systole (starting at the R wave) tends to be more constant with heart-rate change than the length of diastole, triggering at the end of the T wave (end systole) seems to have fewer problems with irregular heart rates than does triggering by percentage of the RR interval.

EBT scanners use prospective triggering. Before the scan begins, a point (or points) in the heart cycle is chosen for the sweep to occur, and only those data are acquired and recon-

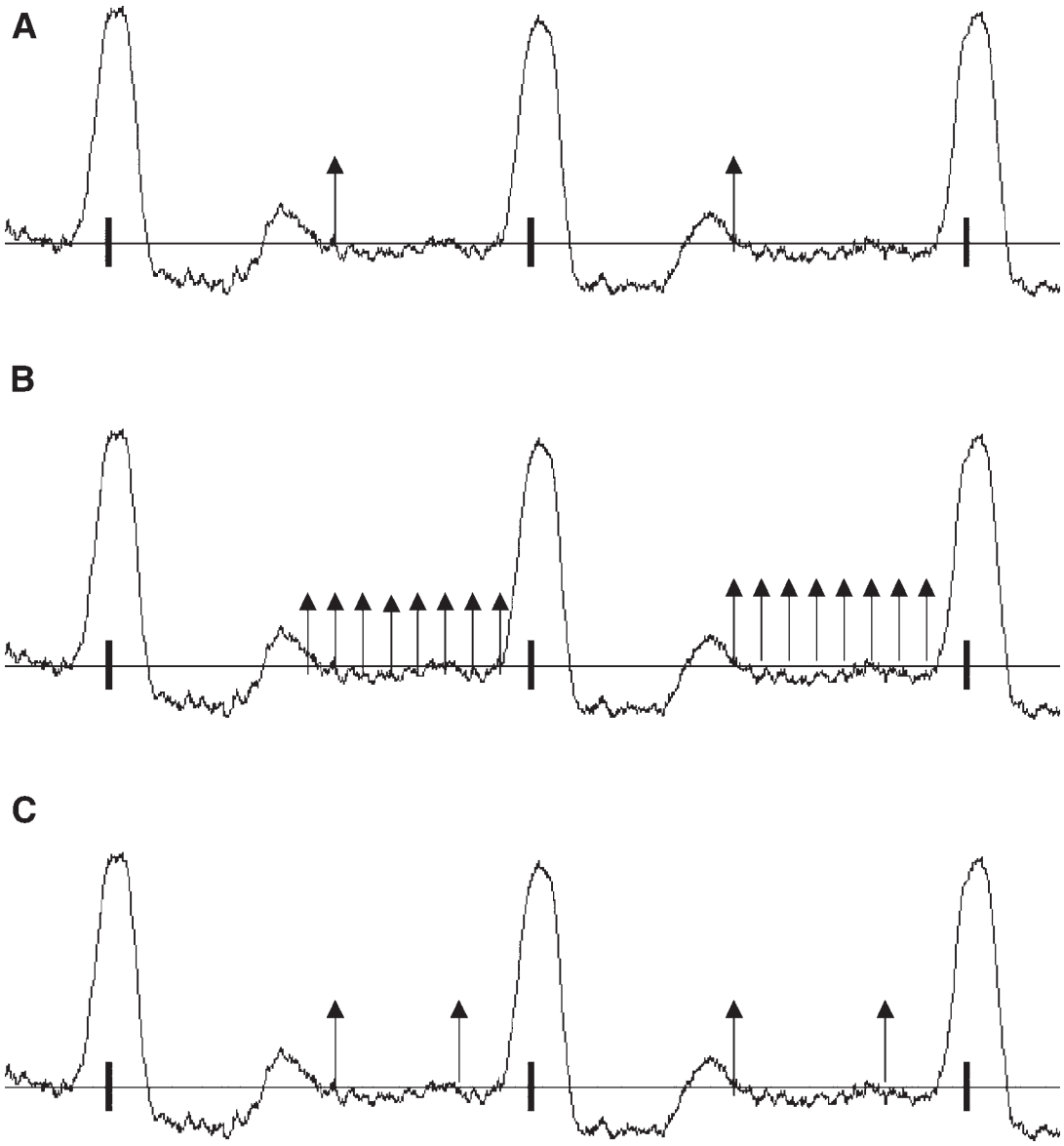

Fig. 3. (A) ECG trace—single trigger at end systole (represented by ↑). **(B)** ECG trace—triggering eight phases of the heart. (**C**) ECG trace—triggering at two positions within the heart cycle.

structed. Therefore, all exposure of the patient contributes to the study images; high temporal resolution and accurate triggering algorithms make this both feasible and efficient.

The simplest ECG-triggered scan mode has one sweep after the trigger at a prespecified time after the R wave; patient-table motion is after the sweep, so the patient table is at rest in the new position in time for the next scan. This is the scan mode used for a coronary calcium study or for an electron beam angiography (EBA) study (contrast in the arteries) with dual 1.5-mm, single 3-mm, or dual 3-mm collimation. Table increment is usually equal to the collimation, although smaller increments giving overlapping images are used in some cases.

Instead of just one sweep after the trigger, more than one sweep can be used, with each sweep immediately following its predecessor. The patient table moves to its next position after the set number of sweeps has been performed. This allows studies of the heart at multiple phases in the same study time as a single-phase study, and allows the clinician to appreciate changes in the heart over the heart cycle. Multiple phases is a cine mode including patient-table motion, and is used with all collimations. The high temporal resolution gives minimal motion artifact at any phase.

A further generalization of this mode allows the user to position sweeps at any point throughout the heart cycle. For example, this might be used to scan at approximately end systole and end diastole to obtain a measurement of ejection fraction.

Figure 3 shows a representative ECG trace with several possible triggering modes indicated.

For any ECG-triggered mode with patient-table motion, if the time required for the number of sweeps requested plus the patient-table motion is less than the RR interval, the table is positioned for the next location when the time to start the next

Table 1
Selected Specifications

Specification	*C150*	*C300 / C150 High Resolution*	*e⁻Speed*
Sweep speeds-thin slices	100 ms	100 ms	33, 50, and 100 ms
Sweep speeds-multitarget	50 ms	50 ms	33, 50, and 100 ms
Thin slice collimation	1.5, 3, and 6 mm	1.5, 3, and 6 mm	dual 1.5 mm, single 3 mm, dual 3 mm
Multitarget collimation	dual 7 mm	dual 7 mm	dual 7 mm
In-plane spatial resolution	7 line pairs (lp)/cm (100 ms)	9.5 lp/cm (100 ms)	10 lp/cm (50 ms) 13 + lp/cm (100 ms) 7 lp/cm (33 ms)
Power	83 kW (130 kV, 625 mA)	83 kW (130 kV, 625 mA)	140 kW (140 kV, 1000 mA) 116 kW (130 kV, 895 mA)
Trigger times (%RR)	40–80 (all) 0 (50 ms only)	40–80 (all) 0 (50 ms only)	0–99 (all)
Trigger times (seconds after R)	0.273–0.999	0.273–0.999	0.060–1.500

set of sweeps arrives; if the patient table is not in place for the next heart cycle, one heart cycle is skipped.

A multitarget cine study, typically covering the entire heart cycle, can also be performed. Sweeps every 33, 50, or 100 ms from the first target cover the time between first and second R waves; the second target is used to cover the third to fourth R-wave interval, the third target with the fifth to sixth R-wave interval, and the last target with the seventh to eighth R-wave interval. If more coverage is needed, the patient table is then moved 40 mm or more and additional scans taken.

A flow study is used to follow a bolus of contrast. One or more sweeps is taken each heart cycle without patient-table motion. If all four targets are used, then one sweep is taken on each of the targets during one heart cycle, allowing the measurement of perfusion over 76 mm of the body using the same contrast bolus. Some heart cycles are usually skipped to sample the bolus curve but minimize dose. This is the standard mode for the measurement of myocardial perfusion.

All of these modes can be triggered manually or by time instead of by ECG; however, cardiac studies are almost always ECG-triggered.

If the table moves while the beam sweeps continually, the result is a continuous volume scan, similar to a helical or spiral scan with a mechanical CT. Data can be acquired with any of the available collimations. Images are reconstructed at any location along the scan; the slice width in the image is determined from the collimation and the table travel corresponding to the sweeps included in the reconstruction. As an EBT scanner has a fixed mA for a given kV, in order to change the exposure in an image without changing the slice width, the speed of the patient table must also change. For example, in a study using 3-mm collimation and reconstructing a 3-mm slice, twice the exposure in the 3-mm slice requires the patient table to move half as fast.

A continuous volume scan is the normal scan mode for studying the noncardiac vessels, the lungs, abdomen, or any other organ not requiring ECG triggering. Of course, the axial modes described previously with a timed or manual trigger may also be used; however, the study time is longer than with continuous volume scanning but without patient-table motion during data acquisition.

A PreView™ (General Electric, formerly Imatron) (scout or localization scan) is created using a special sweep while the patient table moves continuously. The beam is swept very rapidly along the tuning target; near three o'clock and again near six o'clock, the beam moves onto the X-ray target for a short arc at the slower 50-ms speed. X-rays reaching the patient are generated only when the beam is on the X-ray target. A transmission image is synthesized from the source points along these arcs, giving views in both anterior/posterior and lateral directions. From these views the user not only can set the location in the patient direction to start and stop acquiring data, but also can set the center point and field of view for the reconstructed images in the studies to follow. Because PreView images are made from a number of source positions through a process of tomosynthesis, the plane of focus can be changed from the default value of the isocenter, or multiple planes can be computed and viewed. The depth of field is sufficiently large that isocenter is normally chosen.

IMAGE QUALITY

EBT is designed for high temporal resolution: 33, 50, or 100 ms in a single image without requiring data from more than one heart cycle. This temporal resolution is short enough that images can be acquired at any point in the heart cycle with no or almost no motion artifact.

Maximum in-plane resolution ranges from 7 line pairs (lp)/cm at 33 ms to 13 lp/cm at 100 ms. An extensive set of reconstruction kernels allow the optimization of resolution to the clinical study being performed. Narrowest slices are 1.5 mm. Because the arteries are typically moving from 20 to 100 mm/sec *(2)*, resolving the arteries clearly requires a high temporal resolution imaging system.

SPECIFICATIONS

As of this writing there are three EBT models in clinical use: C150, C300, and e⁻Speed. Some critical parameters such as sweep speed, slice width, and resolution are summarized in Table 1.

DOSE

As an EBT scanner irradiates a patient over a 210Yarc (essentially from two o'clock to ten o'clock), dose near the surface of the patient is not uniform. For a patient lying supine, dose is a factor of five lower anterior than it is posterior. The result is that dose to the breast is minimized during a cardiac examination.

Prospective triggering also means that a patient receives only the dose necessary for the images desired for the study. The user can choose to acquire data over the entire heart cycle or over only a part of it, and the scanner acquires just those data.

Table 2 gives the CT dose index (CTDI) *(3)* information for a scan that would be typical of a calcium or coronary angiographic single-phase study for an EBT scanner, as well as an estimate of the effective dose (E) if the entire heart is covered. The 32-cm body CTDI is quoted with a 3-mm slice (or dual 1.5-mm with e-Speed) and a single sweep. The CTDI as defined is an estimate of the single-slice dose taking into account the tails of the dose profiles from the adjacent slices. E is estimated *(4)* by multiplying the CTDIvol (weighted CTDI adjusted for table speed) times the distance covered (which is the dose-length product) × 0.017.

Table 2
Dose Information

Parameter	*C150/C300*	*e-Speed*
CT dose index (CTDI) at A (center)	1.8 mGy	1.9 mGy
CTDI at B (12 o'clock)	1.2 mGy	1.2 mGy
CTDI at C (6 o'clock)	7.9 mGy	6.3 mGy
CTDI at D,E (3 or 9 o-clock)	5.6 mGy	5.1 mGy
CTDIw (weighted average)	3.7 mGy	3.6 mGy
Typical cardiac studies covering entire heart with one sweep each heart cycle:		
kV	130	140
mA	625	1000
Sweep time	100 ms	50 ms
Collimation	3 mm	3 mm
Table increment/heart cycle	3 mm	3 mm
Effective dose	0.7 mSv	0.7 mSv
Sweep time	100 ms	50 ms
Collimation	1.5 mm	dual 1.5 mm
Table increment/heart cycle	1.5 mm	3 mm
Effective dose	1.4 mSv	0.7 mSv

TYPICAL APPLICATIONS

Typical cardiac studies with EBT include coronary calcium *(5)*, EBA of the coronary arteries *(6,7)*, cardiac anatomy *(8)*, wall motion studies *(9)*, and perfusion studies *(10)*. EBT scanners are also excellent vehicles for CT angiography of the noncardiac vessels, as well as for lung studies *(11)* . The references given are to published studies as examples; no attempt has been made to make this a complete bibliography. Figure 4 is an example of an axial image showing coronary calcium; Fig. 5 shows a 3D volume rendering of an EBA study, and Fig. 6 shows a maximum intensity projection image of the same data set.

Studies of coronary calcium use an ECG-triggered axial mode with one phase/heart cycle. Most recently the preferred trigger point is near end systole, a time of minimum motion of the right coronary *(1)* with a 3-mm slice, 130 kV, and no contrast. Historically, most coronary calcium scanning has been done with triggering either near 80% or more recently near 40% of the RR interval. All EBT scanner models have a calcium scanning mode that matches slice width, reconstruction parameters, and kV to the early data on C100/C150, so that the Agatston *(12)* score for calcium is independent of EBT model; thus, historical data can be used for clinical comparison. Other scoring methods such as volume score *(13)* and mass score *(14)* have been proposed and implemented that may be more independent of the scan mode.

Coronary angiographic studies (electron beam angiography, or EBA) use ECG-triggered axial modes with one or more phase/heart cycle and intravenous contrast injection. If a single phase is used, then the scanner is triggered at the point of minimum motion; if multiple phases are being used, then the point of minimum motion is usually included as one of the phases.

In order to optimize the contrast, circulation time is determined before an EBA study, using a short bolus of contrast and a flow study looking at a given level. Typically a scan is triggered at the start of contrast injection (for a baseline value without contrast); some 6–10 heart cycles later a sequence of one sweep every other heart cycle begins through the expected peak of the contrast (18 s or so) followed by a sweep every 3 heart cycles or so to follow the contrast wash out. The circulation time (the time between injection and arrival at the coronary arteries) is then determined from the change in the CT number in the aorta or left ventricle with time. EBA data acquisition starts after a delay from the start of injection that is approximately equal to the circulation time. If the circulation time study uses all four targets and the dual 7-mm collimation, it becomes a perfusion study.

The length of the EBA injection should be approximately equal to the time required to complete the series, estimated from the distance to cover (which when divided by the table increment gives the number of heart cycles needed) and the starting heart rate. For most patients, the heart rate will increase during the study. Optimization of the contrast is very important; in addition to correct circulation time and injection duration, it may involve a saline chaser and variation of the contrast flow rate during the injection. The best studies come with the contrast signal uniform in time and the ratio of the average CT number of the contrasted vessels to the X-ray noise in the images as large as possible.

Most EBA studies are analyzed on a workstation making use of reformats, maximum intensity projections, and volume renderings to appreciate the details of the arteries. Multiple-phase EBAs make arterial motion obvious, although any individual phase can be analyzed. Clinical evaluation of the data frequently includes observing 2D and 3D renderings "beat" as the display cycles through the phases. With an appropriate choice of scan parameters, the study may cover a sufficient portion of the heart cycle to allow the estimation of ejection fraction. Valve motion can also be visualized.

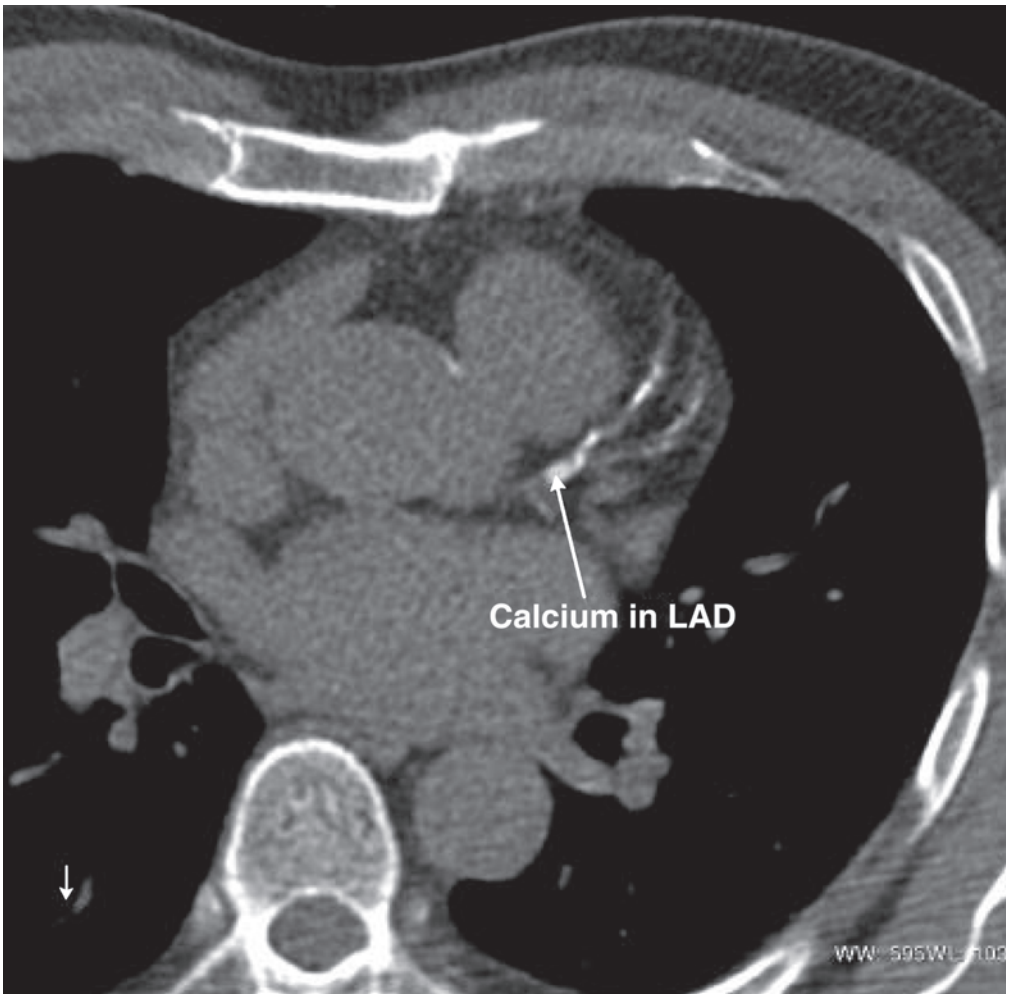

Fig. 4. Calcium example; 3-mm slice; 130 kV; 895 mA; 50 ms.

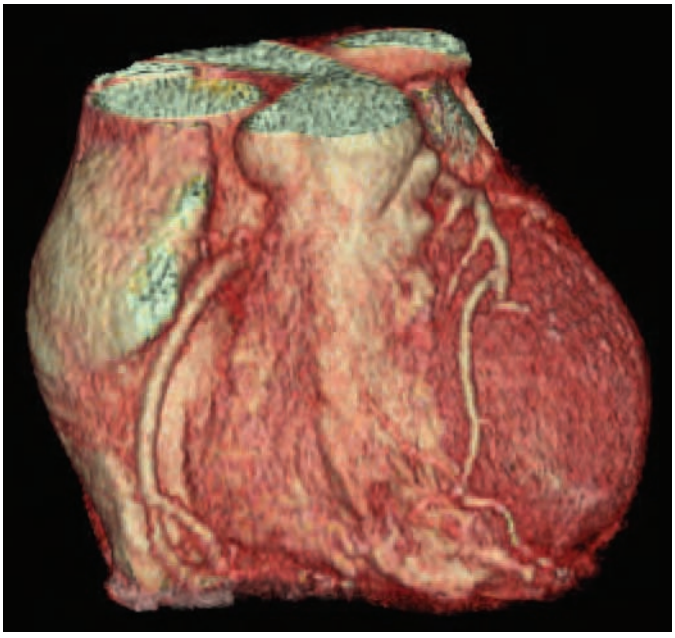

Fig. 5. Three-dimensional rendering of an electron beam angiography study; dual 1.5-mm slice; 140 kV; 1000 mA; 50 ms.

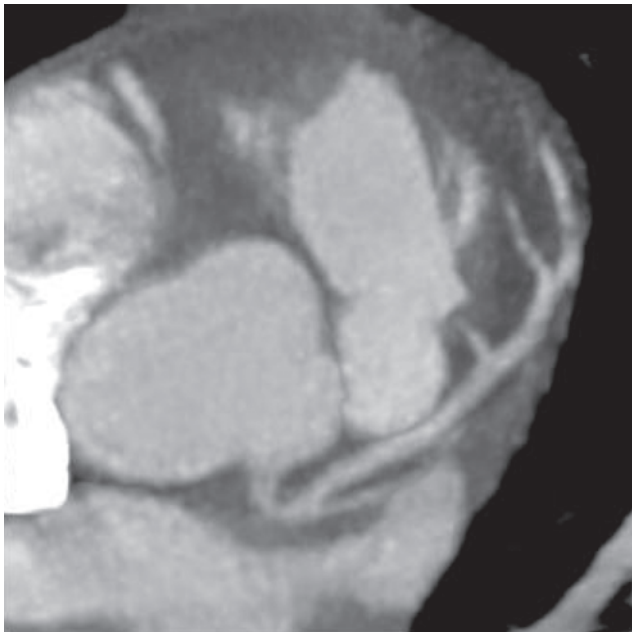

Fig. 6. Maximum intensity projection image of left anterior descending artery; same data set as Fig. 5.

Sufficient cardiac functional information is often gathered from a multiphase study. In addition, there are two ways to perform a dedicated wall-motion or function study. One is essentially an EBA using dual 7-mm collimation covering the entire heart cycle, with a 15-mm patient-table motion every other heart cycle. An alternative, avoiding patient-table motion, is to move the beam sequentially through the different targets instead of moving the patient table.

A perfusion study is a circulation time study with 76-mm coverage, using all four targets and the dual 7-mm collimation. This mode allows the measurement of perfusion throughout the myocardium using one bolus of contrast. There are three gaps of approx 5 mm each (between the targets) that are not fully illuminated.

A complete EBCT cardiac exam consists of a PreView for localization, a calcium study, a flow study used for both perfusion and circulation time determination, a multiphase EBA, and possibly a wall-motion study using residual contrast from the EBA.

The EBCT patient table can tilt 25° and slew ±25° (side to side in the horizontal plane). Tilt and slew allow the acquisition of wall-motion or perfusion data in approximately the short axis position (tilt 15°, slew –15° to –25°) or the long axis position (tilt 0°, slew +25°), approximating traditional cardiac views.

Continuous volume is the mode of choice to analyze the noncardiac vessels, yielding clear motion-free images. For example, using on the e$^-$Speed dual 1.5-mm collimation , 50-ms sweeps, and a 3-mm table increment, the patient table moves at 54 mm/s, allowing most areas of interest to be covered in 10 s or less.

FUTURE DIRECTIONS

EBCT scanners are designed for excellent cardiac imaging. There is no inherent limit on how fast an EBCT scanner can sweep the beam, as there are no moving parts, so sweep times can become even shorter, improving temporal resolution. Potential temporal resolution is limited only by available X-ray power (signal-to-noise in the images) and data acquisition speeds. Future directions are likely to include using a detector with smaller elements for thinner slices, as well as increasing the number of detectors in the patient direction for increased coverage with each sweep. As cardiac imaging continues to grow in clinical importance, EBT is an ideal modality for improved applications and efficient usage.

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3 Scan Techniques for Cardiac and Coronary Artery Imaging With Multislice CT

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INTRODUCTION

Cardiac imaging is a demanding application for any noninvasive imaging modality. On the one hand, high temporal resolution is needed to virtually freeze cardiac motion and thus avoid motion artifacts in the images. On the other hand, sufficient spatial resolution—at best submillimeter—is required to adequately visualize small and complex anatomical structures like the coronary arteries. The complete heart volume has to be examined within a single short breath-hold time to avoid breathing artifacts and to limit the amount of contrast agent, if necessary. The motion of the heart is both complex and very fast. Some estimates of the temporal resolution needed to freeze cardiac motion in any phase of the cardiac cycle are as low as 10 ms. In 1984, electron beam computed tomography (EBCT) was introduced as a noninvasive imaging modality for the diagnosis of coronary artery disease *(1–4)*. Its temporal resolution of 100 ms allows for relatively motion-free imaging of the cardiac anatomy in the diastolic phase, even at higher heart rates. Because the EBCT at that time was limited to axial scanning for electrocardiogram (ECG)-synchronized cardiac investigations, a single breath-hold scan of the heart required slice widths of at least 3 mm. The resulting axial resolution was therefore limited and not adequate for 3D visualization of the coronary arteries. With the advent of subsecond rotation, combined with prospective and retrospective ECG-gating, mechanical single-slice helical or spiral CT systems with superior general image quality entered the realm of cardiac imaging *(4,5)*. Since 1999, 4-slice CT systems, which have the potential to overcome some of the limitations of single-slice cardiac CT scanning, have been used to establish ECG-triggered or ECG-gated multislice CT (MSCT) examinations of the heart and the coronary arteries in clinical use *(6–10)*. As a result of the increased scan speed with four simultaneously acquired slices, coverage of the entire heart volume with thin slices within one breath-hold became feasible. The improved axial resolution provided much more accurate CT imaging of the heart and the coronary arteries *(11–14)*. Recent clinical studies have demonstrated the potential of MSCT to differentiate and classify lipid, fibrous, and calcified coronary plaques *(15)*. Despite these promising advances, the 4-slice CT scanner technology still faces some challenges and limitations with respect to motion artifacts in patients with higher heart rates, limited spatial resolution, and long breath-hold times *(12)*. In 2001, a new generation of MSCT systems with simultaneous acquisition of up to 16 slices was introduced *(16,17)*. With submillimeter slice acquisition and gantry rotation times shorter than 0.5 s, both spatial and temporal resolution are improved, while examination times are considerably reduced.

In this chapter, we present the basic technology of MSCT cardiac scanning with a special focus on recommended scan techniques for different clinical applications. We will also discuss the technology advances and improved clinical performance of state-of-the-art 16-slice CT equipment compared with 4-slice CT scanners.

TECHNOLOGY PRINCIPLES

TECHNOLOGY OVERVIEW

CT examinations of the heart should be performed in a single, short breath-hold scan with high temporal resolution to eliminate cardiac motion and high, preferably isotropic, spatial resolution. It has become increasingly apparent that submillimeter slices are necessary to adequately visualize small and complex cardio-thoracic anatomy and the coronary arteries.

In 1984, EBCT was introduced as the first cross-sectional noninvasive imaging modality that could visualize the cardiac anatomy and the coronary arteries *(1)*. During this period, mechanical scanners typically had single-slice detectors and a minimum gantry rotation time of 0.75–1.0 s, and were not considered of value for strictly cardiac imaging. With presently available EBCT scanners, temporal resolution of 100 ms provides motion-free images of the cardiac anatomy in the diastolic phase of the cardiac cycle even at higher heart rates *(2)*. Cardiac anatomy can be covered in a single breath-hold of 30–40 s with slice widths of 3 mm, but this limits the diagnostic

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