A Versatile Computer Simulation of the Left Ventricle

Tom R. Miller, David M. Epstein,* and Kenneth J. Goldman†

The Edward Mallinckrodt Institute of Radiology, Washington University School of Medicine, St. Louis, Missouri

A versatile mathematical simulation of a beating left ventricle, as seen in a gated cardiac blood-pool study, is described. The basic model consists of a fifth-degree polynomial rotated about its long axis. Motion is mimicked by sinusoidally varying the lengths of the long and short axes. The quality of clinical images is simulated through addition of Poisson noise and camera blur, determined from the measured line-spread function for Tc-99m. To achieve added realism, the modeled ventricle can be inserted in place of the patient's actual ventricle in clinical images. The dimensions, activity, and ejection fraction can all be adjusted. In addition, regional wall-motion abnormalities can be generated in any of the walls. To illustrate applications of the model in testing and evaluating cardiac computer programs, the performances of two ejection-fraction programs are compared, and use of the model in evaluating digital filtering algorithms is discussed.

J Nucl Med 25: 1362–1366, 1984

In recent years many new computer methods have been developed to analyze data obtained from the gated cardiac blood-pool study. With any new processing technique there is a need to test the software for errors in programming, and to evaluate the performance of the computer methods in the clinical setting. Initial testing usually does not use mathematical images that accurately simulate the characteristics of the clinical data. Further evaluation of new computer methods is often performed by correlation with results obtained at cardiac catheterization. While this procedure is an essential step in evaluating most new computer techniques, assembly of an adequate number of cases is often difficult, and precise correlation, such as in wall-motion grading, may be uncertain.

To supplement the existing methods of testing and evaluating new computer programs for cardiac analysis, a versatile mathematical model of the beating left ventricle was developed. Illustrations are presented of important potential uses of this model in evaluating ejection-fraction programs and digital filters.

MATERIALS AND METHODS

To simulate a beating left ventricle in a gated cardiac blood-pool study, a series of 32 images are generated, each with a 64- by 64-pixel matrix, in which the ventricle is modeled by a polynomial equation with the parameters of the model varying sinusoidally to achieve motion. Regional abnormalities of wall motion can be introduced by modifying the motion equation for selected transaxial slices of the left ventricle. The quality of actual nuclear medicine images is simulated by adding background activity, applying a blurring function, and introducing Poisson noise.

To achieve added realism, the modeled left ventricle can be inserted in place of the true ventricle in actual clinical studies, while retaining the normal adjacent structures.

Model construction. The static model consists of a polynomial curve rotated about a vertical axis of symmetry (Fig. 1). Empirical investigation revealed that a simple fifth-degree polynomial closely approximates the ventricular shape at the apex and along the lateral wall. That is,

$$y = x^5, \qquad (1)$$

where x is the perpendicular distance from the ventricle's long axis (y-axis) and y is the distance from the apex. The polynomial is truncated superiorly, giving a horizontal mitral-valve plane. The intensity of the activity within the ventricle can be adjusted to simulate high- or low-count images. No correction is made for photon attenuation within the ventricle.

To understand the generation of temporal motion and abnormalities of regional wall motion, the entire model may be thought of as composed of two stacks of semicircular disks (Fig.2). Temporal motion is achieved in the vertical direction by varying the length of the long axis of the ventricle, leaving the point of cutoff (the valve plane) fixed and moving the x-axis vertically. This can be visualized as eliminating disks at the valve plane as systole proceeds. Horizontal motion is simulated by varying the radii of the semicircular disks between image frames. Expressed mathematically, Eq. (1) is modified such that

Received Dec. 19, 1983; revision accepted June 29, 1984.

For reprints contact: Tom R. Miller, MD, PhD, Mallinckrodt Institute of Radiology, 510 South Kingshighway, St. Louis, MO 63110.

* Present address: Dept. of Pediatrics, Mt. Sinai Hospital, New York, NY.

† Present address: Brown University, Providence, RI.

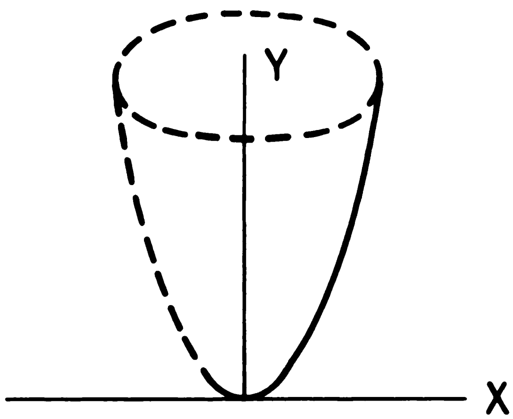

FIG. 1. Static ventricular model is fifth-degree polynomial rotated about its axis of symmetry (y-axis).

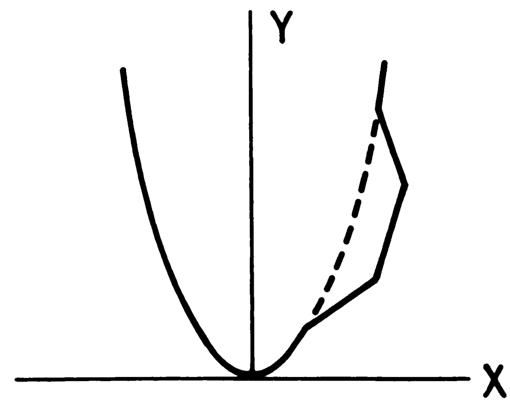

FIG. 3. Generation of dyskinetic segment in lateral wall is shown. Normal position of lateral wall at end-systole is shown by dashed line.

$$x = f(t)y^{1/5}, \tag{2}$$

where f(t) represents the radial variation with time. Both the vertical displacement and the radial motion are applied in proportion to a frame's temporal "distance" from end-diastole. Comparison with actual left-ventricular volume curves in normal patients reveals that realistic motion, with smooth directional changes at end-systole and end-diastole, may be simulated for both vertical and radial motion with two half-cycle cosine curves having different periods and equal amplitudes, one for systole and one for diastole. By adjusting the relative periods of these cosine curves, the user can vary the end-systolic frame number. The ejection fraction (EF) is determined by setting the amplitudes of the cosine functions.

Abnormalities of wall motion are produced by altering the systolic and diastolic length (or width) parameters in part or all of each wall. Thus, apical abnormalities are simulated by varying the number of disks, while septal and lateral wall-motion abnormalities are simulated by varying the radii of the semicircular disks in the left and right stacks, respectively. For example, dyskinesis may be modeled by specifying that systolic length (or width) be greater than diastolic length (or width) in the affected wall. Septal and lateral wall-motion abnormalities are slightly more complex than apical defects, since their position and length along the wall may vary. These abnormalities are produced by varying the position and number of semicircular disks whose radii are changed. Added realism is achieved by specifying the fraction of the abnormality that is to be fully abnormal; the remaining portion of the abnormality is then gradually tapered off to normal in a linear fashion. Figure 3 illustrates the model's behavior in constructing a dyskinetic lateral wall segment. The abnormality is centered at 50% of the length of the wall from the apex. The middle third of this region is fully abnormal, while the upper and lower thirds are tapered off to normal. By adjusting the magnitude, position, length, and percentage fully abnormal, a wide range of hypokinetic, akinetic, and dyskinetic abnormalities may be produced in the lateral, septal, and apical walls.

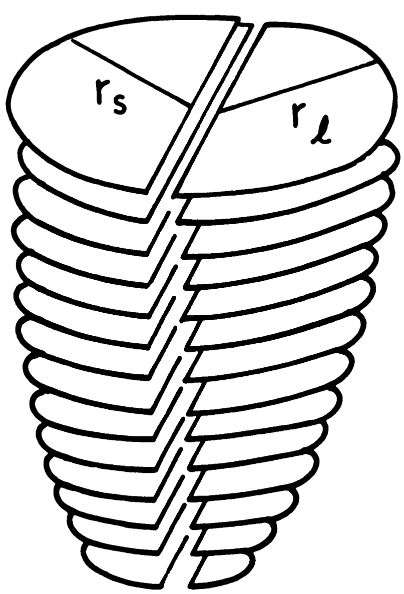

FIG. 2. Model is shown as two stacks of semicircular disks with radii r_l (lateral) and r_s (septal).

The ejection fraction and ventricular time-activity curve are determined as the model is generated by summing the counts in each frame before addition of background activity.

Model degradation. Next, background activity, camera blur, and noise are added to the images. A background level set by the user is added to each frame of the undegraded model. A gradual shading of the background plane from a slightly higher value in the upper left to a slightly lower value in the lower right may be used, if needed, to produce a more realistic approximation of actual background activity in the chest.

The line-spread function of a standard-field-of-view scintillation camera, equipped with a low-energy, all-purpose collimator, was measured at 140 keV with use of a line source of Tc-99m at a depth of 9 cm in a Plexiglas scattering medium (FWHM 11 mm or 2.4 pixels, FWTM 31 mm or 4.4 pixels) (*1*). This line-spread function, representing the one-dimensional camera blurring function, was transformed to a two-dimensional convolution mask (FIR filter) of size 5 × 5 as previously described (*2,3*). This filter is the point-spread function that is passed over the model to simulate camera blur. A larger filter size is not required, since the filter coefficients are less than 10% of their central values at the edges of the mask. Finally, Poisson noise is added to each frame with use of a random number generator (*4*). Noise is added to each pixel according to the counts in that pixel.

Model insertion. A degraded model can be inserted in place of a patient's left ventricle in actual clinical studies obtained in the left anterior oblique (LAO) projection. First, a region of interest is manually drawn around the left ventricle in the clinical study. Then the computer deletes the image within the region and replaces it with the simulated ventricle. Patient studies and models are matched with respect to left-ventricular size, background level, and frame of systole. Satisfactory separation of the left and right

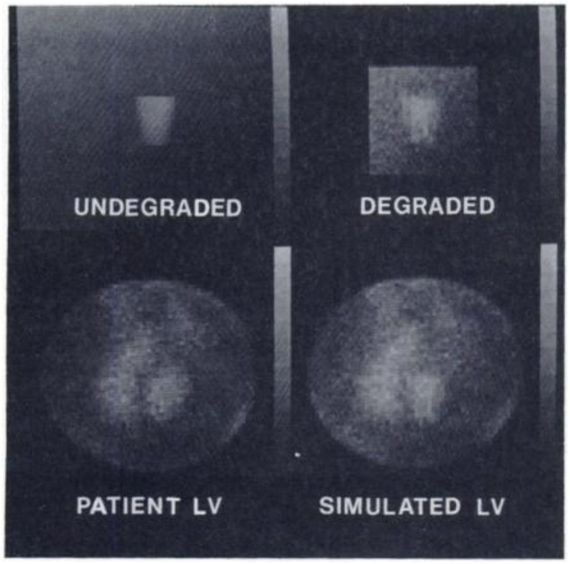

FIG. 4. Normal left-ventricular model is shown at upper left, with shaded background. Same model after degrading with blur and noise is at upper right. Lower left image is from actual clinical study, while at lower right simulated ventricle replaces patient's actual ventricle.

ventricles, and orientation of the long axis of the ventricle parallel to the y-axis of the image, are critical requirements in choosing the clinical studies.

Applications of the model. To illustrate applications of the model, ejection fraction and digital filtering programs used in clinical practice were tested by comparing the results obtained with the undegraded model with the results of the computer program when applied to the degraded model inserted into clinical images.

Edge detection and ejection fraction. The true time-activity curves and ejection fractions of undegraded, isolated models were determined as the models were generated. Following degradation and insertion of the models into clinical images, the curves and ejection fractions were computed with use of two clinical programs. Program 1 uses a very simple, automatic algorithm to detect the ventricular edge in each image frame (*5*). Following nine-point smoothing of the image, this algorithm searches radially from the center of the ventricle as defined by the operator. Preliminary edge detection is signalled by a change in sign of the difference between adjacent pixels along each radial. The final edge is generated by smoothing out sharp discontinuities around the edge. Program 2 uses a manually drawn, end-diastolic region of interest based on digitally filtered images using a Wiener filter—a digital filter with

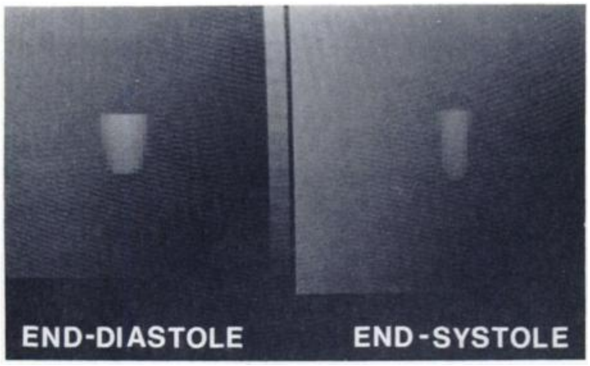

FIG. 5. Simulated ventricle with dyskinetic apex is shown at end-diastole and at end-systole.

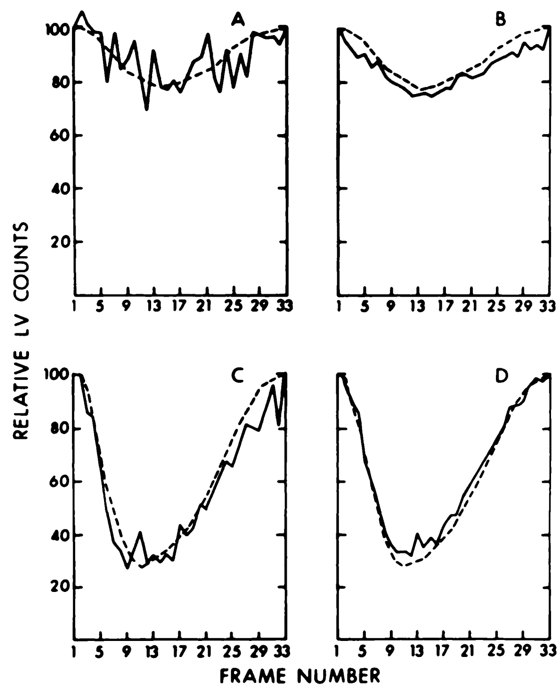

FIG. 6. Time-activity curves are shown as calculated directly from undegraded models (dashed lines) and as determined with use of two programs (solid lines). A: Curves obtained with Program 1 for ventricle with ejection fraction of 22%. B: Results with Program 2. C and D: curves obtained with Programs 1 and 2, respectively, applied to model of Fig. 4 but with constant background.

significant edge-sharpening properties (*2,6*). This fixed region of interest is then used to generate the time-activity curve.

Digital filtering. Degraded images with both high and low total counts were generated and used to test the performance of the Wiener filter (*2,6*) in visually evaluating images from gated cardiac blood-pool studies. Quantitative results using isolated, globally hypokinetic ventricles are reported elsewhere (*7*).

RESULTS

Figure 4 shows the steps in generating a normal left-ventricular model. An undegraded end-diastolic image is shown in the upper left panel. In the second image, camera blur and Poisson noise have been added. The lower left panel shows the first frame of an actual clinical study collected in the 35° LAO projection for 3.5 million total counts. At lower right the degraded model has been inserted into this clinical image in place of the patient's left ventricle. The left-ventricular ejection fraction of the model was 72%, the ventricular width at the base at end-diastole was nine pixels, and the end-diastolic length from base to apex was 13 pixels. Maximum activity within the simulated ventricle (110 counts/pixel) was set to match the activity in the actual ventricle in the clinical study. The average background was 50 counts, with slight shading from upper left (higher counts) to lower right (lower counts).

In Fig. 5 a simulated ventricle with a dyskinetic apex is shown, to illustrate the generation of wall-motion abnormalities.

Figure 6 illustrates use of the model in evaluating two programs for determination of left-ventricular ejection fraction. The true ventricular time-activity curve and the curve generated by Program 1 are shown in Panel A for a simulated ventricle with ejection fraction of 22% and pixel counts corresponding to a total image collection of 4 million counts. The results with use of Program 2 are shown in Panel B. The ventricle shown in Fig. 4, based on a 3.5-million-count clinical study (but without background shading),

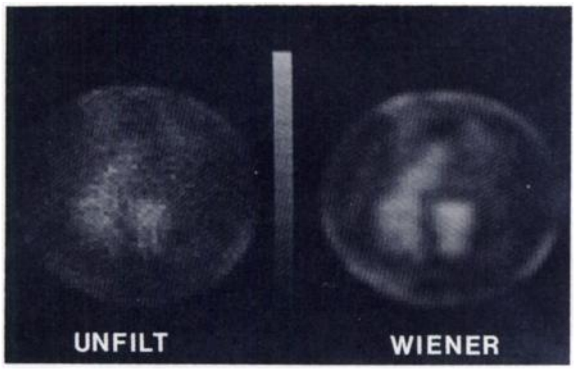

FIG. 7. Left: End-diastolic frame from study of Fig. 4. Right: Same image after Wiener filtering.

was used with Program 1 to obtain the data in Panel C, and with Program 2 in Panel D. For the first model, the results with Programs 1 and 2 were 23% and 24%, respectively. For the second model with an EF of 72%, Program 1 yielded 72% while Program 2 gave 74%. While these results are very similar, note the much closer tracking of the true time-activity curves (dashed lines) by Program 2 in both cases, with especially poor performance of Program 1 in the study with low ejection fraction (Panel A). These findings are in agreement with actual clinical experience with these two programs, except near end-systole, where left-atrial activity leads to underestimation of ejection fraction with Program 2 in the clinical studies.

In Fig. 7 the end-diastolic frame from the study shown in Fig. 4 is displayed along with the same image following digital filtering with the Wiener filter.

DISCUSSION

There are at least three important applications of a mathematical model of the left ventricle.

(1) In testing a new computer program for errors (debugging), simulated clinical data are needed in which the true values of the parameters to be measured by the program are precisely known;

(2) In evaluating the performance of a new processing technique, a realistic simulation of clinical data can serve as an important adjunct to the use of "proven" cases, which are often obtained by correlation with cardiac catheterization;

(3) In training new radiologists, a wide range of normal and abnormal images can be generated to aid in teaching visual perception, understanding functional images, and operating analysis programs.

To be useful in these applications, the present left-ventricular model was designed with three requirements in mind. First, it must approximate the structure of the human left ventricle; second, it must be capable of producing global and regional wall-motion abnormalities in any of the walls and with variable severity; and third, the model must realistically simulate the blur and noise inherent in actual nuclear medicine images. The present model has the flexibility to meet the first two requirements and, as seen in the figures, its images are indeed very realistic in appearance. In fact, when a clinical image is displayed in cine mode side-by-side with the same image containing, instead, the mathematical ventricle (as in Fig. 4), experienced observers frequently cannot tell which is the true and which is the altered image.

Since our model meets the requirements of the applications described here, other physiologic and physical features of cardiac imaging were not incorporated in it, since they would only add to the programming complexity and the time required to generate the simulations. These features include photon attenuation, nonvertical chamber orientation, a more complex ventricular shape, and a sequential conduction wave moving through the ventricle.

With use of this model, the performance of clinical programs for determination of ejection fraction can be evaluated under conditions in which the true ejection fraction and the entire time-activity curve are precisely known. The accuracy of ejection-fraction determinations can be tested over a wide range of ventricular function. The effects of different collection times can be evaluated by generating simulated ventricles with differing numbers of total counts. The effect of photon attenuation on ejection fraction is not incorporated in the model.

Use of the model is illustrated here by comparing the performance of two clinical programs applied to images with widely differing ejection fractions. The program using manual delineation of the ventricular border was found to produce a much smoother time-activity curve in this simulation than the program incorporating automatic edge detection. Clinical experience with these programs was quite similar to the results obtained with use of the modeled ventricles. These findings, however, are not meant to indicate general superiority of any particular method of ejection-fraction determination; they are presented only to illustrate use of the model in assessing two particular ejection-fraction programs that use widely different computational methods.

In testing ejection-fraction programs, use of the simulated left ventricle does not replace the need for correlation with catheterization results. However, it does serve as a useful supplement because a wide range of images can be simulated. Interinstitutional comparison of programs could be faciliated through exchange of a library of simulated cases.

One of the most important applications of this model is in evaluating image-processing algorithms (see Fig. 7). Since the original, undegraded image is precisely known, the effect of a filtering method on the simulated clinical image—degraded by blur and noise—can be quantified by comparing the filtered image with the original image pixel-by-pixel (*8*,*9*).

To assess quantifiably the value of a new image-processing scheme applied to gated cardiac blood-pool studies, it is necessary to evaluate the ability of observers to grade regional wall-motion abnormalities in images of varying quality. With this model, very realistic, beating left ventricles can be generated, with wall-motion abnormalities placed in any wall or combination of walls. Thus, evaluation of new processing methods can be undertaken, such as with use of the receiver operating characteristic (ROC) approach, by comparing wall-motion grading with and without application of the processing algorithm (*7*). Recognition and understanding of artifacts generated by image processing is greatly facilitated because the true characteristics of the image are precisely known.

A final, and potentially valuable use of this model is in teaching. A wide range of realistic images can be generated to train physicians and technologists in the operation of clinical programs and in the understanding of cardiac images.

ACKNOWLEDGMENTS

The authors thank K. S. Sampathkumaran and Drs. Daniel R. Biello and Barry A. Siegel for their valuable suggestions, and are grateful to Andrea Sykes and Linda Beck for their assistance in the preparation of the manuscript.

This research was supported in part by a National Institutes of Health Specialized Center of Research in Ischemic Heart Diseases HL17646.

REFERENCES

1. METZ CE, KUNIO D: Transfer function analysis of radiographic imaging systems. *Phys Med Biol* 24:1079–1106, 1979

2. MILLER TR, SAMPATHKUMARAN KS: Digital filtering in nuclear medicine. *J Nucl Med* 23:66–72, 1982
3. MCCLELLAN JH, CHAN DS: A 2-D FIR filter structure derived from the Chebyshev recursion. *IEEE Trans Circ Syst* CAS-24:372–378, 1977
4. SNYDER DL: *Random Point Processes.* New York, John Wiley and Sons, 1975, p 62
5. BIELLO DR, SAMPATHKUMARAN KS, GELTMAN EM, et al: Determination of left ventricular ejection fraction: A new method that requires minimal operator training. *J Nucl Med Tech* 9:77–80, 1981
6. GOODMAN JW, BELSHER JF: Fundamental limitations in linear invariant restoration of atmospherically degraded images. In *SPIE Vol. 75 Imaging Through the Atmosphere*, Society of Photo-Optical Instrumentation Engineers, 1976, pp 141–154
7. MILLER TR, GOLDMAN KJ, EPSTEIN DM, et al: Improved interpretation of gated cardiac images by use of digital filters. *Radiology:* in press
8. PRATT WK: *Digital Image Processing.* New York, John Wiley and Sons, 1978, pp 174–182
9. KING MA, MILLER TR, JACOBS DA, et al: Nonstationary image processing in the frequency domain. In *Digital Imaging: Clinical Advances in Nuclear Medicine.* New York, The Society of Nuclear Medicine, Esser P, ed. 1982, pp 127–141