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HIGH TEMPORAL RESOLUTION ECG-GATED SCINTIGRAPHIC ANGIOCARDIOGRAPHY

Michael V. Green, Harold G. Ostrow, Margaret A. Douglas, Richard W. Myers, Richard N. Scott, James J. Bailey, and Gerald S. Johnston

National Institutes of Health, Bethesda, Maryland

The cardiac blood pool is visualized with high temporal resolution during a complete, average, cardiac cycle. The technique yields both qualitative and quantitative measures of cardiac performance.

A method is described that produces a collection of scintigraphic images (typically 80–100), each 10 msec in duration, that depicts a single, average cardiac cycle. This image sequence constructed from 1,200–1,500 heartbeats may be analyzed to yield estimates of various left ventricular function parameters such as ejection fraction, mean and peak fractional ejection rate, and mean and peak circumferential fiber shortening. Various temporal quantities can be measured such as left ventricular ejection duration and time-to-peak flow.

Left ventricular wall motion abnormalities can be detected in the difference image created by subtracting the end-systolic image from the end-diastolic image or by presenting the image sequence in cineangiogram format.

METHOD

A scintillation camera equipped with parallel-hole collimation is positioned over the patient's heart in a left anterior oblique orientation modified by a 15–30 deg superior rotation relative to the transverse plane (*1*). This position allows the scintigraphic images of the cardiac chambers to be separated.

A bolus of ^{99m}Tc-human serum albumin (10 mCi) is rapidly injected into the antecubital vein. The first transit of the radionuclide through the heart is used to determine cardiac output and to help delineate cardiac anatomy.

After the tracer is equilibrated in the patient's blood volume, three ECG leads are attached. The ECG signal is continuously processed by a trigger device which emits a pulse (R-marker) each time the R–S transition of the QRS complex is encountered.

Digitized x–y scintillation coordinates and R-markers from the ECG trigger device are then accumulated with a minicomputer system (12K, 16 bits/word) in LIST mode. These data (typically consisting of about 5 million scintillation events per 20 min) are recorded on magnetic tape and transferred to a large-scale digital computer facility (200K, 36 bits/word).

Here the LIST mode data are sorted into a sequence of images or frames as depicted in Fig. 1. The LIST mode data are searched for the first R-marker (Beat 1). The x–y data from scintillations that occurred during the first 10 msec following this marker are sorted into the first picture; the second 10 msec of x–y data are sorted into the second picture, and so on until the next R-marker is detected. Following this R-marker (Beat 2) the first 10 msec of x–y coordinates are sorted additively

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For reprints contact: Michael V. Green, National Institutes of Health, Dept. of Nuclear Medicine, Bldg. 10, Room 1B38B, Bethesda, Md. 20014.

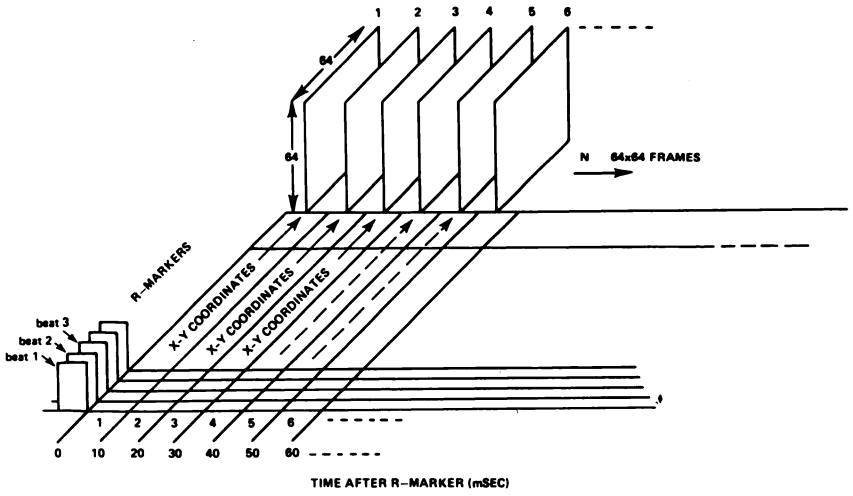

FIG. 1. Picture file creation. 1,200–1,500 cardiac cycles (beats) are examined. Upon completion of framing, each picture typically contains 50 × 10^3 events of which about 10–20% originate in left ventricular region. Picture file represents single, average cardiac cycle.

into Picture 1, the second 10 msec of x–y coordinates into Picture 2, and so on until all R–R intervals have been examined. The final result is a collection of sequential scintigraphic images that represents an average cardiac cycle. This picture file is written back onto magnetic tape and returned to the minicomputer system for analysis.

The picture file can be manipulated to yield a left ventricular "volume" curve from which parameters describing left ventricular function can be obtained. First, an end-diastolic and an end-systolic image are created (Fig. 2A and B, respectively). The end-diastolic image is formed by summing the first three pictures in the file and is thus comprised of scintillation data that occurred during the first 30 msec following each R-marker. The end-systolic image is created by summing the three consecutive images in the file that bracket the time at which the minimum occurs in the (uncalibrated) volume curve (310 msec after the R-marker in Fig. 2E).

Figure 2C shows a region of interest over the left ventricle at end-diastole that has been identified to the minicomputer system. The total count in this region (approximately 10^4 counts at end-diastole) is plotted as a function of picture number, i.e., time during the average cardiac cycle, in Fig. 2E. The peak before the final falloff of this curve indicates the point of minimal R–R interval length for all the beats examined. Beyond this point (to the right of) fewer and fewer beats contribute to the count-time curve. Finally, the point of maximal R–R interval length is reached beyond which no counts are added to the curve. The mean R–R interval length is thus represented by a point about midway down the final slope of the curve.

A substantial portion of the total count in the ventricular region of each image (approximately 50% at end-diastole) is made up of events due to activity in the vessels of the lungs, chest wall, and myocardium. To correct for this background offset, a second region of interest (Fig. 2D) is defined by light pen around the ventricle at end-systole (*2*). On the assumption that left ventricular background is time-independent, the mean count per cell over this region is calculated and this value is subtracted from each cell in the first region (Fig. 2C) over the entire cycle. The result is a curve (Fig. 2F) whose ordinate values are proportional to ventricular volume.

DISCUSSION

Representative quantities that can be computed from this curve include the following:

1. Ejection fraction, $(V_{ED} - V_{ES})/V_{ED}$. In the case shown the ejection fraction was 0.63 (mean ±1 s.d.). Normal (*3*): 0.64 ± 0.07.
2. Peak fractional ejection rate (PFER) can be computed using a suitable polynomial fit to the systolic portion of the volume curve. PFER is defined as maximum systolic dV/dt where $V = V(t)/V_{ED}$ and V(t) is ventricular volume at time t. In the example, PFER was 3.14/sec. Normal (*3*): 2.84 ± 0.50/sec.
3. Peak circumferential fiber shortening (PCFS) can be estimated from the volume curve if it is assumed that the left ventricle is a contracting prolate spheroid whose major and minor axes are in constant proportion. Expressed in end-diastolic circumferences per second, PCFS is:

$$PCFS = [\tfrac{1}{3}(V^{-2/3})\, dV/dt]\ max.$$

In the case shown, this model gave a PCFS of 1.27/sec. Normal (*3*): 1.28 ± 0.33/sec.

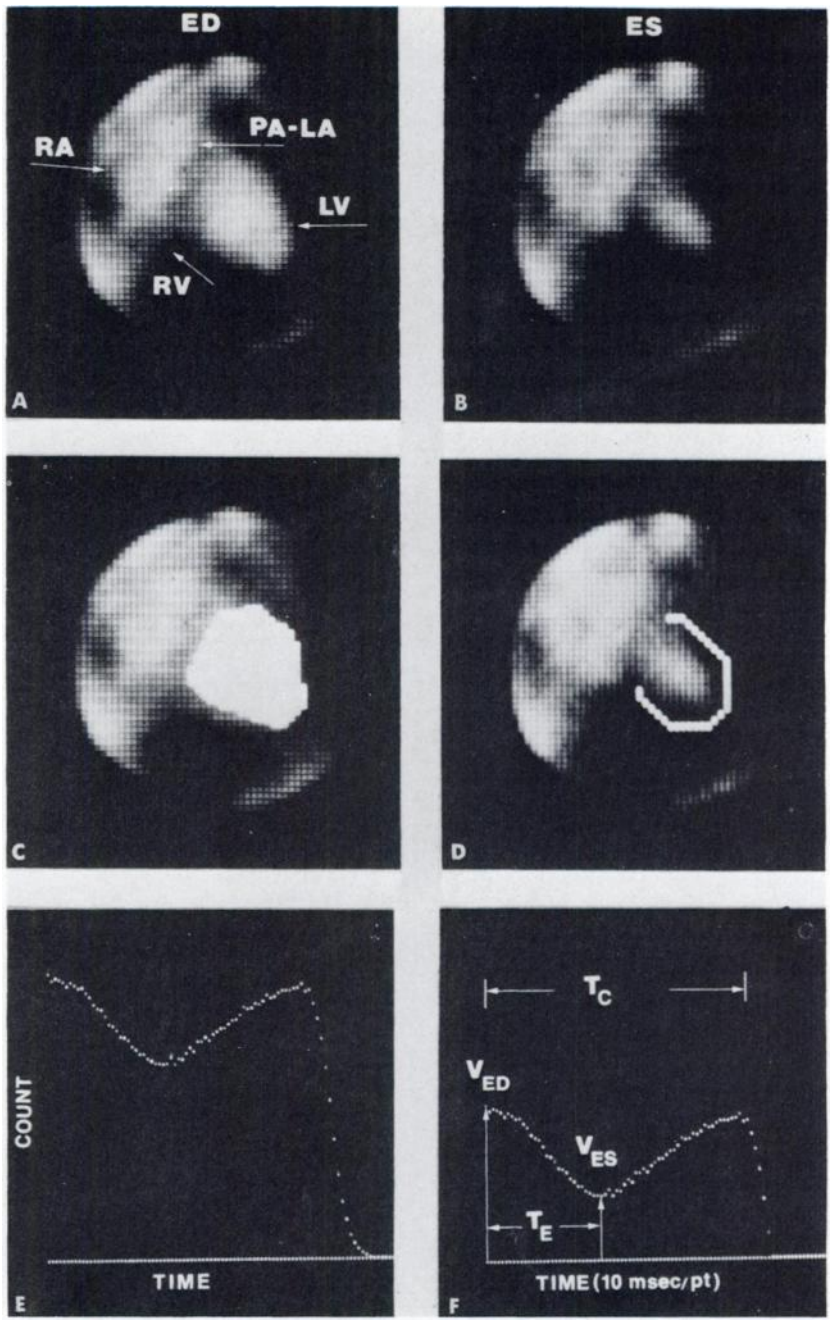

FIG. 2. Results obtained in patient with aortic insufficiency. (A) End-diastole. RA, right atrium; PA-LA, pulmonary artery-left atrium; LV, left ventricle. (B) End-systole. (C) Left ventricular cursor (D) Background cursor. (E) Raw volume curve. (F) Background-corrected volume curve. T_C, minimum cardiac period; V_{ED}, end-diastolic volume; V_{ES}, end-systolic volume; and T_E, systolic duration.

4. The time between ejection onset and peak systolic flow was 170 msec. Normal (*3*): 167 ± 45 msec.

Qualitative assessment of left ventricular wall motion as well as cardiac anatomy can be achieved by presenting the picture file in cineangiogram format (i.e., cyclic projection of the picture file to simulate continuous beating of the heart). Wall motion abnormalities can also be detected in the difference image formed by substracting the end-systolic image from the end-diastolic image.

The method outlined here has several advantages over conventional gated studies (*4–7*). The cineangiogram display allows a qualitative assessment of ventricular performance in a format that is not possible with methods which produce only end-diastolic and end-systolic images. The data in Fig. 2F can also yield estimates of time-dependent functional parameters (*3*). In addition, the end-systolic image can be constructed without knowledge of the T-wave. Instead the temporal location of end-systole is obtained directly from the volume curve.

The basic methodology is currently being investigated in a baboon preparation with a surgically implanted flowmeter around the ascending aorta. In these studies the effects of R–R variation and im-

proper cardiac chamber separation can be assessed.

Comparative studies in patients undergoing cardiac catheterization are also underway to evaluate alternative ventricular background correction methods. Also under investigation are independent methods (*6,7*) that can be used to calibrate the volume curve by providing knowledge of absolute end-diastolic and end-systolic volumes.

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