The Use of Phantoms for Quality Control in Gated Cardiac Studies

Ellinor Busemann-Sokole* **and Trevor D. Cradduck**

Victoria Hospital Corporation, London, Ontario, Canada

A cardiac phantom suitable for quality control of gated blood pool studies is desired. Three commercially available cardiac phantoms purport to fulfill the needs of monitoring the total imaging system. The Vanderbilt phantom has considerable drawbacks, arising mainly from the attenuation of ventricular and background activity from the rotation of an attenuator, which consequently renders it unsuitable for quality control of both the hardware and software of the system. The Jake phantom, though of different design, also suffers from the same drawback in that the counts arising from the background chamber vary in both the region of the left ventricle and the normally adjacent "background" region. Otherwise, it is a satisfactory model for checking overall system performance. The Veenstra cardiac phantom produces a good simulation of left and right ventricular wall motion and stroke volume changes, and constant overlying background activity. Of these phantoms, the Veenstra phantom offers the most versatility for routine quality control of the overall system.

Gated blood pool studies for non-invasive determination of ventricular ejection fraction have become routine in most nuclear medicine departments. There is no absolute standard protocol for these determinations. As a consequence, ejection fraction values for the same heart function can vary considerably. The ejection fraction measurement relies on the synchronization of the ECG trigger and computer gate, the consistency of the framing rate over the R-R interval, the mechanism of ectopic beat rejection, and the software algorithms used for the ejection fraction calculation. A quality control program should include the monitoring of all of these parameters, but should not be so complicated that it becomes prohibitive in its application. On the one hand, a phantom can be used to check the performance of the hardware, and on the other hand, it can be used to generate a standard set of cardiac data that can be used to check software algorithms and the performance of the operators using those algorithms. Such a phantom should mimic the anatomical and physiological characteristics of the heart as closely as possible, but remain simple to use and provide reproducible data.

A number of phantoms have been proposed for the purpose of quality control of cardiac studies (*1–6*). Those described by Nickles (*1*), Hurst et al. (*2*), Shulz et al. (*3*), and Kan and Hopkins (*4*) involve pumps and movement of radioactivity between chambers. The phantom described by Bennett et al. (*5*) employs a rotating disc on which a pattern of absorbers or radioactive sources can be mounted. The phantom of Hudson et al. (*6*) consists of a pendulum source that oscillates over a lead plate with a waisted aperture. The applicability of these phantoms varies, but none fulfills the criteria of checking both hardware and software of the camera–computer system. In addition, they are not generally available to the nuclear medicine community.

The phantom first described by Price et al. (*7,8*), and known as the Vanderbilt cardiac phantom, is now produced commercially.* Recently, two other commercially manufactured phantoms have become available: the Veenstra phantom† and the Jake heart phantom‡ (*9*). All are purported to provide a check of both the hardware and software system and seem simple to use. Therefore, they appear suitable for routine quality control of camera–computer systems used for cardiac studies. The application of these phantoms for quality control is described here.

MATERIALS AND RESULTS

Vanderbilt Cardiac Phantom

Figure 1 is a schematic diagram of the Vanderbilt cardiac phantom (*7,8*). The left ventricle (LV) and atrium are simulated by two hollow prolated spheroids of 40-ml and 20-ml capacity, respectively, stacked one above the other, with their major axes orthogonal. After filling with a homogeneous mixture of radioactivity, these chambers are mounted onto a base which contains a motor that rotates the spheroids at a speed that can be varied continuously between 20 and 200 revolutions per minute. The rotating spheroids simulate LV and atrial wall motion. A simulated ejection fraction is obtained by placing a metal attenuator adjacent to the left ventricular spheroid at the major axis. The attenuator rotates with the spheroids and attenuates the photon beam when passing between the camera and spheroid. Three calibrated attenuators give simulations of three specific ejection fractions (25%, 50%, and 75%).

A hollow background chamber simulates a fixed shape right ventricle, atrium, and aorta. It may be filled with the same or a different concentration of radioactivity as the spheroids. The background chamber is designed to be positioned behind the spheroids. This means that photons emanating from the background chamber will be attenuated by the rotating spheroids and attenuator when viewed by the camera.

For reprints contact: Trevor D. Cradduck, Dept. of Nuclear Medicine, Victoria Hospital Corporation, London, Ontario, Canada.

*Present Address: Dept. of Nuclear Medicine, Academic Medical Center, University of Amsterdam, Amsterdam, The Netherlands.

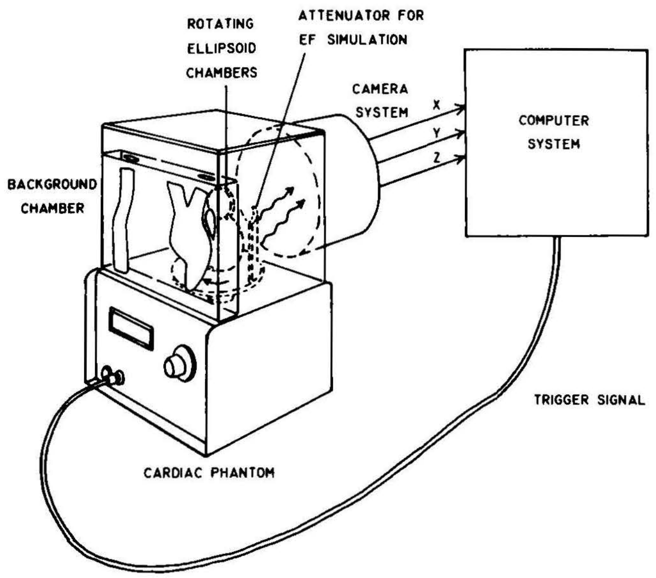

FIG. 1. Schematic representation of the Vanderbilt cardiac phantom.

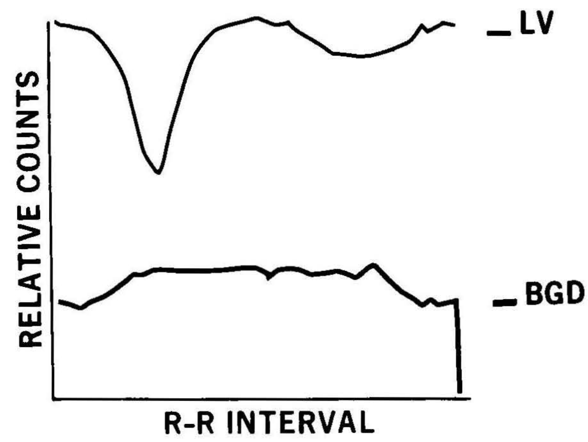

FIG. 2. Curves from the Vanderbilt phantom for regions of interest placed over the left ventricle (LV) and a background region (BGD) adjacent to the left ventricle. The simulated stroke volume curve occurs during the first half of the R-R interval. The dip in the second half of the LV curve is due to self-attenuation. The dip at each end of the BGD curve is caused by the metal attenuator as it passes in front of that portion of the background chamber.

The trigger pulse supplied by the phantom is a TTL-level signal intended for input directly into the computer. The trigger pulse is generated once per revolution of the spheroids. As shown in Fig. 2, the method of attenuation is such that only the first 180 degrees of the revolution simulates a stroke volume curve, i.e., when the attenuator passes between the LV spheroid and the camera. The remaining portion of the R-R interval gives superfluous data. When setting up the phantom for data acquisition, the displayed rotation speed, labeled "beats per minute", needs to be doubled to correspond to the equivalent simulated heart rate. The data acquisition parameters must be similarly adjusted: either the total number of frames per R-R interval and total number of counts need to be doubled if comparison is to be made with the clinical situation, or a fixed time interval must be assigned to each frame and the total number of frames selected so that data are acquired only during the first half of the R-R interval, thereby collecting only the relevant stroke volume data.

The method of attenuation of LV activity and the positioning of the background chamber behind the LV spheroid leads to a situation where the perceived background counts are not constant. If one takes a single LV region at end-diastole and a background region in the normally accepted position from 3 to 6 o'clock on the lateral border of that LV region, the measured background counts vary as the attenuator rotates through these regions (Fig. 3). Moreover, the variation of background region counts is out of phase with respect to the variation of background counts in the LV region itself. This phenomenon constitutes a severe deficiency for the evaluation of software algorithms. Each background subtraction algorithm results in a different ejection fraction value, whether it is point by point subtraction of the background curve from the LV curve, subtraction of a value averaged over the cycle, or a single background value taken at end-systole.

Without the background chamber, the measured results are "correct" in the sense that the ejection fraction numbers agree with the calibration values of 25%, 50%, and 75%. Difficulties arise when the background chamber is added to the phantom and the associated background correction must be made. Most software algorithms assume a constant contribution from background in both the left ventricle and the "background" regions

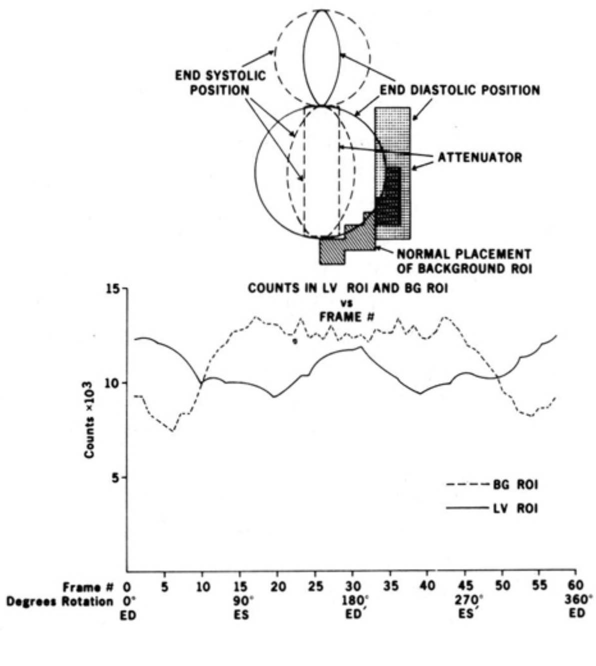

FIG. 3. Time-activity curves obtained over the left ventricular ellipsoid and adjacent background region when only the background chamber is filled with radioactivity. The left ventricular chamber is filled with water. The dotted line was derived from the hatched area indicated by "normal placement of background ROI"; the solid line was derived from the circular area indicated as the "end-diastolic position" for the lower chamber.

of interest throughout the cycle. The placement of the background region and the method of background subtraction may differ from one algorithm to another. However, if the contributions of counts from the background chamber to the two regions of interest vary with time, it is difficult to conceive that any algorithm which assumes constancy of those counts can give valid results. Even if the algorithm does not assume a constant background, the fact that the background counts in the two regions of interest do not display the same variation with time will also corrupt the results. Our experiments also demonstrated that the size and placement of the background region could have a very significant effect on the result. Variations of size or position by just one pixel gave different results. In one particular series of experiments using a fixed, manually drawn region of interest over the LV and a background region consistently chosen in exactly the same position just to the right hand side of the ventricular border, we obtained values of 30%, 55%, and 78%, corresponding to the nominal values of 25%, 50%, and 75%.

Trials with two edge detection algorithms (*9,10*), both of which depended primarily on the zero-order second differential of the counts, demonstrated that the edges so chosen correlated poorly with the symmetrical outline of the LV and were judged to be at variance with the region that would have been selected manually. It is assumed that other edge detection algorithms would probably suffer from similar problems.

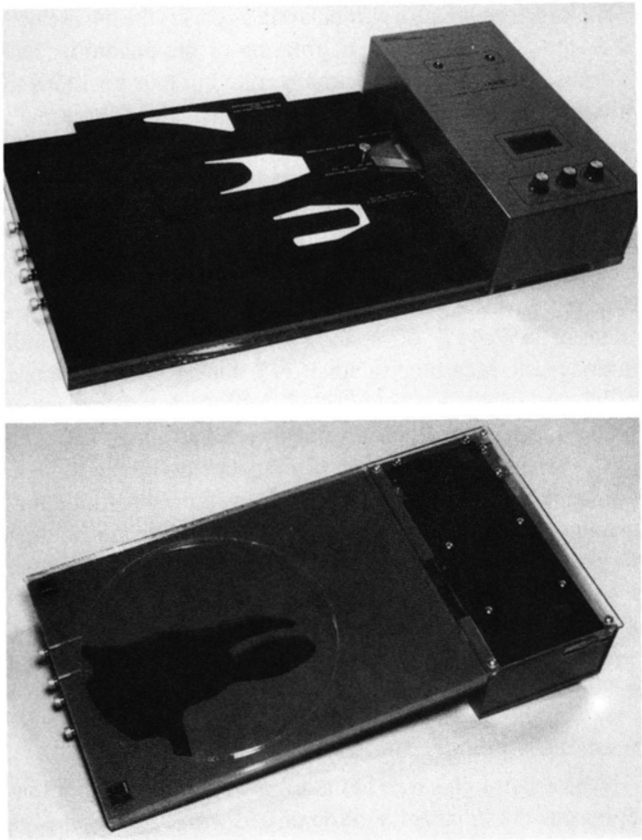

FIG. 4. Photos of the upper and lower views of the Jake phantom. The lower view shows the circular background chamber and cut-out, giving the silhouette of the heart. The upper view includes the three slides with outlines of the masks enclosed in each.

Jake Cardiac Phantom

The Jake phantom (Fig. 4) uses several sheets of Lucite to create a double-layered sandwich (*11*). One layer of the sandwich has a circular chamber that provides for a uniform background source. The other layer of the sandwich has a chamber cut out in the form of the left and right ventricles, the atria, and large vessels. When filled with radioactivity, this chamber gives a uniform silhouette superimposed on the background field.

A number of slides are provided which have shaped masks such that, when slid backwards and forwards in a piston-like motion over the heart chamber, different ejection fractions and wall motion defects are simulated. The slide for normal wall motion can be attached to give one of four quoted ejection fractions—25%, 45%, 60%, and 80%. The slides that simulate an anterioseptal defect and an apical defect are calibrated to give ejection fractions of 38% and 45%, respectively.

An ECG millivolt signal is provided for triggering purposes and the level of this signal can be adjusted. This adjustment was extremely critical on the model we tested. The ECG trigger is generated when the slide is fully retracted (end-diastole) and the R-R interval spans a full stroke of the slide. This allows one to detect time delays between the gate and the commencement of data acquisition.

The heart rate is varied over a range from 40 to 120 beats per minute by means of two uncalibrated controls that vary the rate of movement of the slide during systole and diastole. The desired heart rate must be achieved by adjusting each of these controls while watching a digital read-out. Although this design lends itself to independent variation of the rate of emptying and filling, which is useful when attempting to make an accurate simulation of a stroke volume curve, it is impossible to make this setting on an a priori basis. A study must be collected before one can observe the respective fractions of the R-R interval occupied by systole and diastole. Once set, the heart rate appears to remain stable to within about $\pm$ 3 beats per minute.

The Jake phantom is lightweight (no shielding is included), easy to fill, and simple to operate. Six different ejection fractions can be simulated and three different wall motions modeled. It does, however, suffer from the same deficiency as demonstrated by the Vanderbilt phantom—the background chamber is located distal to the attenuator slide. As a consequence, the "background" counts from the normally accepted background region from 3 o'clock to 6 o'clock vary over the heart beat and, further, the "background" counts in the region of the LV vary. The result is that a plot of background counts displays similar variations with time to those of the Vanderbilt phantom shown in Fig. 3. Ejection fraction algorithms will have difficulty dealing with this abnormal situation.

The two edge detection programs (*9,10*) that we tested were able to identify the edge of the left ventricle, but since there is no significant waisting of the ventricular cavity at the valve plane, the path plotted across that region varied from study to study depending on the count content.

The degree to which a particular mask covers the area selected as the LV controls the calibration of the phantom. The "normal" wall motion slide can be fitted in four positions to simulate four different ejection fractions. The "abnormal" masks have only one associated ejection fraction. It is difficult to determine exactly how these calibrations were made and it is difficult to replicate them in practice. Using manually selected regions of interest and the same algorithm that was used for the Vanderbilt phantom above, measured values of 36%, 63%, 89%, and 70% were determined for nominal values of 25%, 45%, 60%, and 80% using the "normal" wall motion mask. Measured values of 39% and 54% were obtained in the case of the two "abnormal" wall motion masks when the expected or calibration values were quoted as 45% and 35%, respectively. The wide discrepancy between the 38% nominal and 54% measured values using the mask simulating an anterioseptal defect is, perhaps, explained by the fact that the major portion of that particular mask is located over the position of the background region of interest and that portion of the LV adjacent to it. The very wide discrepancies and the reversed order of magnitude with the "normal" mask cannot be explained in these terms.

Veenstra Cardiac Phantom

The Veenstra phantom has a hollow cardiac chamber that represents the RV and LV. When filled with a homogeneous mixture of radioactivity, wall motion and stroke volume changes for both RV and LV are achieved by the movement of metal jaws above the cardiac chamber (Fig. 5), which attenuate the peripheral activity of the cardiac chamber activity. The three selectable LV ejection fractions can be adjusted. Calibration of the actual RV and LV ejection fractions defined by the position of the jaws is required from static images obtained at end-diastole, when the jaws are fully extended, and at end-systole, when the jaws are at minimum closure.

A hollow background chamber simulates atria, great vessels, liver, and spleen. This chamber is positioned above the cardiac chamber and jaws, so that the photons from activity in the background chamber are not attenuated by the cardiac chamber and metal jaws when viewed by the camera.

Two trigger pulses are available: a TTL level signal for direct input into the computer and a millivolt level signal for input into the ECG detection monitor. A trigger pulse is generated when the jaws are fully extended and the R-R interval spans a normal stroke volume cycle. No adjustment of acquisition program parameters is required when setting up this phantom for data acquisition.

The design and movement of the metal jaws and positioning of the background chamber in front of the jaws eliminates the drawbacks observed with the Vanderbilt phantom. Background counts emanating from the background chamber within and adjacent to the LV region are constant over the R-R cycle (Fig. 6). Positioning of the background region of interest is not as critical as it is for the case of the Vanderbilt phantom. The data produced can be submitted to the testing of ejection fraction methods using both edge detection algorithms and different background subtraction algorithms.

The Veenstra phantom includes some shielding which makes the device relatively heavy. It is easy to load and simple to operate. Three heart rates and three ejection fractions can be chosen under push button control, but these settings can be altered by means of adjustable potentiometers.

The Veenstra phantom is calibrated in use by the operator.

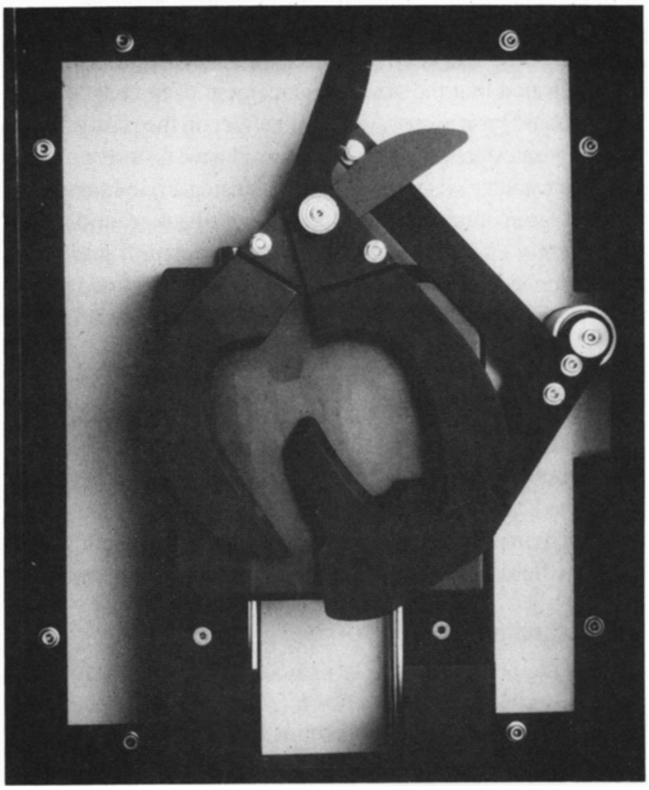

FIG. 5. The Veenstra phantom has metal jaws that open and close to simulate the beating heart. The background chamber that simulates lung, liver, and large vessel activity is placed between the moving jaws and the scintillation camera.

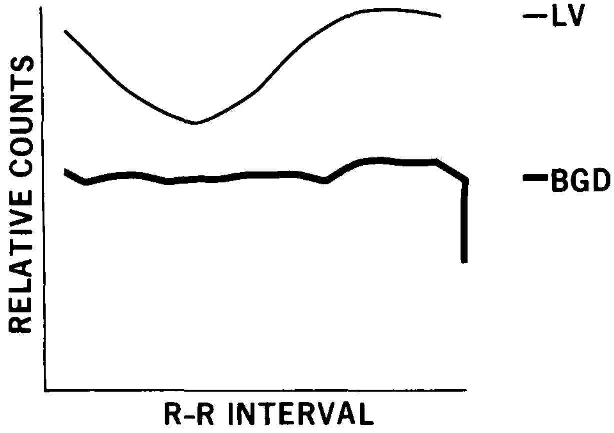

FIG. 6. Curves from the Veenstra phantom reveal that the counts from the normally accepted "background" region of interest remain constant and are independent of the motion of the jaws. This implies that most of the routine ejection fraction algorithms are able to give reasonable results with this phantom.

With the background chamber removed, the moveable jaws are set first at the widest position (end-diastole) and then closed (end-systole). In each of these positions, static images are obtained from which one may calculate the ejection fraction by defining suitable regions of interest. Using the same algorithm to measure the ejection fraction from the gated study, as was used for the Vanderbilt and Jake phantoms, ejection fraction values of 50%, 68%, and 70% corresponding to static calibration values of 54%, 71%, and 75%, respectively, were obtained. The three pre-set ejection fraction values, which depend on the degree of jaw closure, may be adjusted by means of multiturn potentiometers so that a wide range of ejection fraction values may be simulated.

Only normal wall motion may be simulated so that, even if the ejection fraction is in an abnormal range, no akinesis and only generalized hypokinesis of the ventricular walls may be simulated.

DISCUSSION AND CONCLUSION

All of the phantoms described are easy to use. Preparation requires only the filling of hollow chambers. Further handling of the chambers does not involve the risk of radioactive contamination.

The Vanderbilt phantom simulates stroke volume changes in such a way that it cannot easily be used for most software algorithms used to determine ejection fraction. The attenuation of background activity could be circumvented by placing the background chamber in front of the cardiac chambers. However, the phantom is not constructed to facilitate such positioning and would be markedly improved if this modification of design is made. At the present time, it is not sufficient to view the phantom from the rear since this makes the controls inaccessible and, more importantly, causes the cardiac chambers to be switched left-to-right and right-to-left. This invalidates most ejection fraction and edge detection algorithms. Although ejection fraction algorithms that depend on area determination are rare, this phantom would fail in these cases too. In addition, by using two attenuators, one at each side of the major axis of the left ventricular ellipsoid, and triggering every 180 degrees, the R-R interval would encompass a single simulated heart beat so that normal data acquisition parameters could be applied. This modification, as well as an ECG level signal, is now offered as an option by the manufacturer.

The Jake phantom is capable of simulating hypokinesis or akinesis of a ventricular wall. None of the three phantoms accommodates an automatic method of testing ectopic beat rejection. It would be desirable for a phantom to have a capability to simulate ectopic beats, either on a random basis or in a reproducible manner. This could be performed in a fairly uncontrolled manner by changing the R-R interval during the course of acquisition. Both the Veenstra and Jake phantoms provide trigger signals that simulate the ECG trigger signal so that the synchronization of the ECG monitor trigger and computer gate can both be tested. This is important since many monitors have delayed gate outputs. Without modification of the trigger signal or purchase of the option referred to above, the Vanderbilt phantom does not check this aspect of the hardware.

Ejection fraction values reasonably close to those calibrated may be obtained with the Vanderbilt phantom. The values measured with the Jake phantom demonstrate wide discrepancies from the calibration values supplied. This would tend to make this phantom unsuitable for cross-comparison of systems and/or algorithms. The Veenstra phantom is, in effect, self-calibrating in the sense that no calibrated values are supplied and it is the responsibility of the operator to calibrate the phantom with each use. It has the added advantage that right ventricular wall motion is also simulated.

Considerable interest has been shown in determining actual ventricular volume. To do this, it is desirable to have a phantom, the volume of which can be accurately measured in order to check the software used. All three phantoms contain a fixed volume of activity. They are not designed so that software algorithms can be checked over a range of volumes, and therefore fail to meet the requirements of such a determination.

Some of the more desirable quality control procedures can be performed using less expensive equipment. In order to check the R-wave trigger and accuracy of gate synchronization, the method of Wery et al. may be used (*12*). A check of the ECG gating mechanism and framing rate could easily be made using an ECG simulator and flood source (*13*), or, alternatively, an old phonograph turntable with a cam and microswitch.

A fourth cardiac phantom has recently been advertised§ and, although we have only been able to judge this latter device from the sparse advertising material available, it would appear that no cardiac phantom as yet fulfills all of the criteria for an ideal phantom. This is not surprising since the heart is an extremely difficult organ to model both physically and electrically. The latter is important because, in performing nuclear medicine studies of the heart, we are dependent on the electrical signals for external triggering coincident with a mechanical condition. The performance of such a complicated organ will indeed be difficult to reproduce using a simple, inexpensive phantom that is practical to use for routine quality control purposes. Nevertheless, for routine quality control of gated cardiac studies which depend on the evaluation of ventricular ejection fraction, the phantoms described fulfill some of the requirements.

FOOTNOTES

*Amersham Corp., Capintec Inc., U.S.
†Veenstra, The Netherlands
‡ADC Medical, U.S.
§Anzai Sogyo Co. Ltd., Japan

ACKNOWLEDGMENTS

This work was supported in part by grants from the Education and Research Foundation, Society of Nuclear Medicine, New York, New York, and the Physicians' Services Incorporated Foundation, Toronto, Ontario. We are also indebted to Veenstra, The Netherlands and ADC Medical, U.S., who so

willingly provided models of their phantoms for purposes of this study.

REFERENCES

1. Nickles RJ. A dynamic phantom for radionuclide cardiology. *J Nucl Med* 1979;20:547–49.

2. Hurst RR, Logan KW, Holmes RA. A simulated heart model using computer image analysis for validating cardiovascular nuclear medicine parameters. In *Nuclear Cardiology: Selected Computer Aspects,* Sorenson JA, ed, New York: Society of Nuclear Medicine, 1978:55–62.

3. Shulz E, Adams R, Aamodt L, et al. A precision pump for simulated cardiographic studies. *J Nucl Med* 1981;22:643–44.

4. Kan MK, Hopkins GB. Inexpensive ventricular phantom for ECG-gated equilibrium studies. *J Nucl Med* 1980;21:880–82.

5. Bennett GW, Brill AB, Fairchild R, et al. A general purpose dynamic phantom for gated, computer-aided gamma camera evaluation. In *Medical Radionuclide Imaging,* Vol. 2, Vienna: International Atomic Energy Agency, 1981:35–46.

6. Hudson FR, Davis JB, Mott GT. A phantom for simulating fast dynamic studies on a gamma camera. *Phys Med Biol* 1977;22:87–89.

7. Price RR, Rollo FD, Steckley R, et al. A new quality control phantom for gated acquisition systems. *J Nucl Med* 1980;21:P88.

8. Rollo FD, Price RR, Patton JA, et al. A dynamic cardiac phantom for quality control of gated acquisition systems. In *Medical Radionuclide Imaging,* Vol. 2, Vienna: International Atomic Energy Agency, 1981:31–32.

9. Reiber JHC, Lie SP, Simoons ML, et al. Clinical validation of fully automated computation of ejection fraction from gated equilibrium blood-pool scintigrams. *J Nucl Med* 1983;24:1099–1107.

10. Stern D, Research Systems, Inc., Denver, CO, April, 1981, Personal communication.

11. Gupta S, Herrera N, Davies T et al. A new phantom for evaluation of multigated radionuclide studies. *J Nucl Med Tech* 1983;11:165–68.

12. Wery R, Hill J, Dworkin HJ. A method for the determination of ECG gate signal delays. *J Nucl Med* 1981;22:546–48.

13. Erickson JJ, Rollo FD. Quality control of nuclear medicine computer systems. In *Digital Nuclear Medicine.* Erickson JJ, Rollo FD, eds, Philadelphia: JP Lippincott, 1983:218–36.