Calculation of the Left Ventricular Ejection Fraction Without Edge Detection: Application to Small Hearts

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Quantitative gated SPECT (QGS) software has been reported to overestimate the left ventricular ejection fraction (*LVEF*) in patients with small hearts. This finding is caused by the inaccurate detection of the endocardial surface of the left ventricle (LV) due to low resolution and partial-volume effects. In this article we develop a method to calculate the *LVEF* from gated SPECT data without edge detection and compare it with the QGS method of calculating the *LVEF*. **Methods:** The short-axis images were transformed to the prolate spheroid coordinate system, and detection of the layer of maximum counts (a surface area of maximum counts) was made. First, the volume enclosed by the layer of maximum counts (V_{max}) was calculated; then the corresponding ejection fraction [$(LVEF)_{max}$] was calculated. The *LVEF* was calculated by multiplying the $(LVEF)_{max}$ by a constant factor, which was determined from a series of calculations made using QGS on larger hearts. In computer simulations the end-diastolic left ventricular volume (EDV) and the targeted LVEF (*tLVEF*) were varied to produce LVs of different sizes. The LVs were modeled by 2 confocal hemiellipsoids with 7 different EDVs. The *tLVEF* was increased from 25% to 75%, in 5% step-size increments, for a total of 11 different ejection fractions. These datasets were then smoothed, creating a total of 77 smoothed sets. The smoothed images were processed by the QGS method and by our method. In patient studies, 58 patient datasets were processed by the QGS method and by our method. No attenuation correction was performed on these datasets. The patients were divided into 2 groups: 44 patients with large hearts (EDV $\geq$ 80 mL) and 14 patients with small hearts (EDV $<$ 80 mL). **Results:** In computer simulations, the QGS method and our method performed well when imaging large EDVs (EDV $\geq$ 80 mL). Our method derived better results than did the QGS method for small EDVs. In patient studies the *LVEF* calculated by our method matched well with the QGS *LVEF* in the 44 patients with large hearts. The correlation coefficient between them was found to be 0.957. Of the 14 patients with small hearts, the *LVEFs* of 5 patients were severely overestimated by the QGS method compared with the results obtained with our method. **Conclusion:** It is possible to calculate the *LVEF* without edge detection. Compared with QGS *LVEF*, our method gave better results for small LVs in computer simulations.

Key Words: gated SPECT; left ventricular ejection fraction; small hearts

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Gated SPECT (gSPECT) offers the possibility of simultaneous measurement of heart perfusion and myocardial function. Values of the left ventricular ejection fraction (*LVEF*), heart wall thickening, and heart wall motion are important factors in the diagnosis of coronary artery disease and the prognosis of patient recovery. Also, the determination of the *LVEF* using nuclear ventriculography is an important tool for diagnosing dysfunction of the left ventricle (LV), which is related to several diseases of the myocardium. A particular problem in using gSPECT is the low spatial resolution that results in overestimation of the *LVEF* in small hearts. Most methods implemented on commercial SPECT systems use edge-detection schemes in the calculation of the *LVEF*. However, edge-detection schemes are especially susceptible to partial-volume effects, which become more severe when imaging small hearts. The calculation of the *LVEF* in small hearts can be improved using methods, such as the one proposed in this article, that are less sensitive to partial-volume effects and enable calculation of the volume (V_{max}) enclosed by the layer of maximum counts (a surface of maximum counts).

gSPECT using ^{99m}Tc-labeled pharmaceuticals (*1–4*) and ^{201}Tl (*4,5*) has been used to estimate functional parameters for quite some time. However, because of the very low resolution of gSPECT, which is comparable to the thickness of the heart wall, accurate measurement of parameters such as the ejection fraction is not possible. To calculate functional parameters such as the *LVEF*, heart wall thickening, and heart wall motion from gSPECT, edges of the heart are determined using the current commercial quantitative gSPECT software package ([QGS]; Cedars-Sinai Medical Center, Los Angeles, CA) (*6,7*). Then, on the basis of the position of the edges, all functional parameters are determined. Because of the low spatial resolution of approximately 15 mm, the endocardial edges at the opposite sides

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of the LV cavity overlap. As a result the position of the calculated edge appears to be closer to the center of the cavity. (The temporal resolution has less of an effect because most gSPECT studies are digitized into 16 time frames of approximately 0.05 s, depending on the heart rate.) This is particularly a problem for end-systolic configuration because the volume of the LV is at its smallest and the endocardial edges are at their closest points. This is most problematic in low-resolution images and images of patients with small hearts. Incorrect determination of the position of the edges results in underestimation of the end-systolic volume (ESV), which results in overestimation of the *LVEF* (*8–10*). In this article, this effect is referred to as small heart error.

Instead of calculating endocardial edges, we propose a method for calculation of the *LVEF* based on the calculation of the layer of maximum counts. The layer of maximum counts is defined as follows: If one determines the maximum counts along each radius in the short-axis slices, a circumferential curve that inscribes an area that includes a cross section of the intraventricular blood pool and an internal portion of the LV wall is derived. If we do this for each short-axis slice from its apex to its base, a circumferential surface is defined. This surface is referred to as the layer of maximum counts.

In this article, a technique that involves determining the position of the layer of maximum counts surface and using it to calculate the *LVEF* is developed and implemented. The technique does not require detection of edges for the calculation of functional parameters such as the *LVEF*. The technique suffers less from resolution effects because in short-axis slices there is a greater distance between the opposing walls of the layer of maximum counts of the LV than there is between the opposing walls of the endocardial surface. The layer of maximum counts is usually closer to the midventricular surface than to the endocardial surface.

MATERIALS AND METHODS

Calculation of V_{max}

To calculate the V_{max}, the short-axis image of LV was transformed into the prolate spheroid coordinate system (Fig. 1) (*11*). In the prolate spheroid coordinate system the horizontal ξ-axis direction lies along the transmural direction, so it goes "through" the LV wall. The φ-axis (φ-axis, Fig. 1) is the circumferential angle that ranges from 0 to 2π and the θ-axis is the azimuthal angle that ranges from 0 to π. In each short-axis image the location of the maximum counts $\xi_{max}(\theta,\varphi)$ was searched for along the ξ-axis for fixed coordinates θ and φ in the prolate spheroid coordinate system. Because the heart is not entirely ellipsoid, but rather is truncated at some θ_{max}, θ_{max} is dependent on the value of φ. The value $\theta_{max}(\varphi)$ was found by thresholding the image at 50% (*11*).

The V_{max} is calculated by:

$$V_{max} = \iiint J d\varphi d\theta d\xi$$

$$= \int_0^{2\pi} d\varphi \int_0^{\theta_{max}(\varphi)} d\theta \int_0^{\xi_{max}(\theta,\varphi)} d\xi [C^3(\sinh\xi)^3(\sin\theta)(\cos\theta)^2 + C^3(\cosh\xi)^2(\sinh\xi)(\sin\theta)^3], \quad \text{Eq. 1}$$

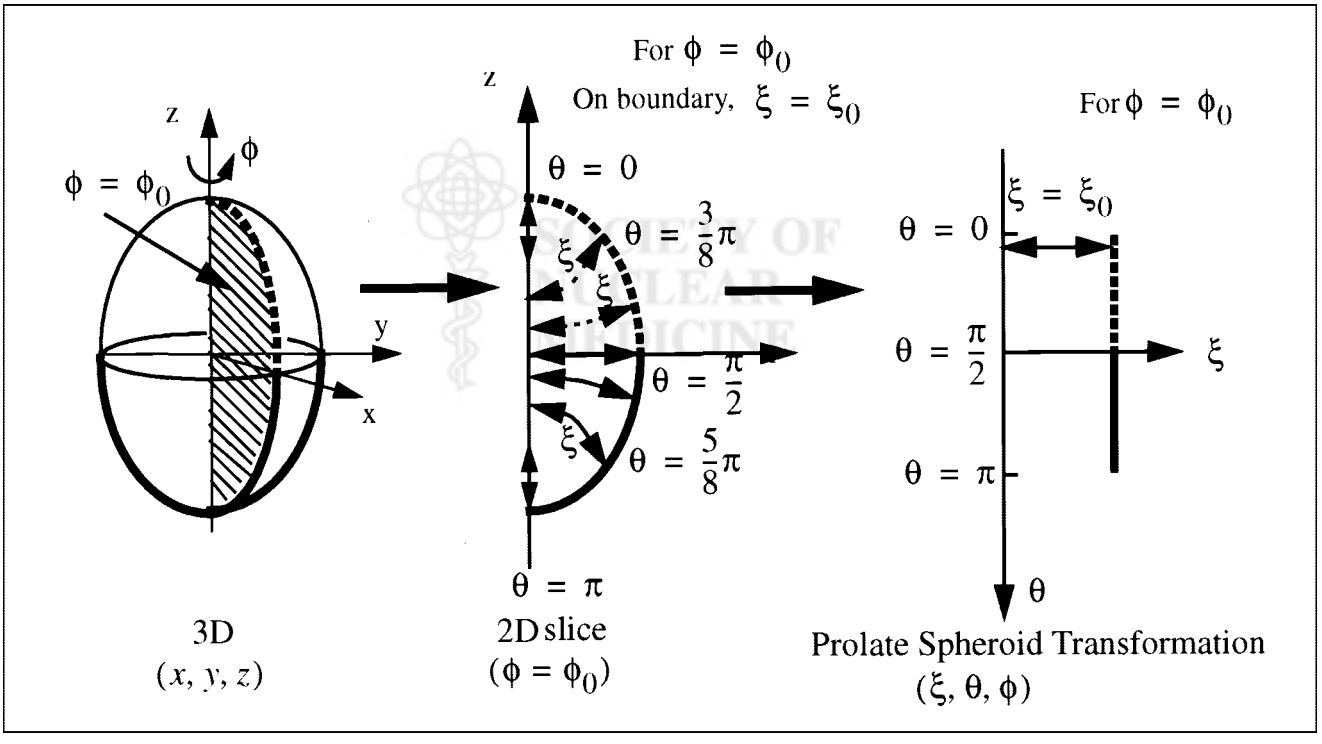

FIGURE 1. Prolate spheroid transformation of ellipsoid. (Left) A 3D ellipsoid in Cartesian coordinate system. (Center) Half of 2D vertical slice through ellipsoid. (Right) Prolate spheroid transformation of 2D slice in Center.

where J is the Jacobian of the prolate spheroid transformation. The focal length C is a parameter related to the prolate spheroid transformation. It was found that the calculated volume V_{max} is not sensitive to the value of C within the range of physiologic heart sizes (*11*).

Estimation of LV Cavity Volume and *LVEF*

Assuming that the layer of maximum counts is located at a fixed position within the wall of the LV myocardium, due to certain smoothing processes, the LV cavity volume V_c is estimated by calculating the V_{max} and then subtracting the volume of the myocardium enclosed by the layer of maximum counts. The part of the LV myocardium that is enclosed by the layer of maximum counts has a volume of V_{myo-en}, whereas the entire volume of the LV myocardium is denoted as V_{myo}. The ratio $\alpha = V_{myo-en}/V_{myo}$ is assumed to be constant for each particular image smoothing process. It follows that:

$$V_{max} = V_c + V_{myo-en} = (1 + \alpha\beta)V_c, \qquad \text{Eq. 2}$$

where $\beta = V_{myo}/V_c$.

The *LVEF* is defined by:

$$LVEF = \frac{(V_c)_{ED} - (V_c)_{ES}}{(V_c)_{ED}}, \qquad \text{Eq. 3}$$

where the subindices ED and ES stand for quantities measured at end diastole and end systole. Using Equation 2 we can rewrite Equation 3 as:

$$LVEF = \frac{(V_{max})_{ED} - (V_{myo-en})_{ED} - [(V_{max})_{ES} - (V_{myo-en})_{ES}]}{(V_c)_{ED}} \qquad \text{Eq. 4}$$

or

$$LVEF = \frac{[(V_{max})_{ED} - (V_{max})_{ES}] - [(V_{myo-en})_{ED} - (V_{myo-en})_{ES}]}{(V_c)_{ED}}. \qquad \text{Eq. 5}$$

The myocardium is assumed to be incompressible because the tissue is incompressible and the tissue volume changes caused by changes in blood volume within the myocardium are negligible factors. This means that it is assumed that the volume of the LV enclosed by the layer of maximum counts is equal for all time gates between end systole and end diastole, and in particular $(V_{myo-en})_{ED} = (V_{myo-en})_{ES}$, so that the difference of these 2 terms in Equation 5 is 0. Using this fact, Equation 5 can be rewritten as:

$$LVEF = \left[\frac{(V_{max})_{ED} - (V_{max})_{ES}}{(V_{max})_{ED}}\right]\left[\frac{(V_{max})_{ED}}{(V_c)_{ED}}\right] = (LVEF)_{max}[1 + \alpha\beta_{ED}]. \qquad \text{Eq. 6}$$

Equation 2 is used to obtain the expression $[1 + \alpha\beta_{ED}]$, where $\beta_{ED} = (V_{myo})_{ED}/(V_c)_{ED}$ and α is assumed to be constant throughout the cardiac cycle. The $(LVEF)_{max}$ is calculated directly from the volumes enclosed by the layer of maximum counts:

$$(LVEF)_{max} = \frac{(V_{max})_{ED} - (V_{max})_{ES}}{(V_{max})_{ED}}. \qquad \text{Eq. 7}$$

Our approach to calculating the *LVEF* is as follows: First the $(LVEF)_{max}$ is calculated and then modified by the factor $1 + \alpha\beta_{ED}$ (Eq. 6). Coefficients α and β_{ED} are known factors in the computer simulations. However, in patient studies they had to be determined. As pointed out earlier, α depends on the imaging method (data acquisition, reconstruction, smoothing, and so forth). For gSPECT images reconstructed with the same software, the reconstructed images were filtered and smoothed by similar processes, so it is reasonable to assume that α remains constant for all patients imaged with the same device. The coefficient β_{ED} stands for the volume ratio of the LV myocardium and the LV cavity at end diastole, which is in a narrow physiologic range. Plotting the *LVEF* found by another source (such as QGS with large hearts) versus the $(LVEF)_{max}$, the slope can be found, which should be the value of $1 + \alpha\beta_{ED}$ for a given imaging system. This same slope should also apply for small hearts as well.

Computer Simulations

Four sets of digital phantoms with different ratios of $V_{myo}/V_c = \beta$ were used to verify Equation 2 and determine the value of α. All of the phantoms consisted of 2 confocal hemiellipsoids with uniform LV myocardial counts. Each set had a fixed β, whereas the targeted cavity volume was increased from 30 to 150 mL, at 15-mL step-size increments. The pixel size was 0.532 cm, which is the common pixel size used in the PRISM 3000XP SPECT system (Marconi Medical Systems, Cleveland, OH). Each phantom was smoothed by applying the 3-dimensional (3D) analog of the familiar "S9" 2-dimensional (2D) low-pass filter (27 total voxels for the real space kernel):

$$K_{ijk} = \begin{matrix} 1 & 2 & 1 \\ 2 & 4 & 2 \\ 1 & 2 & 1 \end{matrix} \quad \begin{matrix} 2 & 4 & 2 \\ 4 & 8 & 4 \\ 2 & 4 & 2 \end{matrix} \quad \begin{matrix} 1 & 2 & 1 \\ 2 & 4 & 2 \\ 1 & 2 & 1 \end{matrix}. \qquad \text{Eq. 8}$$

For this particular filter the cutoff frequency (the point at which the filter is at one half the maximum power) is 0.195 cycle/pixel, or 0.3665 cycle/cm. V_{max} was then calculated and plotted versus the targeted V_c.

Ford et al. (*10*) developed a series of mathematic LVs by varying the end diastole, the LV end-diastolic volume (EDV), and the targeted *LVEF*. We developed similar phantoms to evaluate our method of calculating the *LVEF* and compared our results with the QGS results. As before, the LV was modeled as 2 confocal hemiellipsoids with uniform myocardial counts. A constant ratio was maintained between the axes of the inner hemiellipsoid during the cardiac cycle. The volume of the myocardium was kept constant during the cardiac cycle. The EDV was set to equal the volume of the myocardium ($\beta_{ED} = 1$, which corresponded to one of the heart sizes of the Mathematical CArdiac Torso phantoms). The targeted *LVEF* (*tLVEF*) was chosen such that it would change from 25% to 75% in 5% increments to derive a total of 11 *tLVEFs*. For each *tLVEF*, 8 frames of the cardiac cycle were generated using the formula (*10*) $V_i = EDV[tLVEF\ (\cos(i\pi/4) - 1)/2 + 1]$, where i ranged from 0 to 7. The 3D kernel in Equation 8 was then used to smooth these datasets. There were 7 sets of targeted EDVs (before smoothing): 20, 40, 60, 80, 100, 140, and 180 mL Therefore, the total number of datasets was 77. Each dataset was $64 \times 64 \times 25 \times 8$.

The *LVEF* was calculated using our method and using the QGS method on the Marconi Odyssey V4.0D/3.2 computer (Marconi Medical Systems). In calculations of the *LVEFs* using our method, α and β_{ED} were known from the digital phantom used in the simulation: $\alpha = 0.427$ and $\beta_{ED} = 1.0$, so the $LVEF = 1.427(LVEF)_{max}$. An appropriate Marconi header was added to these datasets before QGS was used to determine the *LVEFs* for these datasets.

Patient Studies

Fifty-eight sets of gSPECT cardiac patient data were used to study the *LVEFs* obtained by QGS and by our new method. The patient population included 29 male and 29 female subjects. Most of the patients were referred for the gSPECT study for evaluation of potential ischemic cardiac disease. The findings for most of the patients showed that there was no evidence of ischemia. There were 6 cases with abnormal results: 1 with apical ischemia with associated dyskinesia, 1 with 2 large areas of ischemia involving the mid-to-distal anterior wall and the lateral wall, 1 with a moderate-sized infarct of the posterolateral wall, 1 with an inferior wall infarction with minimal periinfarct ischemia in the proximal inferior wall, 1 with a mildly dilated LV with mild diffuse hypokinesia, and 1 with a dilated LV showing apical ischemia with global hypokinesia.

The cardiac gSPECT data were acquired as follows: Each patient was injected with approximately 888 MBq ^{99m}Tc-methoxyisobutylisonitrile. For 23 patients, stress was induced on a treadmill using the standard Bruce protocol. For the other 35 patients, stress was induced pharmacologically. Of these, 13 were stressed using adenosine, 9 were stressed using dipyridamole, 12 were stressed using dobutamine, and 1 was stressed using atropine. One hundred twenty projections over 360° were acquired using a 3-head PRISM 3000XP SPECT system (Marconi Medical Systems) with low-energy, high-resolution, parallel collimators. For each projection angle, the data were acquired and digitized into sixteen 64×64 frames over the cardiac cycle. The acquisition was zoomed so that the pixel size was 0.532 cm.

The gSPECT data were reconstructed using the filtered backprojection algorithm with a 1-dimensional ramp filter. A 3D low-pass filter (Butterworth filter: order, 5; cutoff frequency, 0.17–0.21 cycle/pixel) was applied after the projections were reconstructed. No attenuation corrections were made when reconstructing these datasets.

The QGS program was then used to process the reconstructed data. Also, to enable use of our method, the maximum *LVEF* [$(LVEF)_{max}$] was calculated. The threshold used in our calculations to determine the θ_{max} (φ) needed for the volume calculations in Equation 1 was one half of the maximum myocardial counts. By cutting out the short-axis LV within a 13-pixel-radius cylinder, we eliminated the potential high counts that may have occurred outside the LV. The focal length C used in the prolate spheroid transformation was 6 pixels for all patient studies.

The patients were divided into 2 groups. The first group consisted of 44 patients with normal or large-sized hearts ($EDV \geq 80$ mL). On the basis of this group, the value of $1 + \alpha\beta_{ED}$ was calculated by linear regression. The second group consisted of 14 patients with small hearts ($EDV < 80$ mL).

RESULTS

Computer Simulations

In the simulation performed to verify Equation 2, V_{max} was calculated and plotted versus the targeted V_c for 4 sets of phantoms with different ratios of $V_{myo}/V_c = \beta$ (Fig. 2). Each set was fitted to a line through the origin ($y = kx$). The correlation coefficient was >0.99 for each set. From the slope of the fitted line, the value of α was calculated by $\alpha = (\text{slope} - 1)/\beta$. For $\beta = 0.8$, 1.0, 1.5, and 3.0, the calculated values of α were 0.443, 0.427, 0.429, and 0.401, respectively. The error for α was <10%.

These results verify that the relationship between V_{max} and V_c is indeed a straight line that passes through zero as predicted by Equation 2. Also, if we assume that the slope of the line is related to $[1 + \alpha\beta_{ED}]$, the results show that α (V_{myo-en}/V_{myo}) remains fairly constant for different heart sizes and shapes (i.e., different values of β). This result is crucial to the derivation of the result given in Equation 6, which indicates that the layer of maximum counts occurs at almost a fixed depth in the myocardium during the cardiac

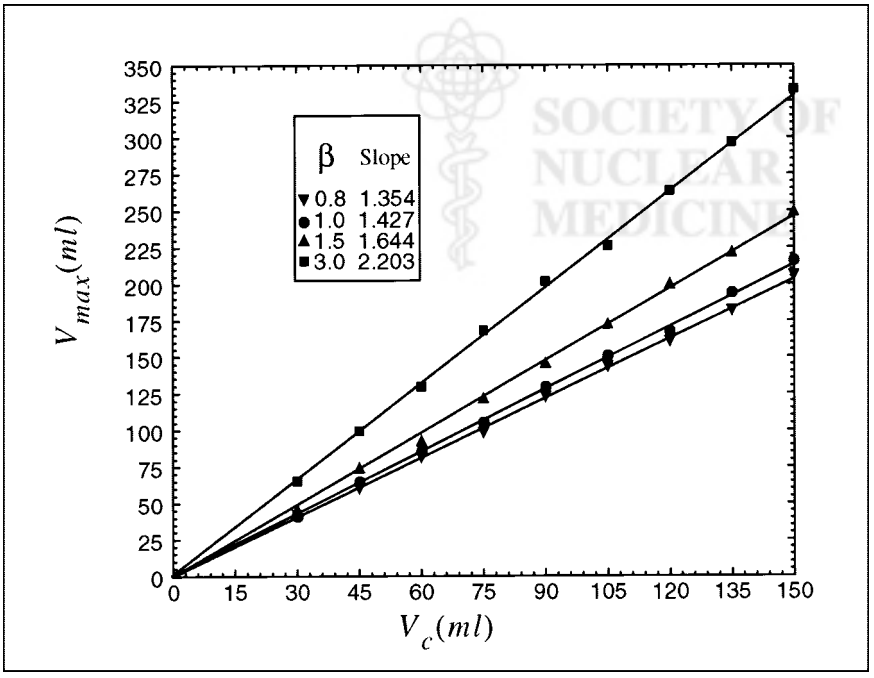

FIGURE 2. Plot of calculated volume V_{max}, volume enclosed by layer of maximum counts, vs. targeted LV cavity volume V_c for 4 sets of phantoms with different ratios between volumes of LV myocardium and volumes of LV cavity ($V_{myo}/V_c = \beta$).

cycle. The constant value of α makes it possible to estimate the *LVEF* from V_{max} in Equation 2 and the $(LVEF)_{max}$ in Equation 7 even though the value of β changes during a cardiac cycle.

In the simulations performed to calculate the *LVEF*, 77 smoothed datasets were processed by the QGS method and by our method. For large phantoms ($EDV \geq 80$ mL), the QGS *LVEF* and the *LVEF* calculated by our method closely approximated the *tLVEF*. Errors in the QGS *LVEF* increased as the cardiac volume decreased ($EDV \leq 80$ mL) (Fig. 3A; Fig. 4A). Conversely, the *LVEF* obtained with our method was closer to the *tLVEF* than the QGS *LVEF* was in the studies of small hearts (Fig. 3B; Fig. 4B).

To determine how well the calculated *LVEFs* matched the *tLVEF* for each EDV, we calculated the slope of the linear least-squares fit to the plot of the calculated *LVEFs* versus the *tLVEFs*. A linear least-squares fit was performed for each dataset in Figures 3A and 3B using XMGR (copyright 1996–1998; ACE/gr Development Team; http://plasma-gate.weizmann.ac.il/Xmgr/). To achieve perfect agreement between the calculated *LVEFs* and the *tLVEFs*, the slope of the fitted line should be unity. If there is a large discrepancy between the fitted slope and unity, the calculated *LVEFs* will lack agreement with the *tLVEFs*. For EDVs = 20, 40, 60, 80, 100, 140, and 180 mL, the slopes for the QGS results were 1.289, 1.255, 1.166, 1.055, 1.018, 0.9823, and 0.9528, whereas the slopes for our results were 1.085, 1.096, 1.040, 1.078, 1.062, 1.023, and 1.046. Our method gives better results when imaging small hearts ($EDV < 80$ mL).

Patient Studies

Values of the *LVEFs* from 58 patients were calculated using the QGS method and our method. Figure 5 shows the QGS *LVEFs* of the 44 patients with large hearts versus the $(LVEF)_{max}$ calculated by our method. The linear coefficient $1 + \alpha\beta_{ED}$ of the least-squares fit was 1.226. Using Equation 6, we arrived at $\alpha\beta_{ED} = 0.226$. We calculated the difference between the *LVEFs* obtained using our method and the QGS *LVEFs* and plotted the histogram in Figure 6.

For the 14 patients with small hearts ($EDV < 80$ mL), the *LVEFs* obtained using our method were plotted versus the QGS *LVEFs* in Figure 7. There were 5 patient studies in which the *LVEFs* calculated by QGS were determined to be overestimated by >9% compared with the *LVEFs* obtained using our method. The EDVs for these patients were 62, 52, 69, 59, and 49 mL. For the patients with small hearts, the average heart size $[(V_c)_{ED}]$ was 64 ± 11 mL. Therefore, these 5 patients were not necessarily the ones with the smallest hearts, but their hearts did tend to be of the smaller sizes and had ejection fractions of >70%. Larger hearts with higher ejection fractions tended to fit closer to the fitted regression line.

DISCUSSION

A method for calculating the *LVEF* without calculating the edges of the endocardial wall of the LV in gSPECT images was developed. The method improves the accuracy of calculating functional parameters from gSPECT in patients with small hearts. The method requires calculating the regression slope of V_{max} versus V_c for large hearts using software such as the QGS software package. The slope of this regression curve was used to develop a formulation for the calculation of the *LVEFs* of smaller hearts, using the layer of maximum counts. Compared with QGS, our method gave better results for small LVs in computer simulations. In patient studies, QGS overestimated the *LVEFs* in small hearts by >9% compared with those obtained with our method. This study concentrated on the calculation of the *LVEF* but the extension of this method to calculating

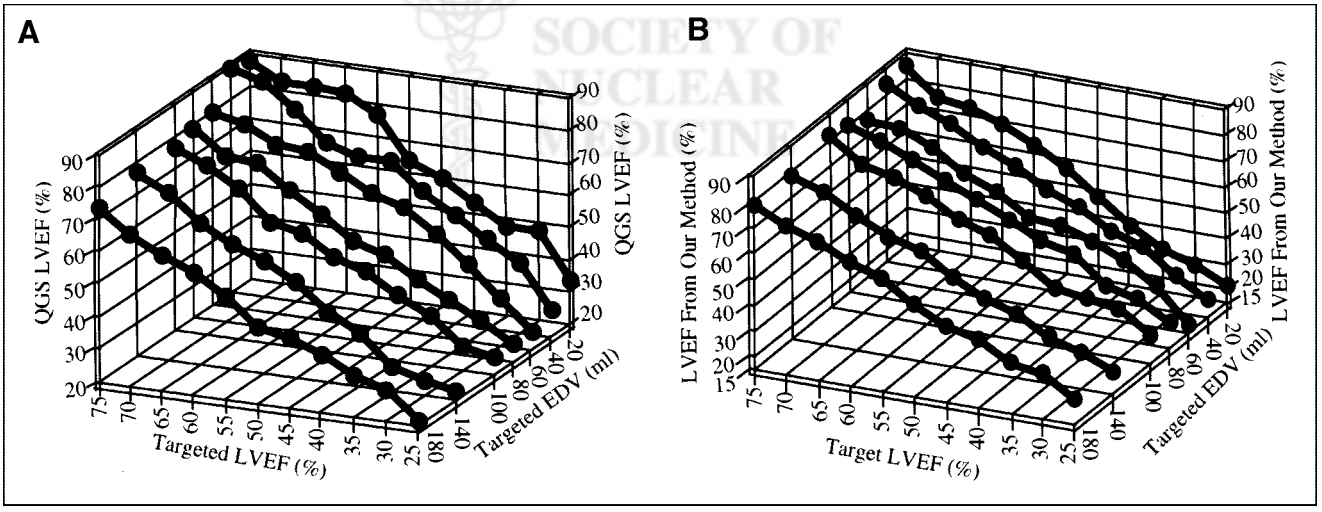

FIGURE 3. Plots of QGS *LVEF* (A) and *LVEF* by our method (B) vs. targeted *LVEF* (*tLVEF*) and targeted end-diastolic volume (EDV). For phantoms with large EDVs ($EDV \geq 80$ mL), QGS *LVEF* and *LVEF* by our method closely approximate *tLVEF*. However, for small hearts ($EDV < 80$ mL), our method produced better results than did QGS method.

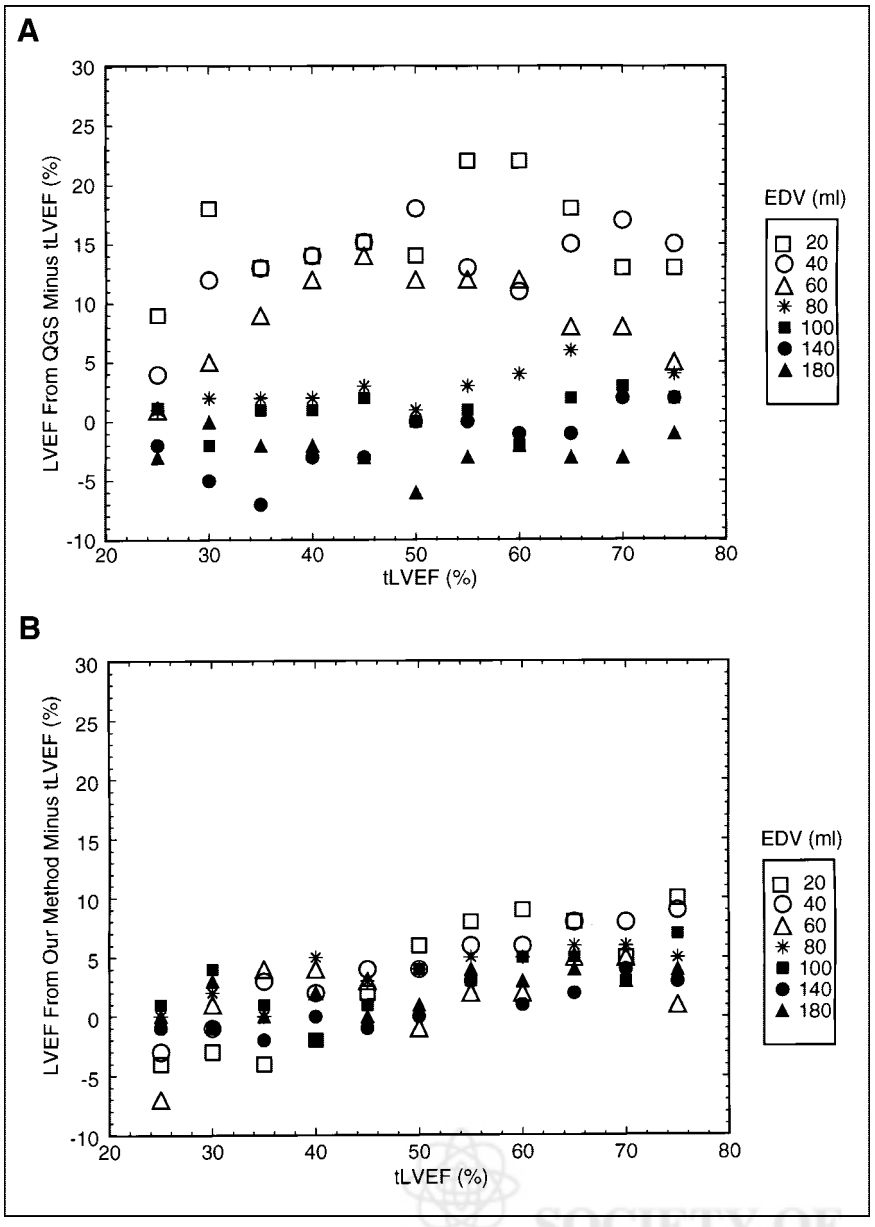

FIGURE 4. (A) Plot of (difference between QGS *LVEF* and targeted *LVEF* [*tLVEF*]) vs. *tLVEF*. (B) Plot of (difference between *LVEF* calculated by our method and *tLVEF*) vs. *tLVEF*.

other parameters such as heart wall thickness would not be difficult.

Our method is based on 2 assumptions. The first assumption is that the layer of maximum counts remains at a fixed position in the LV myocardium for 1 imaging system during the heart cycle. Using the computer simulation results presented in this work, the ratio between the volume enclosed by the layer of maximum counts and the volume of the LV myocardium (α) is constant when the same smoothing filter is applied to each phantom. This is true for any heart size (Fig. 2), which means that α can be treated as a constant in any single imaging system in which each patient dataset is processed using the same reconstruction method. The second assumption is that the volumetric ratio between the LV myocardium and the cavity at end diastole is within a narrow physiologic range for all patients. The strong linear agreement and close correlation ($r = 0.957$) between the *LVEF* calculated by our method and that calculated by the QGS method (Figs. 5 and 6) proves that this is a good assumption for the subset of patients studied in this investigation.

If the second assumption is invalid, such as in cases of systolic or diastolic dysfunction, our method will still result in relatively small errors. For normal hearts we obtained $\alpha\beta_{ED} = 0.226$, where β_{ED} is the ratio between the myocardium volume and the EDV. For a heart with disease (systolic or diastolic dysfunction), β_{ED} may change significantly from normal values, but the calculation of the *LVEF* using Equation 6 will be affected to a much less extent because α is <1 (0.226 if we assume $\beta_{ED} = 1$). For example, for a heart with a systolic dysfunction such as dilated cardiomyopathy (*LVEF* de-

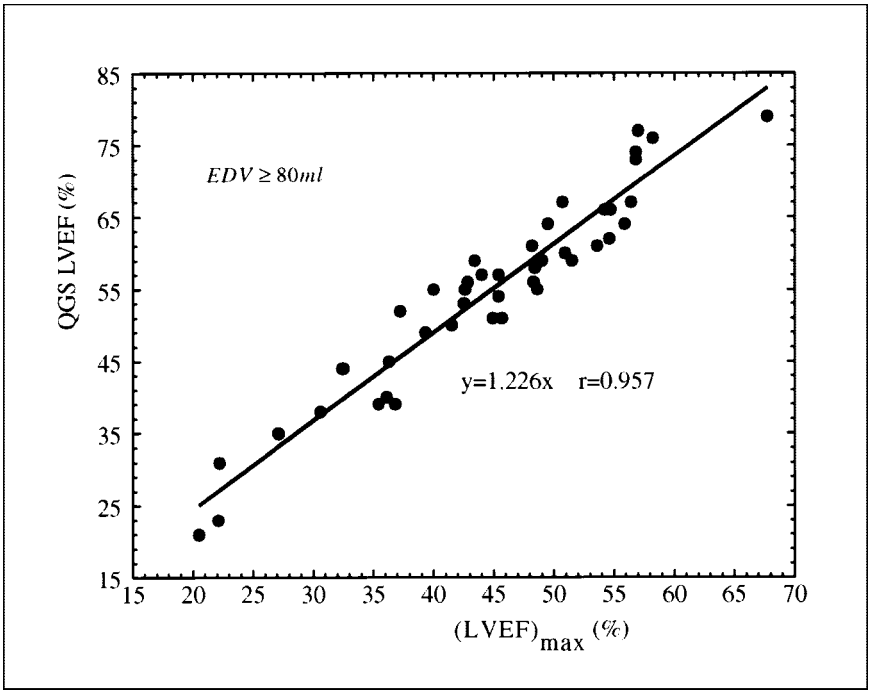

FIGURE 5. For 44 patients with large EDVs (EDV ≥ 80 mL), plots of QGS *LVEF* vs. $(LVEF)_{max}$ by our method. Slope of fitted line is used to calculate *LVEF*.

creases, LV EDV increases), β_{ED} can decrease to half of its normal value because of the increased EDV. In this case the value for $1 + \alpha\beta_{ED}$ could be 1.113 instead of 1.226. The relative error of the *LVEF* caused using $1 + \alpha\beta_{ED} = 1.226$ is 10%. On the other hand, for a heart with diastolic dysfunction such as hypertrophic cardiomyopathy or restrictive cardiomyopathy (*LVEF* increases, LV EDV decreases), β_{ED} can increase to twice its normal value because of the decrease of the EDV or the increase of the volume of the myocardium. In this case the value for $1 + \alpha\beta_{ED}$ could be 1.452 instead of 1.226. The relative error of the *LVEF* caused using $1 + \alpha\beta_{ED} = 1.226$ is −16%.

In images obtained with gSPECT, the edges of the LV are blurred by the smoothing process when very noisy data are reconstructed, which results in the small heart error described earlier. Contrary to the conventional segmentation method used in the *LVEF* calculations, our method detects the layer of maximum counts of the LV myocardium, which is less affected by the small heart error. The position of the layer of maximum counts is basically not affected by this error. We found that if the

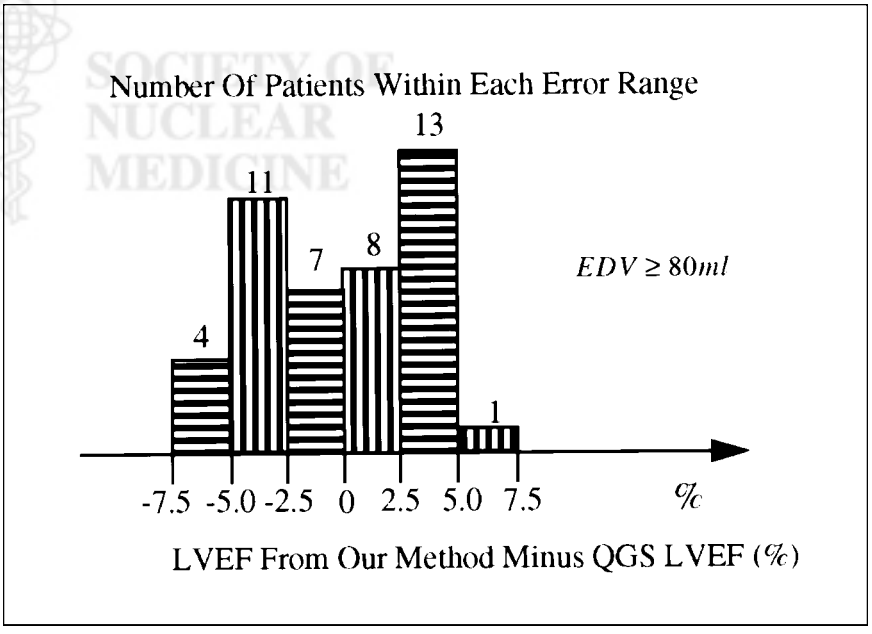

FIGURE 6. Histogram shows difference between *LVEFs* obtained with our method and those obtained with QGS *LVEF* method for 44 patients with large hearts (EDV ≥ 80 mL).

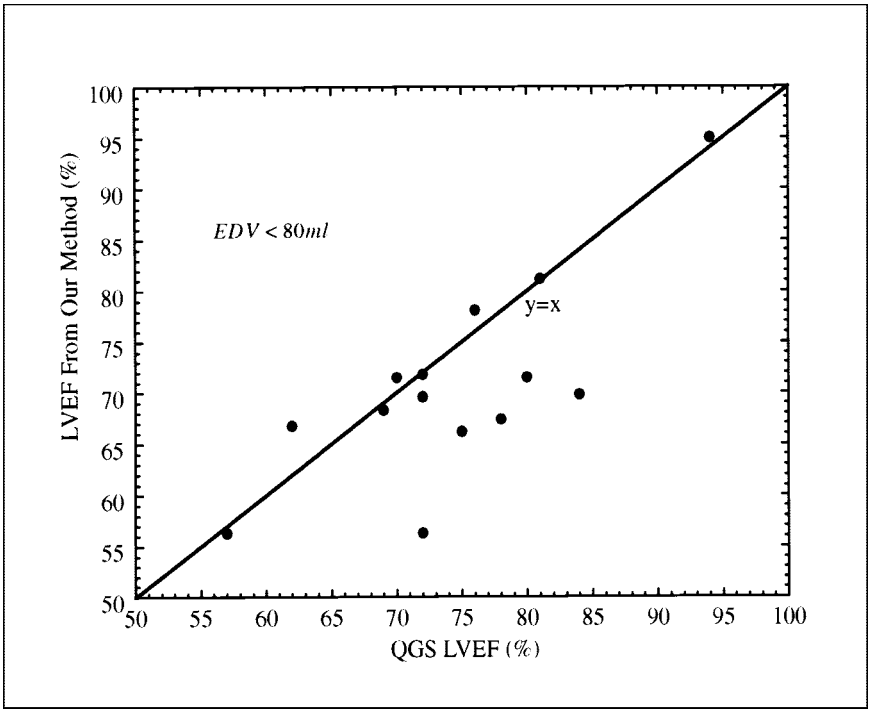

FIGURE 7. For 14 patients with small EDVs (EDV < 80 mL), plots show *LVEFs* by our method vs. QGS *LVEFs*. Five patients had severely overestimated *LVEFs* (>9%) obtained by QGS *LVEF* method compared with *LVEFs* obtained by our method.

EDV > 80 mL the measurements of the calculated *LVEF* were not affected by the small heart error. On the other hand, when the positions of the edges were used to determine the ejection fraction (QGS), the accuracy dropped rapidly when small hearts were simulated. This is in agreement with previous work (*10*).

It is anticipated that in severe perfusion defects it will be easier to calculate the area of the layer of maximum counts than it is to determine edges that are missing, which are needed in standard software for calculating the *LVEF*. We point out that this is conjecture and still needs to be verified. For normal hearts, the area of the layer of maximum counts is as easy to calculate as the enclosed volume. We have shown in computer simulations that the area correlates perfectly in a linear fashion with the volume (*11*). With abnormalities in the LV some interpolations must be done to define the LV cavity in standard methods. With area calculations this is not necessary because it is implicitly assumed that the location of the maximum counts in the defect will be the same as those where the tissue is normal. It is expected that this will not be true in all cases, but it is expected to hold true in most cases. However, this needs to be verified for defects of various sizes and at various locations. This might be very important in the case of hearts with large defects where accurate estimation of the *LVEF* is very difficult to obtain with standard methods. Therefore, when using areas it is expected that the accuracy of the *LVEF* measurements will be better in cases of severe abnormalities, if not at least as good as volume calculations of normal hearts using conventional software. For the same reason, we expect this method to work better in cases of dyskinesia.

CONCLUSION

We have presented a method to calculate the *LVEF* without edge detection. Compared with QGS *LVEF*, our method gave better results for small LVs in computer simulations. In patient studies, our method gave results similar to those of the QGS method when imaging patients with large hearts. On the basis of our computer simulations, one can infer that our method gave more accurate measurements of the *LVEF* when imaging patients with small hearts, whereas the QGS method overestimated the *LVEF*.

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