Volumetric Quantification of Atherosclerotic Plaque in CT Considering Partial Volume Effect

Jamshid Dehmeshki*, Member, IEEE, Xujiong Ye, Hamdan Amin, Maryam Abaei, XinYu Lin, and Salah D. Qanadli

Abstract—Coronary artery calcification (CAC) is quantified based on a computed tomography (CT) scan image. A calcified region is identified. Modified expectation maximization (MEM) of a statistical model for the calcified and background material is used to estimate the partial calcium content of the voxels. The algorithm limits the region over which MEM is performed. By using MEM, the statistical properties of the model are iteratively updated based on the calculated resultant calcium distribution from the previous iteration. The estimated statistical properties are used to generate a map of the partial calcium content in the calcified region. The volume of calcium in the calcified region is determined based on the map. The experimental results on a cardiac phantom, scanned 90 times using 15 different protocols, demonstrate that the proposed method is less sensitive to partial volume effect and noise, with average error of 9.5% (standard deviation (SD) of 5–7 mm³) compared with 67% (SD of 3–20 mm³) for conventional techniques. The high reproducibility of the proposed method for 35 patients, scanned twice using the same protocol at a minimum interval of 10 min, shows that the method provides 2–3 times lower interscan variation than conventional techniques.

Index Terms—Coronary artery calcification, expectation-maximization, partial volume effect, proportion map, volume measurement.

I. INTRODUCTION

CORONARY artery disease (CAD) currently is the leading cause of death in humans [1], [2]. Calcification of the coronary vessel wall is regarded as a marker of advanced coronary atherosclerosis [3], [4]. Early identification of CAD in patients can reduce morbidity and/or mortality. One marker for CAD is coronary artery calcification (CAC). The presence of CAC indicates underlying CAD. Recent experimental investigations have suggested that calcifications in atherosclerotic lesions should be considered as an active process [5]. Increasing coronary calcifications indicates CAD progression. Evidence suggests that the calcium score has a significant predictive value for subsequent cardiac events in both symptomatic and asymptomatic patients [6], [7]. The amount of calcification correlates with the amount of plaque present [8]. Therefore, accurate identification of the calcium amounts in atherosclerotic plaque areas may allow effective treatment to prevent further progression of CAD.

Advances in computed tomography (CT) scanning techniques have provided a means for quantifying calcium in the coronary arteries.

Accuracy and reproducibility have become two of the main issues in the quantification of CAC. High reproducibility is very important for follow-up studies to track the patient’s coronary calcification development [9]–[11]. One of the known methods of quantification of calcium in the coronary arteries is the Agatston method [12]. In this method, a threshold of 130 Hounsfield unit (HU) is applied to the CT image. Applying the threshold typically facilitates the identification of all voxels above the threshold as containing calcium. A scoring system is often used to rate the severity of the calcification, based on the number of voxels above the threshold multiplied by a weight based on the highest intensity within the calcification. For example, if the highest intensity is between 130 and 200 HU, then the weight is 1; if between 200 and 300 HU, the weight is 2; and if over 300 HU, the weight is 3. The values of the threshold and the weights are based on empirical studies of coronary scans, using electron beam computed tomography (EBCT), and the subsequent outcome for the patients.

Several reports have demonstrated that the Agatston scoring method is not suited for reliable and reproducible quantification of coronary calcified atherosclerosis plaque for multiple clinical purposes such as plaque progression assessment [13]. A critical weakness of this method is that the segmentation solely makes use of fixed voxel intensity as threshold without considering any complicated scenarios in CT imaging such as partial volume effects (PVE) and spatial information.

It can be seen that the weighting factor $F$ in the Agatston scoring method is chosen based on the maximum intensity. Obviously, the use of the maximum intensity without consideration of the spatial information is highly sensitive to noise which makes the method inaccurate and less reproducible. To solve this problem, alternative methods such as measurement of calcium volume and calcium mass have been proposed and are gradually gaining clinical acceptances [14]–[18] as means of greater precision and reproducibly in measuring calcified plaques. The volume of the calcium is estimated by multiplying the number of voxels above the threshold (e.g., 130 HU) by the volume of each voxel. The mass of the calcium may be estimated by weighting each voxel above the threshold (e.g., 130 HU) according to its intensity, and summing up the weights.

However, the above methods still suffer from considerable inter-scan variability. For example, any change of alignment between the scanner and the scanned object can affect the number of voxels which fall above the threshold and/or the maximum measured intensity within the calcification. Also, tests of phan-
The same explanation relates to the volume underestimation of the same cylinder in the second scan. As seen in Fig. 1(b), along the cylinder boundary, voxels with intensities closer to 130 HU (such as 127 HU or 123 HU) may still contain a small proportion of the calcium component. So, by considering the calcium proportion component in each voxel rather than using the whole voxel, it can provide an accurate means of volume calculation.

The statistical models of PVE have been studied extensively in the literature [19]–[23]. Among these studies, the main objective of the method introduced in [19] was for object segmentation (such as calcium) rather than quantification. In this paper, an extension of this method is employed not only to detect the coronary calcium region, but also to quantify the calcium amount. More specifically, a global statistical model is built to estimate the distributions of calcium and noncalcium (blood vessel), taking into account information from neighbouring voxels. A mixture of tissues statistic models are then defined considering spatial information for PVE calculations in which a maximum probability method is used to obtain the optimum proportion of the calcium component in each detected calcium voxel. Our comparison results show that the accuracy and reproducibility of the proposed method are superior to the voxel based 130 HU thresholding method using 16-slice multidetector row CT (MDCT) in both of the cardiac phantom and real patients’ data. The high reproducibility of the proposed method for the patient data particularly demonstrates the effectiveness of the new method to deal with motion artifacts.

III. QUANTIFICATION OF CALCIFIED REGION

A partial voxel based segmentation method is presented for accurate detection and quantification of CAC from CT images. This method processes the CT image to identify calcified areas. The method selects a 3-D calcified region and excludes any other calcified regions not forming part of the selected calcified region. The region is selected by applying a threshold (130 HU), identifying connected regions that exceed the threshold, and then selecting one of the connected regions as the region for which calcification is to be quantified. Statistical parameters, such as mean and standard deviation of intensity, are calculated both for the selected calcified region and the noncalcified background. Modified expectation maximization (MEM) algorithm is used to calculate the statistical parameters. The MEM algorithm is applied iteratively and can be performed until the estimated statistical parameters converge to a predetermined degree between successive iterations. Based on the estimated statistical parameters, the estimated partial content of calcium per voxel is calculated in the calcified region. The estimated partial content values are processed to generate a map of partial volume of calcium in the selected calcified region. The volume of calcium in the calcified region is determined based on the map Fig. 2 provides a flow diagram outlining the above key steps.

A. Region Identification

The region identification is performed by the thresholding method to segment the image into foreground and background. The foreground areas are grouped into one or more calcified regions. If more than one region is found, the method selects a

![Intensity maps for one calcified cylinder of QRM cardiac phantom scanned twice in (a) and (b).](image1.png)

![Intensity map of one cross section of the calcified cylinder from the first scanning and from the second scanning.](image2.png)
region and defines an enlarged region including a background area around the selected region but excluding nonselected regions.

Fig. 3 illustrates a calcified region in a scan image. A predetermined threshold (e.g., 130 HU) is applied to each voxel in the image. The voxels having an intensity that exceeds the threshold are referred to as foreground, and the remainder as background. The foreground areas are grouped into one or more discrete regions using a simple 3-D binary region-growing or 3-D labeling technique (based on 26-connectivity of the propagation process) [24].

A calcified region is initially defined as containing only the seed point. Region growing is performed by iteratively adding adjacent or neighboring foreground voxels to the region until there are no more foreground voxels adjacent to the region.

If more than one region is found in the image, further regions may be identified by applying the similar 26-connectivity region-growing from other seed points not belonging to any of the regions already found. Fig. 3(b) shows one identified calcified region.

The region is then relatively enlarged using a distance transform technique to obtain an enlarged region. For each identified calcified region, the distance transform map is calculated and the maximum distance of the region related to region boundary is obtained. The enlarged region is then defined as \( s \cdot f_{\text{max}} \), where \( s \) is the enlarged factor, \( f_{\text{max}} \) is the maximum distance value of the region in the distance transform map. In this study, \( s = 2.0 \). Fig. 3(c) is the distance transform map of (b).

Foreground voxels not forming part of the original region are removed from the enlarged region to obtain a final region. This final region is used as a mask to define the maximum potential extent of the calcified region. Fig. 3(d) shows the subimage obtained from overlapping the enlarged mask on the original image (a). Voxels outside this region are not taken into account when estimating the extent and/or properties of the calcified region.

### B. MEM Algorithm

MEM algorithm [19] can be used iteratively to estimate the probability that each voxel in the enlarged region mask [e.g., Fig. 3(d)] represents calcium. A statistical model can be constructed to estimate such parameters based on the MEM algorithm. An intensity image \( Y = \{y_i, i = 1, 2, \ldots, I\} \) of region mask with \( I \) voxels of intensity \( y_i \) and \( K \) different classes, \( L = \{1, 2, \ldots, K\} \), is provided. A special case includes two classes or tissue types: calcium and noncalcium (i.e., blood vessel).

It is noted that the use of a threshold of 130 HU in Agatston method [12] to identify calcifications is based on the density of two standard deviations above the average density of blood in the aorta. Due to the above fact and by simplifying the model, the ranges of image intensities corresponding to the non-calcium (blood vessel) and calcium are modelled as Gaussian distributions.

Image intensity in CT imaging is spatially dependent. For instance, voxels with the same intensity may have different structural properties. A mixed statistical model that considers spatial properties is employed for the distribution of voxel intensity \( p(y_i | \vartheta) \) as follows:

\[
p(y_i | \vartheta) = \sum_{k \in L} a_k(i) \cdot p_I(y_i | \varphi_k) \quad i = 1, 2, \ldots, I \tag{1}
\]

where, for each \( i \)

\[
p_I(y_i | \varphi_k) = G(\mu_I, \sigma_I) = \frac{1}{\sqrt{2\pi}\sigma_I} \exp \left( -\frac{(y_i - \mu_I)^2}{2\sigma_I^2} \right)
\]

which is a Gaussian distribution with parameters \( \varphi_k = (\mu_k, \sigma_k) \) and \( \vartheta = (\varphi_1, \varphi_2, \ldots, \varphi_K) \). \( a_k(i) \) is a spatial prior probability with spatial constraints imposed by
a Markov random field (MRF) and Gibbs random field (MRF-GRF) [25].

A MRF \( F = \{ F_1, F_2, \ldots, F_k \} \) defined on the set \( S \) is a lattice indexing the voxels in the given image \( Y \), in which each random variable \( F_i \) takes a value \( l_i \in \mathcal{L} \). The probability density of the MRF \( F \) can be given by the Gibbs distribution

\[
p(F) = Z^{-1} \exp[-U(F)]
\]

where \( U(F) = \sum_{c \in C} \nu_c(F) \) is the energy function. The energy function is a sum of clique potentials \( \nu_c(F) \) over all possible cliques in the enlarged region mask, and \( Z \) is a normalization term. According to the Hammersley Clifford theorem, the conditional probability can be derived using MRF-GRF equivalence as follows:

\[
p(l_i | N(i)) = \frac{\exp[-\sum_{l_i \in L} \nu_c(l_i)]}{\sum_{l_i \in L} \exp[-\sum_{l_i \in L} \nu_c(l_i)]}
\]  

(2)

where \( N(i) \) is the neighborhood of voxel \( i \).

Assuming the spatial prior distribution \( a_l(i) \) in (1) is given by the MRF conditional probability \( p(l_i | N(i)) \) in (2), according to the Bayesian probability theory, the posterior probability \( p(\varphi_l | y_i) \) can be obtained as

\[
p(\varphi_l | y_i) = \frac{p(y_i | \varphi_l) \cdot p(l_i | N(i))}{p(y_i | \theta)}.
\]  

(3)

Here, the potential function in (2) is defined as

\[
\nu_c(l_i) = \beta \cdot p(\varphi_l | y_j), \quad j \in c
\]

where \( \beta \) is a positive constant which controls the size of clustering. The posterior probability \( p(\varphi_l | y_i) \) represents the probability that the given voxel \( i \) belongs to one class \( l_i \in \mathcal{L} \). Equation (3) can be used to estimate the highest probability of the reconstructed label image \( \mathcal{L} \) based on the observed intensity value and the image model as defined in (1) and (2). The model parameters \( \theta \) can be obtained to solve (3).

To adapt the model defined in (1) and (2) so that the spatial information is considered by using MRF-GRF model, a modified version of the two-step EM algorithm (i.e., an MEM algorithm) may be used to estimate parameters of the model and classify voxels of each group simultaneously. For example, for a given \( \varphi_l = (\mu_l, \sigma_l) \), the unique solution \( \varphi_l^{m+1} = (\mu_l^{m+1}, \sigma_l^{m+1}) \) can be derived as

\[
\begin{align*}
\mu_l^{m+1} &= \left( \frac{\sum_{i=1}^{N} y_i \varphi_l^{m}(y_i | y_j \varphi_l^{m})}{p(y_i | y^m)} \right) \\
\sigma_l^{m+1} &= \left( \frac{\sum_{i=1}^{N} (y_i - \mu_l^{m+1})^2 \varphi_l^{m}(y_i | y_j \varphi_l^{m})}{p(y_i | y^m)} \right)
\end{align*}
\]

(5)

where \( a_l^{m}(i) \) in each step can be approximately calculated by assuming

\[
a_l^{m}(i) \approx p(l_i | N(i)) = \frac{\exp[-\sum_{l_i \in L} \nu_c(l_i)]}{\exp[-\sum_{l_i \in L} \nu_c(l_i)]}.
\]

The process converges after sufficient iterations, and may be halted after a predetermined number of iterations and/or once a predetermined convergence criterion is met.

The MEM algorithm iteratively calculates a statistical classification of the voxels based on the model parameters of the previous iteration and updates the parameters accordingly. The use of MRF-GRF as a spatial constraint can improve the voxel-based image classification performance of the MEM algorithm, especially in the presence of noisy image data.

C. Estimate of Partial Content of Calcium

MEM algorithm is used to estimate the statistical model parameters in (5) and to detect the calcium region by taking into account PVE problem. In this section, the estimated model parameters are employed to calculate the proportion of calcium in “partial voxels,” which are a mixture of calcium and noncalcium (blood vessel) (see Fig. 4).

A model of calcified material with probability distribution of \( p_1(\mu_1, \sigma_1) \), and a model of non-calcified material with probability distribution of \( p_2(\mu_2, \sigma_2) \), are associated with a particular voxel, where \( \mu \) and \( \sigma \) are the mean and standard deviation of the Gaussian model. These parameters can be estimated using MEM [15]. The distribution for the combined intensities follows a linear combination of two Gaussians

\[
p(y_i | a^l) = G (\hat{a}^l \mu_1 + (1 - \hat{a}^l) \mu_2, \sqrt{\hat{a}^l \sigma_1^2 + (1 - \hat{a}^l) \sigma_2^2})
\]

where \( \hat{a}^l \) is the proportion of calcium in voxel \( i \), and the proportion of noncalcium (blood vessel) in the voxel is \((1 - \hat{a}^l)\).

According to Bayes’ theorem, given voxel \( i \), the statistical distribution of the proportion of calcium component in this voxel can be calculated using the following equation:

\[
p(a^l | y_i) = \frac{p(y_i | a^l) \cdot p(a^l)}{p(y_i)}
\]  

(6)

where \( p(y_i) \) is a normalizing constant, and \( p(a^l) \) is the prior probability of the \( i \)th voxel having proportion \( a \) of calcium. \( p(a^l) \) can be calculated in at least two different ways.

Method 1: Assuming the prior probability is modelled as a uniform distribution in the range [0,1], i.e., \( p(a^l) = 1, \forall i \). According to an embodiment, only partial voxels are considered to determine \( p(a^l | y_i) \). The profiles of \( p(a^l | y_i) \) at different
intensity values $y_k$ are calculated based on the estimated statistical values. Fig. 5 shows profiles of $p(a^i | y_k)$ at four different exemplary intensity values ($y_k = 50$ HU, 100 HU, 130 HU, and 200 HU) of a CT image. In each profile in Fig. 5, the horizontal axis represents the proportion of calcium in the voxel, and the vertical axis represents the probability of the voxel containing the proportion $a$ of calcium.

Method 2: $p(a^i)$ (MRF) can be based on the neighbourhood of a voxel. In (6), the initial prior probability $p(a^i)$ is updated based on $p(a^i | y_k)$ of the neighbourhood. Only neighbouring sites have direct interactions with each other, and they tend to have the same class labels. Based on a technique related to the Gibbs distributions, the prior probability can be derived as follows:

$$p(a^i) = \frac{\exp\left[\sum_{c \in e} v_c(a^i)\right]}{\sum_{a} \exp\left[\sum_{c \in e} v_c(a)\right]}$$

where $v_c(a) = \beta \cdot p(a^i | y_k)$, $s \in c$, over all possible cliques $c$. An exemplary set of conditions may be as follows: $a = 0.1 \times k$, $k \in [0, 10]$, and $\beta = 0.5$.

In this study, method 2 was employed. To calculate the amount of a certain calcium in one voxel, one can determine the highest probability of $a$, namely

$$a_{\text{optimal}} = \max_{a} P(a^i | y_k).$$

Hence the total volumetric size of the calcium plaque is defined as

$$V = V_{\text{voxel}} \times \sum_{i} (a_{\text{optimal}})$$

where $V_{\text{voxel}}$ is the volume of one voxel.

Fig. 6 shows an example of the calcium plaque and its corresponding proportion map. Fig. 6(a) is the original calcium plaque subimage while Fig. 6(c) is the corresponding HU values of voxels containing the calcium plaque. The potential calcified region is initially identified using the algorithm described in Section III-A. The MEM algorithm is then applied on the enlarged coarse region to calculate each class parameters $\varphi_k(\mu, \sigma)$ based on (5). Next, the proportion of the calcium at each voxel is calculated by using (8). Fig. 6(b) is the calculated proportion map and Fig. 6(d) is the proportion values of voxels in the proportion map which corresponds to the intensity values shown in Fig. 6(c). From Fig. 6(c) and (d), it can be seen that the proportion values in the core part of the calcium area (HU values over 188 HU) are 1 signifying that those voxels contain only a calcium component; while the proportion values on the surrounding area are less than 1 (such as 0.30 for the voxel with 114 HU) meaning that those voxels only contain a partial calcium component.

### IV. EXPERIMENTAL RESULTS

Two studies were conducted to compare the accuracy and reproducibility of the proposed method with traditional one that was described earlier. In the first study, a cardiac phantom (QRM, Moehrendorf, Germany) study was performed to show how the proposed method gives accurate volume measurements under different intensity contrasts and radiation doses. Considering the motion artifact, in the second study, reproducibility was further evaluated in 35 patients who were scanned twice.
using the same protocol at a minimum interval of 10 min. The standard deviation (std) of the total calcium volume size between two series was calculated to demonstrate the effectiveness of the proposed method in the presence of motion artifacts. The following are the results of experiments.

A. QRM Cardiac Phantom Study

The cardiac phantom consists of two parts: an anthropomorphic phantom body and heart calibration insert (Fig. 7). The phantom body contains artificial lungs and a spine insert surrounded by soft tissue equivalent material chosen with respect to real attenuation characteristics of X-ray in the thorax. At the anatomic position of the heart is a cylindrical hole of 100 mm diameter which encompasses the heart calibration insert. Fig. 8 shows a sketch of the phantom body with the calibration insert. The calibration insert itself contains nine cylindrical objects that vary in size and hydroxyapatite density which is organized on three series. Each of the series contains three different sizes being 1, 3, and 5 mm and three different densities of hydroxyapatite (CaHA) that are 200, 400, and 800 mg/cm³.

The phantom was scanned 90 times on a 16-slice Multi-Detector Row CT (MDCT) (GE LightSpeed Pro) using 15 different protocols. Acquisition protocols were chosen to provide a large spectra of acquisition parameters that simulate different clinical situations to minimize and maximize the partial volume effect and vary the noise level (Table I). The phantom was placed in three different angular positions (0°, 45°, and 90°) to the z-axis and each protocol was used to scan the phantom twice. A sample slice containing six simulated calcium plaques is shown in Fig. 9(a). Note that in this study, only the calcium insets with a diameter larger than 1 mm are considered. The plaques had three different HU of about 270 HU, 550 HU, and 1100 HU. The actual or true values for the plaques in the phantom are given in Table II.

Fig. 9(c) and (d) clearly highlights the dramatic difference between the proposed and conventional methods. Fig. 9(c) is the proportion map of one calcium plaque (C2) using the proposed method while Fig. 9(d) is the same calcified region detected by the conventional method. It is observed that in the proposed method shown in Fig. 9(c), some voxels near the plaque boundary have values less than 1.0, indicating a mixture of materials (calcium and noncalcium), the number being the fraction of calcium. However, in the conventional method shown in Fig. 9(d) those voxels are incorrectly labeled as 100% calcium illustrating a tendency to overestimate the area.

The tendency to overestimate the volume as shown in Fig. 9(c) and (d) is also apparent when averaged over 90 scans as shown in Table II. Volume measurements using the proposed method have a low bias or systematic error (the difference between true value and mean volume calculated by the proposed method is about 1.1 – 5.6 mm³, the average error of 9.5%), which is almost independent of the plaque size and density. In contrast, the conventional method has larger systematic errors (up to 86 mm³, with an average error of 67%) that depend on both plaque size and density. Fig. 10 shows the relationship of the accuracy of the plaque averaged volume measurements over 90 scans with the plaque size and density for the two methods.

For the reproducibility, the proposed partial voxel based method demonstrates reasonably good reproducibility, with a standard deviation of about 4.1 – 6.3 mm³, regardless of plaque size and density. In contrast, the conventional method is less reproducible for all plaques (3.1 – 19.7 mm³). Fig. 11 shows the relationship of the reproducibility of the plaque averaged volume measurements over 90 scans with the plaque size and density for different methods.

Fig. 12 shows a comparative result of volume measurements for plaque A1 using the two methods. It can be seen that the pro-
Table II

<table>
<thead>
<tr>
<th></th>
<th>True value</th>
<th>Volume (without PVE / Conventional)</th>
<th>Volume (with PVE / Proposed)</th>
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<tr>
<td>plaque</td>
<td></td>
<td>mean</td>
<td>std</td>
</tr>
<tr>
<td>A1</td>
<td>98.2</td>
<td>184.2</td>
<td>19.7</td>
</tr>
<tr>
<td>A2</td>
<td>21.2</td>
<td>55.2</td>
<td>6.3</td>
</tr>
<tr>
<td>B1</td>
<td>98.2</td>
<td>148.0</td>
<td>13.9</td>
</tr>
<tr>
<td>B2</td>
<td>21.2</td>
<td>40.6</td>
<td>6.2</td>
</tr>
<tr>
<td>C1</td>
<td>98.2</td>
<td>105.5</td>
<td>5.5</td>
</tr>
<tr>
<td>C2</td>
<td>21.2</td>
<td>23.0</td>
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</table>

Fig. 10. Averaged volume measurements of 90 different acquisitions with three different densities for different size plaques. (a) Small plaques. (b) Large plaques.

B. Real Clinical Data Study

The in vivo experiments in this section aim to illustrate, with cardiac motion, how the proposed method gives high reproducibility for the volumetric measurements of CAC.

Thirty-five patients, with known or suspected ischemic heart disease or at risk of developing ischemic heart disease were enrolled in the study. Each patient underwent two unenhanced CT scans using the same parameters (120 kV, 200 mA, slice thickness 2.5 mm) at a minimum interval of 10 min. The study was approved by the local Ethics Committee and the informed consent was obtained from all patients.

Data obtained were analyzed independently by two radiologists to determine volumes and scores from the first and the second series, from which 12 patients present calcium. For each series, both the proposed partial voxel based method and the conventional method are used to calculate each calcium volume. Figs. 13 and 14 are two examples of the corresponding calcium...
Fig. 11. Averaged reproducibility of 90 different acquisitions verses density for different plaques. (a) Small plaques. (b) Large plaques.

Fig. 12. Volume measurements for plaque A1 using both the proposed partial voxel based method and the conventional method.

Fig. 13. Example of plaque A scanned twice using the same protocol at a minimum interval of 10 min.

Fig. 14. Example of plaque B scanned twice using the same protocol at a minimum interval of 10 min.

Table III

<table>
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<tr>
<th>Volume measurement with PVE</th>
<th>Volume measurement without PVE</th>
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<tr>
<td>Vol. Scan 1</td>
<td>Vol. Scan 2</td>
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<td>66.76</td>
<td>69.9</td>
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Table IV

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<th>Volume calculations using different methods (units for volume is mm³) for plaque B as highlighted in Fig. 14</th>
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<td>Proposed volume measurement with PVE</td>
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<td>------------------------------------</td>
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<tr>
<td>Vol. Scan 1</td>
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<td>226.8</td>
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Fig. 15. Example of the absolute volume differences between two scans for 20 corresponding plaques.

3.14 mm³ (plaque A) and 1.10 mm³ (plaque B) by using the proposed method, while using the conventional method, the differences are 19.67 mm³ (plaque A) and 10.10 mm³ (plaque B) respectively. Fig. 15 gives an example of the absolute volume difference between two scans for 20 individual plaques. It is noted that due to the cardiac motion, it is very difficult and time consuming to find one to one correspondence between the plaques in the two scans. For example, one plaque in Scan 1 might be observed to split into several pieces in Scan 2. Table V shows the total calcium volumetric measurements using the two methods. The deviation of the total calcium volume size between two series, calculated considering PVE, is about 2–3 times less than the calcium volumetric measurements without PVE. Fig. 16
notably up to double that of the proposed method. While, in the phantom study, the proposed method consistently calculated values which were very close to the known true values, the other method calculated values of up to double the true values. This implies an accuracy of the proposed method in the patient scans.

The proposed method has shown a statistically significant improvement over the conventional thresholding based methods when applied to the phantom and patient scans. This could result in more accurate prediction of future cardiac events and permits more accurate monitoring of the effects of risk factor modification in individual patients.

Moreover, in this study, a linear combination of tissues statistical model is used to calculate the proportion of the calcium amount in the coronary arteries.

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VI. DISCUSSION AND CONCLUSION

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The experimental results on a QRM cardiac phantom with 90 scans using 15 different protocols and 35 patient data with repeated scans have demonstrated that the newly proposed technique gives better results compared with conventional predefined thresholding based methods in terms of accuracy and reproducibility. Significantly, the high reproducibility for the patient data demonstrates the effectiveness of the new method in the presence of motion artifacts. The deviation of the total calcium measurements is about 2–3 times less than that calculated using the traditional method. More interestingly, although the true volume of calcified plaque is obviously not available in patient data, volumes measured by the conventional method are

VI. DISCUSSION AND CONCLUSION

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