Voxel based calcification fraction quantification in atherosclerotic plaques using serial UTE

Jinnan Wang1, Niranjan Balu2, Thomas S Hatsukami2, Chun Yuan2, and Peter Börnert1
1Philips Research North America, Briarcliff Manor, NY, United States, 2University of Washington, 3Philips Research Europe

Introduction
Although MR has been shown to provide accurate and reliable atherosclerotic plaque components identification, the detection of high risk components is usually compromised when they coexist with calcification. Speckled calcification, which presents in over 35% of advanced plaques, usually coexists with high-risk plaque components at an arbitrary fraction. As a result, the MR signal from high risk components is masked by reduced signal from speckled calcification (Fig.1). The signal from the high-risk components can be corrected, in theory, if the fraction of calcification at a particular location is known. Currently, however, there is no technique available capable of accurate estimation of the calcification fraction on the voxel level. To address this gap, a serial UTE based magnetic resonance imaging and processing approach is proposed in this study to accurately quantify the calcification fraction on a voxel level.

Methods
Six human endarterectomy specimens were scanned using a clinical 3T scanner (Philips Achieva, R2.61, the Netherlands) with a solenoid coil for improved SNR. Seven 3D radial UTE images with different TEs (TE = 0.1, 0.2, 0.4... 6.4ms) were acquired to measure the calcification fraction at each location. Other imaging parameters were: T1-FFE, TR 25ms, FA 12°, FOV 28×28×28mm³, resolution: 0.25×0.25×0.25mm³, 40% radial sampling rate, fat saturation, imaging time of each acquisition: 4min26sec.

The signal intensity value at each voxel is given by a bi-exponential decay model:
\[ S_i = \lambda_1 \exp(-TE/T_{1\text{short}}) + \lambda_2 \exp(-TE/T_{1\text{terminal}}), \quad i = 1,2,...,7 \quad (Eq.1) \]

Where \( \lambda_1, \lambda_2 \) are the relative signal weightings of the short and regular \( T_1 \) components and \( T_{1\text{short}}, T_{1\text{terminal}} \) are the values of short and regular \( T_1 \) components. Knowing the proton density difference between calcification and soft tissue is \( \approx 1.5 \), calcification fraction at each voxel can be determined based on the ratio between different \( \lambda_2 \).

Due to the signal variation at each voxel, Eq.1 cannot always be reliably estimated by a least square approach. A non-linear mixed effect (NLME) model (4) was adopted to account for non-constant correlation among the different voxels, making the model more robust to the presence of noise. The NLME model, each observation is represented as a combination of a fixed response, an individual response and a random error:
\[ S_{ij} = f(\kappa_j) + e_j, \quad e_j \sim N(0,\sigma^2) \]
\[ \kappa_j = g(b_j), \quad b_j \sim N(0,\sigma^2_j) \quad (Eq.2) \]

where \( j \) represents different voxels, \( f(\kappa_j) \) is the original function, \( g(b_j) \) is the parameter estimation function, \( e_j \) is the residual error, \( \kappa_j \) is the vector including all parameters in Eq.1, \( b_j \) is the fixed effect parameter, \( b_j \) is the inter-voxel variability. As the mean value is determined on a population basis, no initial value needs to be provided for NLME modeling. The slice with the maximum calcification area from each specimen was analyzed using the NLME model. Only pixels contained calcification were included in the analysis using interactive region of interest (ROI) selection. The calcification fraction of all voxels was plotted in a histogram to examine the calcification distribution. Pearson’s correlation between MR signal intensity at TE=0.1ms and measured calcification fraction was calculated to examine the agreement between the two.

Results
Successful NLME fitting was obtained for over 95% of all included voxels, compared to ~50% when the least square approach is used. An example comparison of fitting results is shown in Fig.2: the NLME model demonstrated more successful fitting on the same dataset. Color-coded calcification map overlaid on the original MR images showed robust calcification fraction measurement without noticeable fitting noise (Fig. 3b). Furthermore, the signal variation pattern on calcification maps corresponds well to that on the original MR images.

When plotted against the original signal intensity on MR images, a good correlation \((r=-0.73, \ p<0.01)\) was obtained (Fig.4). When the calcification fraction is high (over 70%), there was a good correlation between high calcification and low signal intensity, indicating that the signal was dominated by calcification. When the calcification fraction was low, however, the distribution became more spread out, indicating the signal was more dominated by the other tissues coexisting in the same voxel.

In conclusion, the feasibility of measuring calcification fraction on a voxel level has been demonstrated in this study. Robust calcification fraction maps can be generated using the NLME model. This approach can also be applied to other applications where the accurate quantification of calcification on a pixel/voxel level is needed.

References: