In Vivo 3D High Resolution Apparent Diffusion Coefficient (ADC) maps of Carotid Artery Atherosclerosis Plaques Using 3D singleshot Inner Volume Stimulated EPI (3D ss-IV-STEPI) Technique

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INTRODUCTION: The extent of lipid accumulation and the presence of intramural hemorrhage have been found to be associated with the degree of plaque vulnerability and risk of plaque rupture. However, lipids and hemorrhage are often difficult to distinguish from other plaque components by MRI, and the reported appearance of these tissues in different MR studies is inconsistent [(1-2)]. Ex-vivo studies have found that diffusion weighted imaging (DWI) may provide a tool for the characterization of these components [(3-6)]. We have reported in-vivo ADC values of human carotid artery [(7)]. To improve the sensitivity and specify of plaque component identification in ADC maps a high resolution 3D DWI technique is desirable. A 3D single shot Inner-Volume, STimulated EPI (3D ss-IV-STEPI) [(8)] sequence has been developed to acquire a high resolution DWI of a localized region such as a carotid artery. This technique acquires the total k-space of a limited 3D volume after a single diffusion-preparation. An inner volume excitation technique is used to reduce the time required for EPI readout of each complete k-space and thereby reduce blurring and susceptibility artifacts. In this work we present the first in vivo high resolution 3D DWI of the human carotid artery using the 3D-ss-IV-STEPI sequence.

METHODS: Figure 1 shows the pulse sequence diagram of 3D ss-IV-STEPI. The pulse sequence consists of two main sections: diffusion-prepared driven-equilibrium (DPDE) and 3D data acquisition. DPDE (90^o-G_D-180^o-G_D-90^o_{tipup}) precedes the stimulated echo imaging sequence. The slice-selection gradient of the 180^o pulse is applied in the phase-encoding direction to achieve the inner volume imaging. Diffusion gradients (G_D) were applied before and after the 180^o pulse. Before being tipped up to the longitudinal axis the DPDE transverse magnetization in each voxel is dephased by a dephasing gradient (indicated by right arrow: → in Fig. 1) to remove the signal dependence upon the tipup RF pulse phase. The residual transverse magnetization is suppressed by a spoiler applied after the tipup pulse. The data acquisition part consists of multiple EPI segments. After completion of each EPI echotrain, the residual transverse magnetization can be rewound or completely spoiled by a spoiler gradient. Because the centers of k-space are acquired during the first few stimulated echoes, with rewinding or with spoiling the DWI signal intensity undergoes simple exponential decay with respect to the *b* value, as

$$M^+ \cong M^+_0 \cdot e^{-TE/T_2} \cdot e^{-b\cdot D} \qquad (1)$$

To evaluate the feasibility of this new sequence, carotid arteries of five volunteers were scanned with 3D ss-IV-DWSTEPI centered at the bifurcation apex. The imaging parameters were: receiver bandwidth = 1.086 kHz/pixel, imaging matrix =160x41x16 with isotropic 1 mm voxel dimensions, TR = 3s, 32 averages and 16 slices. The 1 mm isotropic voxels were zero-filled-interpolated to 0.5 mm voxel spacing in all directions. Total scan time was 4:24 min for two b-values (10, 500 sec/mm^2). After two b value images were acquired, the ADC map was calculated and displayed using IDL. T1w and T2w images were acquired with two or three interleaved TSE with our modified version of the double inversion preparation sequence [(9)].

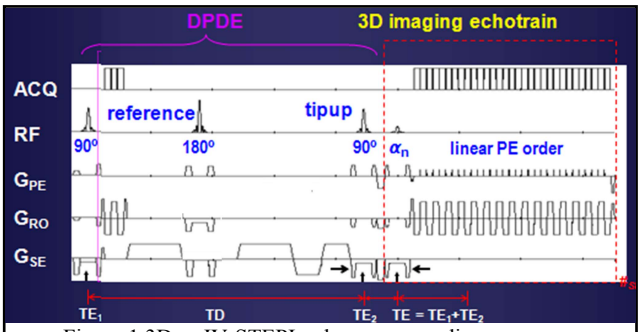

Figure 1 3D ss-IV-STEPI pulse sequence diagram

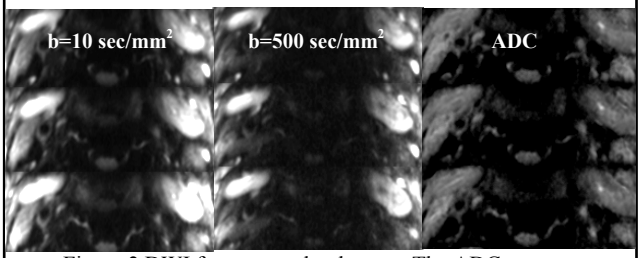

Figure 2 DWI from normal volunteer. The ADC map were calculated using b=10, 500 sec/mm^2.

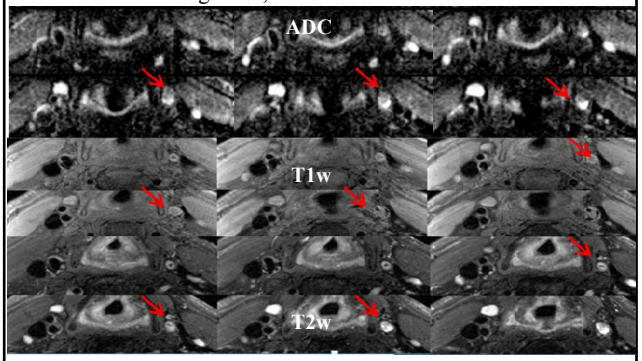

Figure 3 ADC maps, T1w and T2w from a patient who has visible plaque which indicate by red arrows.

RESULTS: DW Images of the carotid artery from a normal subject using 3D ss-IV-STEPI sequence are displayed in figure 2. Each column shows DWI (b=10, b=500. sec/mm^2) and ADC maps from identical locations. All DW images and maps demonstrate clear wall definition and minimal susceptibility artifact. Figure 3 shows the ADC maps, 2D T1w and T2w images from a patient who has a visible plaque. The plaque area is shown as bright signal in the T1w, T2w images. Some plaque with high water content, such as necrosis or old hemorrhage may be very bright or of intermediate intensity with T2 weighting. The ADC of the plaque area (red arrow) was calculated as 1.18×10^{-3} mm^2/sec. The mean ADC value of the vessel wall from five subjects was $1.38 \pm 0.45\ 10^{-3}$ mm^2/sec. This value is close to the ADC value reported previously from an ex-vivo sample [(4-5)].

DISCUSSION: The results obtained indicate that 3D DW imaging could increase the sensitivity and specificity of plaque component quantification in ADC maps. 3D high resolution and limited FOV readout results in reduced susceptibility artifacts while yielding relatively low image SNR on each shot. The averaging of multiple images would be useful to improve SNR. Motion during multiple averaging can cause phase errors and blurring. Because it is a 3D single shot acquisition, each image is free from ghosting due to phase errors caused by motion during the diffusion gradients. However, such phase errors will cause the phase at each voxel within each image to vary, depending on the motion that occurs during the diffusion encoding gradients. The effects of phase errors are eliminated by using magnitude averaging. However, various neck motions such as swallowing, respiration, and blood pulsations may still occur and introduce inconsistency among the different averages. To reduce the effects of neck motion on the ADC calculation we interleaved the acquisition of the low and high b-value images for all averages. It is possible to reduce further the motion sensitivity of the ADC calculation by registering each DW image prior to averaging the magnitude images.

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