Characterization of Carotid Plaque in Three-Dimensional Ultrasound by Registration with Multicontrast MRI

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Introduction The ability of MRI in carotid plaque component identification has been well-established. Modern advances in MRI have allowed the acquisition of multiple images of the same plaque using different contrast mechanisms. This "multicontrast MRI" technique has made the identification of many types of tissue possible. On the other hand, ultrasound (US) imaging has been developed as a cost-effective assessment tool, which can be used to screen a large number of symptomatic and asymptomatic subjects. However, classification of carotid plaque components in US images is associated with large observer variability [1]. Nevertheless, given the cost effectiveness of US, it is important to establish to what extent US features can predict subsequent MRI findings in order to develop an appropriate screening strategy utilizing both modalities. The prerequisite that makes this assessment possible is to establish a spatial correspondence between the MR and US images. This study aims at developing a MR-US registration technique so that component information from a MR image can be compared to US features at the same location.

Methods ***MRI Acquisition*** A multicontrast protocol [2] for carotid MRI was used to obtain 2-dimensional T1-, T2-, proton density-, contrast enhanced T1-weighted black-blood images and 3-dimensional time-of-flight bright-blood images. Subjects were asymptomatic with 16-49% stenosis in their carotid artery as determined using duplex ultrasound [3]. ***US Acquisition*** The 3D US volume was acquired using a VL13-5 transducer on the iU22 US system (Philips Healthcare, Andover, MA). The transducer acquired contiguous 2D transverse images, which were then reconstructed into a 3D volume. ***MR and US Segmentation*** For each MR axial image, a reviewer segmented the lumen, the outer wall and the component boundaries (lipid-rich necrotic core (LRNC), calcification, loose matrix and intraplaque hemorrhage (IPH)) using an in-house image analysis software CASCADE [4]. In US, the media-adventitia boundaries (MAB) were segmented using a semi-automated research plugin in QLAB (Philips Healthcare, Andover, MA). The user draws approximate boundaries in a few key frames in an US volume. The plugin then optimizes and automatically determines MAB in all the remaining US frames. ***Surface Registration*** The outer wall contours in MR and the media-adventitia contours in US were reconstructed to 3D surfaces. The carotid bifurcations on these two surfaces were manually aligned. The optimal rotation required to register the two surfaces is determined by the iterative closest point (ICP) algorithm [6]. This transform (translation and rotation) was applied on the US volume, producing a 3D US image that is registered to the MR image. This registered US image was resliced to match the orientation of MR axial slices (Fig. 1). ***Mapping of Components on US Image*** The component contours obtained from MR axial slices were superimposed on the US axial slices that was obtained by reslicing. Using CASCADE, a user can modify the registration result by shifting the US axial slices. After this manual adjustment, the component contours were reconstructed into surfaces. The grayscale average and the standard deviation of the pixel enclosed by each component were computed.

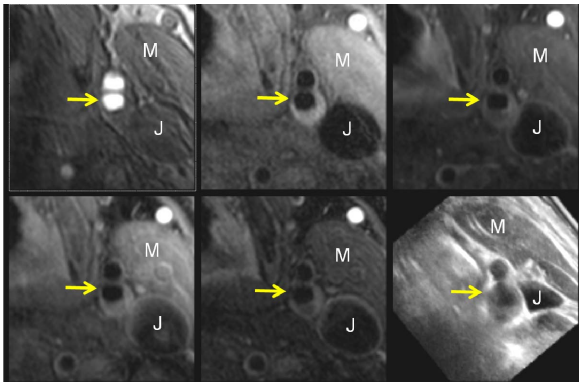

Fig 1: A corresponding slice of a 3D ultrasound image (lower right) is displayed with 5 contrast weightings in a MRI exam.

	Subject 1		Subject 2		Subject 3	
	w/o	w Sh	w/o	w Sh	w/o	w Sh
ECA	0.72	1.02	1.15	1.17	0.59	0.58
ICA	0.47	0.43	1.00	0.85	0.54	0.33
CCA	0.34	0.35	0.90	0.73	0.51	0.52
Total	0.46	0.52	0.99	0.87	0.54	0.47

Table 1: Registration error (in mm) without (w/o) and with (w Sh) manual shifting adjustment.

	Subject 1	Subject 2	Subject 3
Calcification	96.3 (36.9) [18,192]	N/A	70.7(49.4)[0, 254]
LRNC	82.6 (32.6) [16, 167]	77.9 (20.5)[22,157]	N/A
Loose Matrix	45.7 (18.4) [6, 141]	N/A	N/A
IPH	N/A	73.1(12.3)[37,117]	N/A

Table 2: Average, standard deviation (in parenthesis) and range (inside [ ]) of each plaque component of each subject.

Results ***Registration Error*** We quantified the registration error by finding a closest point from the MR surface to the US surface. We measured the errors before and after the manual in-plane adjustment. It is not always possible to correct the alignment of the ICA and ECA simultaneously. In this case, we only align the ICA without considering the registration accuracy of the ECA. This was done because plaque components were only studied at the CCA and ECA branches in a MR image. Table 1 shows that the registration accuracy of ICA always improves after manual shifting adjustment, and the registration accuracy of CCA either stays the same or improves after the adjustment. ***US Intensity of Plaque Components*** Table 2 shows the average, standard deviation and range of the grayscale value of pixels enclosed by each component in each subject. The property of the calcified plaque in Subject 1 is different from that in Subject 3. The calcified plaque of Subject 3 is larger (Fig. 2a vs. b) and the surface of which reflected most of the energy of the US beam, producing an acoustic shadow distal to the calcified plaque surface. This results in the average US intensity being low and the standard deviation high. The properties of the LRNC region in Subject 1 and 2 are similar. Results in Subject 2 provided further evidence that it is difficult to distinguish between LRNC and IPH [7,8]. It is unexpected that the US intensity for loose matrix is low, as fibrous tissue is usually more reflective (Fig 2c).

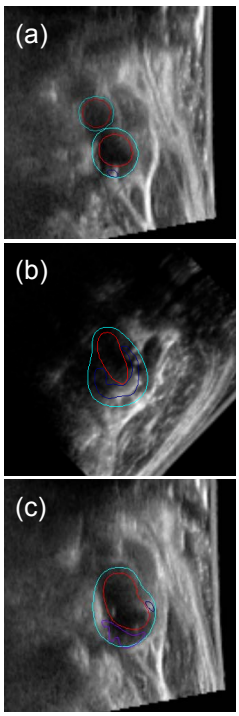

Fig. 2 Plaque Components (Calcification – Dark Blue, Loose Matrix – Purple), lumen (red) and outer wall (light blue) boundaries as appeared in US images. Calcification in (a) Subject 1 and (b) Subject 3. (c) Loose matrix appears dark in an US image.

Discussion and Conclusion Due to the cost effectiveness of US, it would be beneficial to establish a screening strategy in which US is used to identify subjects suspicious of having vulnerable plaques, who are then further studied using MRI. To evaluate the viability of such screening strategy, it is crucial to establish to what extent US can identify unstable plaque components. To make this assessment possible, we have developed a MR-US registration technique that establishes the spatial correspondence of different plaque components in MR and US image. Our registration method is shown to have an average error below 1mm. Our findings in terms of average intensity of each component are in large consistent with previous US characterization studies. We also found that there is a good correspondence between hyperechoic structures and calcifications, although US is not good at localizing a large calcified plaque because of the US shadowing artifact.

Reference [1] Arnold JA et al., 1999 Stroke 30(1), 61–65. [2] Yuan C et al., 2006 NMR Biomed 19(6), 636–654. [3] Roederer G et al., 1984 Bruit 8, 174–178. [4] Kerwin W et al., 2007 Top Magn Reson Imaging 18(5), 371–378. [5] Zhu S et al., 1996 IEEE TPMI 18(9), 884–900. [6] Besl P J et al., 1992 IEEE TPMI 14(2), 239–256. [7] ECST Collaborative Group 1995 Lancet 345, 209–212. [8] Bock R et al., 1992 Diagnostic vascular ultrasound pp. 225–236.