Magnetic Resonance Elastography for Measuring the Compliance of Occlusive Vascular Disease

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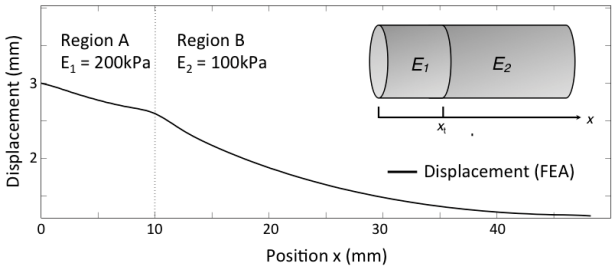


Fig. 1. Displacement along the center of a simulated artery using FEA. Regions A and B correspond to the proximal cap and soft core of a CTO respectively, with estimated young's moduli of E_1 and E_2.

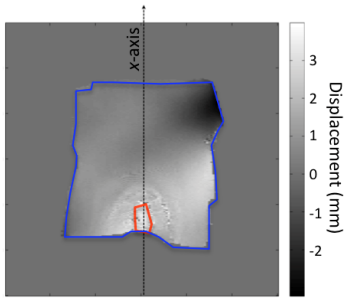
TARGET AUDIENCE: Interventional MR researchers and clinicians investigating novel MR techniques to aid the guidance of interventions.

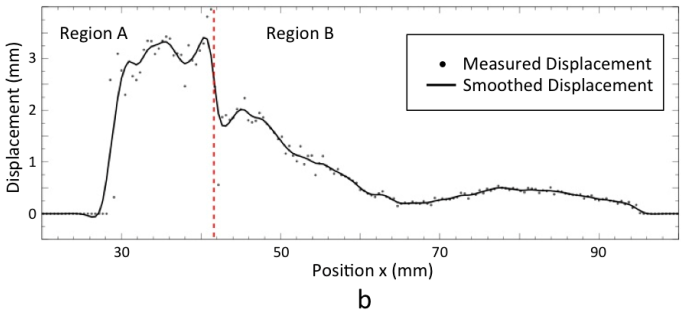
PURPOSE: Chronic total occlusions (CTOs) have proven to be the most difficult type of arterial occlusions to cross by percutaneous intervention. As a result, the presence of one or more CTO in either the coronary or peripheral arteries usually leads to bypass surgery or for other forms of medical therapy. Although X-ray fluoroscopy is the most prominent imaging modality used to guide percutaneous revascularization, it is limited by poor soft tissue contrast [1]. This makes it challenging to characterize critical properties of the occlusion such as the nature of its composition, which is integral to guidewire selection. Guidewire selection may be an important factor in success as, although a soft guidewire may be insufficient to penetrate hard occlusions, a stiffer guidewire is more likely to perforate the vessel wall. A technique known as MR Elastography (MRE), which has been used to measure the stiffness of various tissues, could provide valuable information about the stiffness and geometry of an occlusive arterial lesion. In particular, it would be helpful to have information about the thickness and stiffness of the hard proximal fibrous cap (PFC) seen in many CTOs. This study explores quasi-static MRE for determining the stiffness and extent of regions in CTOs. Compression loading is applied to the surface an occluded artery phantom. The displacement due to compression is analyzed using a phase contrast MRI pulse sequence, and the regions within the samples are identified.

THEORY: The response of an arterial lesion to deformation can be determined using the equations of continuum mechanics [2]. Deformation of tissue is typically dependent on complex three-dimensional interactions between stresses and strains, and usually requires the use of finite element analysis (FEA) to obtain an approximate solution. A simplified 1-D analysis provides an analytical solution representing the idealized response of a lesion to deformation. In this way, a lesion can be modeled as a cylinder surrounded by tissue that provides a resistive shear force, but allows for unrestrained expansion in the radial direction. Supposing the surrounding tissue restrains the artery elastically in the longitudinal direction according to Hooke's law, the shear force on the artery is related to the displacement and stiffness of the surrounding tissue. Solving the resulting differential equation for displacement as a function of position yields a hyperbolic equation. The derived equation can be used to predict the displacement in a region of tissue with constant stiffness. The interface between two regions of tissue corresponding to the PFC and soft core of a CTO should be visible as a change in displacement slope based on a change in the parameters of the hyperbolic curve. The displacement response of a simulated CTO with a PFC and soft core is shown in Fig. 1 as determined by FEA, and verified with the derived hyperbolic equations.

METHOD: A CTO phantom consisting of a soft agar/gelatin mixture embedded with a hard inclusion of agar to mimic the hard fibrous cap of an artery was subjected to cyclic deformation within an MR scanner. The loading apparatus was designed as a pneumatic system connected to an electronically controlled valve in order to time cyclic displacements with the MRI scanner acquisition. Displacement was delivered to the surface of the phantom by a 5 mm diameter plastic rod. A stimulated echo imaging pulse sequence (STEAM) was used for this study, since it allowed transients due to compression to come to rest within a window of time called the mixing time [3]. The pulse sequence consisted of three 90° RF pulses, with the second and third pulse separated by the mixing time *Tm*. The piston of the pneumatic cylinder compressed the sample within the mixing time and remained activated until after the readout for each cycle. Phase images were acquired in the coronal plane, through the center of the lesion. Two phase images were subtracted from each other in order to eliminate errors due to B_0 inhomogeneity, and the resulting phase was converted to displacement.

Fig. 2. a) Displacement map showing displacement in the +*x*-direction through a cross section of the CTO phantom. The PFC is shown outlined in red, while the surrounding tissue is in blue. b) Displacement data plotted along the *x*-axis shown in (a), as well as a mildly smoothed displacement curve determined using a smoothing spline in Matlab with a *p*-value of 0.9.





RESULTS: A displacement image was obtained through a cross section of the CTO and agar gel as shown in Fig. 2a. Displacement along a single column of data was taken through the center of the image labeled as the *x*-axis, and is shown in Fig. 2b. The shape of the displacement curve indicates a relatively constant section in Region A, followed by a sudden drop and gradual change towards zero in Region B. The region of relatively constant displacement represents the hard cap of the lesion, and the sudden drop indicates a transition to a softer gel. From this transition, the interface between the hard and soft gels is shown as the red dashed line to be around 13 mm from the surface of the occlusion, which is in good agreement with the measured hard cap length of 12.5 mm.

DISCUSSION/CONCLUSION: The theory and simulation presented showed that certain key features indicated a transition between tissues of high and low stiffness. This included a change in slope in the displacement plot from low values for hard tissue to high values for soft tissue. Sharp changes in the slope of the displacement curve indicated changes in tissue stiffness, and gradual changes were due to overall absorption of displacement with tissue length. A measure of the bulk tissue stiffness in separate regions can be determined by fitting hyperbolic functions derived from theory to the measured displacement. These values of average Young's modulus in each region may be useful for guidewire selection. For more detailed mapping of the stiffness within a lesion, it would be necessary to use methods such as inverse strain mapping [2]. Future work for this study will focus on transitioning towards validation of the presented theory through static elastography on ex vivo samples of occluded arteries.

REFERENCES: [1] Strauss B *et al*., J of Interv Cardiol, 2005;18:425-436. [2] Samani A *et al*., IEEE Trans Med Imaging, 2001;20:877 – 885. [3] Plewes D *et al*., Phys Med Biol, 2000;45:1591–1610, Jan 2000.