Validation of Distensibility Measurements from MRI-based Pulse Wave Velocity

G. Acevedo-Bolton[1], J. Leung[1], C. Hardy[2], C. DeVincent[3], D. Saloner[1,4], and A. Meadows[3]
[1]Radiology, UCSF, San Francisco, CA, United States, [2]GE Global Research, Niskayuna, NY, United States, [3]Radiology, [4]Radiology, VA Medical Center, San Francisco, CA

Introduction:

The use of Fourier-velocity-encoding(FVE) M-mode MRI to measure pulse wave velocity (PWV) in arteries has great potential in determining vessel health as PWV provides an indirect measure of distensibility [1]. Previous reports have shown compelling results indicating the utility of this method in vivo. However, the actual values of arterial compliance in vivo are generally unknown, and because of the difficulty in obtaining arterial specimens following in vivo imaging, a systematic validation of the technique has not yet been performed. The aim of this study is to develop an in-vitro model for validating the FVE based distensibility calculations.

Methods:

A computer-controlled gear pump was used to generate a pulsatile flow of water through a custom built flow loop. The waveform could be programmed to generate a time-dependent shape that matched the flow waveform that is generally measured for blood flow in the ascending aorta in vivo. Four latex tubes with varying wall thickness and hence differing elasticity were evaluated. The PWV was determined for each tube by connecting a 2 foot length of the tubing into the flow loop and subjecting that length of tube to the pulsatile flow input. The latex tube was centered in the bore of a GE Signa HDx 1.5T magnet and 1D M-mode imaging was performed using the cardiac surface coil. The FVE protocol was performed using a 2cm diameter pencil beam, 12 windings, 4 interleaves, venc=75cm/s, TE=6ms, flip angle=45deg. PWV velocity was measured and used to calculate distensibility using the formula $D=1/(\rho\ v^2)$ where ρ=water density and v=pulse wave velocity. Young's modulus (E) for each tube was obtained using mechanical testing both with a linear stretch system and a biaxial stretcher. Distensibility was also calculated from the measured value of E using the formula $D= 2(a_0/h)/E$ where a_0=internal radius of latex tube and h=wall thickness. The distensibility value measured using mechanical testing was plotted against the inverse of the square of the PWV.

Results:

Distensibility increased with decreasing wall thickness in both sets of calculations (Fig 1). The measured pulse wave velocity was found to be proportional to the square root of the distensibility estimated from the measured Young's modulus, as expected. Various features of the waveform can be used in the calculation of the PWV, and we followed the prescription of Hardy where the foot of the systolic peak was used. In this flow model, and in the absence of background tissue, we found that measure to be sensitive to noise. This can be ameliorated by suitable postprocessing and by using multiple waveform features to analyze the passage of the pulse.

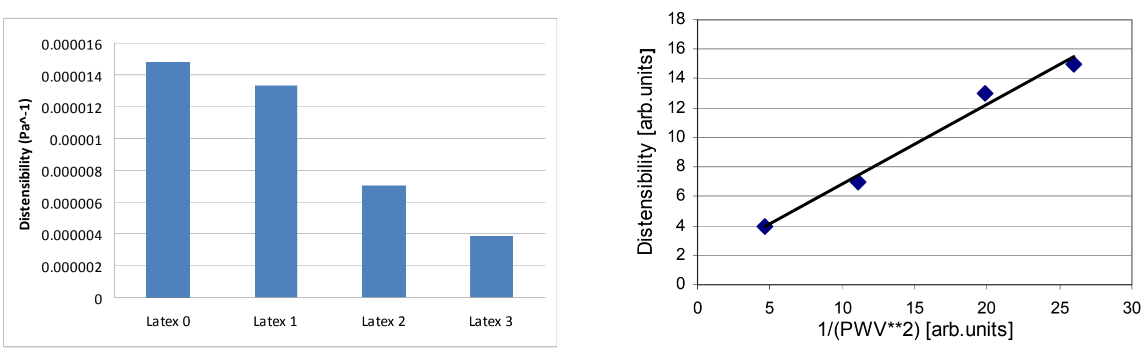

Figure 1. A) Distensiblity calculated from Mechanical testing of Young's modulus. Increasing latex number denotes an increasing wall thickness. B) Distensibility values calculated from mechanical testing are plotted against the inverse of the square of the PWV obtained from FVE M-mode MRI.

Conclusions:

The distensibility determined from the PWV as measured by MRI showed a linear correlation with distensibility calculated from the mechanical testing. This validation increases confidence in the ability of this technique to accurately assess distensibility in vivo, and hence the presence of arterial disease.

Reference:

1)Hardy CJ, et al. *MRI determination of Pulse Wave Velocity in the Carotid Arteries* Proc ISMRM 16 (2008)