Validation of finite-element stress analysis of Abdominal Aortic Aneurysm using dynamic MRI

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Introduction

An abdominal aortic aneurysm (AAA) is a life-threatening dilatation of the aorta. In clinical practice, the maximum transversal AAA diameter is used to assess its rupture risk and to decide whether or not surgical repair is required. There is however a strong indication that knowledge about the AAA wall stress can provide more accurate rupture-risk prediction than the maximum diameter [1]. In recent years, we have developed a finite-element analysis (FEA) methodology to derive the patient-specific AAA wall strain and stress from dynamic MRI acquisitions [2]. In this paper, we describe how we have validated our finite-element calculations.

Methods

10 male patients (average age 75) with AAA diameter of more than 5.5 cm, all scheduled for surgical repair, were scanned with cardiac triggered 2D and 3D Steady-State Free Precession (SSFP) MRI (Philips Intera 1.5T R10.4, B-TFE, TE 4.5 ms, TR 2.2 ms, flip angle 50 degree, 25 (2D) or 50 slices (3D), voxel size 1.34x134 mm^2, slice thickness 3 mm (2D) or 6 mm (3D), 15 phases/cardiac cycle) [2]. The blood pressure in the aorta was monitored during imaging using a pressure wire mounted on a catheter. After rigid registration of the 2D and 3D MR image data, the AAA outer wall was automatically segmented for the end-diastolic phase, using the deformable model approach of [3] driven by features from all MRI data. Figure 1 shows an example of an MR image with a 3D outer wall segmentation.

The thickness of the AAA wall could not be estimated from the acquired MRI data, due to its limited resolution and due to the fact that the image intensity of the wall is very similar to that of thrombus connected to the wall. Therefore a wall of constant thickness of 2 mm was assumed. A 3D volume mesh of the wall was created, consisting of 15-node quadratic Crouzeix-Raviart tetrahedrons (total around 20,000 nodes per AAA wall) [4].

Biomechanical wall-stress simulations were performed with the Sepran finite-element analysis software (Sepra, Den Haag, The Netherlands) using the non-linear wall material model of [5]. The AAA blood pressure measured during imaging was used as boundary condition for the simulations. In most published wall-stress simulations, the initial stress in the wall at end diastole is ignored, i.e. the full difference between zero and systolic pressure is applied to the end-diastolic zero-stress wall. This will lead to overestimation of the AAA wall motion. We compared this approach with the one proposed in [6], which does estimate and include initial stress. Figure 2 shows an example of simulated stress (colour overlay of stress on the 3D AAA model).

The simulated AAA outer wall motion was compared to the motion measured from the dynamic 2D/3D SSFP MRI data as follows. The end-diastolic 3D outer wall segmentation was first converted to one outer wall contour per image slice. The resulting contours were automatically propagated to all other phases in the cardiac cycle using the approach of [7]. For each of the phases and slices, the average in-slice distance between simulated and measured contours was thereafter used to evaluate their motion similarity.
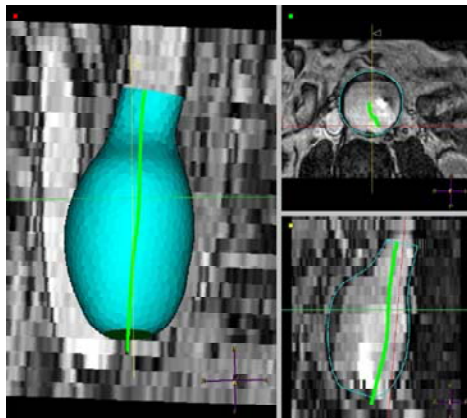

Figure 1 – MRI with AAA wall segmentation

Figure 2 – Wall stress
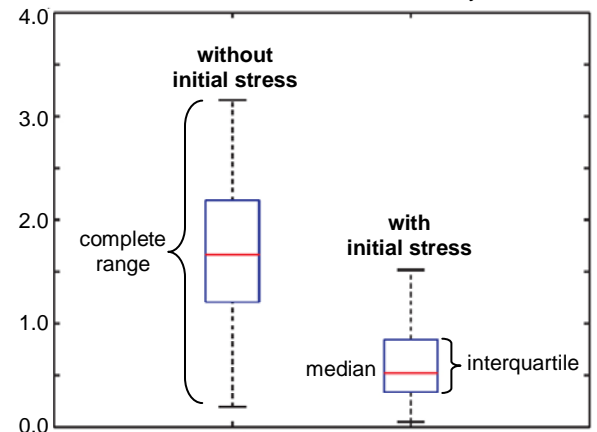(red = high, blue = low)

Figure 3 – Error between measured and simulated motion

Results

The simulated wall motion equals the measured motion much better when taking the initial wall stress into account. With initial stress, the (median) distance between simulated and measured contours is only 0.5 mm versus 1.7 mm without initial stress (see figure 3).

Conclusions

We have presented the validation of our AAA finite-element wall stress calculations. We found a good correspondence between simulated and measure AAA wall motion when taking the initial stress in the AAA wall at end diastole into account during simulation.

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