**Proceedings of the ASME 2013 Summer Bioengineering Conference**
**SBC2013**
**June 26-29, 2013, Sunriver, Oregon, USA**

**SBC2013-14716**

# EXPERIMENTAL VALIDATION OF A COMPUTATIONAL ALGORITHM FOR THE ZERO PRESSURE GEOMETRY DERIVATION OF BLOOD VESSELS

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## INTRODUCTION

Patient-specific computational assessment of biomechanical parameters such as peak wall stress is a promising tool for rupture risk assessment of blood vessels. However, this assessment is dependent on image based modeling of the vasculature [1] and on either structural or fluid-structure interaction analyses performed with numerical models to compute the stress and strain in the vascular wall. Protocols have been successfully derived to develop 3D models of normal and pathological vessels from individual Computed Tomography (CT) or Magnetic Resonance Imaging (MRI) [2]. While the image based models used for these simulations are essentially in a pressurized state (gated to diastolic pressure), the application of physiologic systolic and diastolic pressures to compute stresses and strains is debatable. Therefore, the derivation of a "simulation ready" computational geometry is of great importance to the research community as the accuracy of the computational results is dependent on it.

We have developed a pull-back algorithm for reconstructing the unloaded vascular geometry corresponding to zero intraluminal pressure, termed here as the zero pressure (ZP) geometry. Details of the algorithm can be found in [3, 4]. In short, our two stage algorithm is designed to utilize the initial computational mesh of the image-based (IB) geometry (ideally it represents the diastolic geometry) for reconstruction of the ZP geometry. In the first stage, the algorithm develops an initial approximation of the nodal coordinates of the ZP geometry mesh by extrapolating the displacements of each wall node obtained from the structural simulation of the IB geometry at diastolic pressure. In the second stage, a fixed point iterative algorithm is implemented to make incremental corrections to the nodal coordinates and to minimize the iteration error between the zero pressure geometry (pressurized at diastolic pressure) and CT image based geometry. The iteration error is essentially an average of the relative measure of the nodal distances between these two geometric configurations.

In this work we present an experimental framework for the validation of the ZP computational algorithm with the objective of proving that it is successful in reconstructing the zero pressure geometry of a blood vessel originating from a clinical image based geometry at diastolic pressure.

## MATERIAL & METHODS

We used a phantom model of idealized axisymmetric abdominal aortic aneurysm (AAA) geometry (Fig. 1a) and applied the proposed algorithm on the pressurized configuration of this phantom to reconstruct the zero pressure (ZP) geometry. Our strategy was to: 1) obtain micro CT images of the AAA phantom at unpressurized (0 mmHg) and pressurized conditions, the latter with internal pressures of 80 mmHg, 120 mmHg and 140 mmHg; 2) develop image based models and meshes from micro CT image data; 3) apply the proposed algorithm to the pressurized geometry to reconstruct the ZP geometry; and 4) compare and validate the reconstructed ZP geometry with the CT image based geometry obtained at the unpressurized condition. A schematic of the experimental setup is shown in Fig. 1b.

The phantom was developed using a silicone elastomer (Applied Silicone Corp., Santa Paula, CA) with a modulus of elasticity, E = 1170 kPa. The measured average wall thickness was 0.1365 cm, the maximum aneurysm diameter was 4.7 cm and total length of the phantom was 26 cm. As shown in the schematic, a balloon (initial thickness = 0.02 cm) was inserted into the AAA phantom and pressurized using a hand pump until the desired intraluminal pressure was achieved in the aneurysm sac, monitored using a pressure gauge. The phantom was then inserted into a μCT scanner (Skyscan 1076 in vivo scanner, Bruker Corporation, MA) and imaged at the unpressurized condition (0 mmHg) and then repeated for the three different pressurized conditions (80 mmHg, 120 mmHg, and 140 mmHg). The scan resulted in 2,996-3,010 images with a pixel resolution of 35 µm and slice spacing of 35 µm. Scanned images were obtained as image files (.png), which were then converted to DICOM files before the commercial software Mimics (Materialize NV, Belgium) was used for automatic segmentation and lumen surface generation. Our in-house code 'AAAMesh' [5] was then used to develop the mesh with hexahedral elements (Fig. 1c). A uniform wall thickness of 1.365 mm was assumed for these models during mesh generation. The element size was optimized through a mesh sensitivity study, which resulted in a mesh size in the range 40,242 – 41,952.

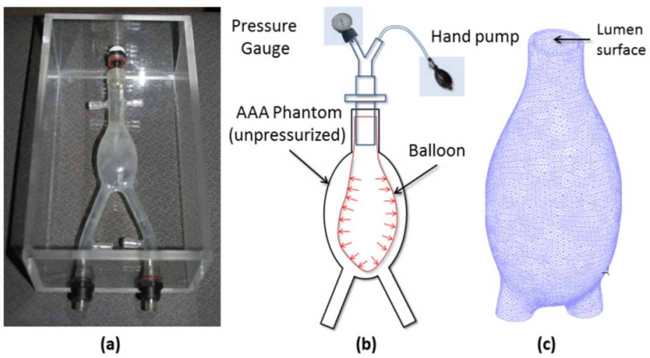


**Figure 1: a) Phantom replica of AAA; b) Schematic diagram of experimental setup for zero pressure algorithm validation; c) Image based computational geometry of AAA phantom with hexahedral mesh used for structural simulations.**

The pressurized phantom meshes were subject to the pull-back algorithm for reconstruction of the ZP geometry. The algorithm was coded in Matlab 2012 and the structural simulations performed with the finite element solver ADINA 8.8 (Adina R&D Inc., Watertown, MA). Although the phantom wall and the balloon form a composite laminate structure, the effective Young's modulus calculated using classical mechanics did not change significantly (1,171 kPa), given the relatively thin balloon compared to the phantom's wall thickness.

For each of the pressurized meshes, the iterative algorithm was executed until the iteration error was less than 0.5%. The lumen surface of the ZP geometries was then compared with the lumen surface of the AAA phantom at the unpressurized condition, qualitatively and quantitatively. The relative matching of lumen surfaces was visualized using a part comparison feature in 3-Matic (Materialise NV, Belgium). A directional Hausdorff distance (**dHd**) was further calculated between the nodes of the two lumen surfaces to quantify the mismatch. dHd is a measure of closeness between two sets of points (e.g., P and Q). Intuitively, dHd finds the point p (from set P) that is furthest from any point in set Q and measures the distance from p to its nearest neighbor in Q. However, this calculation was performed from P to Q and again from Q to P and the maximum of those two values is reported as dHd.

## RESULTS AND DISCUSSION

The ZP algorithm was applied to the three pressurized phantom meshes (at 80mmHg, 120 mmHg, and 140 mmHg) and our algorithm converged in 6, 7 and 8 iterations, respectively. The convergence of the algorithm is presented in Fig. 2a with the iteration error (%) as a function of the number of iterations required to achieve convergence. Figure 2b illustrates, qualitatively, a comparison between the ZP geometry obtained from the phantom at 80 mmHg (ZP geometry/80mmHg) and the lumen surface of the phantom geometry at the unpressurized condition (Phantom@0mmHg). From Fig. 2c, it can be seen that the two surfaces matched well on the right and left sides but a discrepancy was noticed on the anterior surface (right hand side of Fig. 2b). The ZP geometry was found to underestimate the phantom geometry by a maximum distance of 2.28 mm (measured in 3-Matic). A qualitative description of the discrepancy is shown in Fig. 2d with the distribution of the distance between the two surfaces. Similar qualitative behavior was observed for the ZP geometries obtained from the other two pressurized conditions (120 mmHg and 140 mmHg). The dHd calculated for the three ZP geometries were 2.3 mm, 3.0 mm, and 3.3 mm, indicating that the phantom pressurized at 80 mmHg resulted in the derivation of the most accurate ZP geometry.

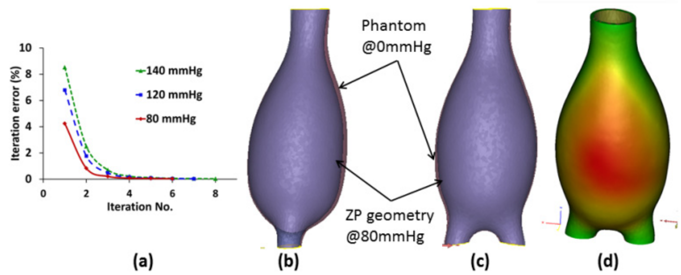


**Figure 2: a) Convergence of ZP algorithm applied to phantom geometries pressurized at 80, 120 and 140 mmHg; b) front view and c) right view of superimposed lumen surfaces of ZP geometry/80mmHg and Phantom@0mmHg; d) spatial distribution of nodal distances between the two surfaces (red represents maximum distance).**

It should be noted that the aforementioned discrepancy on the anterior surface is due to the phantom being in contact with the CT scanner tray during the scanning process. In the structural simulations the inlet and outlet boundary nodes were fixed, therefore nodal displacements in the sac region was forced to be uniform on all sides. We believe the contact of the phantom with the tray led to the mismatch between the actual and predicted ZP geometries. A future study should be performed where a rigid surface should be brought in contact with the anterior wall of the phantom to mimic the experimental setup before applying any intraluminal pressure.

## CONCLUSION

In this work we present the experimental validation of our pull-back algorithm, developed to obtain the ZP geometry of a blood vessel starting from an image based mesh at diastolic pressure. Using a compliant AAA phantom, we demonstrated that the algorithm is capable of reconstructing the ZP geometry of the AAA using intraluminal pressures of 80 mmHg, 120 mmHg, and 140 mmHg, with an iteration error less than 0.05% and a maximum directional Hausdorff distance of 3.3 mm.

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