Hemodynamic Impact of Transcatheter Aortic Valve Deployment Configuration

Eric Sirois

Kewei Li

Joe Calderan

Lisong Ai

Wei Sun

Tissue Mechanics Laboratory,
University of Connecticut,
Storrs, CT 06269

1 Introduction

Transcatheter aortic valve (TAV) replacement holds promise for a large number of patients who otherwise have limited or no treatment options. However, it also poses various challenges, due to its unique disease treatment mechanism. Successful TAV deployment and function are heavily reliant on the tissue-stent interaction [1,2]. For patients with aortic stenosis, heavy calcium deposition on the valve leaflets and the aortic root can also cause distortion of TAV geometries, resulting in a valve of an elliptical shape [3–5] instead of a nominal circular shape. In a recent study by Schultz et al. [5], the geometry and apposition of the TAV after implantation in 30 patients with aortic stenosis were evaluated using multislice computed tomography. The results indicated that none of the TAV frames reached exactly nominal designed dimensions. The difference between the orthogonal smallest and largest diameters of TAV cross section at the ventricular end was 4.4 mm, which was clearly an asymmetric elliptical shape [5]. In this study, we sought to develop computational models of valve hemodynamics under clinically relevant deployment scenarios.

2 Methods

The geometry of a “generic” TAV has been described in detail [6]. It is a thin pericardial bioprosthetic valve with similar design features to TAVs used commercially. This model has been extended previously to the development of elliptical valves, with eccentricities of 0.3, 0.5, and 0.68 and alignment of either the major or minor axis of the ellipse to a native commissure [7]. In the current study, the open configuration for each of these elliptical models was obtained using ABAQUS finite element simulation (Fig. 1(a)). The leaflets were attached to a circular mounting ring (Fig. 1(b)) and then sealed in a 38-mm tube (Fig. 1(c)) in preparation for import into computational fluid dynamics (CFD) software using Hypermesh (Altair Engineering, Troy, MI).

The CFD software STAR-CCM+ was used, with a polyhedral mesh and boundary conditions of 30 liters per minute (LPM) mass flow at the inlet and 100 mmHg pressure at the outlet. A Newtonian fluid was used, with a viscosity of 0.0035 Pa·s and a density of 1056 kg/m³. Turbulence was modeled by specifying intensity and length scale for the k - ϵ model.

The sealed geometry was also used to fabricate physical models through additive manufacturing, which were then used for experimental validation using steady forward flow at 30 LPM. A centerline probe was advanced upstream, measuring pressure at 1-mm spacing.

Fig. 1 (a) Finite element simulation of valve opening, (b) leaflets placed on mounting ring, and (c) mounting ring placed in 38-mm tube

Fig. 2 Validation of simulation using experimental centerline pressure drop curve. Axial position of zero corresponds to test valve annulus.

3 Results/Discussion

For the baseline circular valve, a peak transvalvular pressure of 10.92 mmHg was simulated (Fig. 2), which correlated to an effective orifice area (EOA) of 2.93 cm². In a long-axis view, the peak centerline velocity was 2.16 m/s and the peak turbulent kinetic energy (TKE) was 0.30 J/kg. We observed that EOA dropped for more elliptical conditions, particularly when the minor axis of the ellipse was aligned with the TAV commissures. The changes in flow were also observed as higher transvalvular pressure, higher peak velocity, and higher TKE as compared with the baseline circular condition.

4 Conclusion

Experimental pressure versus axial position curves can be used to validate steady-flow models. Best turbulence modeling was obtained for 10% turbulent intensity and a length scale of 4.6 mm. TAV ellipticity tended to cause increased EOA with higher peak velocity. These models may be used in the future to investigate more complicated clinical scenarios.

Manuscript received September 27, 2013; final manuscript received October 14, 2013; published online December 5, 2013. Assoc. Editor: Jeffrey E. Bischoff.

Acknowledgment

The authors gratefully acknowledge the funding provided by the NSF, NIH, and Edwards Lifesciences, Inc.

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