INTRODUCTION

For patients with severe aortic valve disease, valve replacement via open-chest surgery has long been the preferred method of treatment. For patients with comorbidities or advanced age, the risk of intraprocedural mortality can outweigh the potential benefit of valve replacement and these patients are often denied surgery [1]. Recently, minimally-invasive percutaneous aortic valve (PAV) replacement has emerged as a viable alternative for these patients. Although significant experience with PAV procedures has been gained through multicenter clinical trials, various adverse effects have been detected after device implantation [2]. This study documents an investigation of the hemodynamic environment changes that occur following PAV intervention using a combination of finite element analysis (FEA) and computational fluid dynamics (CFD).

METHODS

PAV-aortic root model. The PAV leaflet geometries used in this study were detailed in a previous study [3]. A 13 mm high rigid stent was added around the valve leaflets, having a stent thickness of 0.5 mm, resulting in a size 23 PAV. The geometry of an aortic root was obtained using ECG-gaited clinical cardiac 64-slice multidetector row CT scans collected at Hartford Hospital, CT, with IRB approval. A 68 year old male with no known comorbidities was selected for this study. The patient’s aortic annulus size was determined from image analysis to be 21 mm. The patient’s aortic valve and root geometries were digitized and reconstructed using custom in-house software.

Finite element simulation of PAV deployment. The geometries developed above were imported into the finite element analysis software ABAQUS (DSS Simulia, Warwick, RI). Aortic root, aortic valve leaflet, and ascending aorta material properties were assigned using linear elastic models with Young’s moduli of 1200-1500kPa and Poisson ratio of 0.45. Note that since the final deformed aortic root and leaflet geometries were determined by the stent expansion, using either linear elastic or nonlinear hyperelastic tissue material properties for the root and leaflet would not have had a significant impact on their final geometries. The loading condition was assigned to expand the aortic annulus region to its full diameter of 23 mm, designed to mimic the clinical practice of over-sizing [4]. The stent was deployed such that approximately one third of the stent was below the aortic annulus. Contact modeling was accomplished using master-slave pairs of the PAV stent and the aortic root wall.

CFD physics modeling. A shear-rate dependant non-Newtonian fluid with a constant density of 1056 kg/m³ was used. To implement the non-Newtonian model, a power-law shear-rate dependant relationship was used [5]. The solution was obtained using the k-epsilon turbulence model, solved by the commercial CFD package Star-CCM+ (CD-Adapco, Plymouth, MI). The velocity-pressure coupling method was the SIMPLE algorithm. A second order differencing scheme was chosen for resolving velocity and turbulence quantities. The unsteady solution was discretized into 1 ms intervals and solved using the Unsteady Reynolds-Averaged Navier-Stokes model.

Implementation of boundary conditions. The initial boundary conditions for this patient were physiologically appropriate velocity inlet (at the aortic annulus) and pressure outlets (at the ascending aorta and coronary arteries) [6, 7]. To develop accurate boundary conditions following...
PAV intervention, the pre-interventional aortic annulus velocity profile ($u$) was scaled in accordance with an annulus diameter increase from 21 mm to 23 mm and a heart rate (HR) increase from 73 bpm to 91 bpm [8]. This scaling was accomplished by setting the net volume change over one cardiac cycle ($\Delta V$), i.e. the stroke volume, to be the volumetric flow rate ($Q$) integrated over time:

$$\Delta V = \int Q_t \delta t = \Delta V_2 = \int Q_2 \delta t_2$$  \hspace{1cm} (1)

By continuity, the post-intervention velocity ($u_{post}$) at any time increment was calculated by:

$$u_{post} = \frac{u_{pre}A_{pre}(HR)_{post}}{A_{post}(HR)_{pre}}$$  \hspace{1cm} (2)

The aortic pressure outlet was converted into a three-element Windkessel model, containing two resistor elements ($R_p$ and $R_c$) and one capacitor element ($C$). Windkessel in the time domain was accomplished for each time ($t$) by [9]:

$$\frac{dPao}{dt} = \frac{(R_p + R_c)Qao}{R_p C} + R_c \frac{dQao}{dt} - \frac{P_{t-1}}{R_p C}$$  \hspace{1cm} (3)

The coronary artery boundary conditions were replaced with unique time-varying resistive elements for each outlet. These resistances were used to prescribe each coronary flowrate ($Q$) as a function of its corresponding reservoir pressure ($P_{res}$) and resistance ($R$) such that $Q = -P_{res}/R$. Reservoir pressure was calculated for each coronary branch as a function of observed pressure ($P$) and velocity. At any time, $P_{res}$ was given as [7]:

$$P_{res} = \left( \int_0^t P(t) + P_{t=0} \right) e^{-(a+b)t}$$  \hspace{1cm} (4)

The values $a$ and $b$ are constants obtained from continuity requirements and direct curve fitting.

**RESULTS AND DISCUSSION**

**Model validation.** The model was validated where possible by comparison with data reported from clinical use of PAVs. Pre-operatively, we observed from the CFD simulation a max transvalvular pressure gradient of 77.9 mmHg and a mean systolic ejection period transvalvular pressure of 45.8 mmHg, which compare favorably with the 78+/−19 and 45+/−12 mmHg observed by Bauer et al. [8]. We also observed a simulated EOA of 0.696 cm$^2$. These values also compare favorably with PAV replacement candidate hemodynamic data reported by Clavel et al. of 47+/−17 mmHg mean pressure gradient and 0.6+/−0.14 cm$^2$ EOA [10] and by Ben-Dor et al. of 43+/−13 mmHg mean pressure gradient and 0.63+/−0.14 cm$^2$ EOA [11]. Thus, the patient examined here was hemodynamically equivalent to other PAV replacement candidates.

Post deployment, we obtained the peak transvalvular pressure gradient of 22.63 mmHg which compared favorably with the 20+/−7 mmHg observed by Bauer et al. [8]. The mean transvalvular pressure gradient observed of 11.52 mmHg was slightly higher than the 8+/−2 mmHg from Bauer et al. [8], well higher than the 3.88+/−1.62 mmHg from Ben-Dor et al. [11], and within the 10+/−4 mmHg from Clavel et al. [10]. Also, we observed a post-deployment EOA of 1.56 cm$^2$, compared with 1.69+/−0.11 cm$^2$ from Bauer et al. [8], 1.97+/−0.41 cm$^2$ from Ben-Dor et al. [11], and 1.61+/−0.40 cm$^2$ from Clavel et al. [10]. These comparisons indicated that the simulated post-deployment hemodynamics were similar to those observed clinically following PAV intervention.

**Flow analysis.** Flow conditions were examined before and after the simulated PAV intervention (Fig.1). Although peak velocity values were similar, the magnitude and length of the central jet were significantly reduced by PAV intervention. Additionally, the displaced native leaflets appear to cause a low pressure region that resulted in virtually no change in coronary artery flow despite a 25% increase in cardiac output following intervention.

**REFERENCES**


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