Published in final edited form as:

J Biomech. 2015 September 18; 48(12): 3026–3034. doi:10.1016/j.jbiomech.2015.07.031.

Comparison of transcatheter aortic valve and surgical bioprosthetic valve durability: a fatigue simulation study

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Abstract

Transcatheter aortic valve (TAV) intervention is now the standard-of-care treatment for inoperable patients and a viable alternative treatment option for high-risk patients with symptomatic aortic stenosis. While the procedure is associated with lower operative risk and shorter recovery times than traditional surgical aortic valve (SAV) replacement, TAV intervention is still not considered for lower-risk patients due in part to concerns about device durability. It is well known that bioprosthetic SAVs have limited durability, and TAVs are generally assumed to have even worse durability, yet there is little long-term data to confirm this suspicion. In this study, TAV and SAV leaflet fatigue due to cyclic loading was investigated through finite element analysis by implementing a computational soft tissue fatigue damage model to describe the behavior of the pericardial leaflets. Under identical loading conditions and with identical leaflet tissue properties, the TAV leaflets sustained higher stresses, strains, and fatigue damage compared to the SAV leaflets. The simulation results suggest that the durability of TAVs may be significantly reduced compared to SAVs to about 7.8 years. The developed computational framework may be useful in optimizing TAV design parameters to improve leaflet durability, and assessing the effects of underexpanded, elliptical, or non-uniformly expanded stent deployment on TAV durability.

Keywords

transcatheter aortic valve; bioprosthetic heart valves; soft tissue fatigue damage; finite element analysis

INTRODUCTION

Transcatheter aortic valve (TAV) intervention is now the standard-of-care treatment for inoperable patients and a viable alternative option for high-risk patients with symptomatic

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CONFLICT OF INTEREST

None.

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aortic stenosis (**AS**) (Haussig, Schuler et al. 2014). Since the first TAV procedure in 2002 (Cribier, Eltchaninoff et al. 2002), TAVs have been implanted in more than 150,000 patients worldwide (Zhao, Jilaihawi et al. 2014). Short and mid-term clinical results are promising: TAV intervention can significantly improve valve hemodynamics and patient quality of life with the added benefits of lower operative risk and shorter recovery time compared to surgical aortic valve (**SAV**) replacement (Kodali, Williams et al. 2012, Milburn, Bapat et al. 2014). However, little is known about the long-term durability of these devices, owing in part to the relative immaturity of TAV intervention and the advanced age and illness of the patients selected for this treatment. To date, TAV intervention is still not considered for lower-risk patients: SAV replacement with either a mechanical or bioprosthetic valve remains the gold standard.

Bioprosthetic SAVs display superior hemodynamics to mechanical valves and eliminate the need for anticoagulant therapy. The caveat for these valves is the durability of the tissue leaflets due to calcification or fatigue-induced structural deterioration over time. The second generation pericardial valve from Edwards Lifesciences, the Carpentier-Edwards Perimount (**CEP**), has an *in vivo* durability of up to 20 years, which is a significant improvement over first generation valves such as the Ionescu-Shiley (**IS**) valve which has since been taken off the market for its subpar durability of approximately 5 years (Gabbay, Bortolotti et al. 1984, Gabbay, Bortolotti et al. 1984, Brais, Bedard et al. 1985, Reul Jr, Cooley et al. 1985, Cooley, Ott et al. 1986, Nistal, Artinano et al. 1986, Nistal, Garcia-Satue et al. 1986). The improved durability of the CEP valve has been attributed to valve design factors, such as the flexible stent which acts as a cushion to reduce leaflet stresses (Singhal, Luk et al. 2013), thicker leaflets, and improved leaflet coaptation (Vesely 2001).

TAV leaflets are made of similar materials as the IS and CEP SAVs, generally either glutaraldehyde-treated bovine pericardium (**GLBP**) or porcine pericardium; however, TAV leaflets must be thinner to permit transcatheter delivery (~0.25 mm compared to ~0.4 mm in SAVs). Furthermore, TAV stents do not allow for deflection to cushion the leaflets. It has been shown through finite element (**FE**) analysis that due to these constraints, TAV leaflets experience higher stresses and strains (Li and Sun 2010) than traditional SAV leaflets, particularly in the presence of aortic calcification resulting in non-circular, asymmetric stent deployment (Sun, Li et al. 2010, Gunning, Vaughan et al. 2014). Therefore, TAV devices can be expected to have reduced durability compared to SAVs. However, while there have been numerous biomechanics studies on TAV devices, they have been confined to the TAV function (Dwyer, Matthews et al. 2009, Li and Sun 2010, Sun, Li et al. 2010, Gunning, Vaughan et al. 2014, Kuetting, Sedaghat et al. 2014) and interaction with the surrounding tissue (Dwyer, Matthews et al. 2009, Capelli, Bosi et al. 2012, Wang, Sirois et al. 2012, Auricchio, Conti et al. 2014, Morganti, Conti et al. 2014, Wang, Kodali et al. 2015) immediately following implantation. The durability and potential failure modes of TAV devices remain unknown. Before TAV intervention can be considered a potential alternative treatment for lower-risk patients with longer life expectancies, there is a pressing need for new strategies to evaluate and improve leaflet durability.

The objectives of this study were two-fold: first, to develop a computational framework for assessing TAV leaflet fatigue under cyclic loading and second, to compare leaflet fatigue in

a TAV and SAV under identical loading conditions and with identical leaflet material properties. This computational approach to assess leaflet fatigue is advantageous in terms of efficiency (i.e. hours of simulation time versus the months it may take to fabricate and test valves via accelerated wear testers), and also because it enables well-controlled, side-by-side engineering design comparisons. For instance, the precise effects of various design features (leaflet shape, free edge height, leaflet thickness, etc.) on TAV leaflet durability can be quantitatively assessed and compared to that of the CEP SAV under the same loading conditions.

METHODS

Constitutive modeling of tissue fatigue

In this study, GLBP was selected as a representative valve leaflet material. GLBP is comprised of stiff collagen fibers embedded in a compliant matrix of elastin and proteoglycans, and thus can be considered a fiber-reinforced continuum. Accordingly, the total tissue free energy, W , was decomposed into isochoric, W_{iso} , and volumetric parts, W_{vol} , as

$$W(\mathbf{C}) = W_{iso}(\bar{\mathbf{C}}) + W_{vol}(\mathbf{C}), \quad (1)$$

where $\mathbf{C}$ is the right Cauchy-Green tensor, $\bar{\mathbf{C}}$ is the deviatoric right Cauchy-Green tensor, and W_{iso} is further decomposed into distinct matrix and fiber contributions denoted with “ m ” and “ f ” subscripts respectively giving

$$W_{iso} = W_m(\bar{\mathbf{C}}) + W_f(\bar{\mathbf{C}}, \mathbf{M}), \quad (2)$$

where $\mathbf{M}$ is a structural tensor describing the fiber orientation.

Fatigued state tissue free energy function—Details on the constitutive modeling of the soft tissue fatigue response were described previously (Martin and Sun 2012, Martin and Sun 2013). Briefly, to incorporate changes to the valve leaflet material properties as a result of fatigue damage, W , was enhanced with the addition of a stress-softening parameter, D_s , and a permanent set parameter, $\mathbf{D}_{ps}$, given by

$$W(\mathbf{C}, D_s, \mathbf{D}_{ps}) = (1 - D_s) W_{iso}^0(\bar{\mathbf{C}}, \mathbf{M}) + W_{ps}(\bar{\mathbf{C}}, D_s, \mathbf{D}_{ps}) + W_{vol}^0(\mathbf{C}), \quad (3)$$

where W_{ps} is the dissipated energy due to the permanent set, and W^0 indicates the initial (un-fatigued) strain energy. At the un-fatigued state, both D_s and $\mathbf{D}_{ps}$ are inactive, i.e. $D_s = 0$ and $\mathbf{D}_{ps} = \mathbf{0}$, thus W reduces to the strain-energy function, W^0 . The parameters D_s and $\mathbf{D}_{ps}$ become active with the onset of fatigue damage induced by cyclic loading.

GLBP tissue fatigue damage evolution was considered to be a function of the peak equivalent strain per cycle as in our previous studies (Martin and Sun 2012, Martin and Sun 2013). The equivalent strain, $\bar{\epsilon}_t$, (Simo 1987) at time $t \in [0, T]$, a scalar quantity

proportional to the distortional energy, was defined for the matrix and fiber constituents distinctly as

$$\Xi_{t_m}(\overline{C}(t)) := \sqrt{2W_m^0(\overline{C}(t))} \text{ and } \quad (4)$$

$$\Xi_{t_f}(\overline{C}(t), M) := \sqrt{2W_f^0(\overline{C}(t), M)}. \quad (5)$$

The peak equivalent strains for each loading cycle were thus

$$\Xi_{n_m}^{peak} = \max_{t \in [\frac{n}{h}, \frac{n+1}{h}]} \sqrt{2W_m^0(\overline{C}(t))}, \quad n=0, 1, 2, 3, \dots, n_{tot_m} \text{ and } \quad (6)$$

$$\Xi_{n_f}^{peak} = \max_{t \in [\frac{n}{h}, \frac{n+1}{h}]} \sqrt{2W_f^0(\overline{C}(t), M)}, \quad n=0, 1, 2, 3, \dots, n_{tot_f}, \quad (7)$$

where h is the frequency and n is the number of loading cycles up to a maximum number of cycles, n_{tot_m} and n_{tot_f} . The number of cycles until failure (n_{tot}) were defined for the matrix and fiber constituents distinctly as (Martin and Sun 2012):

$$n_{tot_k}(\Xi_{n_k}^{peak}) = \begin{cases} \infty & \text{if } \Xi_{n_k}^{peak} < \psi_{min_k} \\ \frac{\beta_k(n_{max_k}-1)}{\Xi_{n_k}^{peak} - \psi_{min_k} + \beta_k} \left(\frac{1 - \exp\left(\frac{\alpha_k \left(1 - \frac{\Xi_{n_k}^{peak}}{\psi_{max_k}}\right)}{1 - \exp\left(\frac{\alpha_{m,f_i} \left(1 - \frac{\psi_{min_k}}{\psi_{max_k}}\right)}\right)}\right)}{1} \right) + 1 & \text{if } \psi_{min_k} \leq \Xi_{n_k}^{peak} \leq \psi_{max_k}, \quad k=m, f, \\ 1 & \text{if } \Xi_{n_k}^{peak} > \psi_{max_k} \end{cases} \quad (8)$$

where α and β are material constants governing the amount of damage incurred by a single cycle at Ξ_n^{peak} , and ψ_{min} and ψ_{max} , define the boundaries of the fatigue damage evolution zone. Note that α , β , ψ_{min} , and ψ_{max} were also defined for the matrix and fiber constituents distinctly. The total amount of damage due to stress softening, D_s , after n tensile loading cycles was defined as

$$D_{s_k}(\Xi_n^{peak}) = \begin{cases} 0 & \text{if } \Xi_{n_k}^{peak} < \psi_{min_k} \\ \sum_{n=1}^n \frac{1}{n_{tot_k}} & \text{if } \psi_{min_k} \leq \Xi_{n_k}^{peak} \leq \psi_{max_k}, \quad k=m, f. \\ 1 & \text{if } \Xi_{n_k}^{peak} > \psi_{max_k} \end{cases} \quad (9)$$

The tissue permanent set, D_{ps} , was considered to be due to damage to the matrix described by (Martin and Sun 2013):

$$D_{ps_{ij}}(\Xi_{n_m}^{peak}) = \begin{cases} 0 & \text{if } \Xi_{n_m}^{peak} < \psi_{min_m} \\ \sum_{n=1}^n \frac{1}{n_{tot_m}} D_{ps_{max}}^n \frac{E_{n_{ij}}^{peak}}{E_{n_{max}}^{max}} & \text{if } \psi_{min_m} \leq \Xi_{n_m}^{peak} \leq \psi_{max_m} \\ D_{ps_{max}}^n & \text{if } \Xi_{n_m}^{peak} > \psi_{max_m} \end{cases} \quad (10)$$

Here the matrix permanent set is scaled by the peak strain ratio, $\frac{E_{n_{ij}}^{peak}}{E_n^{max}}$, to enforce anisotropy, where $E_{n_{ij}}^{peak}$ is the Green strain at $\Xi_t = \Xi_n^{peak}$ in direction ij , and $E_n^{max} = \max \{E_{n_{ij}}^{peak}\}$. The D_{psmax}^n refers to the maximum permanent set Green strain associated with tensile failure of the material. The permanent set was enforced with a plastic stress, S_P (Martin and Sun 2012).

$$S_P = -\frac{\partial(1-D_s)W^0}{\partial E}\bigg|_{(E=D_{ps})} \quad (11)$$

The S_P contribution to the overall tissue response is governed by η in Eqn. 12, which was modified from Dorfmann and Ogden's η function (Dorfmann and Ogden 2004) in order to accommodate for the dissipated equivalent strain associated with the permanent set, Ξ_{ps} , defined by Eqn. 13,

$$\eta = \frac{\Xi_{t_k} - \Xi_{ps_k}}{\Xi_{n_k}^{peak} - \Xi_{ps_k}} \quad k=m, f, \quad (12)$$

$$\Xi_{ps_k} := \sqrt{2W_k^0(E(t))}\bigg|_{(E=D_{ps})} \quad k=m, f. \quad (13)$$

The second Piola-Kirchoff stress tensor may be expressed in the following reduced form:

$$S = (1-D_s)2\frac{\partial W_{iso}^0}{\partial C} + 2\frac{\partial W_{vol}^0}{\partial C} + (1-\eta)S_P. \quad (14)$$

Un-fatigued strain energy function—GLBP was assumed to be an incompressible, anisotropic, nonlinear, hyperelastic material (Sun, Abad et al. 2005), thus the un-fatigued strain energy was expressed by a fiber-reinforced hyperelastic material model based on the work of Holzapfel, Gasser et al. (2000) and Gasser, Ogden et al. (2006), given by

$$W_m^0(\bar{C}) = c_1 \{ \exp[c_2(\bar{I}_1 - 3)] - 1 \}, \quad (15)$$

$$W_{fi}^0(\bar{C}, M_i) = \begin{cases} \frac{k_1}{2k_2} \left[\exp \left\{ k_2(\bar{I}_j - 1)^2 \right\} - 1 \right] & \text{if } \bar{I}_j \geq 1 \\ 0 & \text{if } \bar{I}_j < 1 \end{cases} \quad i=1, 2, \quad j=4, 6, \quad (16)$$

$$W_{vol}^0(C) = \frac{1}{D}(J-1)^2, \quad (17)$$

where $\bar{I}_{1,4,6}$ are the deviatoric strain invariants, $c_{1,2}$ and $k_{1,2}$ are the matrix and fiber parameters respectively, D is a material constant to enforce near compressibility, and J is the

determinant of the deformation gradient. The fiber orientation was defined by $\mathbf{M}_i = \mathbf{m}_{0i} \otimes \mathbf{m}_{0i}$ with $\mathbf{m}_{01} = [\cos\theta, \sin\theta, 0]$ and $\mathbf{m}_{02} = [\cos\theta, -\sin\theta, 0]$.

Finite element modeling of valves

The FE model of a size 25 mm CEP SAV developed and experimentally validated previously by Sun, Abad et al. (2005) was used for this study. Briefly, each leaflet was modeled using large-strain brick (C3D8H) elements in Abaqus 6.13 (Simulia, RI) FE software, and was prescribed a thickness of 0.4 mm to correspond to commercial CEP valves. The leaflets were attached to a wireframe stent with sutures. Both the stent and sutures were constructed with beam (B31) elements and modeled as linear elastic materials. The stent was assigned a Young's modulus of 160.1 GPa and a Poisson ratio of 0.3, while the sutures were assigned a Young's modulus of 4.14 GPa and a Poisson ratio of 0.3. The FE model of a generic 23 mm diameter TAV developed previously by Li and Sun (2010) was also used for this study. The TAV leaflets were also modeled using large-strain brick (C3D8H) elements, and prescribed a thickness of 0.25 mm similar to commercial TAV devices. For both models, local material orientations were defined for the leaflets at each element. The leaflet material properties were defined by the constitutive law, Eqn. 14, which was incorporated into a user material subroutine (UMAT) using the numerical approximation of tangent moduli method proposed by Sun, Chaikof et al. (2008). The contact between two leaflets was modeled using a master-slave contact pair (an option in Abaqus). The SAV stent was allowed to deform to accommodate approximately 5° of stent-tip deflection under the peak pressure load of 120 mmHg (Fig. 2c). The TAV stent was not included in the simulations because the deformation of the stent is assumed to be minimal; thus, the stent-attachment nodes were fixed in space to mimic attachment to the non-deformable stent.

The un-fatigued GLBP leaflet material parameters were determined by fitting the biaxial testing data of GLBP CEP valve leaflets presented by Sun (2003) to Eqs. 15–17. The constitutive model was able to capture the response well (Fig. 1a). The matrix contribution to the overall stress was kept small compared to the fiber contribution during the parameter fitting, because the fibers are estimated to be over 100X stiffer than the matrix materials (Wagenseil and Mecham 2013). The GLBP leaflet fatigue response was defined by the hypothetical GLBP fatigue parameters presented in our previous work (Martin and Sun 2013). The permanent set was assumed to be dependent on the matrix damage rather than the fiber damage on account of experimental observations that permanent sets are more significant perpendicular to the preferred fiber orientation in aortic and carotid tissues (Maher, Early et al. 2012). The two fiber families were assigned identical material properties and oriented symmetrically about the leaflet material axis in the radial direction illustrated on the 2D leaflet schematic in Figure 1b. The leaflet parameters are given in Table 1. The amount of damage per simulated cycle was scaled up to reflect approximately 10×10^6 cycles real-time based on the fatigue model parameters. However, due to limited experimental data, the amount of damage at certain cycle states may not be accurate, and we use the variable, N , to nominalize the simulated cycle state, where N is the scaling factor relating the number of real-time cycles to the number of simulated cycles. Although the precise timing of fatigue

events cannot be predicted, through well-controlled side-by-side comparison, the effects of valve design on the leaflet durability can be assessed.

Valve fatigue simulations and analysis

In a pilot study, systolic opening of the valves was simulated by applying a uniform pressure of 4 mmHg to the ventricular surface of the leaflets (Fig. 2a&b), and diastolic closure of the valves was simulated by applying a uniform pressure of 120 mmHg to the aortic side of the leaflets (Fig. 2c&d). We found that the peak leaflet stresses and strains were consistently lower in the fully open configuration compared to the fully closed configuration (Fig. 3). Additionally, fatigue damage due to 1) cyclic opening and closing, and 2) just cyclic closing was simulated in the SAV model. In the cyclic opening and closing simulation, one cycle consisted of applying a sinusoidal pressure waveform to the ventricular side of the leaflets from a minimum of 0 mmHg to a peak of 4 mmHg, and then applying a sinusoidal pressure waveform to the aortic side of the leaflets from a minimum of 0 mmHg to a peak of 120 mmHg. In the cyclic closing only simulation, one cycle consisted only of applying the sinusoidal pressure waveform to the aortic side of the leaflets. Contour plots of the peak equivalent strain, as well as the matrix and fiber damages at the 9N fatigue state for each fatigue loading condition are shown in Figure 4. The peak equivalent strain was slightly underestimated at the commissures and the central portion of the free edge under the closing only fatigue condition compared to the opening and closing condition; however, the overall resultant damage patterns were nearly identical under both fatigue loading conditions. Given that opening did not significantly impact the SAV leaflet damage and required much more time to run, only cyclic closing was considered in the TAV fatigue simulations, and all comparisons between the SAV and TAV fatigue simulations in the proceeding sections are made between the cyclic closing fatigue simulations. The fatigue simulation results including the leaflet stress, strain, equivalent strain, matrix and fiber damage factors were extracted at each fatigue cycle state for comparison. Values from two layers of boundary elements along the stent-attachments were ignored to avoid potentially inaccurate boundary effects.

RESULTS

The SAV fatigue simulation ran until complete local failure of the matrix ($D_{sm} = 1$) at the 28N fatigue state, which caused a numerical singularity and solution divergence. The TAV fatigue simulation only ran until the 9N fatigue state due to numerical instability. The peak leaflet strain and equivalent strain were compared between the two valves for each cycle (Fig. 5). In each case, the peak values were higher in the TAV compared to the SAV at corresponding cycle states. The initial (0N state) peak strain and equivalent strain values in the TAV leaflets were 26% and 51% higher than the respective SAV leaflet values. The TAV peak strain and equivalent strain values at the 9N fatigued state corresponded to the values observed in the SAV leaflets at the 23N state (Fig. 5). Therefore, the leaflet peak equivalent strain, max principal stress, and max principal strain contours in the fully closed valve configuration were compared between the TAV at the 9N state and the SAV at the 23N state (Fig. 6a). While the peak values were similar, the respective contour plots show that a larger area of the TAV leaflet was subjected to high strains compared to the SAV

leaflet, and the TAV leaflet had higher stress throughout. In each case, the peak equivalent strains, stresses, and strains were located near the commissures and along the stent-attachments, which contributed to the most significant damages in these regions of the leaflets.

The peak equivalent strain, max principal strain and stress contours are also shown for the SAV at the 28N state representing the failure state in Figure 6b. The leaflet failure points correspond to the peak equivalent strain and max principal strain concentrations near the commissures. The SAV leaflet stress contour plot at the 28N state shows that the stress distribution homogenized from the 0N state (Fig. 3).

DISCUSSION

In this study we extended the recently developed computational framework for assessing SAV leaflet fatigue under cyclic diastolic pressurization (Martin and Sun 2013), to include distinct matrix and fiber descriptions. In our previous study, GLBP was treated as a single component material, and as such, damage to the matrix and fiber components could not be delineated. One would expect the threshold for the fibers to be higher than that for the matrix. Thus, under the same loading condition, the fiber reinforced direction may accumulate less fatigue damage than the cross-fiber direction. In addition, it can be seen from Figure 1 and Eqns. 4 and 5 that the stress-strain responses and the respective strain energy components (i.e., peak equivalent strain and damage) for the matrix and fiber constituents are drastically different. All of these factors warrant the development of a fatigue damage model to include distinct matrix and fiber descriptions.

Since the threshold to induce damage separately for each constituent has not been quantified experimentally in the literature, in this study we assumed under the same minimum equivalent strain, it will take 36.8N cycles to completely break the matrix and 50N cycles to completely break the fibers. Alternatively, the minimum equivalent strain threshold to induce damage could be defined separately for each constituent. Clearly, there is a pressing need to collect the relevant fatigue experiment data to calibrate and validate the fatigue model.

Overall, the SAV fatigue simulation results presented in this study were similar to those in our previous study. The peak equivalent strain contours both showed peak values near the commissures and along the suture attachments, and in both cases SAV failure was predicted at the 28N cycle state due to tissue failure at these regions. However, by including separate mechanisms for distinct matrix and fiber damages in this study, matrix damage was revealed in regions of low stress, albeit large strains such as the lower belly region of the SAV leaflet (Fig. 4). This effect is due to the marked anisotropy of the tissue which allows the leaflets to achieve large strains in the radial direction under low stress. By comparing the contour plots of the max principal stress and strain in the SAV leaflets shown in Figure 3 with the matrix and fiber damage contour plots in Figure 4, it appears that matrix damage corresponded to regions of high strain, while damage to the fibers corresponded to regions of high stress. Damage at the nadirs (Fig. 4) due to significant compressive stresses (Fig. 3) was also predicted unlike in our previous study. The leaflet may be particularly susceptible to fatigue

damage under compressive loading, because fibers are typically assumed to only bear load under tension, thus, the weaker matrix components must support all compressive stresses. In these regions where only the matrix is damaged, the fibers may become debonded from the surrounding matrix causing delamination. Thus, while the inclusion of distinct matrix and fiber damage descriptions did not change the predicted failure location for this case, in cases where the leaflets may be undergoing larger bending deformations or compressive stresses, it could have a significant impact on the conclusions of the study.

Comparison of SAV and TAV leaflet fatigue damage

Similar to the recently published FE results comparing SAV and TAV leaflet mechanics at the un-fatigued state (Li and Sun 2010), the leaflet stresses and strains were significantly higher in the TAV compared to the SAV. The TAV leaflet peak equivalent strains remained significantly higher than the SAV leaflet values at corresponding cycle states throughout fatigue life. This translated to elevated peak damage values and larger damage areas in the TAV leaflets at any given cycle state. In both valves, these peak leaflet damages were predicted near the commissures and along the stent-attachments. The SAV leaflets eventually failed in these regions at the 28N cycle state, which corresponds with the known failure regions of SAV leaflets (Trowbridge and Crofts 1987, Schoen and Levy 1999, Butany, Nair et al. 2007, Singhal, Luk et al. 2013). Unfortunately, TAV leaflet fatigue until complete failure could not be simulated. However, considering that the peak strain and equivalent strain values of the TAV leaflets at the 9N state corresponded to the respective peak values of the SAV leaflets at the 23N state (Fig. 5), and it is known that SAVs have a durability of approximately 20 years, we can estimate the durability of TAVs. If we scale the SAV failure point, 28N, to be equivalent with 20 years, 23N represents about 16 years and 82% of the fatigue life. Assuming proportional fatigue damage in the TAV, the 9N state represents 82% of the fatigue life; thus, the expected durability of TAVs is about 7.8 years. This result has important clinical implications: the limited durability of TAVs compared to SAVs must be taken into consideration when evaluating whether lower-risk AS patients should receive TAV replacement versus SAV replacement. TAV durability will likely be further reduced in the setting of non-optimal deployments where the stent is under-expanded or elliptical in shape (Stearns, Saikrishnan et al. 2014), which may disturb normal coaptation and increase leaflet curvatures (Stearns, Saikrishnan et al. 2014), as well as stresses and strains (Sun, Li et al. 2010). TAV leaflet fatigue analysis under various deployment configurations is necessary in order to develop design and deployment strategies for optimizing TAV durability.

Limitations of this study

In this study we assumed a linear progression of the stress softening and permanent set parameters at a given equivalent strain as a function of the number of loading cycles, whereas both values may evolve faster during the initial cycles. As in previous studies (Martin and Sun 2012, Martin and Sun 2013), the fatigue model parameters were not rigorously determined through experiments. Also, GLBP was considered a continuum throughout fatigue life: this assumption may not be valid at advanced fatigued states when there is fiber fracture or debonding, and ignores potential delamination between layers in the tissue. However, delamination is not thought to be a major failure mode of pericardial valves

(Mirnajafi, Brett et al. 2010), due to the relatively homogenous structure of GLBP. Structural valve degeneration due to calcification was also not considered; however, calcification tends to occur in areas of high stress (Schoen, Fernandez et al. 1987), so the fatigue damage areas presented here may also represent areas most susceptible to calcification.

The combined effect of tissue softening and permanent set over cycling created numerical instabilities in the simulations, particularly in the TAV simulations, where damage was accumulating at an accelerated rate, and was highly concentrated. At these damage concentration zones, adjacent elements may have very different D_s and D_{ps} values which creates discontinuities in the material properties and affects numerical convergence. For this reason, TAV fatigue could only be simulated up to the 9N state, whereas in the SAV case, the incremental damage per cycle was low and more homogeneous, and more advanced fatigued states could be simulated (up to 28N). An explicit solver could potentially improve numerical convergence; however, at this time, an implicit code is needed to accurately simulate the tissue fatigue response.

Unfortunately, there is lack of experimental fatigue data on TAVs to verify the simulations results presented in this study. Further analysis of TAV failure mechanisms and thin GLBP fatigue properties is warranted. Despite these limitations, the side-by-side comparison of leaflet fatigue in different settings remains valid.

Conclusions

TAV intervention can significantly improve valve hemodynamics and patient quality of life for inoperable and high-risk AS patients with the added benefits of lower operative risk and shorter recovery time compared to SAV replacement. One of the determining factors limiting the use of TAV devices in younger and healthier patients with longer life expectancies is the unknown durability. Therefore, in this study a previously developed computational tissue fatigue model was implemented to investigate the durability of TAVs compared to traditional surgically implanted bioprosthetic valves. The results of this study suggest that even when properly deployed, the durability of TAV devices will be significantly reduced compared to SAVs to about 7.8 years. The durability of TAVs deployed in non-optimal (elliptical or underexpanded) configurations is expected to be further reduced.

Acknowledgments

Research for this project was funded in part by NIH HL104080 and HL108240 grants. Caitlin Martin is also supported by a NIH F31 HL112632 pre-doctoral fellowship.

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Figure 1.

a) GLBP equi-biaxial response from Sun (Sun 2003) (open circles), with the modified-Holzapfel model fit (black lines), and the contributions from the matrix (blue lines) and fibers (red lines) to the overall response. b) Diagram of the leaflet material orientations and the fiber orientations drawn on the 2D leaflet schematic.

Figure 2. Overlay plots of the un-deformed and deformed a) SAV and b) TAV FE models in the open position, and of the c) SAV and d) TAV FE models in the closed position. * indicates the angle of stent-tip deflection of the SAV.

Figure 3. Contour plots of the max principal stress and max logarithmic strain for each valve in the open and closed configurations at the 0N state. In each valve the leaflet stresses and strains were higher in the closed position with peak values located at the commissures (red) and minimum values at the nadir (blue). All contour plots are shown on the aortic side of the leaflet deformed geometry at peak pressurization.

Figure 4. Contour plots of the peak equivalent strain, fiber and matrix damage in the SAV leaflets due to cyclic opening and closing and cyclic closing only at the 9N state, showing slightly higher peak equivalent strains due to cyclic opening and closing at the free edge, but nearly identical fiber and matrix damage patterns.

Figure 5. The peak a) max principal strain and b) equivalent strain observed in the leaflets for each case and cycle.

Figure 6.
a) Contour plots of the peak equivalent strain, max principal stress, and max principal strain of the TAV at the 9N state and the SAV at the 23N showing similar patterns and peak values. b) Contour plots of the peak equivalent strain, max principal stress, and max principal strain of the SAV at failure (28N state).

Table 1

GLBP material property coefficients

a. Un-fatigued state modified-Holzapfel model parameters					
c_1 (kPa)	c_2	k_1 (kPa)	k_2	θ (°)	D (kPa ⁻¹)
30.03	3.47	74.50	63.19	43.11	1.00e-5
b. Fatigue model parameters					
	ψ_{min} ($\sqrt{\text{kPa}}$)	ψ_{max} ($\sqrt{\text{kPa}}$)	n_{max} (N)	D_{pmax}	α
β					
Matrix (m)	4.67	4.32×10^5	36.80	0.105	1.00
Fibers (ρ)	4.67	4.32×10^5	50.00	n/a	1.00
					10.00