DOI: 10.5604/01.3001.0013.3185

Volume 92 • Issues 1-2 • January-February 2019

Journal of Achievements in Materials and Manufacturing Engineering

International Scientific Journal published monthly by the World Academy of Materials and Manufacturing Engineering

FEA of displacements and stresses of aortic heart valve leaflets during the opening phase

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ABSTRACT

Purpose: Modelling of biomechanical behaviour of heart valve materials aids improvement of biofunctional feature. The aim of the work was assessment of influence of material thickness of leaflets of artificial aortic valve on displacements and stresses during opening phase using finite element analysis (FEA).

Design/methodology/approach: The model of aortic valve was developed on the basis of average anatomical valve shapes and dimensions. Nonlinear dynamic large displacements analysis with assumption of isotropic linear elastic material behaviour was used in simulation (Solidworks). The modulus of elasticity of 5.0 MPa was assumed and Poisson ratio set to 0.45. The rigidly supported leaflets was loaded by pressure increasing in the range 0-55 mmHg in time 0.1 s. Leaflets with material thickness 0.13 and 0.15 and 0.17 mm were analysed. The thickness was simulated with shell finite elements.

Findings: The highest stresses were observed in the areas of fixation of the leaflets near the scaffold and were lower than dangerous value of fatigue of polyurethanes. Increasing the thickness of valve leaflet material in the range of 40 micrometres resulted in reduction of the valve outlet by almost 10 percent.

Research limitations/implications: The FEA was limited to the isotropic linear-elastic behaviour of the material albeit can be used to assess leaflet deformation during dynamic load.

Practical implications: Leaflets design may be start from efficient FEA which helps estimation of material impact on stress and fold formation which can affect local blood flow.

Originality/value: Aortic heart valve leaflet material can be initially tested in dynamic conditions during opening phase with using FEA.

Keywords: Heart valve, Aortic valve, Leaflet material, FEA, Simulation

Reference to this paper should be given in the following way:

O. Bialas, J. Żmudzki, FEA of displacements and stresses of aortic heart valve leaflets during the opening phase, Journal of Achievements in Materials and Manufacturing Engineering 92/1-2 (2019) 29-35.

BIOMEDICAL AND DENTAL ENGINEERING AND MATERIALS

1. Introduction

Diseases related to the cardiovascular system are still the main cause of death in Poland and in the world according to WHO data from 2016 which claims 30% of global mortality (Fig. 1). Despite the passage of years, this condition is not satisfactorily improved, hence there is a need for continuous development of methods of treatment for chronic cardiovascular diseases. Currently, researchers hopes to obtain natural valves made from the patient's stem cells. Another way of cure is taking valves from human donors. Xenogenic transplants are also successfully used. There are also continuously improved mechanical prostheses [1-3].

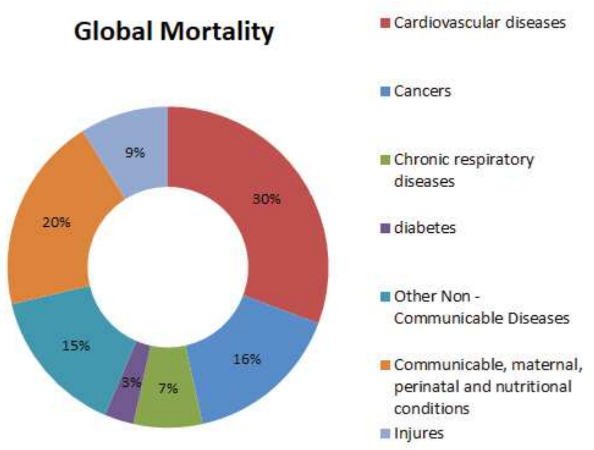

Fig. 1. Most common death causes diagram from WHO 2016 report

Naturally, the heart valves function as one-sided valves, which are designed to open under the pressure of adequate blood pressure and prevent from its turning back.

Mechanical valves have been in use since 1960 and are subject to constant modifications. Essentially, the changes concern their long-term functioning and rely on mechanical and material improvements [2,4-8].

First of all, the use of structural solutions maximizing the valve outlet area in relation to the size of the entire prosthesis, so as to ensure the best hemodynamic of valve operation. This is especially important for patients with a tight ring of the valve, in which the dilatation of the arterial outlet is risky, for example due to the age.

Secondly, creating valvular mechanisms with as little thrombogenicity as possible to reduce the dosage of anticoagulants. When it comes to mechanical, the first valve had a mechanism based on a silicone ball moving within a metal "cage". Currently, a valve mechanism with 1 or 2 discs is in use.

However, knowledge about phenomena responsible for the functioning of each system is invaluable, because it allows to prevent negative occurrence and, consequently, prolongs failure-free operation. One of the factor is the state of stress in the heart valves, which affects the process of material degradation in artificial and natural valves. Thanks to the computer simulation using the finite element method (FEM), it is possible to study the opening of the valve leaflets.

The aim of the study was analysis the influence of leaflets material thickness on stress distribution and aortic valve outlet during the opening phase [9-12].

2. Materials and methodology

The model of the aortic heart valve has been shown in Figure 2. The model was made in the CAD software (Solidworks) in accordance with the dimensions and shape of the natural valve: height – 13 mm, diameter – 19 mm.

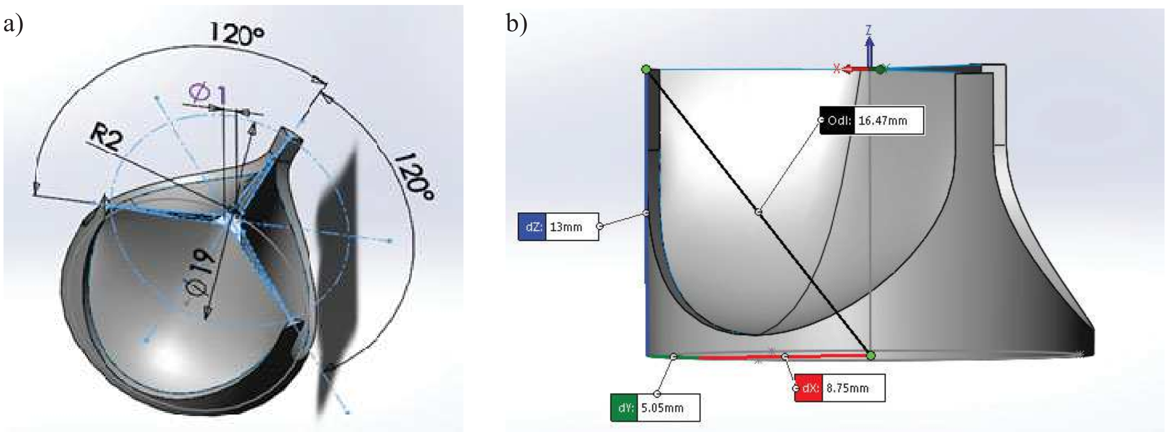

Fig. 2. Heart valve CAD model including dimensions: isometric view (a) lateral view (b)

The scope of the work includes the analysis of behaviour of a valve made of polyurethane with a different thicknesses. The simulation tests were simplified to the surface model with constant thickness of leaflets valve. Large displacements non-linear dynamic analysis with assumption of isotropic linear-elastic material behaviour were used. The model has been rigidly supported on the edge between a leaflets and the scaffold. The load was modelled by the pressure uniformly distributed over the surface of the leaflets with linear increases in the range 0-55 mmHg during systole which results from cyclic pressure changes during hemodynamic cycle [3,13].

The division of the leaflets into triangular surface finite elements has been shown in Figure 3. Simulation conditions and analysis parameters are summarized in Table 1 and Table 2 [14-17].

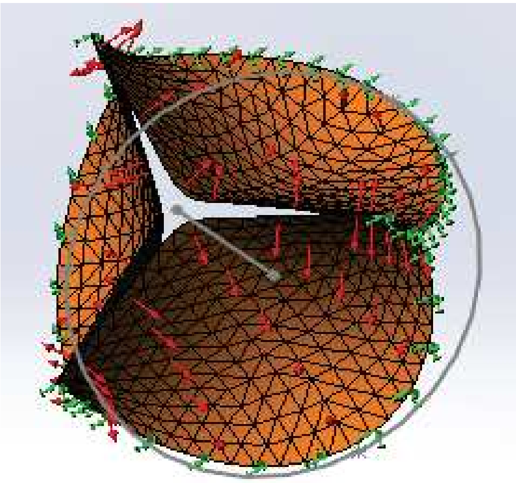

Fig. 3. Numerical model with finite elements division including fixation (green arrows) and pressure (red arrows)

Table 1.
Simulation conditions during analysis

Simulation conditions	
Material model	isotropic linear elastic
Type of analysis	nonlinear dynamic analysis
Solver	FFEPlus
Total number of nodes	2554
Number of elements	1212
Element size	1.08411 mm
Time increase	Auto
Iterative Method	NR (Newton-Raphson method)
Iterative method	Newmark

Table 2.
Parameters of the analysis during simulation

Parameters of the analysis	
Time	100 ms=0.1 s
Pressure	7333 Pa = 55 mmHg
Young module	5.0 MPa
Leaflets thickness	0.13 mm; 0.15 mm; 0.17 mm
Material density	1020 g/cm^3
Poisson ratio	0.45

2.1. Results

The distribution of equivalent Huber-Misses (H-M) stress, strains and displacements in the leaflets during the stage of reaching the maximum value of blood pressure has been shown in Figure 4. In order to assess the impact of the finite element mesh size on the results, an additional test with smaller finite elements was performed for the leaflets material thickness of 0.13 mm. The result had proven that stress concentrations does not result from mesh inaccuracy errors, but from physical phenomena [9,14-17]. However, analysis of influence of mesh size at rigid edge was impossible due to limited computer power.

Based on the distributions of the stress, the criterion areas of stresses and displacements were determined. Criterion areas has been shown in the Figure 5. They occur near to fixation of leaflets at the upper (A) for stresses and in central location (B) for displacements.

The analysis showed that the highest stresses occurred through the whole margins where leaflets is bonded to the scaffold. This phenomena was caused due to limitation of the freedom of deformation in these locations as the scaffold is rigid and polyurethane's leaflet material is elastic. Equivalent H-M stress in the area between "fold belt" and support ranged from 0.7×10^5 Pa to 3.5×10^6 Pa (Fig. 6) at the rigid edge. Maximum value around rigid support is however an FEM artefact which was not evaluated due to computational problems with a smaller mesh size. Hence, the criterion area was set at a distance of one element from the edge to avoid overestimating the stresses in the node near the singularity.

Figures 7-8 show changes in equivalent H-M stress and displacements in criterion areas during increase of pressure in the time of 100 ms. Opening of the leaflets was rapid and was about 8.3 mm in 0.014 s. Then the displacement was reduced to 4.62 mm due to dynamic repulse of leaflet what has been shown in Figure 9 which can slightly change local fluid flow and cause turbulent phenomena.

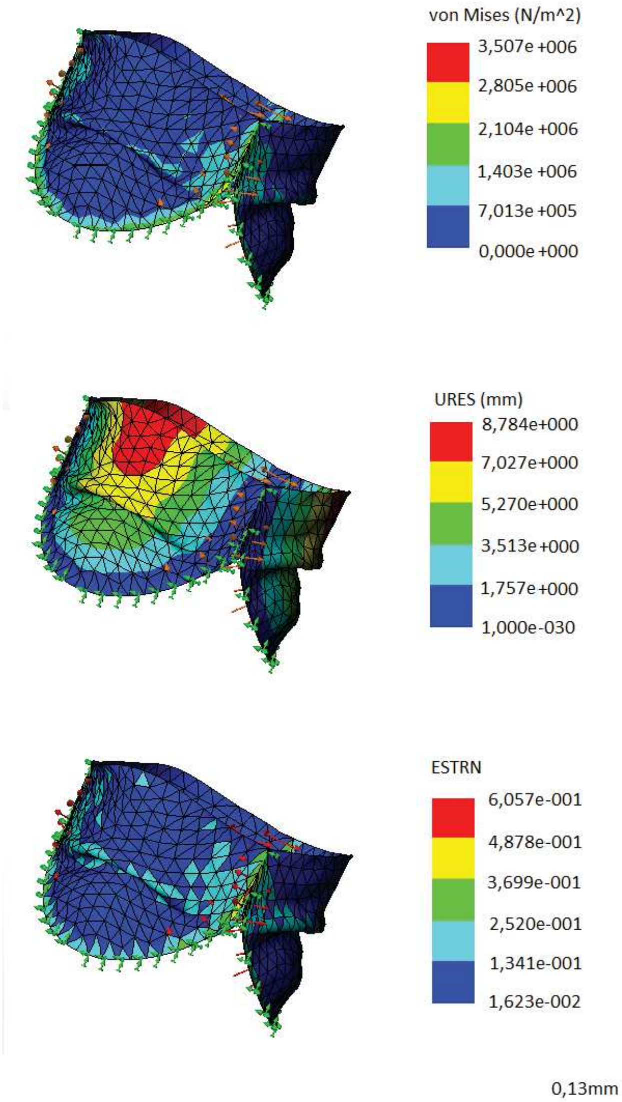

Fig. 4. Distribution of stresses, displacements and strains obtained in last step of computer simulation; leaflet thickness 0.13 mm

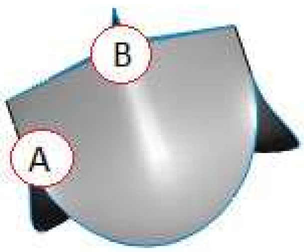

Fig. 5. Criterion nodes localisation

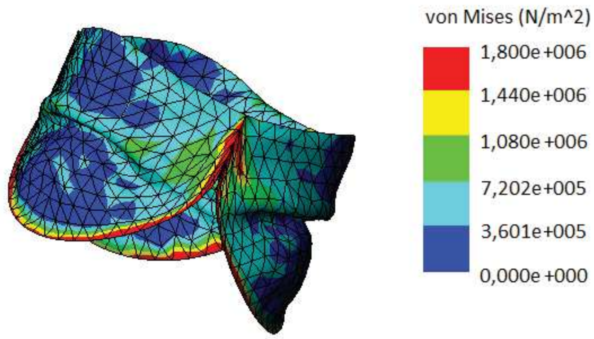

Fig. 6. H-M stress distribution with scale to 1.8 MPa which shows value in the distance of one element from the singularity

The opening was about 8 mm after the repulse. It was for 22 mmHg of pressure. Further increase in pressure from 22 to 55 mmHg influenced the increases in opening in the a little value of 0.3 mm. The largest opening value was 8.66 mm for 0.13 mm thick leaflet. A little oscillations of the leaflets were observed after repulse and its amplitude was slightly lower for a thinner leaflets. Larger oscillations of a stiffer leaflets indicate that they are mainly caused by inertia forces.

3. Discussion

Biofunctional parameters of implant can be estimated before in vitro tests due to progress in computer simulation. However, model simplifications are needed accordingly to the available computer. In the analysed model material non-linear behaviour, unequal blood pressure on the leaflet and compliance of support were omitted. In the simplificated conditions there were estimated influence of leaflet stiffness on stress state in leaflet polymeric material and the valve outlet in dynamic conditions during an increase of blood pressure.

The time of tests has been assumed based on the hemodynamic cycle. Time of pressure increase is about 100 ms during systole in healthy humans [3,13,18]. Just in the first 20 ms leaflets can open rapidly and begin to flap. The short time of that kind of phenomena force to conclusion that toughness resistance is also important and should be included in every case when it comes to cardiovascular implementation. The main requirement for conventional heart valve implants from ISO standard [19] when it comes to pressure is that heart valve substitute must be able to withstand a single maximum cycle consisting of a pressure of 230 mmHg [20], but in the context of long-term functioning, the stress state achieved during the normal cardiac cycle is no less important.

Hemodynamic of the cycle shows that difference between systole and diastole acquires about 50 mmHg (from 80 mmHg to 120 mmHg). Because of that fact consequent from about 50 mmHg [13] stress state is main leading in determination the fatigue of the material and a normal valve outlet. Dynamic behaviour and stress increase of leaflet were similar to observed in the work [21], where 1st principal stress of 1.75 MPa in case of linear model and 3 MPa during nonlinear are calculated. Albeit should be take into account different Young modulus (2 MPa), Poisson ratio (0.495) and leaflet thickness of 0.5 mm, as also as that model is loaded with a pressure greater by 40 mmHg [21].

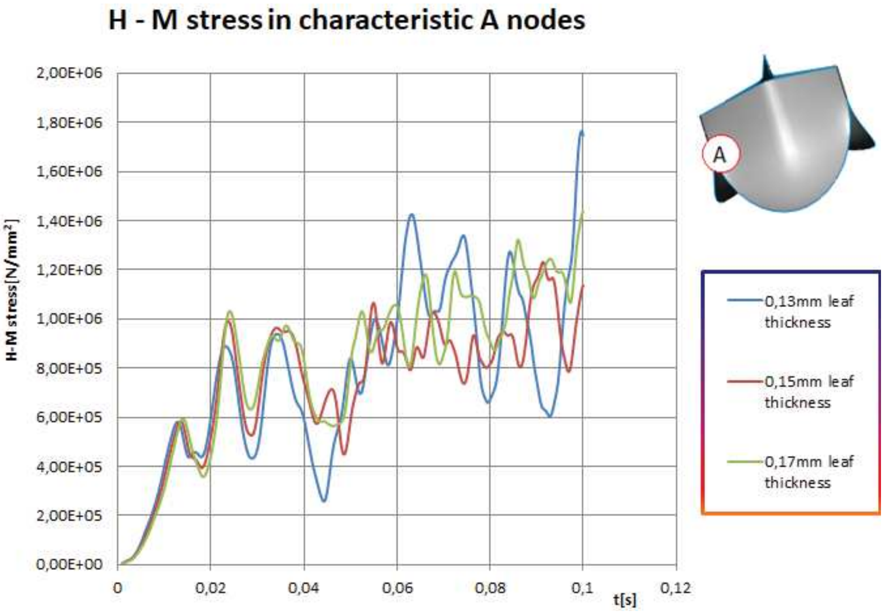

Fig. 7. Huber-Mises stress in area A

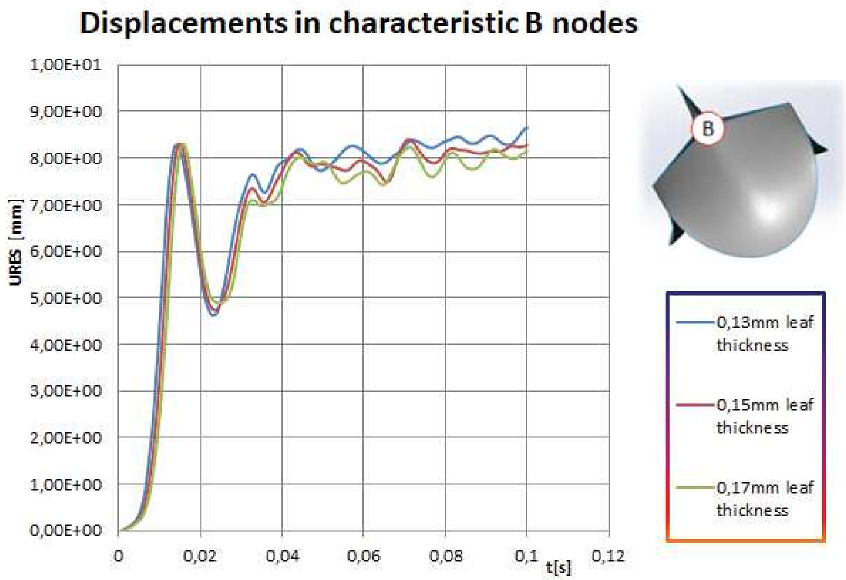

Fig. 8. Displacements diagram in area B

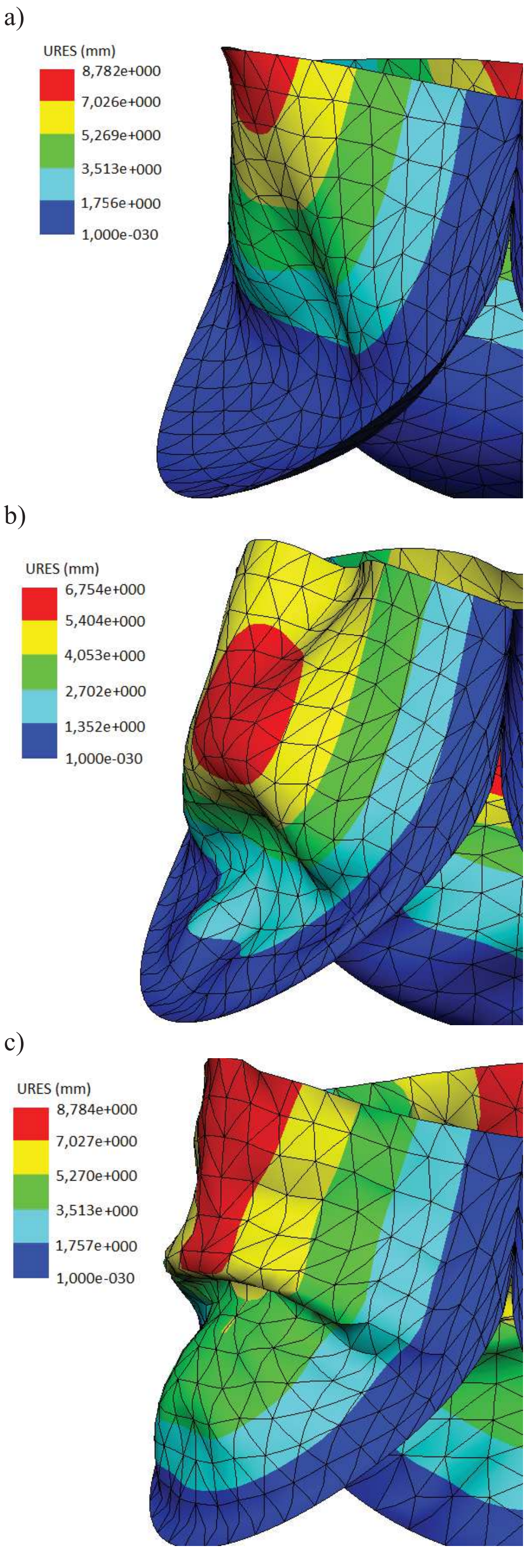

Fig. 9. Flop of the "fold belt" in a) 0.014 s, b) 0023 s and c) 0.1 s

Fold belt which has been shown in Figure 6 occur due to limitation of the deformation freedom of leaflet. In natural valve the aortic wall is deformable. Hence, natural leaflet is deformed not only under blood pressure but also by aortic wall. Stress calculated in the model may apply to artificial scaffold. The scaffold in artificial valve is relatively stiff and was modelled as non-deformable. Nevertheless, maximum value of achieved stress is safe for PU used for heart valve, because its lifetime under cyclic stress (for ordinary PU results are 306128±119093 cycles to failure at 30×10^{6} Pa cyclic stress to failure and can be also improved to over 800 million in the Strathclyde university valve model with PU strengthened by carbon nanotubes during in vitro tests) [22-24].

Linear extrapolation of the stress to those under pressure occurring during 230 mmHg leads to value of 14.6×10^{6} Pa. That means that a hypothetical stresses meets requirements from ISO standard [19,20].

Criteria area responsible for supporting and keeping leaflet during the opening into the scaffold must be considered in the fatigue behaviour, as the phase opposite to the valve closure [25]. Non linear "in-elastic" behaviour should be taken into account in the future simulations during prosthesis improvement.

4. Conclusions

Simulation tests in the assumed conditions of the surface model and the assumed isotropic linear-elastic behaviour of the material during large displacements non-linear dynamic analysis allowed to state that:

- The highest stresses were observed in the areas of fixation of the leaflets near the scaffold, while in the middle of the leaflets the stress was almost non-existent.
- Stress in the material near the scaffold decreased with increasing thickness of the leaflet material.
- Increasing the thickness of valve leaflet material in the range of 40 micrometres resulted in a reduction of the valve outlet by almost 10 percent.
- In the case of a linear-elastic material, folds in the middle of leaflets have been formed that may cause blood flow disorders.

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