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MODEL SIMPLIFICATION AND BOUNDARY CONDITIONS OF ABDOMINAL AORTIC ANEURYSM

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Summary: *The article compares different boundary conditions for models of abdominal aortic aneurysms in order to find a suitable boundary condition which would allow removing the bifurcation from the models. Removal of the bifurcation simplifies the geometry reconstruction and FEA. The best way to do so is by cutting it out by a plane which is not perpendicular to the longitudinal axis of the body.*

Introduction

Abdominal aortic aneurysm (AAA) is defined as a permanent dilatation of the abdominal aorta. Its prevalence varies from 3% up to 7% at people above 65 years (Sonesson et al., 1999. Nichols et al.,1998). When left untreated it can lead to a fatal end-rupture. When rupture occurs, a massive internal bleeding will follow. Overall mortality in such cases is about 50%. Therefore it is vital to predict the risk of rupture in order to plan a surgery.

So far the diameter of the AAA is a widely accepted criterion for such a decision but the wall stress is a better predictor of rupture (Fillinger et. al., 2002). Recently clinical software has appeared which uses FEA of the AAA to estimate its wall stress in order to predict the rupture risk.

Despite a great progress in FEA of AAAs there are still some limitations of its use. First of all, it is necessary to obtain a realistic patient-specific geometry of the AAA. It has been shown that idealized models of AAAs are not able to describe the wall stress distribution realistically (Vorp, 2007; Gholam et. al., 2007). To obtain the patient-specific model of geometry, it is necessary to reconstruct it from CT or NMR images. Such process has of course some limitations due to the resolution of the images, which causes unwanted geometric artifacts mainly in the area of aortic bifurcation. Such artifacts will of course result in unrealistic values of stresses in the influenced areas (see Figure 2). So far the only way of how to deal with it published (Speelman et. al.,2008) was to ignore these extremes by omitting 1% of volume with the highest stress from the evaluation.

Another geometry related problem is the fact the geometry on the CT or NMR images is already loaded by blood pressure. Nevertheless, the FEA needs a stress-free geometry. It is possible to obtain an initial stress-free geometry in several ways (Gee et.al.,2009; Putter et.al.,2007) but they need an artifacts-free geometry for the mathematical solution to be convergent (Gee et.al.,2009).

Both problems mentioned above could be avoided by modeling of the AAA without the bifurcation, which is the most problematic part to be reconstructed and it commonly contains artifacts which lead to the previously mentioned problems.

Another problem is material related. To our knowledge, all the authors, no matter what kind of material model they used in their studies, used the same model for both AAA and connected iliac arteries. This approach includes another error into the analysis because mechanical properties are substantially different for the AAA wall and the healthy iliac arteries. This is another argument why we think it is worth to try to remove the bifurcation from the FEA.

An elimination of the bifurcation, however, brings a problem how to deal with boundary conditions (BC). Generally, a fixed support is used to constrain the deformation at the end surfaces of the model. Such BC´s are indeed very easy to defined and they make also the analysis easier to converge. On the other hand one must be aware such BC´s are not realistic and therefore the results in their close surroundings will not be realistic too. This is actually the main reason why AAA is modeled with bifurcation despite the disadvantages mentioned above.

We have decided to investigate whether or not it is possible to find BC´s which would allow us to eliminate the bifurcation. It must also influence as small area as possible because AAA generally ends at the bifurcation and we want the stress results in the AAA to be as realistic as possible.

Methods

We chose five patients from our database of CT images in order to reconstruct a patient-specific geometry. We used these models for FEA in the ANSYS software with different BC´s and then we compared results to see the range of the influenced area where the stress results are substantially influenced by chosen BC.

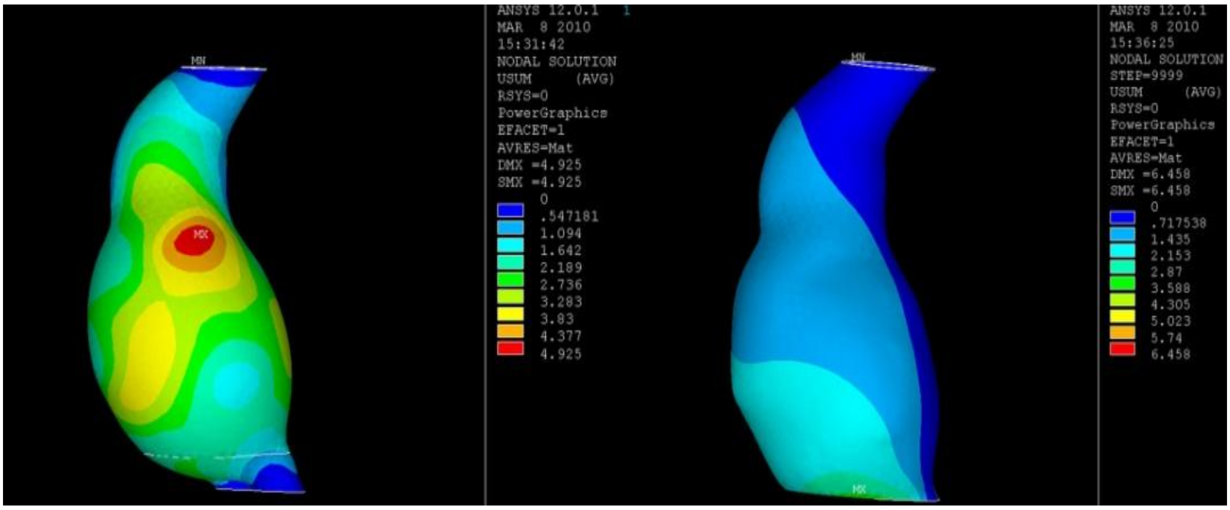

Figure 1 patient no.1 left: total deformation of the AAA [mm].right: difference in the total deformation [mm] between the whole model and the model where the bifurcation is cut away.

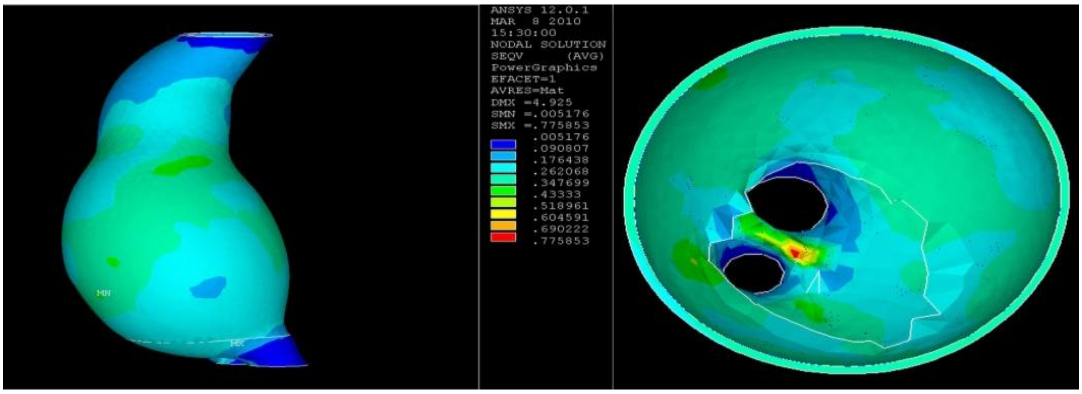

Figure 2 (patient no.1) left: von Mises stress distribution on the AAA [MPa], right: detail of the artifact in the bifurcation area which produces an unrealistic peak stress

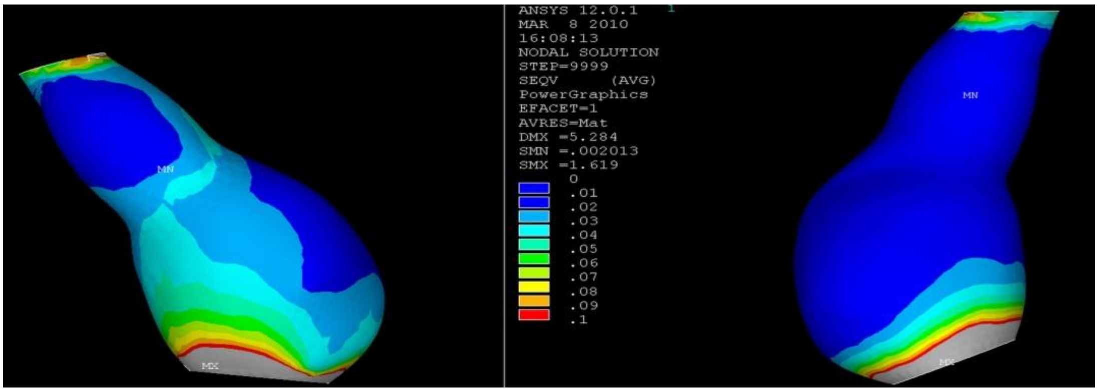

Figure 3 (patient no.1) left: von Mises stress differences [MPa] between the models A and C2. Right: von mises stress differences [MPa] between the models A and C1.

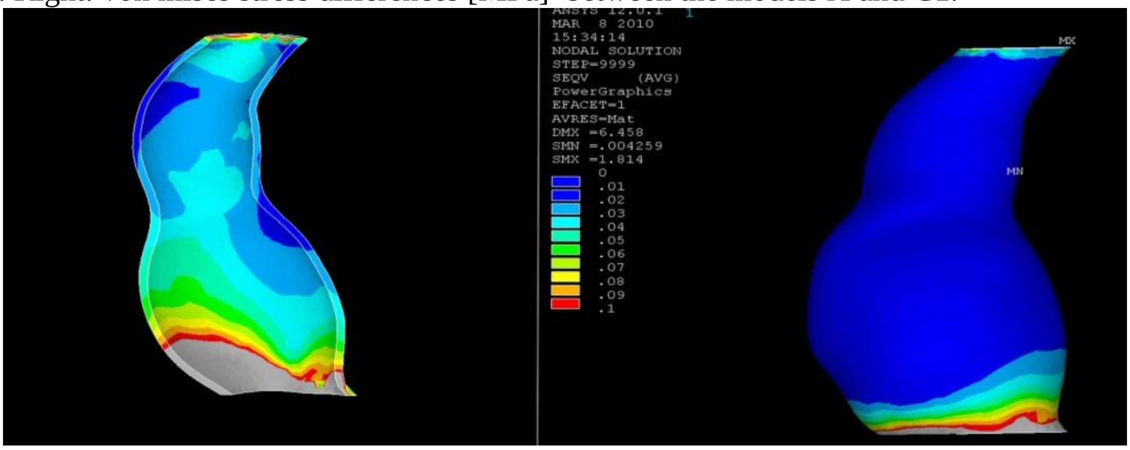

Figure 4 (patient no. 1) left: von Mises stress differences [MPa] between the models A and B2. Right: von Mises stress differences [MPa] between the models A and B1

Geometry reconstruction

To reconstruct the 3D geometry of each AAA, we used the commercially available software A4research from the company Vascops. After having loaded a set of CT images we assigned the values of density to the lumen and the intraluminal thrombus and let the program to generate the luminal and the outer wall surfaces. Unlike some other software, in this program the determination of the surfaces is not based on generating point clouds which need to be smoothened prior to the FEM analysis but on an expansion of a virtual balloon until the border between two levels of intensity is reached. Although such approach produces smooth surfaces it still could contain artifacts such as sharp edges and corners in the bifurcation area which would produce unrealistic values of stress. When the outer wall surface is generated, the program calculates the distance between the luminal and outer surface. if it is greater than the threshold value (set up to 0.5mm), the program adds a thrombus at the inner side of the wall. As the last step the wall thickness is generated based on the distance from the lumen. Where there is no ILT, the wall thickness has been set up to 1.5mm. The thicker is thrombus, the thinner is the wall, up to 30mm of ILT thickness, which generates a 1mm wall[11,12]. Then the volumes defined by these areas are generated.

Although the program A4research contains also a FEM part, its possibilities are limited compared to ANSYS. Therefore we exported the geometry in the STL format to CATIA. As this file format contains only mesh data and we needed to have a possibility to cut out some parts of the model and to remesh it, we had to build a solid body from this file. For that we used CATIA which contains a toolbox for a volume generation from STL files. Although it is possible to reconstruct the wall, thrombus and lumen as well, we needed only the wall for this study. So we rebuilt a solid body of the AAA wall and then exported it to the ANSYS.

Finite element model

We built the FE model in ANSYS workbench interface because it is capable to load a complicated geometry such as that of the AAA wall. A hyperelastic isotropic material model proposed by (Raghavan et. al. 1996) has been used for AAA wall.

Tetragonal quadratic 10 nodes elements have been used, because it is not possible to use hexagonal mesh for such a complex geometry. The element size has been established in order to achieve a reasonable compromise between the accuracy and the speed of computation. So we have chosen the 3mm size of elements, which gave 40 000 up to 80 000 nodes. This number is small enough to keep the computational time in order of magnitude of minutes. According to our experience the stress results show still an acceptable accuracy with such numbers of nodes. Of note we compared only the stress deviations for different BCs so that the errors in stress magnitudes due to an insufficient discretization are not vital. In addition, we used the same discretization for different BCs which guaranteed the stress results to be calculated in exactly the same points and therefore we have completely eliminated the influence of the mesh density on the results.

The reconstructed geometry was considered as load free and the pressure of 20kPa was applied on the inner surface of the AAA wall. The FEM analysis was performed with considering a large strain algorithm.

Regarding BC´s, we used a fixed support for the upper end of the AAA (close to the renal arteries). For the lower end of the AAA we have formulated the following BCs:

Model A: the bifurcation is included and a fixed support is applied on the iliac arteries far enough to be sure that the BC cannot influence the stress results in the aneurysm sac.

Model B: the bifurcation was cut away from the AAA model by a plane perpendicular to the longitudinal axis of the body.

Model C: the bifurcation was cut away as well but the cut surface (at the lower end) was not perpendicular to the longitudinal axis of the body.

By cutting the bifurcation part away by the plane which is not perpendicular to the longitudinal axis tried to cut out as small part of the AAA sac as possible.

For the analysis with the bifurcation cut away we applied two different BC´s:

VARIANT 1: we applied a fixed support in order to show the range of the influenced area.

VARIANT 2 was based on the idea that aorta dilates during the cardiac cycle "in vivo". We assumed the tangential movement of the aorta to be negligible. The axial movement of the aorta in the body is also constrained. In fact there is even some axial prestress "in vivo" but there is no information about its magnitude in an aneurysmal aorta. Therefore we have formulated a BC by setting the tangential and axial movements to zero while the radial displacements remain free.

So we obtained one referential model geometry (A), and two shortened geometries (B and C) for each AAA, each of the latter with two different BCs, giving variants B1, B2, C1 and C2. For 5 patients it counts 25 analyses altogether. Such a high number justifies our effort to keep the number of nodes relatively low.

Results

The outputs from the analysis are obtained in the form of fields of displacements vectors and stress and strain tensors. They are presented in detail for the patient No. 1 in figures 1 to 3. While the displacement magnitude is up to 5 mm (see fig.1) the differences between the individual models do not exceed 2 mm in the region of the aneurysm. Even more important is the accuracy of the resulting stresses, represented as von Mises equivalent stresses. They reach their maximal values of 775 kPa in the bifurcation region but these values are not credible because of the geometrical artifacts in the FE model. In the aneurysm balloon the maximum values are between 400 and 500 kPa. The differences between the individual models exceed 100 kPa only in the very vicinity of the cutting surface of the model and they are lower than 30 kPa in all the aneurysm balloon, which represents an error on the order of several percents only. This comparison is shown in figures 3 to 9 for all the investigated patients. It is obvious that the stress differences are always the greatest at the cutting surface. Although the stress difference drops promptly with the distance from the BC, there is a high stress difference close to the other BC despite the fact that this BC remained the same for all the analyses. The stress differences drop faster when all DOFs are constrained (variant 1 of BCs) and when the cutting surface is not perpendicular to the longitudinal axis (model C).

Discussion

The results show clearly that it is not suitable to cut the bifurcation out by a plane parallel to the upper end surface and to let the radial deformation free, because in this case the model "loses" the information that the AAA sac is ending rapidly and it behaves as if it continued still at same diameter.

Figure 9 shows that the stress differences are much higher in this variant.

We obtained better results when the cut surface was fully constrained. However, we achieved the best results for the cases when the cutting plane was not parallel to the top cross section. There were not significant differences between the BC 1 and BC 2. For the patient no.3, better results were obtained with the BC2 and for the patients no.1 and no.2 the BC 1 is more suitable.

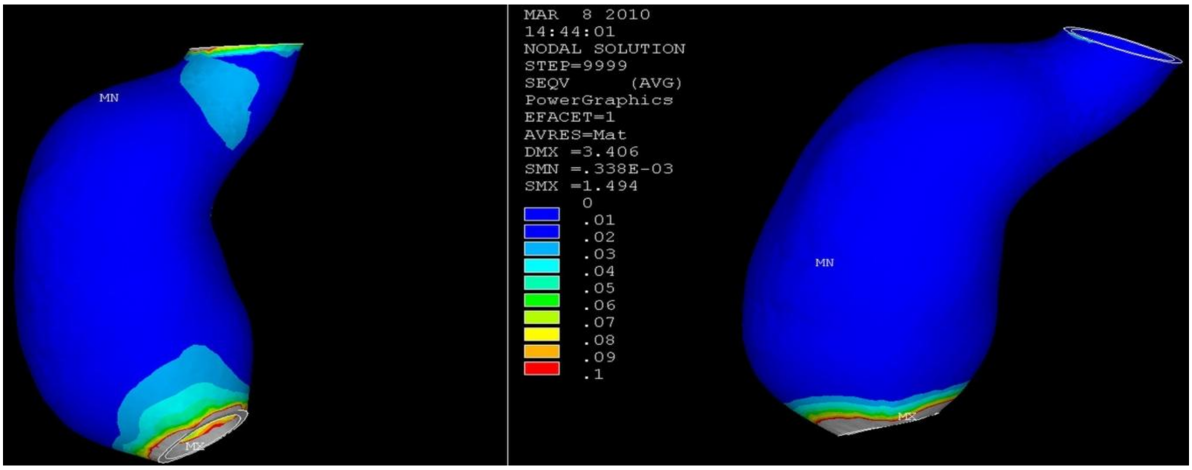

Figure 5 (patient no.2) left: left: von Mises stress differences [MPa] between the models A and C2. Right: von mises stress differences [MPa] between the models A and C1.

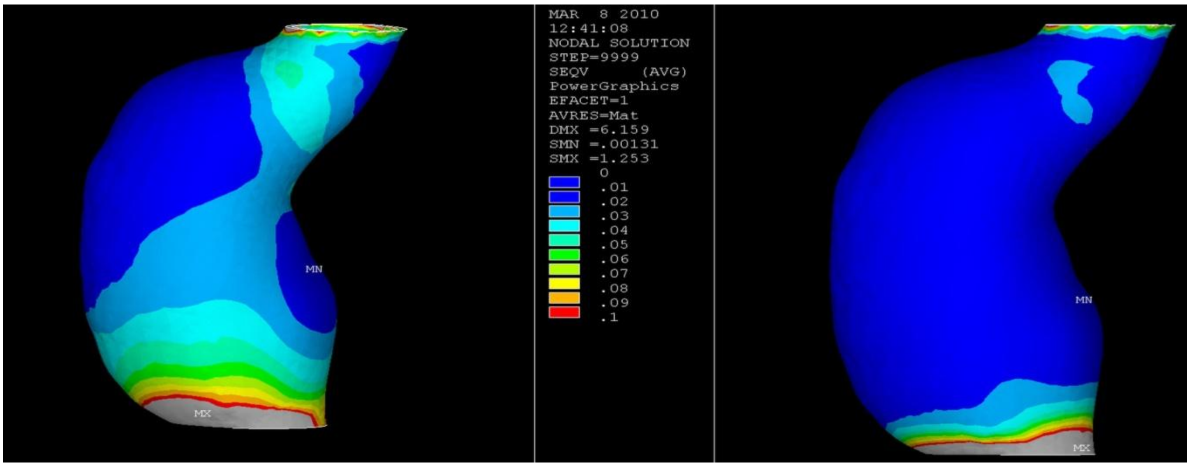

Figure 6 (patient no. 2) left: von Mises stress differences [MPa] between the models A and B2. Right: von Mises stress differences [MPa] between the models A and B1.

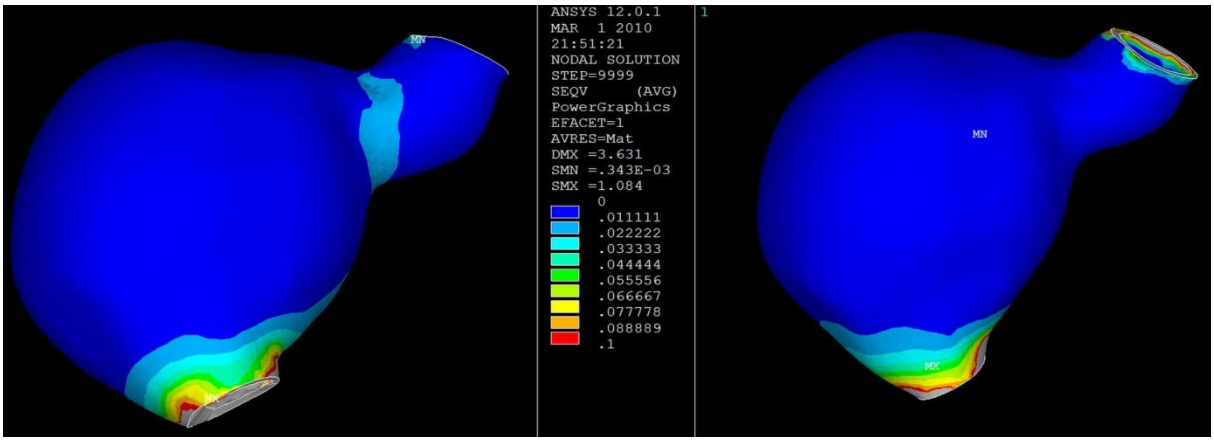

Figure 7 (patient no.3) left: von Mises stress differences [MPa] between the models A and C2. Right: von Mises stress differences [MPa] between the models A and C1.

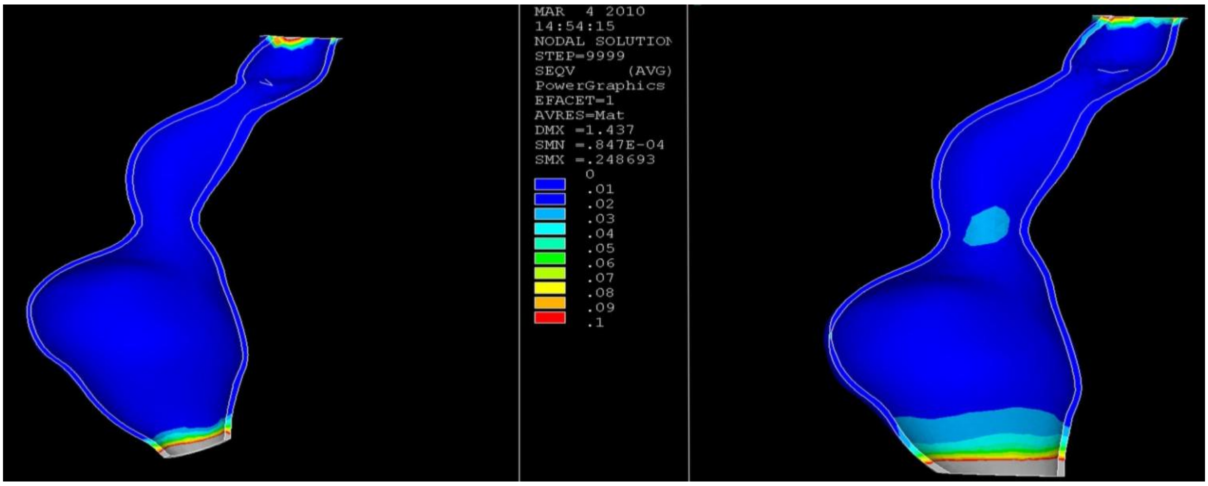

Figure 8 (patient no.4) left: von Mises stress differences [MPa] between the models A and C1. Right: von Mises stress differences [MPa] between the models A and B1.

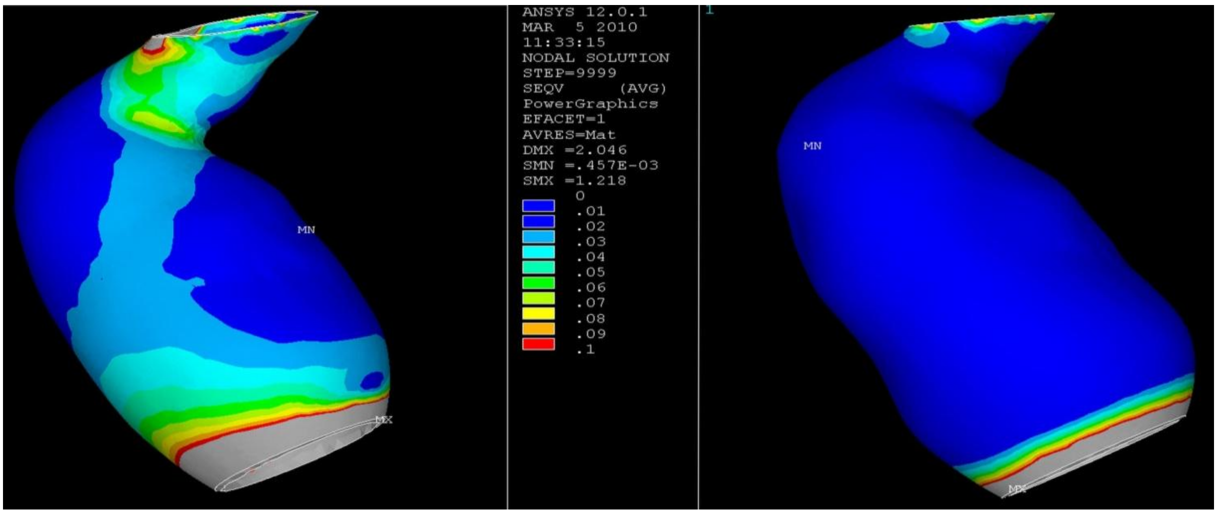

Figure 9 (patient no.5) left: von Mises stress differences [MPa] between the models A and C2. Right: von Mises stress differences [MPa] between the models A and C1.

It is interesting that there are differences in the stress values not only near the bottom BC but also in the area close to the top BC. It is clearly visible in all of the pictures. This result is caused by the fact that different BC caused different movements of the models as shown in

Figure **1**. The differences in the deformation displacements are more than 1mm. The larger aneurysm movement induces more bending of the aortic wall close to the top end surface and therefore higher stresses in this region.

Of note

Figure **1** captures only differences in the magnitude of the displacement vector but not the differences in the directions. In the extreme case, if the movement of the first model was 10mm to the left and the other model moves 10mm to the right, it would show zero difference in magnitude of total displacement. However, we believe the differences in directions of the displacements are negligible and therefore

Figure **1** represents the total differences between the displacements in the compared models.

Conclusions

We showed it is possible to perform realistic FEA with an AAA model without including the bifurcation. However, it is necessary to cut away as few of the aortic sac as possible and it should be done by a plane which is not perpendicular to the axial direction. Then the area influenced by the BC is acceptably small and the removal of bifurcation from the model simplifies also the creation of the unloaded geometry.

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