

An Experimental and Finite Element Model for Traumatic Injury in Aorta

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Abstract— An experimental model of porcine aorta was developed to simulate the pinching phenomenon, i.e. dynamic large deformations due to direct contact, which is considered to be one of the mechanisms of TAR. Thoracic descending aortas, incased in a clear urethane box were pressurized to 1.6 psi (11 kPa). The specimens were fixed at the two ends and were allowed to bend in the middle upon impact. A 5-mm flat indenter was placed in front of the specimen approximately at the mid-point and the setup was decelerated at 55g to cause bending.

A finite element model of the experiment was developed in LS-Dyna (LSTC, CA) using the Arbitrary Eulerian Lagrangian (ALE) formulation for the fluid and a hyper-viscoelastic Lagrangian formulation for the aorta material. The aorta wall was assumed to be single-layer, homogeneous and isotropic in the FE model. To address the inhomogeneity of the material properties of aorta, micro-indentation tests were performed on the wall along the radial direction and the variability in the viscoelastic shear modulus was determined.

Significant W-shaped deformations were observed in the specimens upon impact and dissected sections of aorta near the indenter showed laceration in inner media. The FE model successfully simulated the experiments in terms of global kinematics and internal pressure and the effective strain predicted by the FE model near the indenter was about 30% which agreed with the expected threshold of partial failure for a young human aorta. The results of indentation tests showed that the shear modulus of the aorta wall increases radially toward the outside layers which explains why more injury was sustained in inner layers.

Significance of TAR

Traumatic aortic rupture (TAR) is a significant cause of death in motor vehicle accidents:

- TAR is the cause of fatality in about 20% of motor vehicle crashes [1].
- In the USA and Canada annually around 7500–8000 victims die from TAR [2], which in the majority of cases is the result of an automotive accident.

Mechanisms of TAR

Mechanisms of TAR are still unknown however, several hypotheses have been proposed:

- relative motion between heart and ascending aorta
- intravascular pressure
- the water-hammer effect
- osseous pinch

Previous Studies on Mechanisms of TAR

Mostly involved cadaveric torsos and they achieved limited or no success in producing TAR [3, 4].

Shortcomings associated with the cadaveric models of TAR:

- Pressurization method: leaky vessels
- Lack of muscle tone and drop of organs
- Inter-specimen variability

Objective

To develop an *in vitro* model that can be modeled with FEA to understand the mechanisms of TAR by isolating the effects of pressure, and shear strain due to direct impact.

Materials and Methods

Experimental Model simulating pinching phenomena:

- 200mm of porcine thoracic aorta incased in a urethane box
- Filled with PBS
- Pressurized up to 1.6 psi or 11 kPa (physiological pressure)
- Fixed at 2 ends and allowed to bend in the middle
- 5mm indenter placed approximately at mid-point
- Setup was decelerated at 55g using a highly repeatable impact system

Finite Element Model:

- Developed in LS-Dyna (LSTC, CA)
- Material model for aorta: hyperelastic-viscoelastic derived from Mohan and Melvin (1982) [5] experiments
 - Single-layer
 - Homogeneous
 - Isotropic
- Arbitrary Eulerian Lagrangian (ALE) formulation was used to model the fluid inside aorta.

Inhomogeneity of material properties of aorta wall:

The aorta wall is consisted of 3 layers. An indentation test was performed to determine the variability in the shear modulus of the aorta wall.

- Transverse sections of aorta near isthmus region
- Spherical indenter with 0.2 mm diameter
- Indentation tests on every 0.4 mm in radial direction
- Quasi-linear viscoelastic model assumed [6]

$$P(t) = \int_0^t G(t-\tau) \frac{\partial P^e(h)}{\partial h} \frac{\partial h}{\partial \tau} d\tau$$

- Reduced relaxation function was assumed to be a discrete approximation with Prony series:

$$G(t) = G_\infty + \sum_{i=1}^4 G_i \exp(-\beta_i t)$$

- Nonlinear instantaneous elastic response was assumed:

$$P^e = \frac{8\mu}{3(1-\nu)} \frac{\sqrt{D}}{2} h^{3/2}$$

Results

Experimental tests results:

- Significant W-shaped deformations was observed upon impact (Fig. 2)
- Pressure rise in the aorta was below 80kPa which is significantly lower than threshold of rupture due to pressure(100 kPa) [3]
- Dissected sections of aorta near the indenter showed laceration in the inner media (Figure 3).
- total transection was resulted when the experiment was repeated twice.
- The obtained shear modulus from indentation tests (Figure 4) showed that the stiffness of the aorta wall increases radially from inside layers toward outside layers.

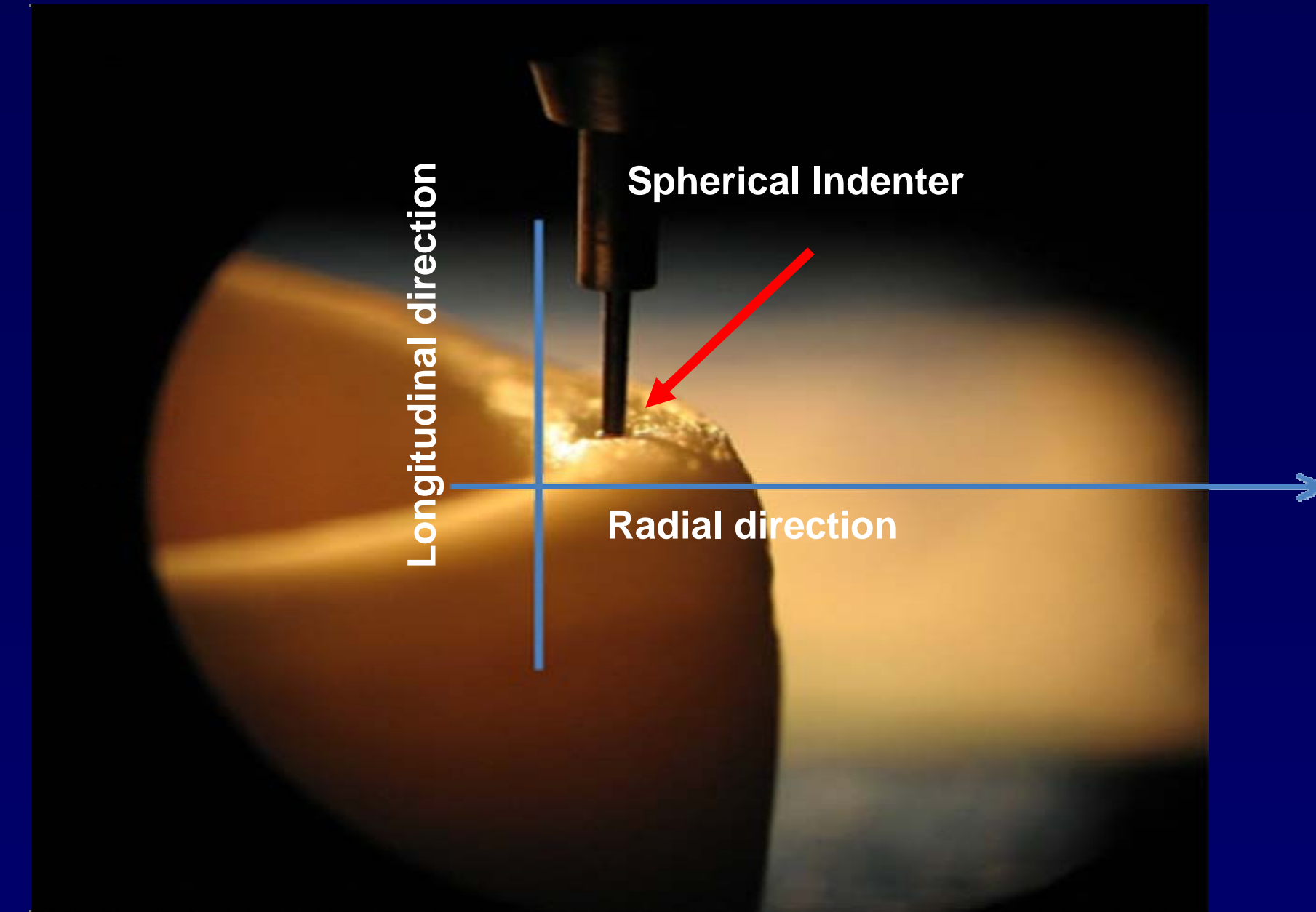


Figure 1. Indentation test setup on aorta wall

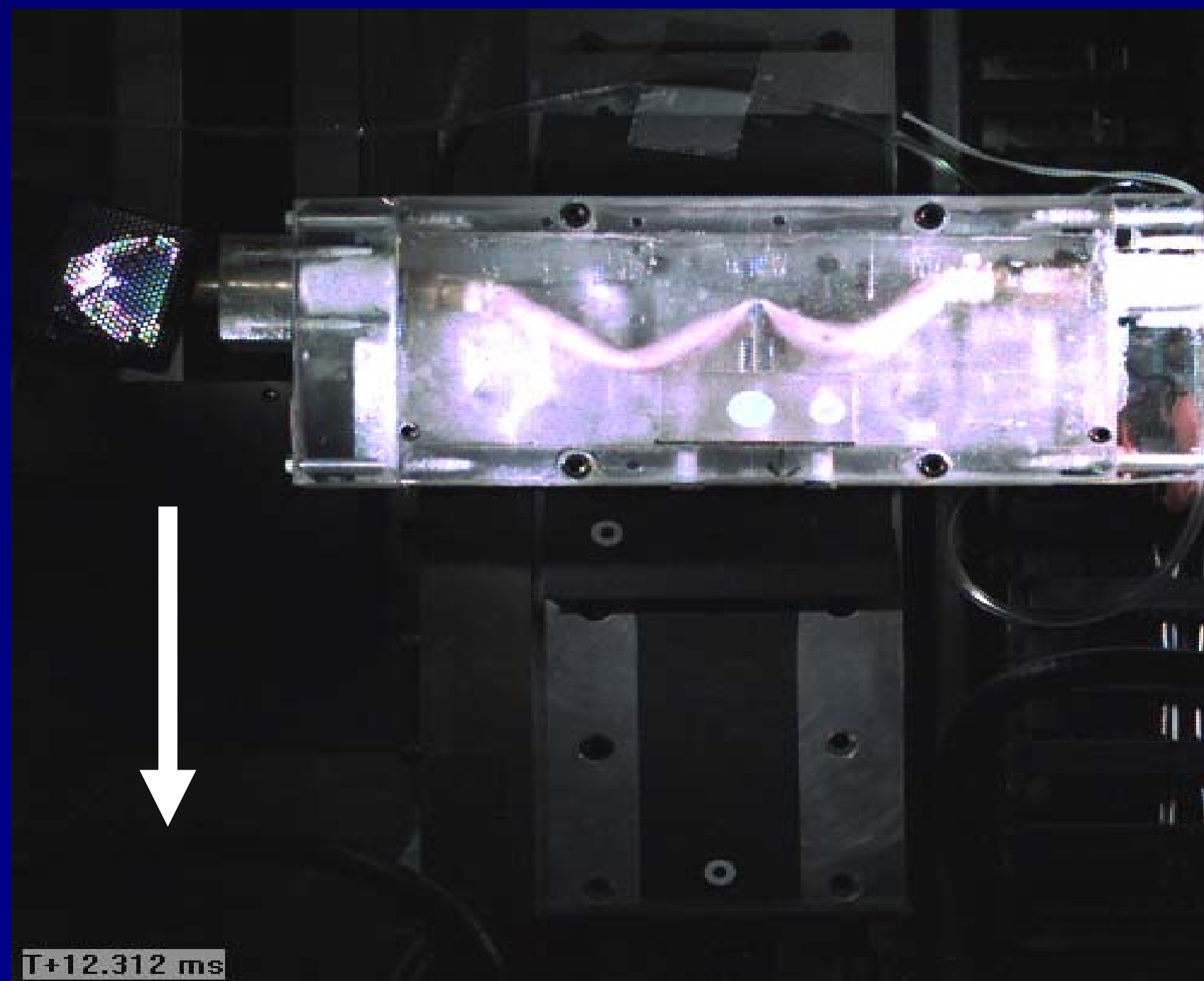


Figure 2. The impact model for pinching of aorta at the time of maximum deflection. The direction of impact is shown.

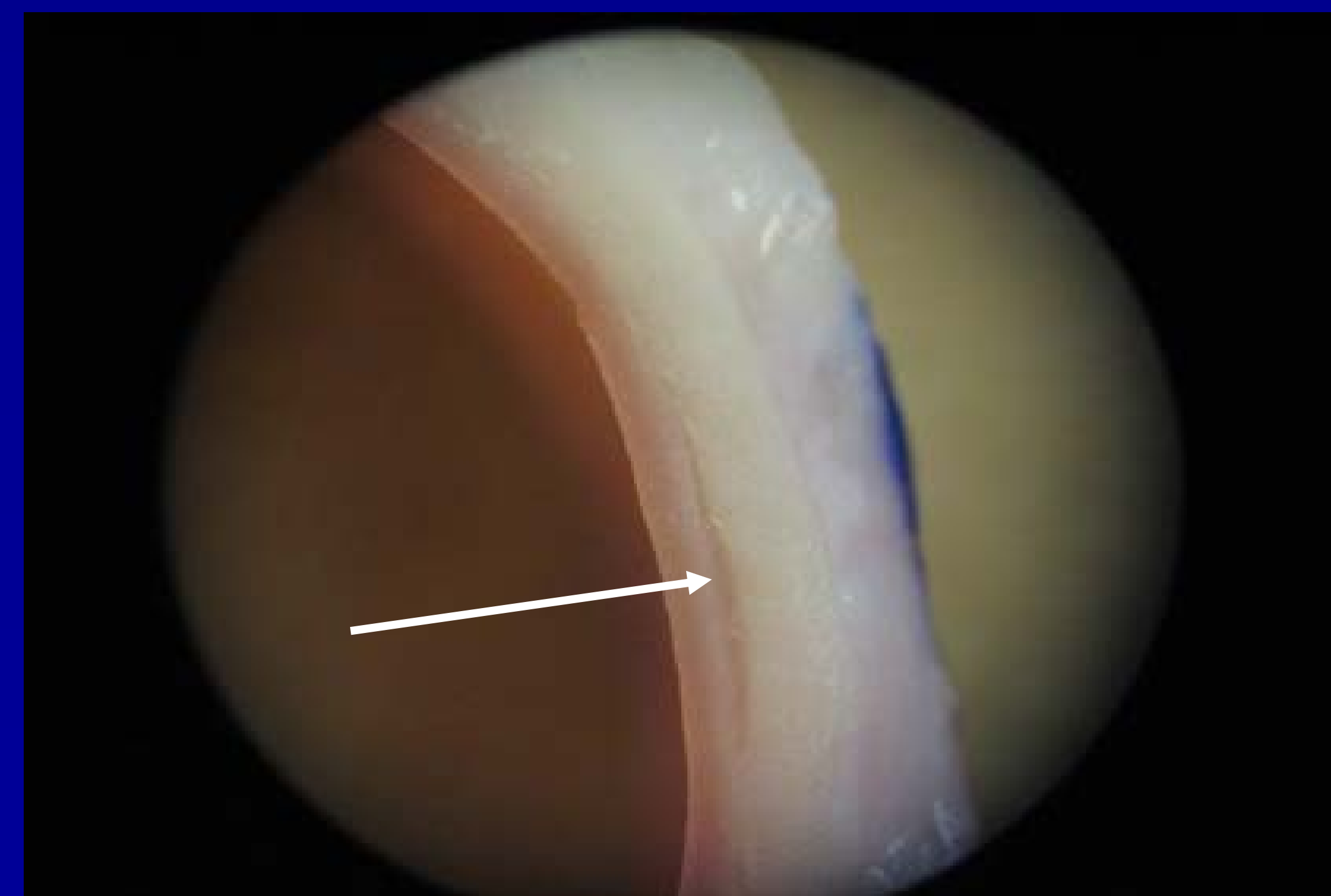


Figure 3. Evidence of local laceration in inner media (shown) near the site of indentation (100x magnification).

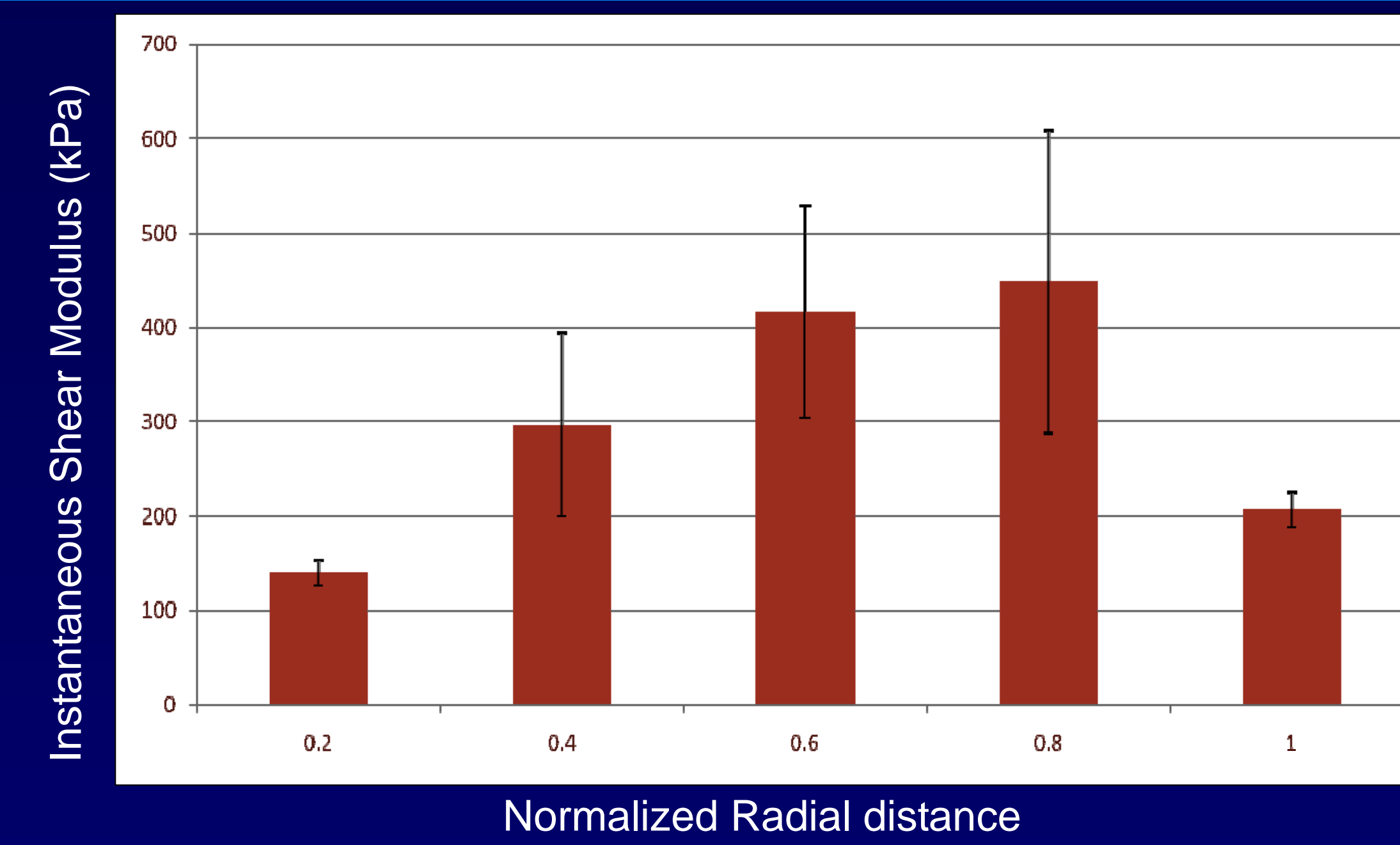


Figure 4. The Instantaneous shear modulus of aorta wall determined from indentation tests

Finite Element results:

- The FE model was able to successfully simulate the experiment in terms of global kinematics (Figure 5)
- Comparison between the maximum proximal and distal deflections in the experimental and FE models showed similar trends (Figure 6).
- The peak deflections in the FE model appear about 5 ms after the experimental results. Also the difference between the maximum distal and proximal deflections was almost double in the FE model.
- These differences were attributed to
 - idealization in the modeling of aorta (homogeneous, isotropic, and constant thickness)
 - approximations in the numerical interface (contact) algorithms.
- The effective strain predicted by the FE model near the indenter was about 30% (Figure 7)

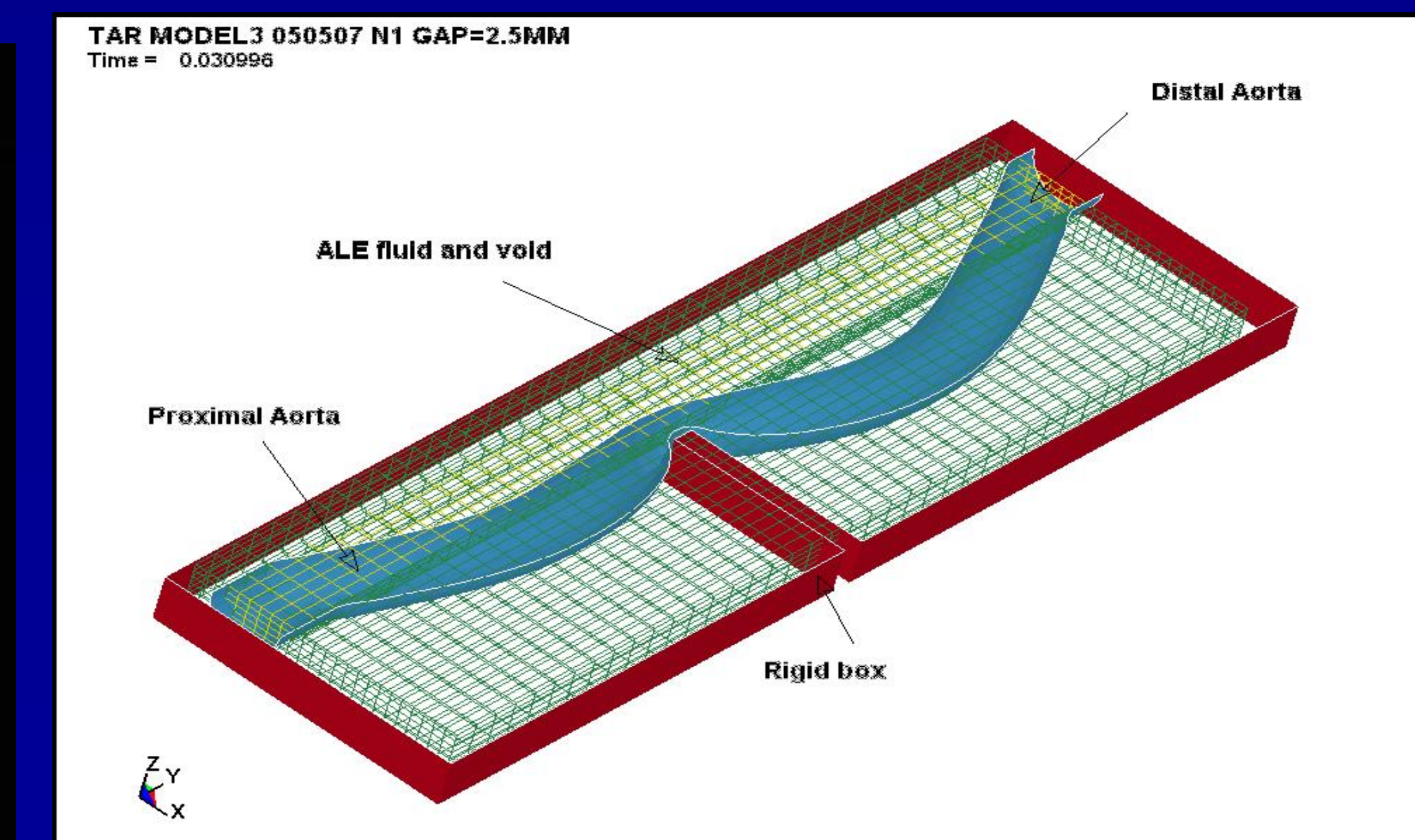


Figure 5. FE model (LSDYNA) of the aorta impact model with local pinching. The model uses ALE formulation and simulates large deformations and contacts in the aorta wall. Effective strain in the indentation site reaches the failure limit.

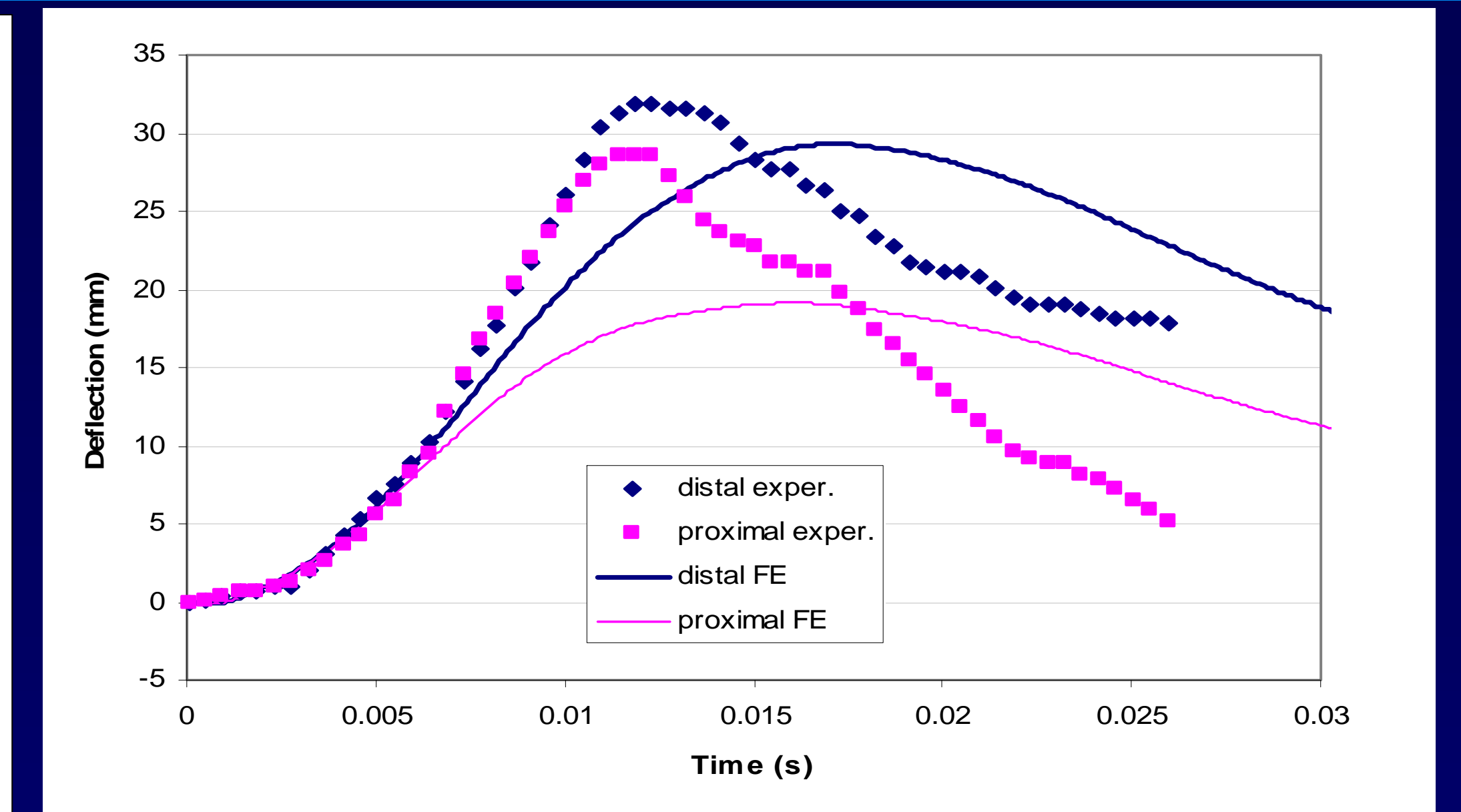


Figure 6. Comparison between the maximum deflections measured in one experiment and its FE simulation.

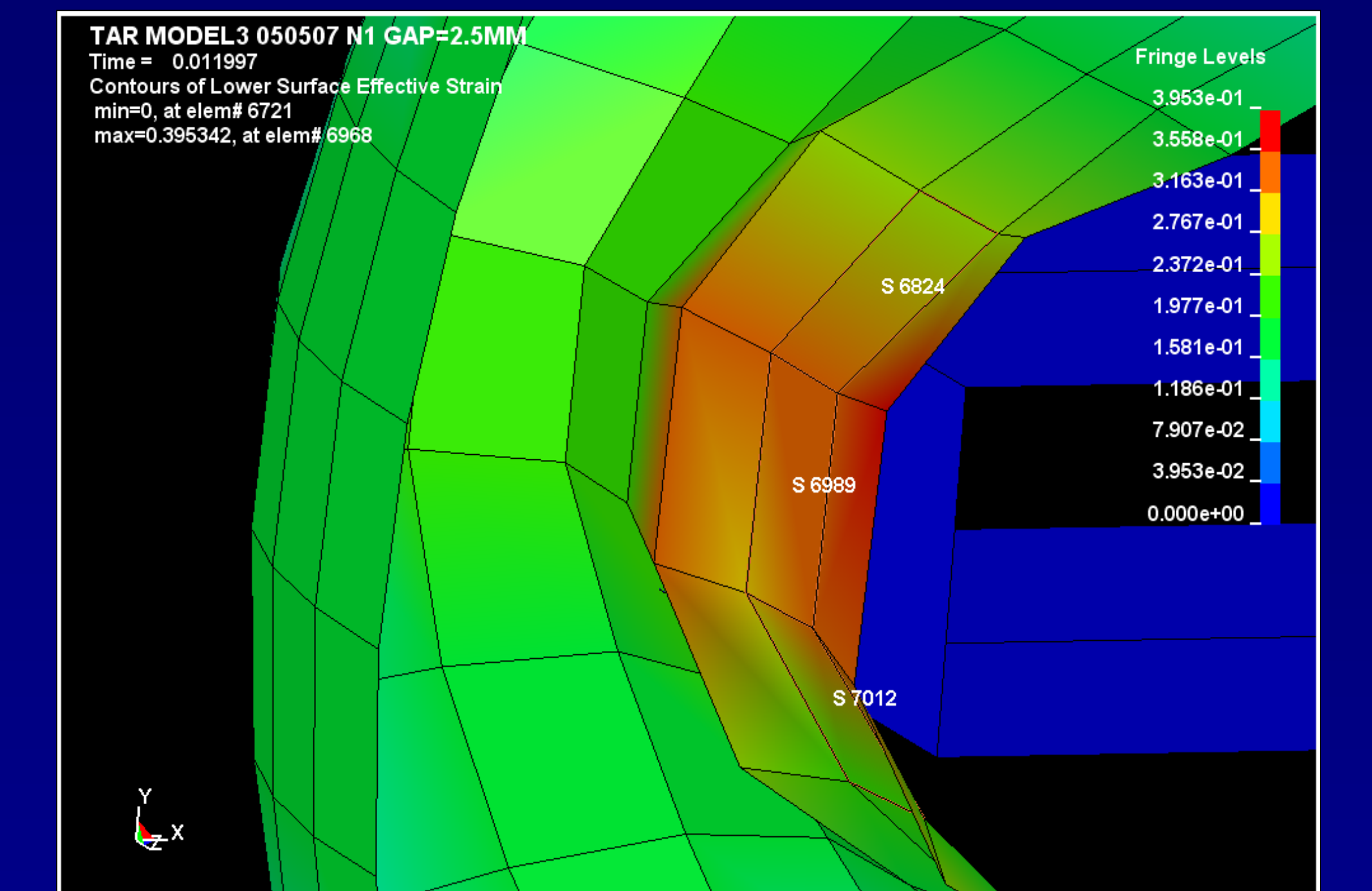


Figure 7. Effective strain around the contact area. The FE simulation showed about 30% maximum strain at the impact site

Conclusions

The models used in this study showed that dynamic local pinching that created about 30% local effective strain caused laceration in the media layer of aorta. This result confirms the role of pinching (direct contact), as a contributing factor in traumatic aortic injury. However this mechanism, alone is unlikely to cause a total transection.

Acknowledgement

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