APPLICATION OF THE PFEM TO THE STUDY OF BLOOD FLOWS AND THEIR INTERACTIONS WITH ARTERY WALLS

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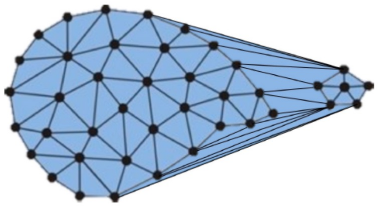
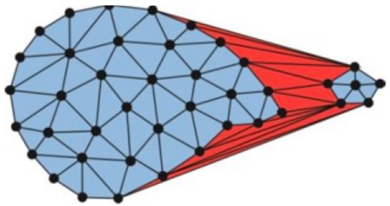
1 The Particle Finite Element Method (PFEM)

Boundaries identification using the α- shape technique

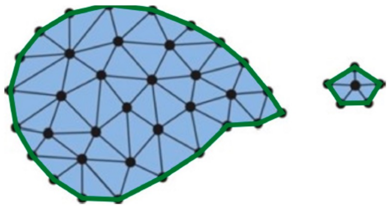
(a) Cloud of particles



(b) Delaunay triangulation

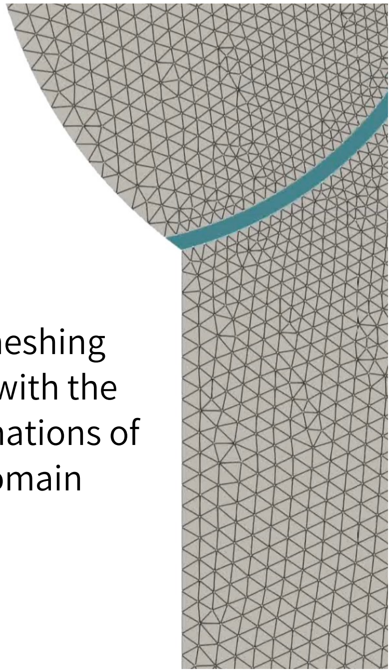


(c) Deletion of spurious triangles (in red)



(d) Identification of boundaries (in green)

Efficient remeshing compatible with the large deformations of the blood domain



M. L. Cerquaglia, *Development of a fully-partitioned PFEM-FEM approach for fluid- structure interaction problems characterized by free surfaces, large solid deformations, and strong added-mass effects,* (2019). PhD thesis, ULiège

2 Fluid-Structure Interaction

Solid model (blood vessel)

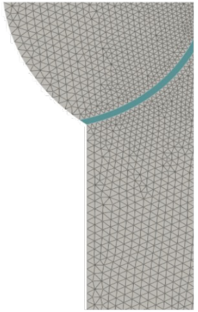
Metafor
Updated Lagrangian / Arbitrary Lagrangian Eulerian

Hyperelastic laws (nonlinear elastic, Mooney-Rivlin, …)

CFD model (blood)

PFEM3D
Particle method

Reconstruction of a mesh based on particles



Dirichlet-Neumann paradigm

1 Predictor displacement $\mathbf{u}_S^\Gamma$

2 $\mathbf{t}_F^\Gamma = \mathcal{F}(\mathbf{u}_S^\Gamma)$ **Fluid**

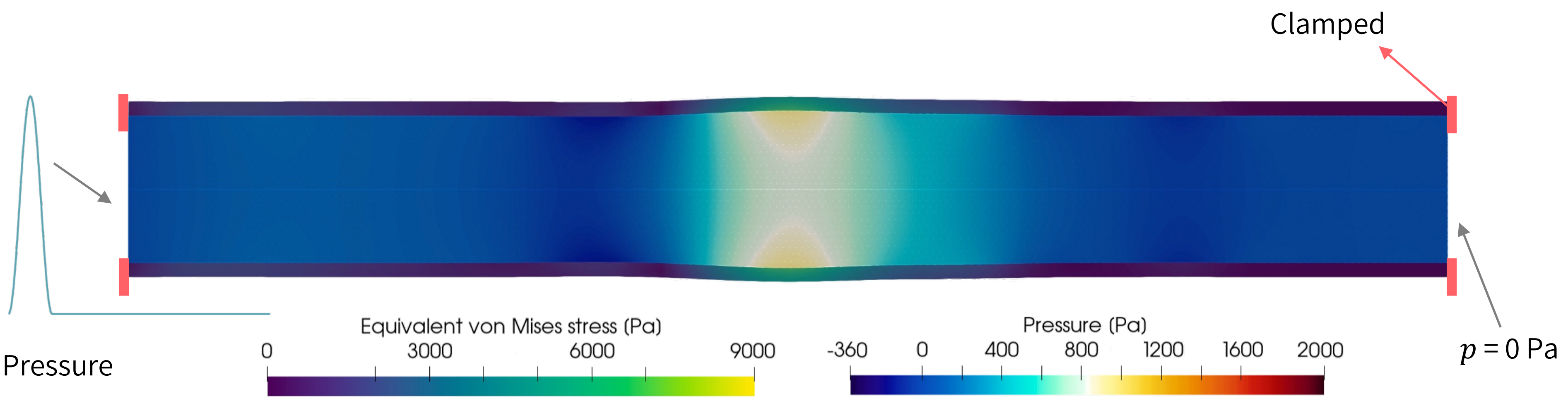
3 Surface tractions $\mathbf{t}_F^\Gamma$

4 $\hat{\mathbf{u}}_S^\Gamma = \mathcal{S}(\mathbf{t}_F^\Gamma)$ **Solid**

S = solid F = fluid Γ = interface

Solved iteratively until $\hat{\mathbf{u}}_S^\Gamma \simeq \mathbf{u}_S^\Gamma$

3 Propagation of a pressure pulse



Verification against Moens-Korteweg theory: $c_0 = \sqrt{\frac{Eh}{\rho d}}$

E = Artery wall Young's modulus
h = Wall thickness
ρ = Blood density
d = Undeformed diameter

4 Blood modeling

- Newtonian: $\tau = \mu\dot{\gamma}$
- **Casson** model

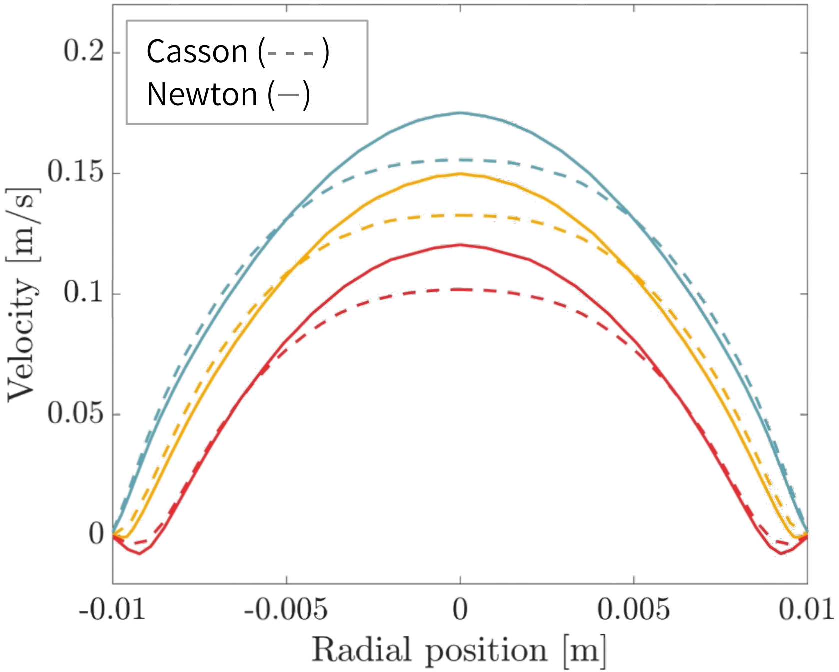
Shear stress τ
Asymptotic viscosity μ_∞
Viscosity μ
Yield stress τ_y
Shear rate $\dot{\gamma}$

$$\begin{cases} \tau = \mu(\dot{\gamma})\dot{\gamma} & \text{if} \quad |\tau| \geq \tau_y \\ \dot{\gamma} = 0 & \text{if} \quad |\tau| < \tau_y \end{cases}$$

$$\mu(\dot{\gamma}) = \left(\sqrt{\mu_\infty} + \sqrt{\frac{\tau_y}{|\dot{\gamma}|}} \right)^2$$

Negligible difference in case of **high shear** and/or **highly pulsatile** flow.

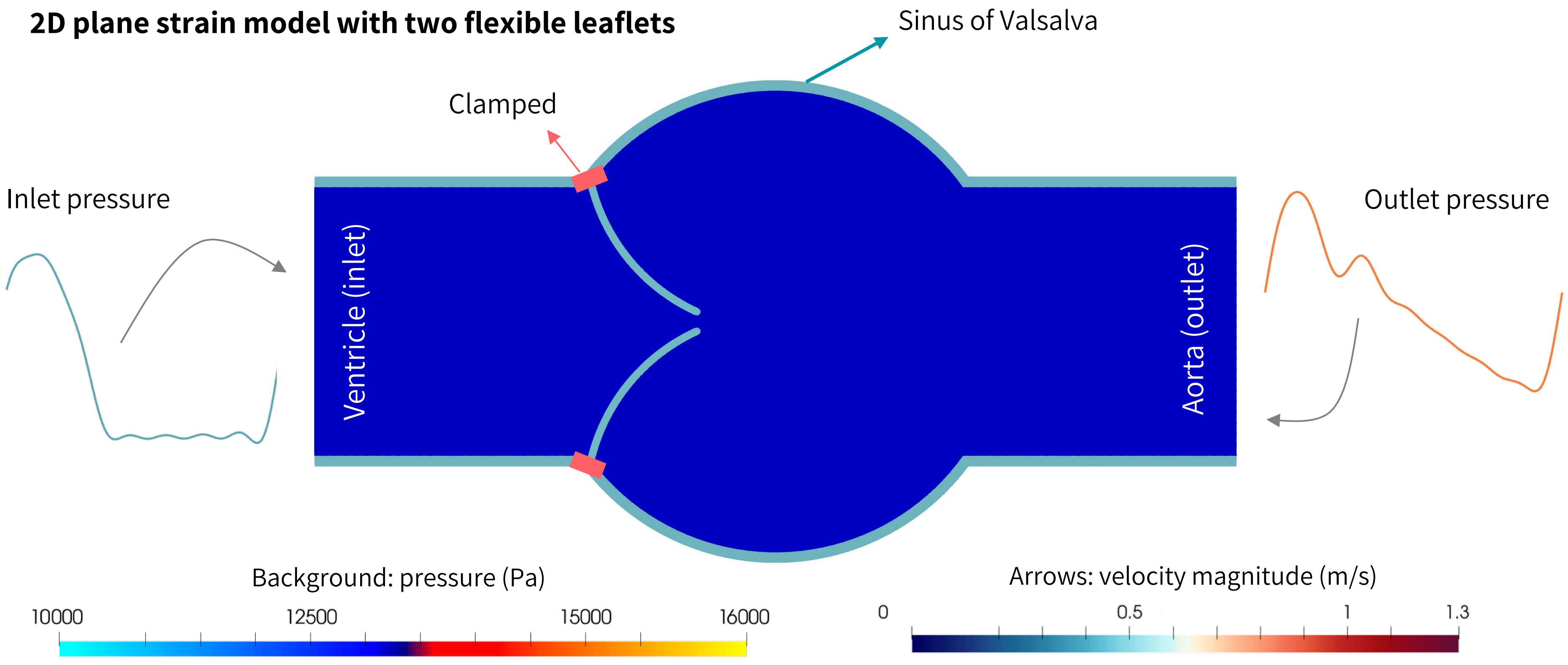
Velocity profiles at successive times of the cardiac cycle in **the inferior vena cava**.



t = 0.4 s (—), t = 0.56 s (—), and t = 0.8 s (—)

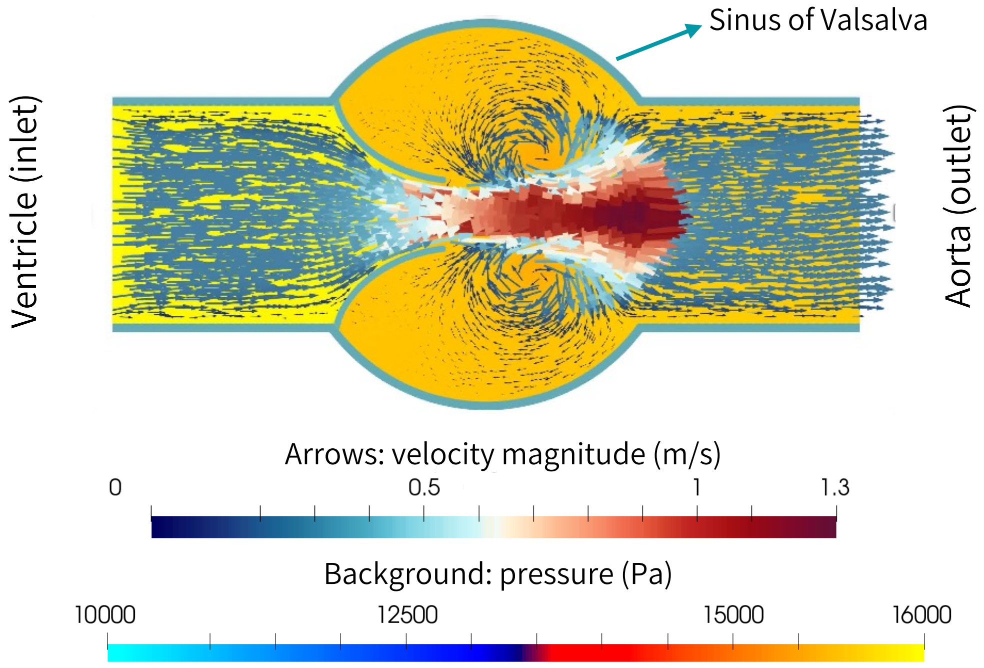
Aortic valve modeling

2D plane strain model with two flexible leaflets

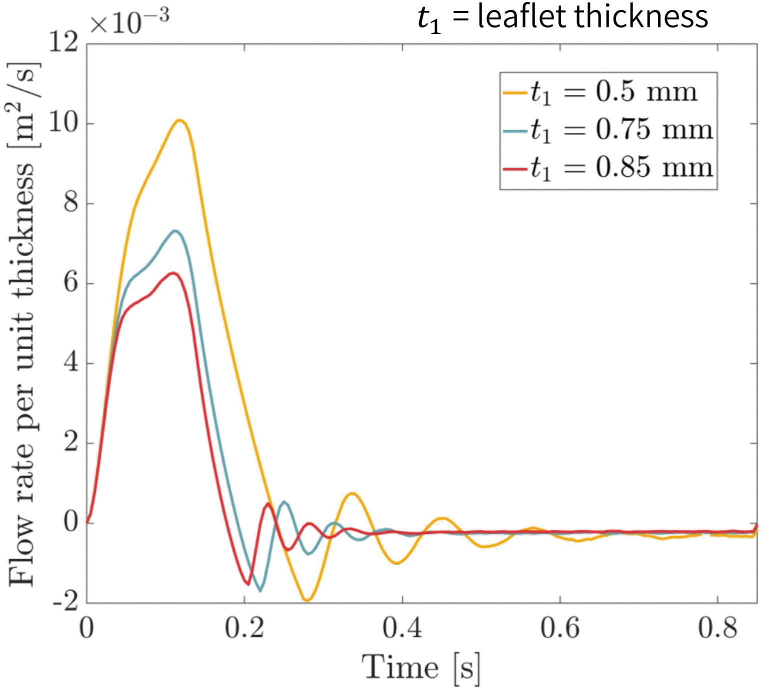


5 Aortic valve modeling

2D plane strain model with two flexible leaflets

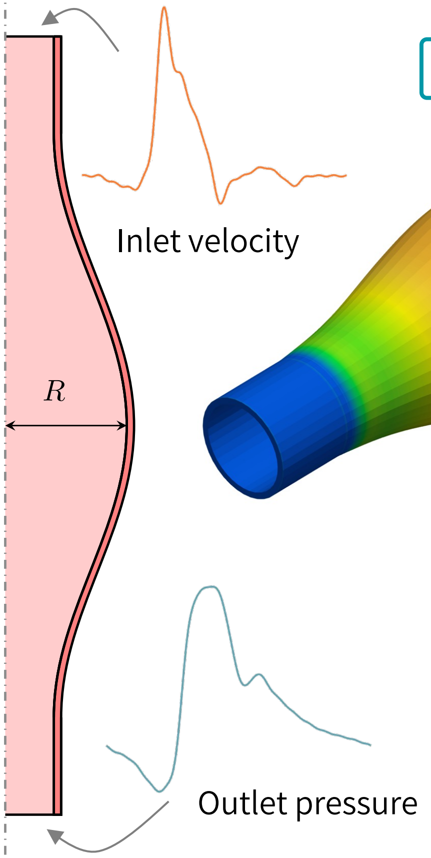


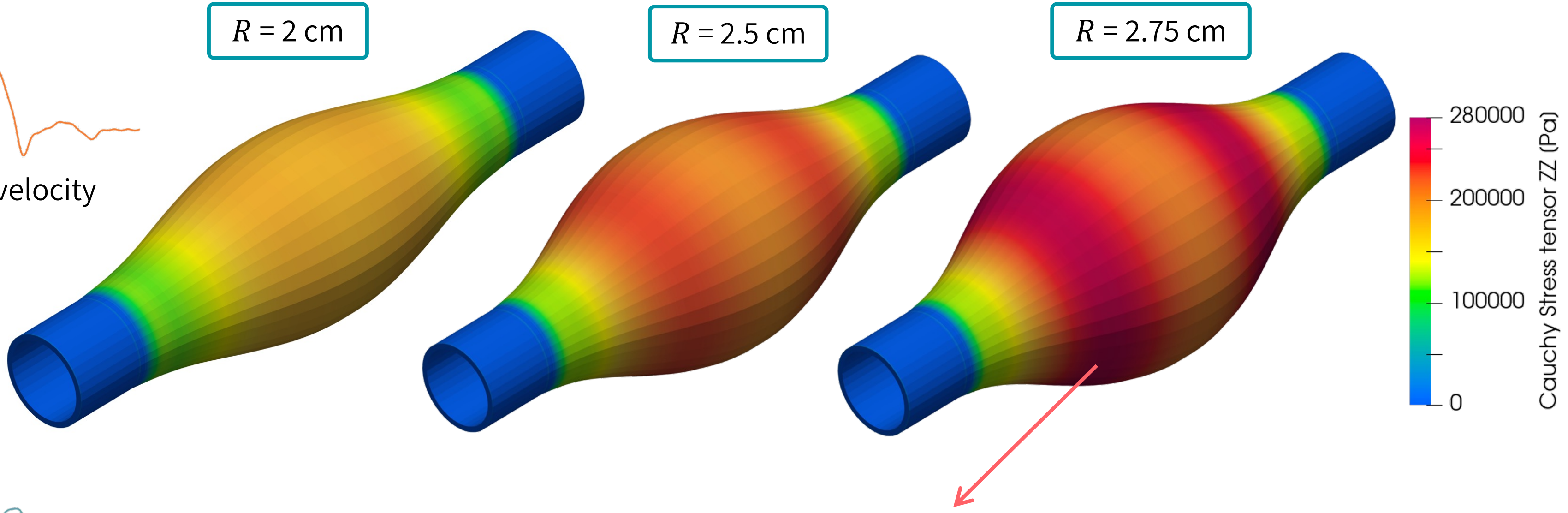
t_1 = leaflet thickness



Impact of calcification on the heart ability to pump blood throughout the body

6 Axisymmetric abdominal aortic aneurysm



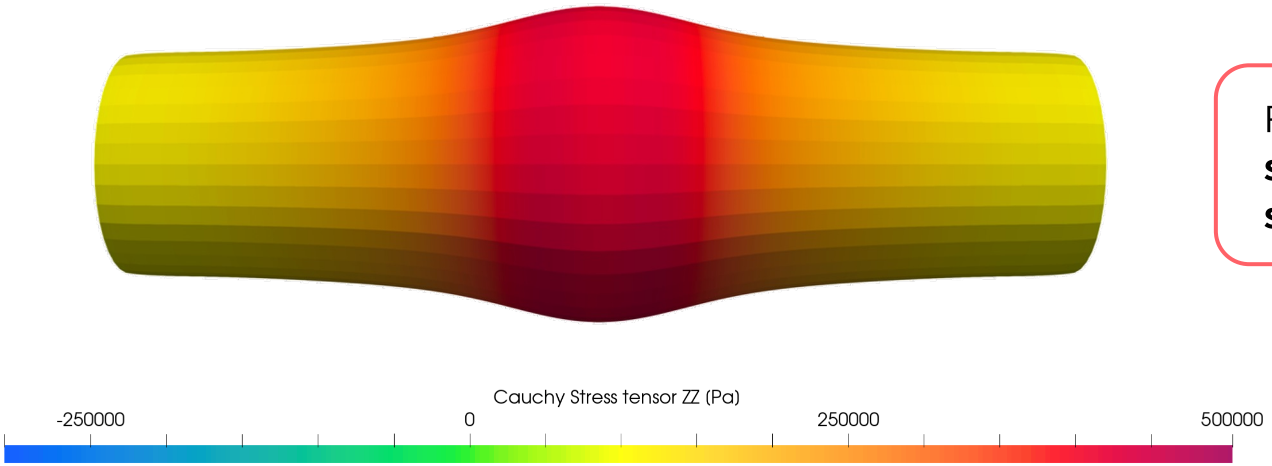


Peak hoop stress around the points of **inflection**

Peak hoop stress increases with R → Importance of the **size** of the aneurysm for rupture risk assessment

7 Aneurysm initiation and stability

E = 0.675 MPa (stiffness of healthy artery)
Thickness reduced by 50% in the middle of the artery:

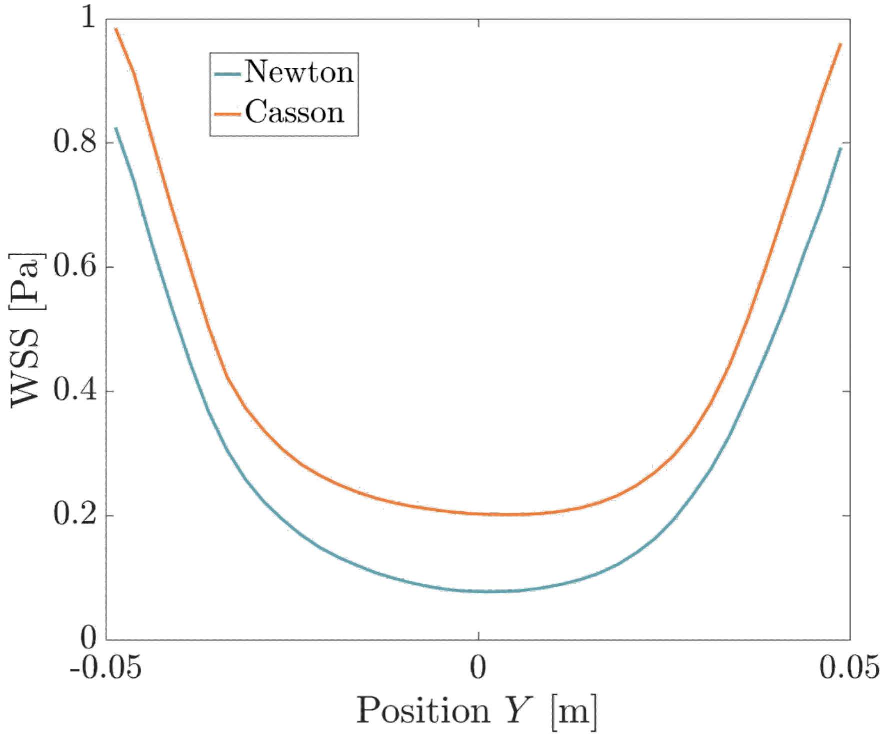


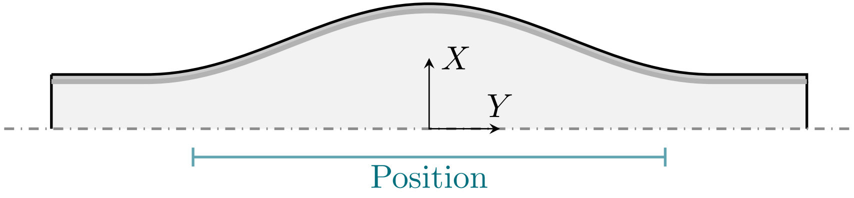
Rupture risk depends on the **stiffness** of the wall and the **size** of the aneurysm.

Initial **weakness** should be compensated by **remodeling** to avoid instability and rupture.

7 Aneurysm initiation and stability

Time-averaged wall shear stress along the aneurysm





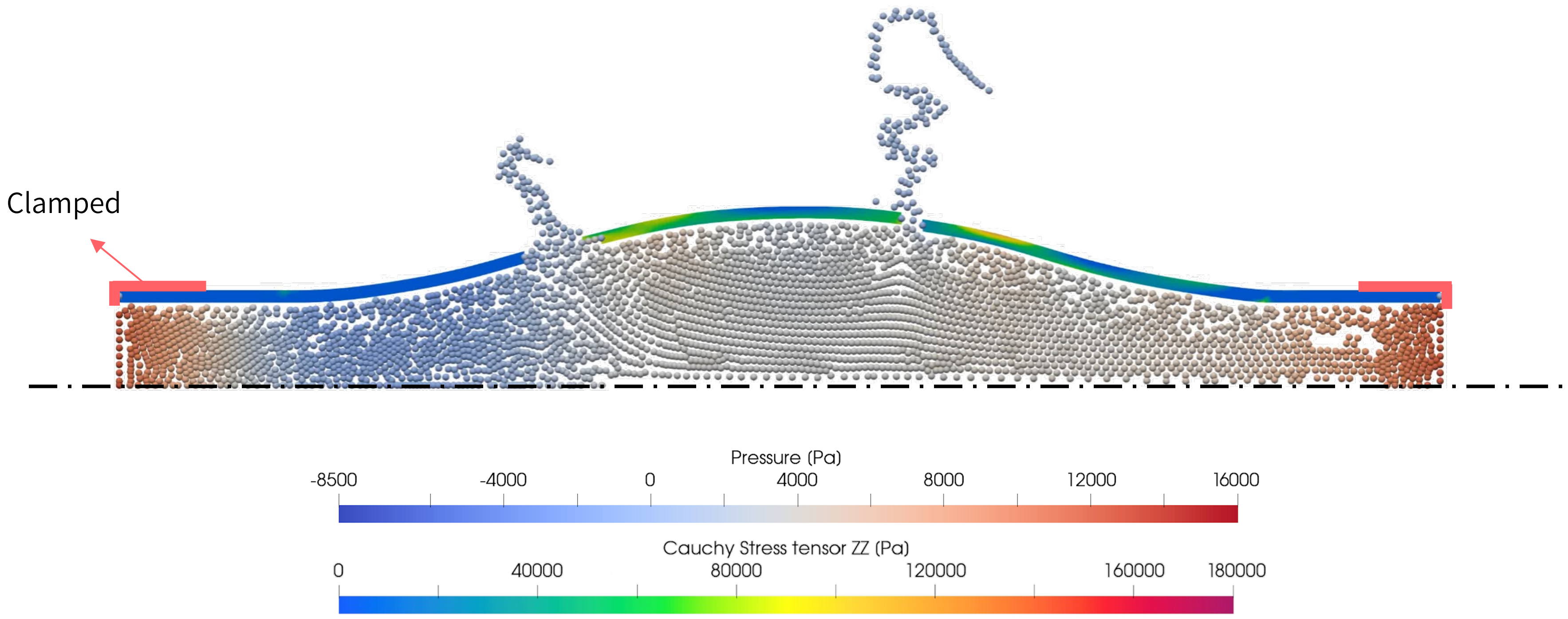
Physiological importance
of the **wall shear stress (WSS)**

→ Low WSS is pathological
(risks of atherosclerosis and aneurysm rupture)

→ Importance of fluid model for accurate diagnosis

8 Abdominal aortic aneurysm rupture

Validation against clinical results: rupture occurs close to an **inflection** point of the aneurysm.



9 Conclusions and perspectives

First application of the PFEM to the study of blood flows and their interactions with artery walls

Numerical improvements?

- Turbulence model
- Boundary conditions to cope with signal propagation

Perspectives?

- More realistic arterial constitutive laws (anisotropy)
- Improved geometries (three-dimensional, patient-specific)

Thank you for your attention!

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