Computational Modeling of High Risk Aortic Manipulation during Open-Heart Surgeries

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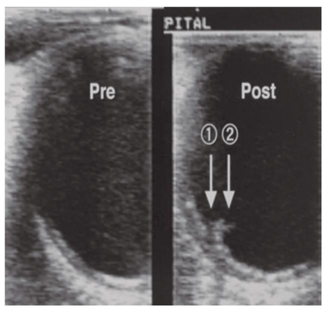
Introduction

Worldwide, approximately 2 million open-heart surgeries are done annually [1]. A major risk factor of these procedures include end-organ ischemia, including stroke, with an incidence of 0.9-13%, depending on the procedural complexity [2]. At least 50% of the peri-operative embolic load is caused by aortic manipulation, including especially aortic occlusion and arterial cannular flow [2,3], as also seen in clinical findings shown here on the right.

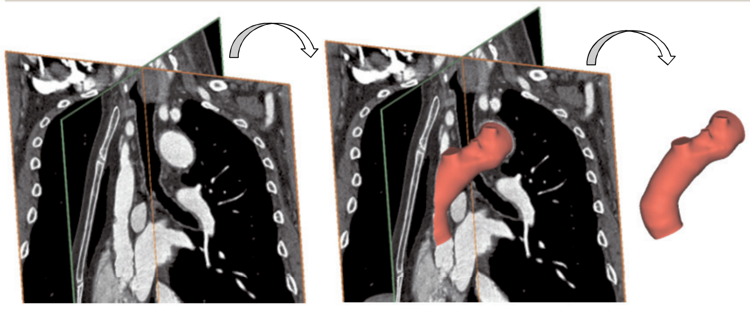


Ultrasound image revealing likely aortic manipulation related injury, showing a new intimal tear (1) and a new mobile lesion (2) – adapted from [4]

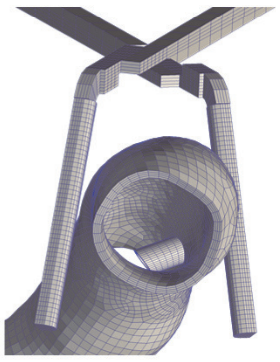
Objectives

Detailed knowledge of the patient-specific impact of aortic manipulation, especially w.r.t. high risk maneuvers such as arterial cannular flow and aortic occlusion is needed to not only aid in clinical decision making and protocol evaluation, but also for improved product design.

Aortic Occlusion of Calcified Aorta

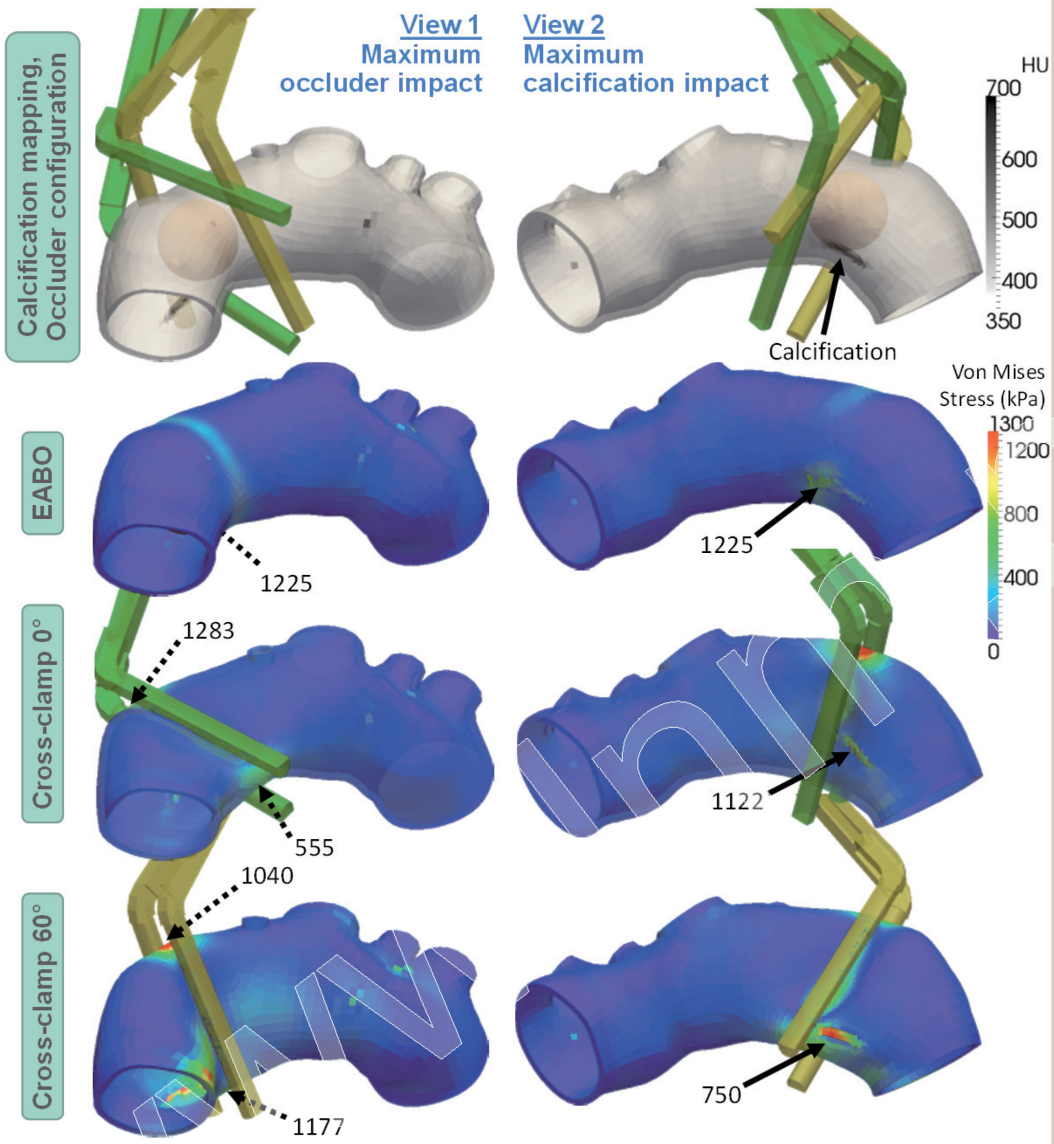


Patient-specific (male, 66 yrs) aortic lumen geometry with major atherosleротic calcification was extracted from medical CT-data.



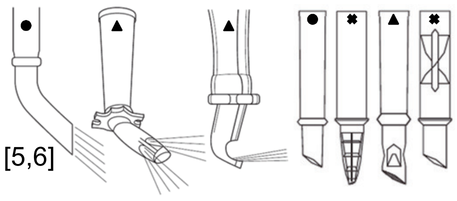
The outer surface of the lumen was meshed selectively to capture the most important features of the vessel deformation during occlusion. On this surface mesh, a pure HEX-mesh could be extruded (4 layers, 2.3 *mm* total thickness) in CUBIT (Sandia Nat. Labs. Albuquerque, NM, USA), including also an arterial cannula mesh. Standard DeBakey cross-clamps was applied at two different orientations rotated around the vessel axis by 60°. At the same location, an endo-aortic balloon (EAB) was also applied.

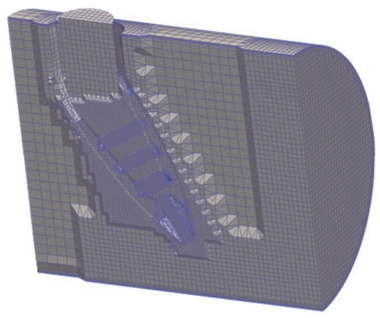
The aortic wall was modelled with an isotropic, hyperelastic Raghavan&Vorp model developed for aneurysmatic arterial walls (α = 0.174 MPa, β = 1.88 MPa), with regional stiffening at the calcified wall, mapped to and relative to the Houndsfield Units (HU) extracted from the CT-data [7]. Prior to occlusion, the aorta was prestressed to approximate the imaged internal pressure of the aorta using a Modified Updated Lagrangian Formulation [8]. The clamps and the EAB was modelled in a way to mimic the actual clinical load exerted on the artery. The structural contact simulation was performed using a reduced integration, linear FE scheme on our in-house FE solver *baci*.



Arterial Cannular Flow influenced by Tip Design

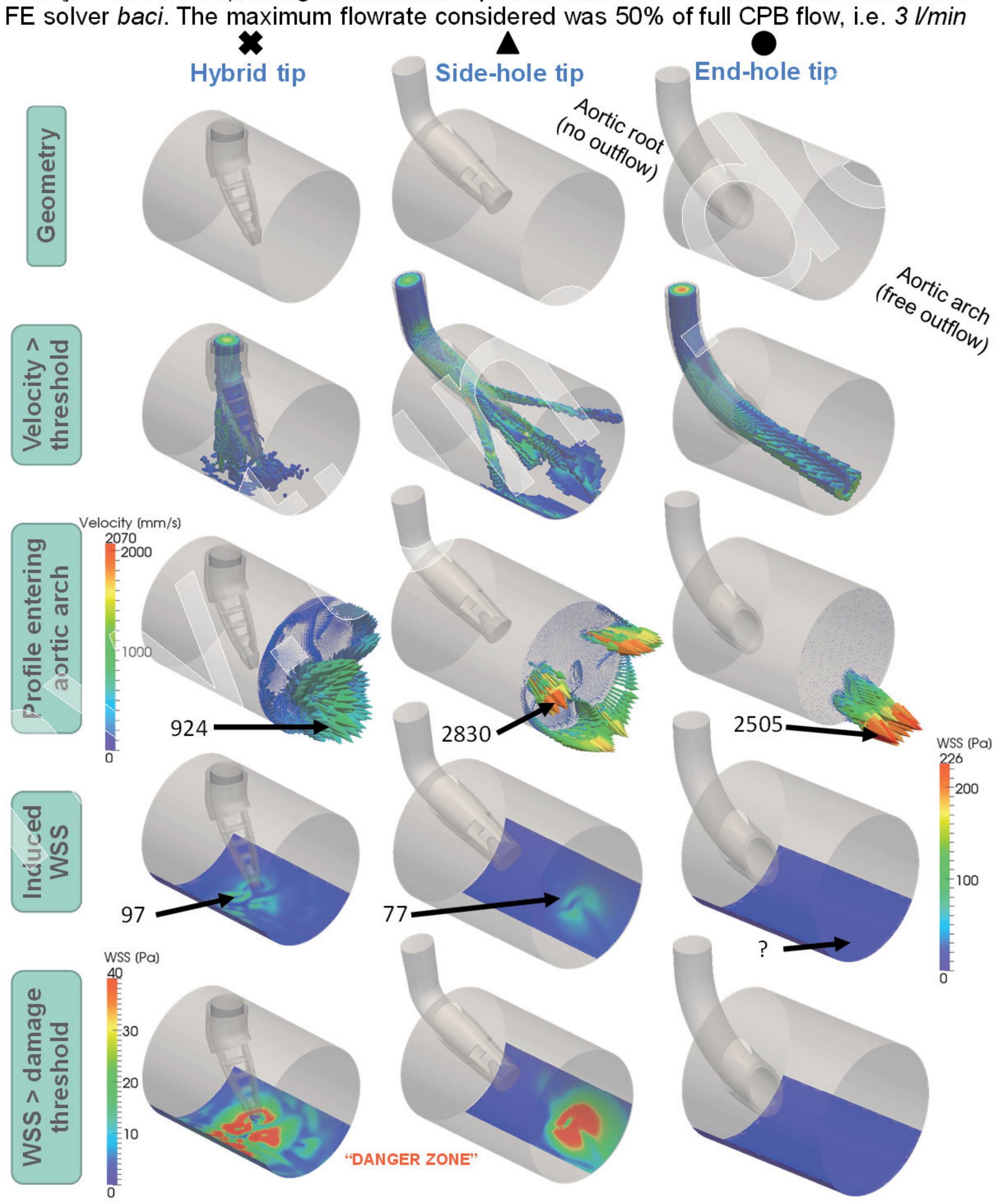
The three main different types of clinically available arterial cannula tip-designs was considered in an idealized arterial segment.





Each fluid domain was discretized in Harpoon (Sharc Ltd., Manchester, UK) with a HEX-dominant mesh with selective refinement and boundary layer meshes to ensure sufficient resolution especially of flow through the cannula, the jet flow through the vessel prior to hitting the wall and the jet flow where it hits the wall.

The blood flow through the cannulas were simulated as an incompressible, newtonian fluid (μ = 0.004 $Pa \cdot s$), using a stabilized, equal-order, linear FE scheme on our in-house FE solver *baci*. The maximum flowrate considered was 50% of full CPB flow, i.e. *3 l/min*



Discussion and Future Challenges

Large deformations caused by **occlusion exerts significant high stresses** in the arterial wall, of which the magnitude and therefore **risk of damage is highly variable, depending unintuitively on the of occluder, occluding location and patient-specific arterial wall constitution.**

The tip design of **arterial cannulas determines the distribution of flow entering the aortic arch**, and therefore the cerebral circulation, **as well as** the amount of damage caused by the **dangerous "sand-blasting" effect** of the high velocity jet against the arterial wall. No design completely fulfills the requirements of safe arterial return.

Future investigations are needed to further **improve models that will help to optimize clinical protocol and aid in device design**.

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