CHALMERS

Fluid-Structure Interaction Analysis of the Forces Causing Stent Graft Migration

Patrik Andersson
Johan Pilqvist

Chalmers University of Technology
Supervisors:
Professor Ragnar Larsson
Associate Professor Håkan Nilsson

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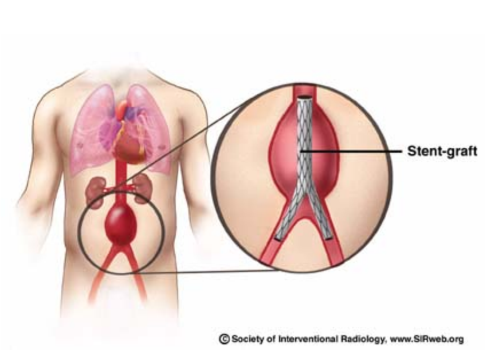
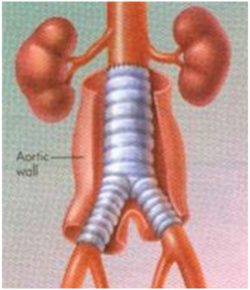
Agenda

Introduction

- Abdominal aortic aneurysm
  - A localized dilation of the abdominal aortic vessel
  - Common for males of 65 years of age and older
- Common treatments
  - Open surgery - Dacron or e-PTFE (gore-tex) tube - Graft
  - Inserted internal reinforcement, relining the vessel - Stent graft
    - Endovascular Aortic Repair (EVAR)

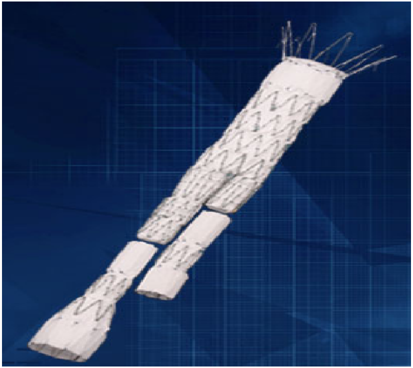




Introduction
Problem description

Difficulties regarding attachment of the reinforcing structure

- Anchor hooks only at upper attachment point
- Lower extensions kept in place only by self expansion
- Stent graft migration → pressurized blood flow in the aneuyrysm → increased risk of rupture

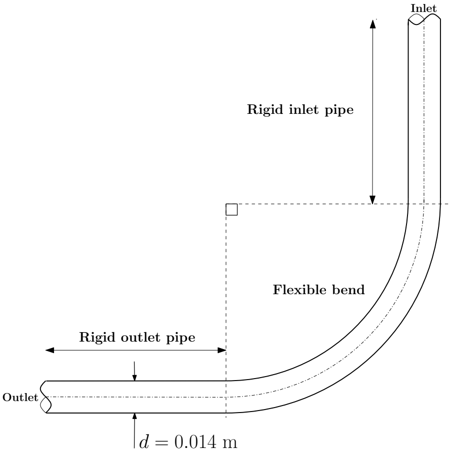


Introduction
Purpose

- In 2004, a study by Zarins et al. showed that 18.8% of 1119 patients experienced stent graft migration
- Forces causing stent graft migration of the non-fixated distal attachments are highly interesting.
- An experimental study performed by Malina et al. demonstrates that these forces range between 2 and 3.4 N.
- Previous studies by e.g. Li & Kleinstreuer show forces of these magnitudes from FSI-simulations.
- Comparison of two different numerical approaches when performing FSI simulations

Introduction
Method

- Two parallel FSI studies, using two different softwares
  - LS-DYNA (LD), FE-based
  - OpenFOAM (OF), FV-based
- Two different flow scenarios
  - Steady flow
  - Pulsating flow
- Simple momentum balance calculation of deflected inviscid flow.



Introduction

Limitations, assumptions and simplifications

- Turbulence not modelled
- No gauge pressure
- Gravity is neglected
- Symmetry
- Fluid medium set to water
- Only one distal extension is simulated
- The flexible bend is in an initial stress free state
- The pulsating flow is assumed to do so sinusoidally with 60 bpm
- Pipe walls modelled as smooth

Introduction

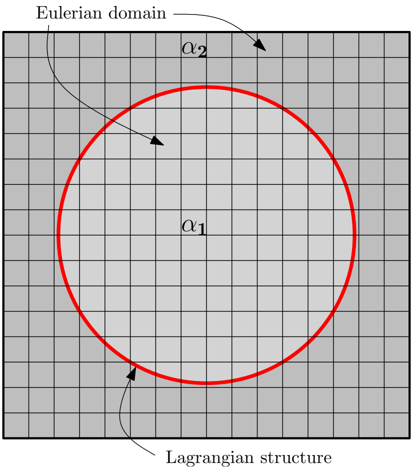
Limitations, assumptions and simplifications

Material properties

- Fluid
  - Properties of water at 20°C
- Structure (extracted from prior studies)
  - Material is considered isotropic and homogeneous, disregarding the metallic mesh of the stent $\rightarrow$ Endovascular Graft (EVG)
  - Young's modulus (E) = 10 [MPa]
  - Poisson's ratio (η) = 0.27 [-]
  - Density = 6000 [kg/m^3]

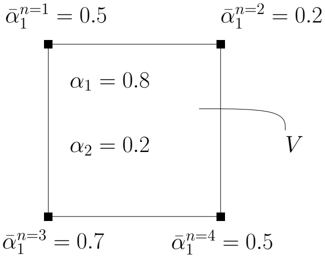
Computational methods
LS-DYNA

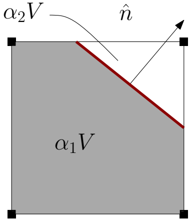
- Lagrangian structure constrained within a fixed, independent Eulerian mesh
- Two Eulerian domains containing the fluid (water) and a dummy material, respectively.
  $\rightarrow$ Multi-Material ALE (MMALE)
- α_1 (water) is the flowing material interacting with the Lagrangian structure



Computational methods
LS-DYNA

- Mixed elements are cut with a plane separating the materials
- The orientation of the plane is controlled by the gradient of the volume fraction field (i.e. distribution of α_1 and α_2), which is governed by the Lagrangian structure
- When the structure moves, the volume fractions are updated and the interface plane is reconstructed accordingly





$$\hat{n} = \left|\left|\frac{\partial \bar{\alpha}_1}{\partial x}\right|\right| \cdot \frac{\partial \alpha_1}{\partial x} = \frac{1}{\sqrt{2}} \begin{Bmatrix} 1 \\ 1 \end{Bmatrix}$$

Computational methods
LS-DYNA

Governing equations

- Fluid
  - Continuity and momentum equation in the Eulerian framework:

$$\frac{\partial \rho}{\partial t} + \nabla \cdot (\rho \mathbf{v}) = 0$$

$$\frac{\mathrm{d}}{\mathrm{d}t}(\rho \mathbf{v}) = -\nabla p + \mu \nabla^2 \mathbf{v}$$

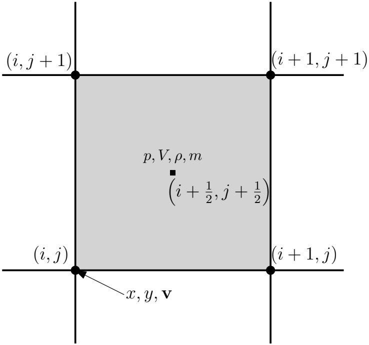
- Structure
  - Conservation of momentum in the Lagrangian framework:

$$\rho_0 \frac{\mathrm{d}\mathbf{v}}{\mathrm{d}t} = \boldsymbol{\mathcal{P}} \cdot \nabla_{\mathbf{X}} + \rho_0 \mathbf{b}$$

$\boldsymbol{\mathcal{P}}$ is the first Piola-Kirchhoff stress tensor.

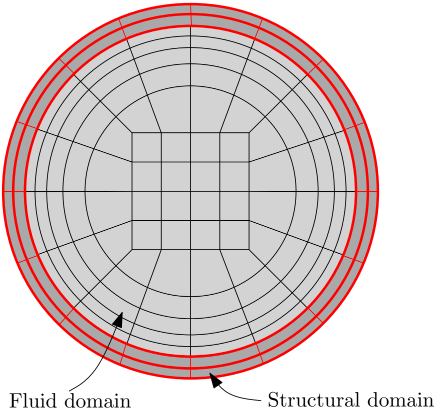
Computational methods
LS-DYNA

- Constraint based coupling algorithm
  - Conservation of momentum but loss in kinetic energy
- Bulk modulus of water set to $2.2 \cdot 10^6$ $(< 2.2 \cdot 10^9)$ Pa
- van Leer MUSCL advection scheme + Half-Index Shift (HIS) advection algorithm



Computational methods
OpenFOAM

- Two separate meshes; one fluid and one structural mesh
- User defined interface
  $\rightarrow$ mesh deformation
- Requirement of volume elements
  $\rightarrow$ increase of material thickness



Computational methods
OpenFOAM

Governing equations

- Fluid
  - Incompressible continuity and momentum equation in the Eulerian framework:

$$\nabla \cdot \mathbf{v} = 0$$
$$\frac{\mathrm{d}\mathbf{v}}{\mathrm{d}t} = -\frac{1}{\rho}\nabla p + \nu \nabla^2 \mathbf{v}$$

  - Linear upwind advection scheme

- Structure
  - Conservation of *incremental* momentum in the *updated* Lagrangian framework:

$$\rho_u \frac{\mathrm{d}\delta\mathbf{v}}{\mathrm{d}t} = (\mathbf{F}_u \cdot \mathbf{\Sigma}_u) \cdot \nabla_{\mathbf{X}} + \rho_u \delta\mathbf{b}$$

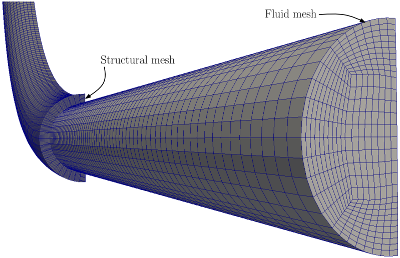
Computational methods

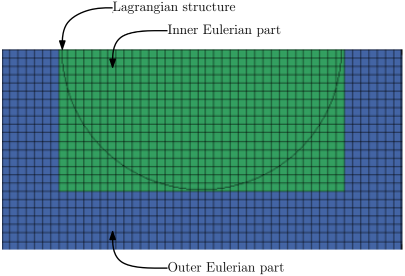
Boundary conditions

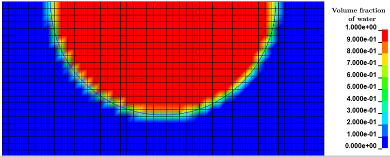
Software	Boundary	Type	Value
OF	Inlet	Steady-state velocity	0.5 [m/s]
		Periodic velocity	$0.5 + 0.5sin(2\pi ft)$ [m/s]
	Outlet	Mean pressure	0 [Pa]
	Rigid walls	No-slip	-
	Flexible walls	Moving wall velocity	-
	Symmetry plane	Symmetry	-
LD	Inlet	Steady-state velocity	0.5 [m/s]
		Periodic velocity	$0.5 + 0.5sin(2\pi ft)$ [m/s]
	Outlet	Zero traction	0 [N/m^2]
	Rigid walls	No-slip	-
	Flexible walls	Moving wall velocity	-
	Symmetry plane	Symmetry	-

Computational methods
Meshes

Two different meshes

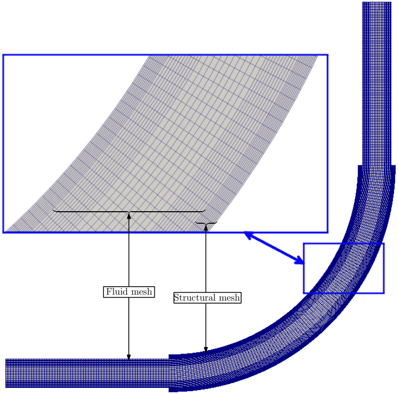


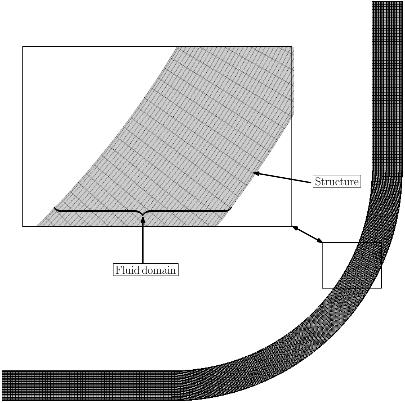




Computational methods
Meshes

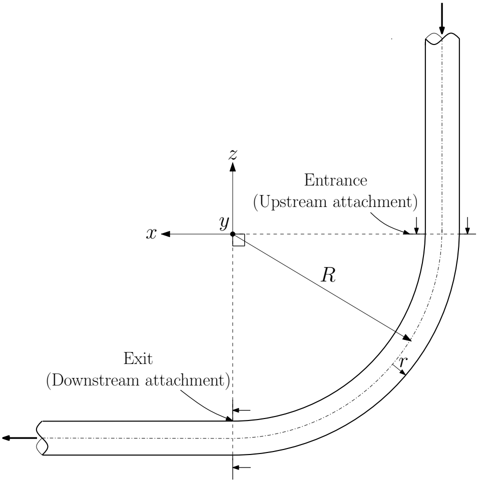
- OF cell count: $108,360$
- LD cell count: $285,768$ fluid elements, $13,616$ structural (shell) elements





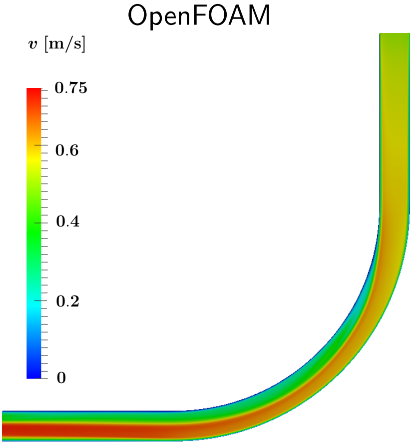
Results
Overview

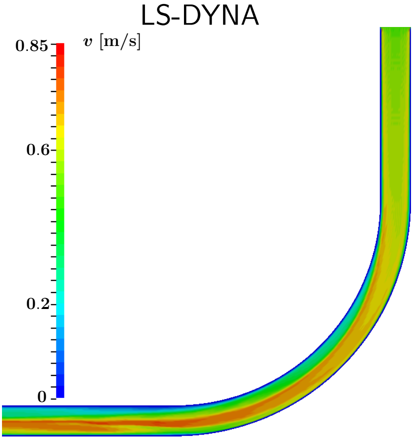
Two coupled analyses on a bent EVG setup using a steady and a sinusoidal inlet velocity boundary condition



Results
Steady FSI

Velocity fields, steady inlet velocity: $|\mathbf{v}_{inlet}| = 0.5$ m/s

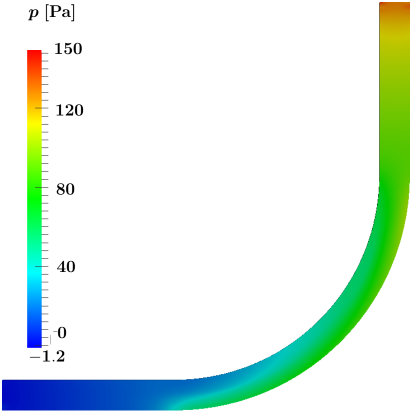




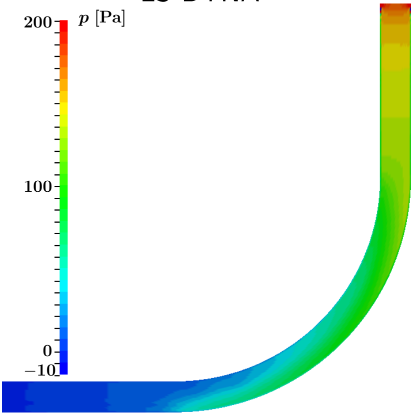
Results
Steady FSI

Pressure fields, steady inlet velocity: $|\mathbf{v}_{inlet}| = 0.5$ m/s

OpenFOAM

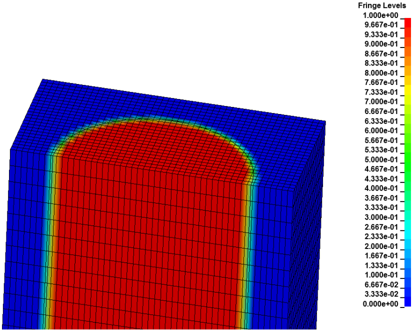


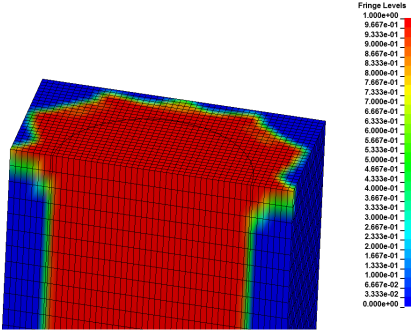
LS-DYNA



Results
Steady FSI

- Differences in magnitude may be due to
  - Different meshes
  - Different boundary conditions
  - Numerical leakage at inlet in LD

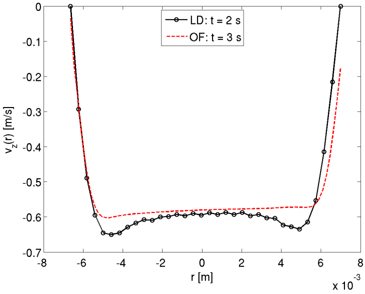


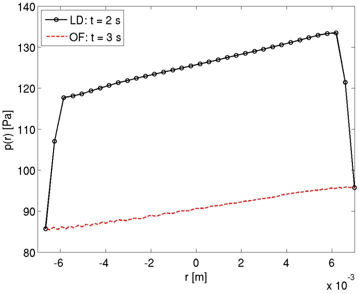


Results
Steady FSI

Cross-sectional velocity and pressure profiles (upstream)

- Boundary layer less developed in LD
  - Lower mesh density in near wall region
  - (Different advection algorithm)
- Volume fraction method used in LD creates uncertainties regarding the near wall solution variables.

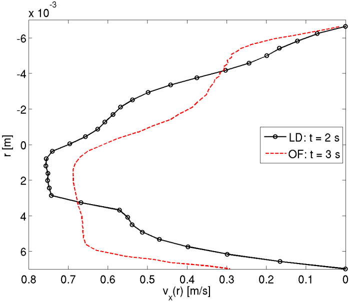


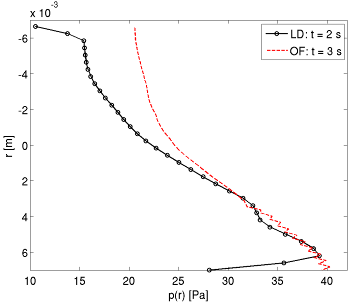


Results
Steady FSI

Cross-sectional velocity and pressure profiles (downstream)

- Laminar flow in OF, occurence of numerical instabilities in LD
    - Different meshes
    - Different advection algorithms
- The flow separation is validated by the pressure gradients

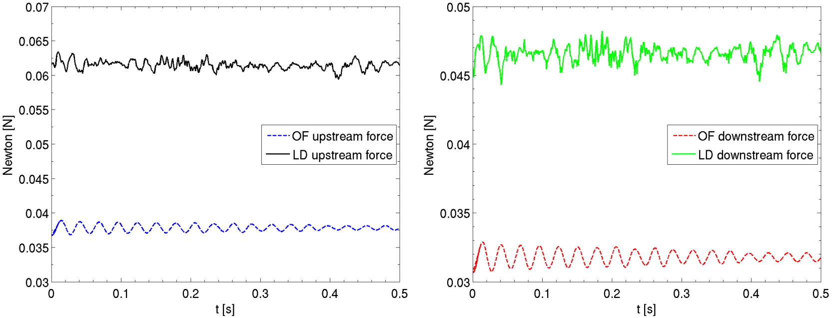




Results

Steady FSI

Upstream and downstream normal forces.



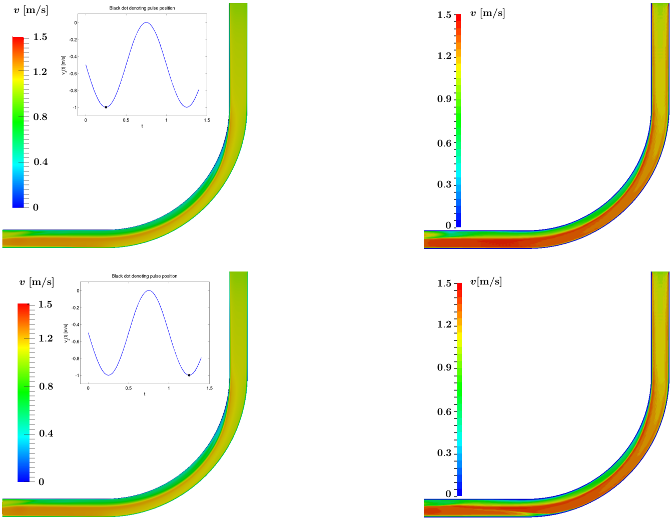
Simple momentum balance calculations show that

$F_{upstream} = F_{downstream} \approx 0.03841$ N

Results

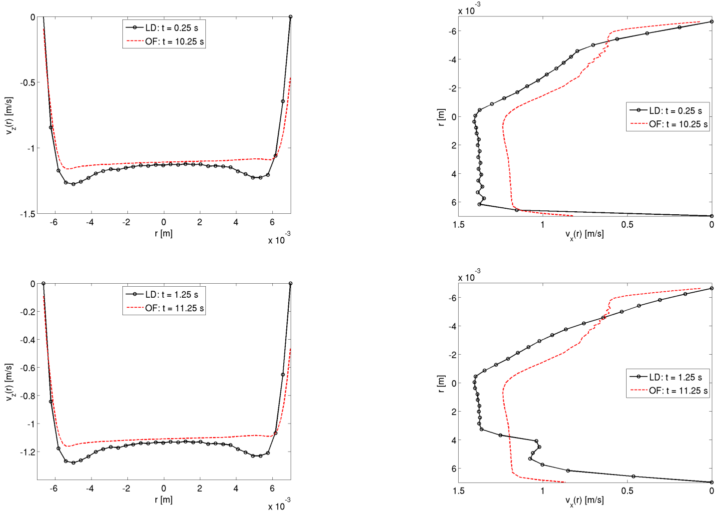
Pulsating FSI

Velocity fields at 0.25 s and 1.25 s, $|\mathbf{v}_{inlet}| = 1$ m/s



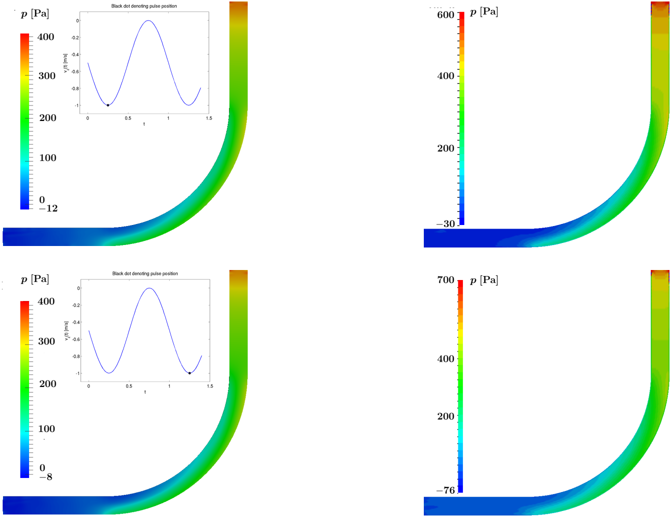
Results
Pulsating FSI

Velocity profiles at 0.25 s and 1.25 s, $|\mathbf{v}_{inlet}| = 1$ m/s



Results
Pulsating FSI

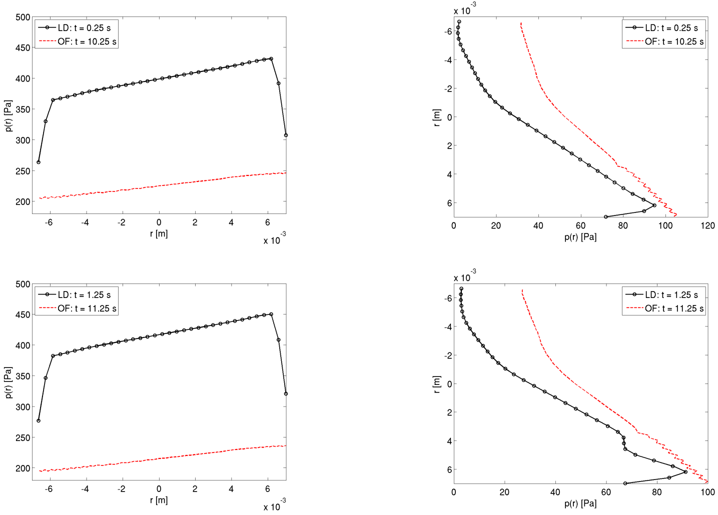
Pressure fields at 0.25 s and 1.25 s, $|\mathbf{v}_{inlet}| = 1$ m/s



Results

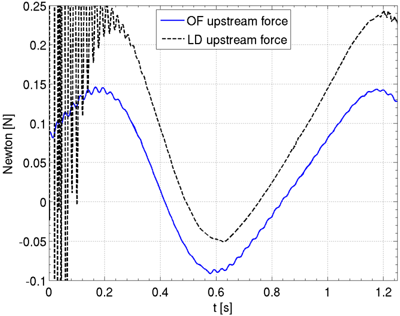
Pulsating FSI

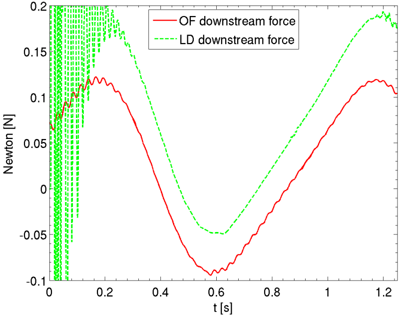
Pressure profiles at 0.25 s and 1.25 s, $|\mathbf{v}_{inlet}| = 1$ m/s



Results
Pulsating FSI

Normal forces in upstream and downstream attachments





Software	Upstream force [N]		Downstream force [N]	
	0.25 s	1.25 s	0.25 s	1.25 s
OF	0.1298	0.1279	0.1058	0.1039
LD	0.2286	0.2278	0.1756	0.1751

Conclusions

Judging by prior studies, FSI simulations are realizable and possible to perform in such manner that a fair evaluation of the forces causing stent graft migration is achieved.

However, this project

- only offers simplified simulations of the problem

and

- has been performed using different softwares than in prior studies.

Nonetheless,

- The results show good promise of utilizing both LD and OF for more complex studies.

Recommendations

There are still several aspects that need to be improved and further tested to reach a more satisfactory level of complexity and accuracy:

- More realistic inlet velocity pulse
- Gauge pressure corresponding to a represantative blood pressure
- Parallel experimental and numerical studies
- Parametric studies
- Non-newtonian blood flow
- Longer simulations
- Different setup in LD (e.g. parabolic inlet velocity, different coupling algorithm)

Thank you for listening!