Accuracy of carotid strain estimates from ultrasonic wall tracking: a multiphysics model study

Abigail Swillens¹, Gianluca De Santis¹, Joris Degroote², Lasse Lovstakken³, Jan Vierendeels², Patrick Segers¹

¹bioMMeda, Institute Biomedical Technology, Ghent University, De Pintelaan 185, 9000 Gent
²Department of Flow, Heat and Combustion Mechanics, Ghent University, Sint-Pietersnieuwstraat 41, 9000 Gent
³Norwegian University of Science and Technology, Department of Circulation and Medical Imaging, Olav Kyrres gt. 9, 7489 Trondheim, Norway

Correspondence:

Abigail Swillens
Ghent University
IBiTech-bioMMeda
De Pintelaan 185
9000-Gent, Belgium
abigail.swillens@ugent.be
Tel: +32 9 332 33 79
Fax: +32 9 332 41 59
Abstract
**Introduction**

Ultrasound imaging of the carotid artery is a common procedure when screening for cardiovascular disease, as the vessel is particularly prone to atherosclerosis and easily accessible with ultrasound probes. However, to reveal abnormalities in carotid flow and wall deformation with increased sensitivity and specificity, improved imaging modalities are desired. Since in-vitro and in-vivo testing of new imaging algorithms often lack ground truth information, ultrasonic image simulation based on fully known and realistic vascular behavior can be expected to support image development.

In this context, we developed a multiphysical simulation environment which integrates advanced numerical methods to calculate complex flow patterns and mechanical deformations on the one hand with an ultrasonic simulator on the other hand. For the ultrasonic image modeling, we relied on the Field II-software [ref], which can simulate with great scanning flexibility images of arbitrary tissue (i.e. both blood and arterial wall) by representing it as an ensemble of point scatterers on which the ultrasound waves reflect. In a first phase, we coupled computational fluid dynamics (CFD) with Field II, allowing simulation of radiofrequency (RF) data from realistic and complex flow fields, by moving the scatterers according to the spatially and temporally interpolated velocity fields.
obtained from CFD. We demonstrated the realism of the simulation environment with color flow imaging and pulsed wave Doppler examples in the carotid artery [ref] and by validating the simulation strategy on an in-vitro flow phantom of the carotid bifurcation [ref]. Fundamental limitations of this approach were the absence of the vessel wall signal and the rigid vessel walls. In a next phase, we integrated fluid-structure interaction (FSI) simulations with the ultrasound simulator [ref], which allowed to simultaneously assess the complex flow field and vessel wall deformation by coupling the numerical solution of a dedicated flow solver and structural solver. Hence, an FSI-Field II integration offers the possibility to simultaneously simulate the RF-signal of the blood pool and the moving vessel wall. This, however, requires a much more complex coupling methodology (derivation of the scatterer displacement) compared to the CFD-Field II coupling, due to the temporally varying fluid volume and the layered vessel wall, as explained in [ref]. We demonstrated the complex FSI-Field II coupling in a 3D straight tube, representative of the common carotid artery [ref].

In this paper, we further advance the realism of the synthetic vascular imaging set-up with the simulation of the 3D blood flow and arterial mechanics of a patient-specific carotid bifurcation model. The model includes the tissue surrounding the vascular wall, which is a stabilizing factor for the fluid-structure
interaction simulation but also results in a more realistic ultrasonic visualization of
the arterial territory. The extensive methodology behind the FSI-Field II coupling
will be briefly discussed, before demonstrating the realism of our multiphysics
modeling with simulated duplex images. Subsequently, we assess the accuracy of
ultrasonic estimation of radial and circumferential strain and related material
properties as derived from a wall tracking algorithm [ref], which is the main focus
of this paper. The performance of wall tracking algorithms was previously
analyzed in a simplified tube configuration (deforming in an axial symmetrical
way) in order to ease the interpretation of the measured distensions (ΔD) and the
corresponding circumferential strain (ΔD/D) estimates [ref]. We demonstrated
that distension and circumferential strain estimates as obtained from wall
tracking measurements are affected by the physics of the ultrasonic image
formation and should be interpreted with care when linking them to the
mechanical properties of the wall tissue [ref]. As wall tracking is applied to more
complexly deforming blood vessels like the carotid artery, we anticipate that its
intricate wall mechanics will even further complicate the analysis. Therefore,
simulated distension and strain estimates (circumferential and radial
components) derived from an ultrasonic wall tracking algorithm will be compared
with the true mechanical deformation and complex 3D strains as known from the FSI-simulation.
Methods

1. FSI-simulations

1.a Numerical approach

A partitioned FSI-approach was followed, i.e. the equations for the flow and structural domain were solved separately with a dedicated flow (*Fluent 12.0.16*, Ansys, Canonsburg, PA, USA) and structural solver (*Abaqus 6.7*, Simulia, Inc., Providence, RI, USA). More information on the applied numerical approaches in these dedicated solvers can be found in [ref]. The solutions for the fluid and structural domain were coupled using in-house code (*Tango*) with Dirichlet-Neumann partitioning (the flow problem is solved for a given displacement of the fluid-structure interface, and the structural problem is solved for a given stress distribution on the wet side of the structure). A converged solution for both the fluid and structural equations and the coupling conditions was found by performing coupling iterations between both solvers, until equilibrium between the fluid and structure was reached. To enhance convergence of the coupling iterations, an Interface Quasi-Newton (IQN) method was used [ref]. Note that the fluid and structural domain have inherently different grid formulations, which was solved by using an Arbitrary Lagrangian Eulerian (ALE) method for the
fluid domain. For more information on the IQN and ALE method, we refer to [ref].

1.b (Meshing the) carotid geometry

The 3D geometry was reconstructed from MRI-scans of a stenosed carotid bifurcation of an 83-year old volunteer. The MRI-scan sequence covered a 2 cm-region centered around the bifurcation. The geometry was artificially prolonged at the in- and outlets (total length of the model was 6.5 cm), in order to obtain fully developed flow in the region of interest (the bifurcation), but also to create a sufficiently long computational phantom for the simulated ultrasonic scanning sequence. An in-house open-source mesh generation code (Pyformex) was used to construct a computational grid for the vascular wall. This allowed to create a 4-layered mesh of the vascular wall (to some extent mimicking the intima-media-adventitia layers), consisting of 31680 first order hexahedral elements. Pyformex was subsequently used to mesh the fluid domain with hexahedrons, resulting in 87522 elements and a matching grid at the fluid-structure interface. The structural domain was further expanded to also include the tissue surrounding the vascular wall. Although the model becomes more computationally expensive, adding the external tissue had a stabilizing
effect for the FSI-simulations and allowed for more realistic ultrasound simulations. The carotid artery was embedded in a cylinder with a radius of 2 cm. Obviously, the surrounding tissue requires meshing as well, which is a challenging undertaking. Meshing this tissue with hexahedral elements is not an option since this element type causes intersections close to the bifurcation. As such, although hexahedrons are preferred for structural analysis (tetrahedrons behave stiffer), a first-order tetrahedral element type was chosen for the tissue volume. To ensure continuity of the hexahedral and tetrahedral meshes at the wall-tissue interface (to avoid interaction problems in the structural solver), the quadrilaterals at the outer surface of the vascular wall were split into triangles to match with the tetrahedrons of the tissue domain. A cross-section of the complete computational grid is shown in fig. 1.

1.c FSI-setup

Fluid domain: At the in- and outlets of the carotid geometry, we imposed physiologically realistic boundary conditions. We measured a velocity profile with pulsed wave Doppler (12L linear array vascular probe, GE Medical Systems, Milwaukee, WI, USA) in the common carotid of a healthy volunteer, which was further applied as a mass flow inlet condition.
Outflow percentages were imposed at the outlets (35% at the external and 65% at the internal carotid). Since the absolute pressure level is undetermined for such a setup, we added a non-invasively measured pressure (varying in time, with a pulse pressure of 40 mmHg) to the obtained fluid pressure distribution. As such, when transferring the interface stress to the structural solver, a realistic pressure value was imposed on the wet side of the structure. Blood was modeled as a Newtonian liquid with a viscosity of 3.5 mPas and a density of 1050 kg/m³.

**Solid domain:** Assuming that the mechanical properties of the vessel wall material can be linearized around the operating pressure, we modeled the vessel wall as a linear elastic material with Young’s modulus of 250 kPa, density of 1200 kg/m³ and Poisson modulus of 0.49 (nearly incompressible). The properties of the surrounding tissue were chosen to obtain a realistic distension degree for the chosen vessel elasticity: a Young’s modulus of 10 kPa and Poisson modulus of 0.3 [ref]. Longitudinal movement of the in- and outlets was prevented.

We refer to fig.1 for a complete overview of the applied fluid and solid boundary conditions. The cardiac cycle of 1s was divided into timesteps of 5 ms and 2 cycles were computed to obtain results independent of transient
effects. The coupling algorithm was executed on one core, the flow solver on eight cores and the structural solver on eight cores of a dedicated machine with two Intel Xeon 2x Quad-core Intel Xeon processors (2.66 GHz).

2. Ultrasound-simulations

2.a Field II

The Field II software [ref] was used to simulate the RF-signals from the fluid and structural domain. This modeling approach allows simulating arbitrary transducers and scansequencing with great flexibility and is based on the spatial impulse response estimation as described by Tupholme and Stepanishen [ref]. This simulation strategy is limited to linear wave propagation and determines the ultrasound field based on the ultrasonic excitation pulse, the temporal impulse responses of the transmitting and receiving transducers, and the spatial impulse response at a given point. For further details on the theoretical background, we refer to [ref]. The RF-signals can be simulated with a high degree of realism because Field II models tissue as a distribution of (random) point scatterers, whose position can be updated for each simulated ultrasound beam. By moving the
scatterers according to flow fields and wall deformations obtained from CFD/FSI, imaging algorithms can be studied in complex conditions [refs].

**Fluid phantom:** We refer to [] for details on how the scatterers can be propagated using FSI simulation results accounting for the temporally varying fluid volume and the applied ALE grid formulation. [ref].

**Wall phantom:** Deriving the scatterer displacements for the wall phantom was less complex due to the Langrangian grid formulation, i.e. the grid displacement corresponds to the material (scatterer) displacement. However, to account for local changes in material (acoustic) properties, random point scatterers were generated for each element of the wall mesh. Further, as can be seen in echo images, the vessel wall also causes specular reflections due to transitions between different tissue types. These cannot be simulated but only mimicked in Field II, by placing scatterers in a structured fashion at the borders of the vessel wall (i.e. tissue/vessel wall and vessel wall/blood). For more details on the coupling methodology for the wall, we again refer to [ref].

**Tissue phantom:** As explained above, the model also includes the surrounding tissue. To reduce computational times, scatterers were not generated for each mesh element of the tissue (as for the wall phantom),
but for the complete cylinder surrounding the carotid artery, comprising the fluid, wall and tissue domain. Afterwards, scatterers created inside the arterial wall and fluid volume were removed. Although the number of scatterers in Field II is determined by the resolution of the imaging system (10 scatterers per resolution cell assuring Gaussian distributed RF-signals), the total amount of tissue scatterers was reduced by a factor 10 for calculation purposes. This was justified by the fact that no visual differences were apparent between images with the full and reduced amount of scatterers. Further, no in-depth evaluation of tissue RF-signals was included in this study.

2.b Imaging setup

Both for the duplex scanning and the wall tracking, a linear array transducer was modeled, with a focal depth position at 2 cm. Each transducer element was divided into four smaller rectangular mathematical elements so that the backscattered signal from each point scatterer was simulated with sufficient accuracy. A dynamic focus and expanding aperture was used on receive to retain constant imaging properties throughout depth. To reduce beam sidelobes, apodization was applied.
**Duplex scanning:** A duplex scan is the superposition of a color flow image (CFI) on a B-mode image. However, the image acquisition requirements of CFI and B-mode are inherently different due to dissimilar spatial and temporal resolution requirements, and therefore compromises have to be made. A 5 MHz centre frequency was chosen for both image acquisitions, but the beam density was doubled for the B-mode imaging and the pulse length was increased for CFI from 1.5 to 4 pulse periods. To achieve these differing imaging properties, an interleaved scanning scheme was applied, switching between color flow and B-mode acquisitions. This resulted in a frame rate of 12 fps. Full details on the imaging setup can be found in table 1.

The color flow velocity estimates were estimated with the autocorrelation method for phase-shift estimation, as proposed for ultrasound applications by Kasai et al [ref]. The axial velocity $v_z$ (cfr. coordinate system on fig.1) was calculated according to: $v_z = \frac{c\text{PRF}}{4\pi f_0} \arctan \left( \frac{\hat{R}(1)}{\text{Re}(\hat{R}(1))} \right)$, with PRF (=pulse repetition frequency) the frequency of emitting ultrasound beams, $f_0$ the centre frequency of the ultrasound pulse and $\hat{R}(1)$ the estimated autocorrelation function at lag 1. The $\hat{R}(1)$ estimate was averaged over an
ensemble of 10 slow-time samples. The phantom was angled 70 degrees
towards the ultrasound scanline to reduce flow transversal to the beam. To
improve the CFI frame rate, we applied a beam interleaved acquisition
scheme as described in [ref], typically used when the Doppler PRF (as
determined by the imaged velocity range) is chosen lower than the
maximal possible PRF\textsubscript{max} (as determined by the imaging depth). We chose a
setup with a PRF of 4 kHz, a PRF\textsubscript{max} of 16 kHz, resulting in an interleave

groupsize of 4 beams.

**Wall tracking:** Vessel wall velocities were estimated with a modified
autocorrelation approach [ref], allowing to determine vessel wall motion
as: \( z[t+\Delta t] = z[t] + v[t] \Delta t \), with \( z[t] \) the position in the vessel wall, \( v[t] \) the
estimated velocity, and \( \Delta t \) the velocity resolution corresponding to the
packet size times the pulse repetition period (3 • 1/1000). A relatively low
PRF of 1 kHz was chosen, because tissue velocities in the carotid artery are
typically in a lower range than blood velocities. The phantom was not
angled since, for this application, ultrasound beams are ideally emitted
perpendicularly to the vessel wall. Compared to the duplex scanning, an
imaging setup with a much higher resolution was required and the 12L
linear array probe (GE Medical Systems, Milwaukee, WI, USA), as used in
the applied distension software [ref] was modeled with a 1.5 period sinusoidal pulse excitation of 8 MHz centre frequency. The RF-signal from the surrounding tissue and fluid domain was neglected for this application. Further details on the imaging setup can be found in Table 1. By tracking the wall motion on both the anterior and posterior side of the blood vessel, the diameter distension curve can be determined as well as its associated distension measure \( \Delta D_{\text{max}} = D_{\text{max}} - D_{\text{min}}, \) with \( D_{\text{max}} \) and \( D_{\text{min}} \) respectively the maximal and minimal diameter during the cardiac cycle (cfr. fig. 5). Assuming planar deformation, radial \( (\varepsilon_{rr}) \) and circumferential strain \( (\varepsilon_{\theta\theta}) \) can be derived from the ultrasonic distension estimation \( \Delta D \) as:

\[
\varepsilon_{rr} = \frac{\partial (\Delta D)}{\partial D}
\]

\text{Equation 1}

\[
\varepsilon_{\theta\theta} = \frac{\Delta D}{D_{\text{min}}}
\]

\text{Equation 2}

With \( \Delta D = D - D_{\text{min}} \) (with \( D \) the vessel diameter at the considered time point) and \( \partial D \) twice the distance between sample points in the vessel wall. Assuming linear elasticity theory and small deformations, also the Young’s modulus can be derived from distension measurements as:
\[ E = \frac{\Delta p \cdot D_{\text{min}}^2}{2h\Delta D_{\text{max}}} \]

with \( h \) the wall thickness and \( \Delta p \) the systolic-diastolic pressure difference.

The wall tracking algorithm is based on an existing application where ultrasound beams are emitted at 8 different scan positions [ref]. We performed distension analysis at 2 locations (in the common and internal carotid artery, indicated in fig.1 & 2), and at 5 depths in the vessel wall (5 nodes are present throughout the wall thickness in a 4-layered mesh model). For the ground truth, the reference FSI-strains were calculated by applying equations 1 and 2 to the Abaqus node displacements. We also studied the complex 3D deformation nature of the blood vessel by visualizing the strains which Abaqus computes from a second order strain tensor obtained from large deformation theory. The ground truth for the Young’s modulus is known from the input to the Abaqus-model (cfr. methods section 1c).


Results

1. Duplex scanning

Duplex images are shown during systolic deceleration and diastole in fig. 2. Complex flow patterns are present throughout the cardiac cycle, and specular reflections are apparent when perpendicularly insonifying the blood vessel. This is particularly visible for the internal and external carotid artery as seen in the B-mode image of fig.2, showing the carotid bifurcation as visualized with the distension imaging setup.

2. Wall tracking

2.a Common carotid artery

The circumferential strain $\varepsilon_{\theta\theta}$ displayed as a function of the depth in the vessel wall (cfr. fig. 3A), shows a decreasing trend in distension from inner to outer wall, both for the simulated ultrasound data (from 7.64 % to 4.41%) and the ground truth (from 7.49 % to 4.31%). While the reference curve shows a $1/D^2$-trend, the ultrasound data follow an S-shape throughout depth, with a slight overestimation of $\varepsilon_{\theta\theta}$ at the inner and outer wall (maximal absolute deviation throughout depth of 0.43%). Much stronger deviations from the ground truth are found for the radial strain at peak systole throughout the vessel wall (cfr. fig. 3C), with a maximal
absolute deviation of 5%. The Young modulus as estimated from the distension measurements on the inner wall was 215 kPa, implying an underestimation of 14 % (note that the Young modulus derived from the Abaqus node displacements on the inner wall entailed an underestimation of 11.6%)

2.b Internal carotid artery

The relation between circumferential strain and depth is no longer S-shaped, but still a decreasing trend from inner to outer wall is noticed, both for the simulated ultrasound data (from 8.83% to 4.47%) and the ground truth (from 7.78% to 4.5%) (Fig. 3B). One particularly notices a strong overestimation at the inner wall (deviation of 1.04%). This local measurement error is even clearer in the radial strain analysis, with a maximal absolute deviation of 6.68%. The estimated Young modulus was 173 kPa, or an underestimation of 30.6%.
**Discussion**

Ultrasonic vascular imaging is still mainly limited to morphological screening of vascular structures and visual interpretation of color flow images, rather than quantitative assessment of arterial (fluid-)mechanical properties. There is, however, a growing clinical interest in measuring arterial stiffness (believed to be an early phenotype of atherosclerosis with prognostic power) [ref], and the uncomplicated nature of ultrasonic wall tracking measurements has gained a lot of attention in this domain. Many parameters based on arterial distension assessment have been investigated including distensibility, compliance, circumferential strain and different definitions of elasticity moduli (e.g. Young’s modulus $E$, incremental elasticity modulus $E_{\text{incr}}$, stiffness index $\beta$).

Wada et al. non-invasively assessed the stiffness index of the common carotid artery in a heterogeneous population of 465 patients and derived reference stiffness values for non-invasive screening based on postmortem analysis on 60 carotid arteries [ref]. Blacher et al showed in a study of 79 patients with end-stage renal disease that the incremental elasticity modulus is a strong predictor of cardiovascular mortality [ref]. In a population study of Riley et al (3321 subjects), the Young’s modulus has been shown to provide more insight into the onset of cardiovascular disease [ref]. On the other hand, Jensen et al.
obtained contradictory results, with the stiffness index of the carotid artery showing no increase with age as generally expected. An overview on the use of non-invasive ultrasound in the assessment of arterial wall dynamics can be found in Reneman et al [ref].

While these mechanical parameters have been widely assessed in clinical trials, it should be noted that they are based on the assumption of planar deformation of the artery. This might be valid in rather healthy and straight arterial segments, but not for more complexly deforming blood vessels like the carotid bifurcation. Our data indicate that the E-modulus is underestimated using distension data, with an underestimation of 14% for the common carotid artery and even 30% in the more complexly deforming internal carotid artery. Hence, reported E-moduli based on wall track studies should be interpreted with caution.

Although the FSI-Field II integration allows comparing simulated ultrasound data with the ground truth behind the image, one should carefully consider which reference data to use. In case of strain estimations, the ground truth in fig.3 was the radial and circumferential strain as calculated from the Abaqus node displacements by equations 1 and 2. However, besides this self-
calculated strain measure, we also visualized (at peak systole) the radial (fig.4-A) and circumferential (fig.4-B) logarithmic strain as obtained from large deformation theory in eight cross-sections of the carotid model, using a local cylindrical coordinate system for the common, internal and external carotid artery. Fig.4-A and to a lesser extent fig.4-B demonstrate that strain distributions are relatively uniform proximal and distal to the actual bifurcation, but strong local strain variations appear in the vicinity of the bifurcation itself, complicating a correct measurement from the relatively simple ultrasound wall tracking data. Note that these logarithmic strain values cannot be directly quantitatively compared to ultrasound estimations as the latter assumes small deformation theory.

In the common carotid artery, the circumferential strain assessed from ultrasound decreased from inner to outer wall (as previously retrieved in vivo [ref]), following an S-shaped trend throughout the vessel wall depth (in contrast with the expected 1/D^2-trend), which disappeared for the internal carotid artery. As we previously demonstrated for a 3D tube [ref], the S-shape finds its origin in the presence of specular reflections which cloud the measurement for sample points close to these reflections. Since the internal carotid artery is slightly tilted towards the ultrasound beam, the specular
reflections disappear (cfr. B-mode in fig. 2) and hence also the S-shape in the strain curve. In our tube model study, the ultrasound measurement coincided with the ground truth at the position of the inner and outer wall. Here, however, we noticed a slight overestimation at these positions in the common carotid artery. We attribute this (small) discrepancy to the more complex, non-axial symmetric deformation of the simulated common carotid artery, as shown in fig.5. At peak systole, the deformation of the common carotid is almost ellipsoidal, with its main axis at about 45 degrees (cfr. case 2 in fig.5). This results in a more complex spatial distribution of scatterer displacements picked up by the ultrasound beam. This hypothesis is confirmed by the observation that, at the beginning of the cardiac cycle, with a more uniform deformation (non-ellipsoidal, cfr. case 1 in fig.5), the overestimation disappears. The situation is worse for the internal carotid artery, with an even stronger distension overestimation at the inner wall. In particular, when analyzing the tracking results for the anterior and posterior side separately (cfr. fig. 6), one finds that the posterior side has a higher overestimation than the anterior side (0.93 % versus 0.65 % at the inner wall). This can be attributed to a local increase in displacement values close to the scanposition, but still contained in the main lobe of the ultrasound beam and thus affecting
the measurement (cfr. case 2 in fig. 6). When more uniform displacement patterns are present (early systole and late diastole), the overestimation by ultrasound disappears. Since the posterior wall shows larger axial displacements than anterior, these are inherently measured with a higher variance. Also note that during diastole, the tracking of the posterior inner wall drifts away as the echo from this location then becomes blurred compared to the anterior inner wall (cfr. B-mode scan of fig.2).

As wall tracking can be done for several material points within the arterial wall, data can be processed to obtain radial strain (under the assumption of axial-symmetric planar deformation). Our data, however, indicate that results are not reliable. This estimate entails a spatial derivation of the distension data, and therefore amplifies inaccuracies in distension data. Inaccuracies in wall tracking will, most likely, also corrupt measurements of compressibility, which can be deduced from distension measurement at different depths in the arterial wall [ref].

Interpretation of distension measurements is even further hampered when considering that the distension and strain estimates are derived from a tracking algorithm applied both to the anterior and posterior vessel wall, while
obtaining just one strain estimate value for the complete cross-section at the end. For the (non-atherosclerotic) common carotid artery, this might be a valid approximation, given the rather homogeneous deformation pattern (cfr. fig. 4). However, further down the bifurcation (e.g. internal carotid), there might be a difference in kinematic behavior between the anterior and posterior side (due to the complex arterial geometry and the presence of the surrounding tissue), affecting the distension measurement (cfr. fig.6). For the simulated case, for instance, the anterior wall deformed less along the ultrasound beam, but more in the lateral and elevation direction. This might be misleading as both the anterior and posterior side might still have the same mechanical properties (the same E-modulus), as is the case in our FSI-simulation setup. Further, one should bear in mind that the wall tracking of the anterior and posterior wall might be affected by the ultrasonic beam formation, with the anterior wall of the internal carotid still quite close to the in-focus position (for the simulated set-up at least), but the posterior side imaged with a broader main lobe due to the diverging beam propagation after focusing (cfr. fig.1). Finally, also out-of-plane motion of the arterial wall might affect the ultrasonic assessment. However, in our simulated setup, no considerable out-of-plane
motion occurred (5.5e-5m at peak systole in the anterior internal carotid compared to an in-focus beamwidth of 4.8e-4m).

Note that the above considerations are maybe of a rather theoretical nature, as it is recommended in clinical practice to perform distension measurements on the common carotid artery, with the artery aligned along the probe in a longitudinal view. Nevertheless, due to anatomical reasons, it might not always be possible to obtain ideal views of the common carotid artery, and it is most unlikely that the assumption of planar deformation applies to the common carotid artery given the non-symmetric embedding in surrounding tissue. Atherosclerosis will even further complicate carotid wall mechanics.

Although our multiphysics simulations offer the opportunity to compare ultrasonic distension measurements with a ground truth, limitations exist both at the level of the FSI-simulations as well as the ultrasound model. While we applied an advanced FSI-code, resulting in a strongly coupled solution for the flow and structure, the imposed boundary conditions (both for the structure and fluid) and the assumed material model can be improved. Impedance boundary conditions for the fluid domain (i.e. fully taking into account the hemodynamic properties of the upstream/downstream vascular bed) and
anisotropic, non-linear material models for the different arterial layers would result in more physiologically/biomechanically correct simulations. Also the ultrasonic simulator shows limitations, as it is restricted to linear wave propagation and does not allow modeling complex wave phenomena typically encountered in medical scanning (e.g. reverberations and aberrations). Further, it does not take into account the true scatterer nature of the tissue (e.g. red blood cells are represented by points instead of discs, showing no inertia, no interaction and no frequency dependent scattering).

To conclude, we successfully expanded our multiphysics simulation tool to the 3D deforming carotid artery, accounting for (the RF-signal from) the surrounding tissue and hence increasing the realism of the simulations. We further demonstrated that carotid strains (and in particular the radial component) estimated from wall tracking algorithms should be interpreted with caution. Both the complex 3D arterial deformation and the spatially varying ultrasonic beamprofile hamper the measurement, resulting in strain estimates and Young moduli deviating from the ground truth, with the error depending on the phase of the cardiac cycle and scanning location.
### Tables

#### Ultrasound set-up parameters for duplex scanning and wall tracking

<table>
<thead>
<tr>
<th>Parameter</th>
<th>CFI/B-mode</th>
<th>Wall tracking</th>
</tr>
</thead>
<tbody>
<tr>
<td>$f_0$=centre frequency</td>
<td>5 MHz</td>
<td>8 MHz</td>
</tr>
<tr>
<td># elements</td>
<td>192</td>
<td>192</td>
</tr>
<tr>
<td>Pitch</td>
<td>246μm</td>
<td>203 μm</td>
</tr>
<tr>
<td>Height</td>
<td>6 mm</td>
<td>3.25 mm</td>
</tr>
<tr>
<td>Focus</td>
<td>2 cm</td>
<td>2 cm</td>
</tr>
<tr>
<td>Dynamic receive focusing</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Expanding aperture</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Excitation</td>
<td>Sinusoidal</td>
<td>Sinusoidal</td>
</tr>
<tr>
<td>Pulseperiods</td>
<td>4 / 1.5</td>
<td>1.5</td>
</tr>
<tr>
<td>PRF$_{max}$</td>
<td>16 kHz</td>
<td>8 kHz</td>
</tr>
<tr>
<td>PRF</td>
<td>4 kHz / 16 kHz</td>
<td>1 kHz</td>
</tr>
<tr>
<td>Packetsize</td>
<td>10 / 1</td>
<td>3</td>
</tr>
</tbody>
</table>
Figure 1: Complete setup of the virtual echographic phantom. A cross-section of the wall and surrounding tissue mesh is shown together with the boundary conditions applied in Fluent and Abaqus. The measurement positions of the distension application are indicated (yellow), i.e. the common and internal carotid artery.
**Figure 2:** Left panel shows a B-mode scan of the setup for the arterial distension imaging. Middle and right panel are the duplex scans resulting from the interleaved B-mode/CFI scanning acquisition, shown during systolic deceleration and diastole, both cardiac phases indicated on the imposed flow curve.
Figure 3: Circumferential and radial strain curves in the common and internal carotid artery. Black refers to strain derived from the Abaqus node displacements and gray is the simulated ultrasound measurement.
Figure 4: A. Radial strain as obtained from large deformation theory in Abaqus, shown in eight cross-sections of the carotid bifurcation, at peak systole. B. Circumferential strain as obtained from large deformation theory in Abaqus, shown in eight cross-sections of the carotid bifurcation, at peak systole.
Figure 5: The mechanical deformation (displacement magnitude) in a cross-section corresponding to the scanning location in the common carotid artery. Phase 1: relatively uniform displacement pattern at the beginning of the cardiac cycle, resulting in a good match of the distension measurement with the ground truth. Phase 2: ellipsoidal displacement pattern at peak systole, resulting in an overestimation of the arterial distension.
Figure 6: Analysis of anterior and posterior tracking in the internal carotid artery (note different color scales for axial displacement). Both for the anterior and posterior side, quite uniform axial displacement patterns are present at the beginning of the cardiac cycle (phase 1). Higher blood and wall velocities cause more strongly varying axial displacements (phase 2), resulting in overestimated distension measures.