Chapter 5

Conclusion

5.1 Summary

For the diagnosis, treatment planning and post-surgical monitoring of cardiovascular disease (CVD), hemodynamic markers have proven to be of great utility. However, non-invasive assessment of the hemodynamics of a patient is still a challenge. Phase-contrast magnetic resonance imaging (PC-MRI) can measure the distribution of blood velocity along two-dimensional planes or in three-dimensional volumes and is limited in accuracy mainly by the image resolution and noise. The local variation in the blood pressure cannot be measured non-invasively, but is required in the clinical practice to evaluate CVD. Other hemodynamic quantities, such as the arterial wall stiffness or wall shear stress can also be relevant as diagnostic quantities and for understanding the onset of CVD, but are not observable with imaging techniques.

This thesis approaches the topic of patient-specific hemodynamics on three different paths.

In Chapter 2 of this thesis a method was presented to improve the accuracy of hemodynamic data recovery from partial 2D PC-MRI measurements by means of solving an inverse problem of the Navier–Stokes equations of fluid flow. Vessel geometries extracted from MRI or CT images are affected by errors due to noise, artifacts and limited image resolution. Small errors in the geometry propagate into the recovered data and lead to large errors in the solution when standard no-slip boundary conditions are used on inaccurately positioned walls. The core idea of this work was replacing no-slip boundary conditions at the arterial walls by slip/transpiration conditions with parameters which were estimated from velocity measurements. Numerical results of synthetic test cases showed an important improvement in accuracy of the estimated pressure differences and the reconstructed velocity fields.

In Chapter 3 a comparison study of different direct pressure gradient estimation techniques was presented. These methods compute relative pressure fields directly from 3D PC-MRI data. The new Stokes estimation method (STE) by Švihlová et al. [Švi+16] was applied for the first time to real phantom and patient data. In com-
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In comparison to the classical Poisson pressure estimation method (PPE), the STE method proved more accurate and more robust to noise and the image segmentation in most cases.

Chapter 4 was dedicated to a numerical validation of the new MAPDD model [Ber+19] for a domain decomposition reduction of vascular networks. This approach considers the vessels as a network of thin pipes in which the flow has the shape of a Womersley flow, connected by arbitrary 3D junction domains where the flow is governed by the Navier–Stokes equations. In the MAPDD model, the thin pipes are replaced by coupling conditions on the junction domains. A strategy to easily implement the MAPDD model with the finite element method was presented and the theoretical results of Bertoglio et al. [Ber+19] were reproduced with numerical simulations in a simple test case. The method was shown to deliver accurate results even for moderately large Reynolds numbers, far from the regime where the theory is valid.

5.2 Perspectives

As was shown in Chapter 3, the investigated direct pressure estimation methods are sensitive to the image segmentation and to the image resolution, especially for high Reynolds numbers. The methodology presented in Chapter 2 using slip/transpiration boundary conditions aims specifically at an improved robustness with respect to errors in the computational domain. Future work should therefore apply the discussed slip/transpiration data assimilation framework to the phantom data of Chapter 3. It is hypothesized that pressure estimates using the data assimilation method are more accurate and less sensitive with respect to the image segmentation and resolution than the PPE and the STE methods. The aortic phantom and realistic aorta geometries in general require handling multiple outlets. Outflow boundary conditions based on the MAPDD method discussed in Chapter 4 can be used to model the effect of the truncated parts of the vasculature. To that end, the MAPDD junction conditions have to be applied to the corresponding boundaries and the parameter describing the length of the truncated pipe, $d_l$ in Eq. (4.3), has to be estimated with the parameter estimation formalism. The small bifurcating vessels in the aortic arch justify the assumption of a Womersley flow made by the MAPDD model. Such a boundary condition is a simpler alternative to Windkessel boundary conditions in terms of modelling complexity, implementation and computational effort when solving the system. Only one parameter per boundary is required and the model is entirely local. The MAPDD junction conditions are guaranteed to be stable with respect to backflow. The accuracy of the model needs to be studied carefully for the proposed application. In addition to 3D PC-MRI data, the data assimilation method can also be applied to 2D measurements. Minimum data requirements should be identified for the data assimilation method.

Turbulence was seen to deteriorate the accuracy of direct pressure estimators in Chapter 3. The Navier–Stokes solver used in the data assimilation framework can
be extended in straightforward manner to include turbulence models, such as large eddy simulation [Pop00]. It is expected that the precision of the data assimilation framework applied to real data of cases of CoA, in particular to the 9 mm and 11 mm CoA phantoms under stress conditions, can be improved by considering turbulence. Future work should examine this hypothesis for clinically relevant cases and determine if the increased demand in computational resources is justified.

One of the limitations of the study presented in Chapter 2 was that the coefficients of the slip/transpiration boundary conditions were assumed constant in space and time. It is expected that the accuracy of the method can be improved by considering a spatial and temporal distribution of the parameters. Future studies should analyze different approaches, such as spatially parameterizing the coefficients, e.g., with respect to the center line. Time-dependence could be implemented, for instance, by considering a sum of shape functions (e.g., sines) with coefficients to be estimated in the parameter estimation framework.

The elasticity of the arterial walls plays an important role in the arterial hemodynamics and should be taken into account in many cases for reliable hemodynamic simulations [NOV11]. A worthwhile path for future work could be exploring the estimation of mechanical wall properties from velocity measurements (instead of displacements [BMG12; Ber+14]) using data assimilation and a fluid–structure interaction (FSI) model [FQV09a; SY16]. If the slip/transpiration model can be reasonably applied to FSI or should be viewed as an alternative approach is not clear at this stage. This question should be discussed in the future.