

Newtonian vs. Non-Newtonian Blood Mimicking Fluids: A Comparative Study in Context of Mitral Regurgitation Jets

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ABSTRACT

Mitral regurgitation is a prevalent heart valve disease with high incidence rates at higher ages, which, in an aging society, has a significant impact on the health care system. To improve medical diagnostics of mitral regurgitation it is crucial to understand the fluid mechanics of the occurring regurgitation jet, whereas accurate physiological flow modeling in a laboratory in-vitro environment is essential to translate findings to the human cardiac blood flow. Blood exhibits non-Newtonian (NN) behavior, and the importance of the blood-mimicking fluid's (BMF) ability to replicate this characteristic depends on the flow conditions under investigation. Mitral regurgitation jets are pulsatile, and the flow velocities range from 0 up to 5 m/s, resulting in a broad range of shear rates that challenge Newtonian assumptions. In this work, fluids of various viscosities are compared in a newly-designed experimental setup mimicking the left heart in context of mitral regurgitation. To clarify the influence of viscosity changes on the jet characteristics three fluids – i.e. distilled water, viscosity adapted water-glycerol, and water-glycerol-xanthan-starch as a NN fluid – are used as working fluids. The dynamics of the jet are split into two phases, the starting phase and an ejecting phase. After consideration of mean vorticity, bulk velocity, circulation and shear profiles, it is recommended to consider the NN-behavior of blood in in-vitro mitral regurgitation set ups.

1. Introduction

Mitral regurgitation is the second most common heart valve disease in Europe (Marx et al., 2020). In physiological flow, the mitral valve opens during the diastole, permitting blood flow from the atrium to the ventricle. During the systole the mitral valve closes and seals the atrium from the ventricle, which allows the ejection of blood towards the aorta. Mitral regurgitation means that the mitral valve is insufficient, which causes a regurgitant flow from the left ventricle to the left atrium during the systole. In clinical routines, reliable mitral regurgitation diagnostics are essential for selecting an appropriate therapeutic approach. However, Leister, Karl, Stroh, et al. (2025) amongst

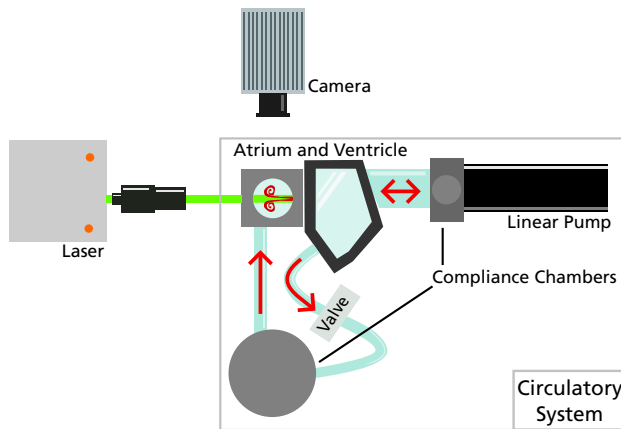
others demonstrated that the standard procedure to evaluate mitral regurgitation severity, which relies on simplified fluid mechanic assumptions, might lead to inaccuracies. Understanding the fluid dynamics of the regurgitant jet is, therefore, crucial for improving the interpretation of diagnostic images. Yet, direct in-vivo access to mitral regurgitation flow is not feasible, making experimental in-vitro modeling a powerful tool to gain valuable insights and hidden flow relations. Consequently, in-vitro test facilities – often called hemodynamic simulators – have been developed in the recent years, which focus on carefully selected phasings within the cardiac cycle such as Karl et al. (2024) or Ginty et al. (2018).

However, to transfer the insights gained in such in-vitro experimental setups, the findings need to fulfill various unalterable boundary conditions. In the context of mitral regurgitation flows one of these side conditions is the choice of a blood mimicking fluid (BMF), which depicts all relevant flow properties of real human blood. It is well known that blood acts as a non-Newtonian (NN) fluid and is of higher viscosity than water. While it is widely assumed that modeling blood as Newtonian is acceptable for macroscopic flows in large vessels (e.g. Yousif et al., 2011), recent CFD studies indicate that in pulsatile conditions the use of NN models has a significant impact on a number of flow parameters; ranging from problem specific parameters like wall shear stress, its gradient along the wall in aortic models to more cross-cutting parameters like the flow distribution (Sarkar et al., 2024; Tyfa et al., 2024). Similarly, NN BMFs have shown a profound influence on flow properties in aortic phantom experiments (Moravia et al., 2022). A reliable conclusion on the validity range – i.e. applicability and limits – of Newtonian simplification has not yet been drawn. In mitral regurgitation, pulsatile jets produce flow fields with a broad range of velocities (Leister, Karl, Stroh, et al., 2025) and consequently shear rates, which in turn challenge Newtonian assumptions.

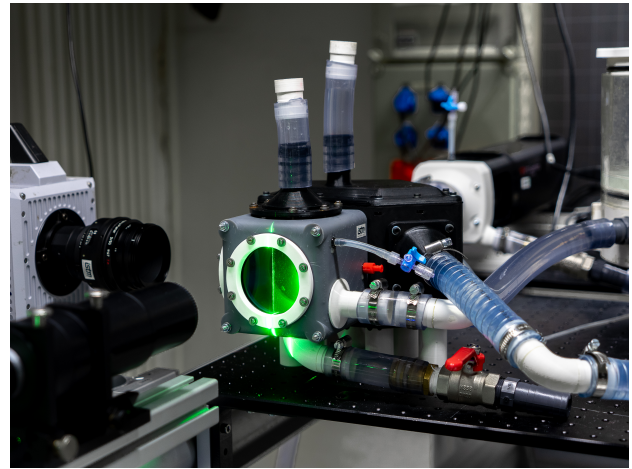
To address this open question in the context of mitral regurgitation, a hemodynamic setup was developed to replicate the physiological flow and pressure in the left heart. This allows optical access to the regurgitation jets. In this work regurgitant jets of BMFs with different rheological properties are compared to assess the impact of viscosity and NN behavior on flow characteristics of mitral regurgitation.

2. Setup

The experimental setup mimics the cardiac flow, described in the introduction; a sketch of the setup can be found in Figure 1a. The heart contraction and thus driver of the flow is realized by a pulsatile linear pump (*Vivitro SuperPump*), attached to the ventricle. The atrium and ventricle are 3D-printed using stereolithography (*Formlabs 3L*) with grey v4 and black v4.1 resin (*Formlabs*), respectively. Two medical mechanical valves (27 mm *SJM Master HP Series Valved Graft*) act as aortic and mitral valve to facilitate a physiological circulation with accurate pressure values. Mitral regurgitation is simplified and modeled as rigid orifices between atrium and ventricle in a circular shape and a diameter of 12.2 mm. Note that two smaller orifices with diameters 4.7 mm and



(a) Overview of the circulatory system and 2D2C high-speed PIV setup.



(b) Setup in the laboratory with Laser light sheet in the atrium.

Figure 1. Systematic sketch and current laboratory image of the hemodynamic setup and PIV setup.

8.7 mm were considered, which showed similar results and are not shown here for a clearer narrative. To regulate the pressure curves, compliance chambers are added to the setup, which were manufactured using a Prusa XL (*Prusa Research a.s.*) with PLA and sealed water-tight with Dichtol (*DIAMANT Polymer GmbH*). They imitate the elasticity of vessels and tissue of the cardiovascular system. The vascular systemic resistance is implemented via an adjustable needle valve. Three pressure probes (*ABP-Series, Honeywell*) and a thermometer (*Pt1000, THERMA GmbH*) monitor the system. The experiments are performed at a heart rate of 70 bpm and physiological pressures, controlled in the atrium (10 – 25 mmHg), ventricle (20 – 120 mmHg) and aorta (80 – 120 mmHg).

The PIV setup consists of a Photron SA4 Fastcam equipped with a 50mm lens (*Interlock Milvus 2/50M, Zeiss*) and an Nd:YLF laser ($\lambda = 527 \text{ nm}$, *Terra PIV, Continuum Electro-Optics Inc.*). Polyamide particles of 20 μm are submerged in the fluid as tracers, leading to a characteristic time response of $\tau_p = 25 \mu\text{s}$ and to a Stokes number of 0.014, with the given maximum velocity of 5 m/s and the outlet diameter of 8.7 mm.

The viscosity of blood is often reported between 3.5 and 5.5 mPas, however, this is only an approximate range due to the NN shear-thinning behavior (further details e.g. Nader et al., 2019). In this work three fluids are compared that are commonly used in hemodynamic setups. The first is distilled water (**water**). As second fluid a mixture of water-glycerol (60 % : 40 %) is used, this is indicated as **glycerol** in the following figures. The third fluid with NN-behavior is a mixture of water, glycerol, xanthan and starch with a mass ratio of 59.98 % : 40 % : 0.01 % : 0.01 %. This fluid is further on referenced as **xanthan**. Details on the use of the fluids and their respective mixture ratios can be found e.g. in Odemis et al. (2024); Leopaldi et al. (2012); Moravia et al. (2022).

Perrira et al. (2022) investigated multiple mixtures of water, glycerol, xanthan gum and starch to mimic blood's viscosity over shear rate. They recommend a mixture of 40 vol% glycerol, 59.98 vol% water, 0.01 wt% xanthan gum and 0.01 wt% starch. This formulation serves as a baseline

for preparing NN mixtures in this work. The viscosities of the glycerol and xanthan BMF were measured using a concentric cylinder rheometer (*ARES G2, UB TA Instruments, Waters GmbH*). Water was not included in the measurements due to the low and well documented viscosity. In Figure 2 the viscosity of water is added from literature (e.g. Korson et al., 1969). The tests were run at room temperature (25°C) and repeated to ensure significance of the results. The dynamic viscosity of the fluids μ over the shear rate $\dot{\gamma}$ is shown in Figure 2. Glycerol has a higher viscosity that matches the range of blood. In lower shear rates, xanthan shows a higher viscosity, which decreases to a plateau after a shear rate of $\dot{\gamma} = 200 \text{ s}^{-1}$. Thus, the xanthan mixture matches the desired NN behavior.

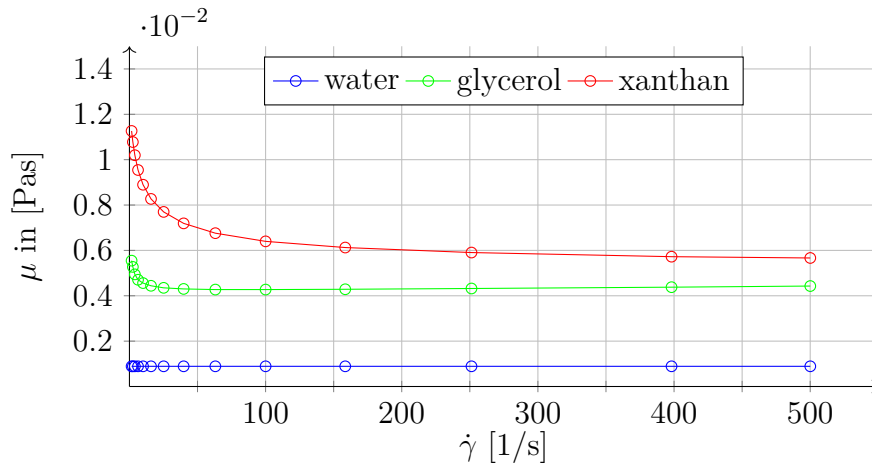


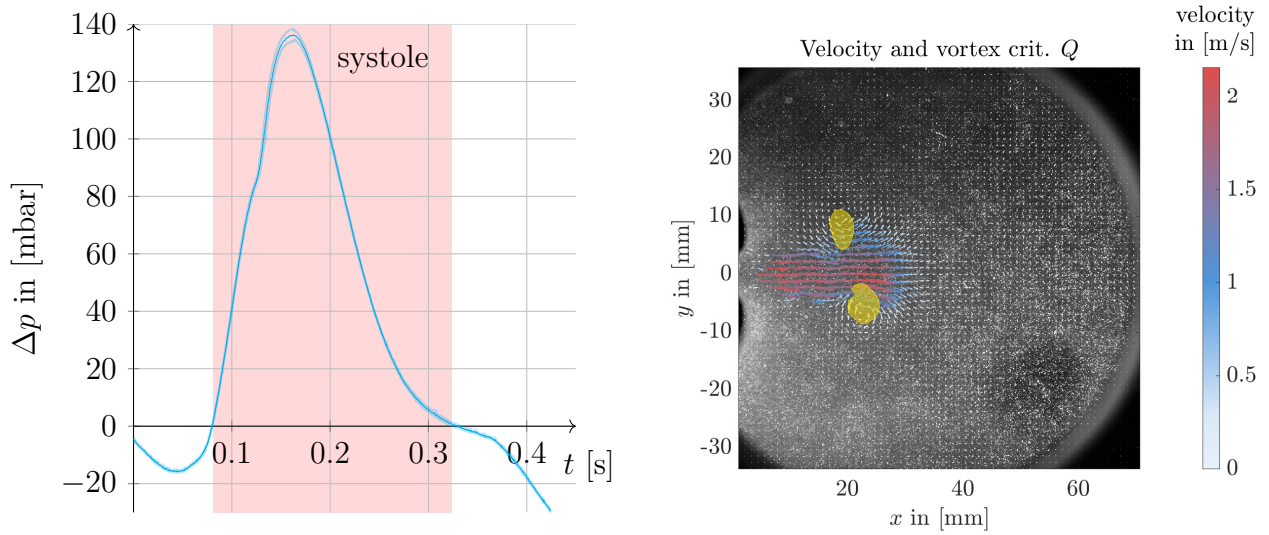
Figure 2. Dynamic viscosity of water, glycerol and xanthan.

3. Results

The setup produces a repeatable jet through the orifice. The pressure curve, in terms of the pressure difference between the ventricle and the atrium, shows a peak at $\Delta p \approx 130 \text{ mbar}$, which matches physiological conditions. Exemplarily, Figure 3a displays the pressure over the orifices during the systole. The standard deviation – indicated in light blue in the figure – is determined from 10 consecutive repetitions of a beat. The resulting flow is a regurgitation jet into the atrium shown in Figure 3b, where the starting vortex of the jet is marked using the Q-criterion.

3.1. The phases of the jet

Since the jet is generated and is collapsing for every heart beat, one can distinguish characteristic phases that happen during every cycle. We here define two main phases. In phase I the jet starts to form. A starting vortex rolls from the edge of the orifice hole, detaches and travels in positive x direction. Here, we define the end of phase I by the time the vortex leaves the field of view. After the pressure peak of the systole the pressure difference decreases, which is defined as phase II. During this phase, the jet is turbulent and semi-converged. The following collapse of the jet shows



(a) Pressure gradient over the orifice in terms of difference between the pressure in the atrium and the ventricle $p_a - p_v = \Delta p$. (b) Snapshot of the regurgitation jet with color-coded velocity and vortex Q-criterion (•), $t \approx 0.076$ s. The systole (■) is around $\Delta t_{\text{sys}} = 0.22$ s long.

Figure 3. Temporal and spatial evaluation of the generated vortex of mitral regurgitation jet through a circular orifice with distilled water as working fluid.

no significant features as the pressure gradient is not as steep compared to the starting phase. Phenomena like a stopping vortex indicated in similar flow problems (Zhu et al., 2024) cannot be observed.

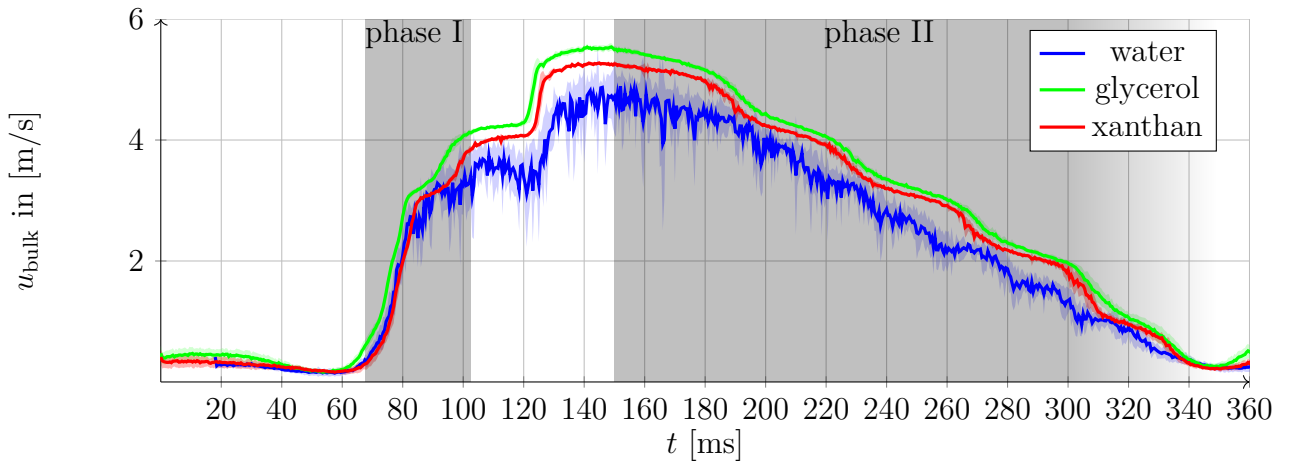


Figure 4. Bulk velocity of all three fluids, shaded standard deviation

A first insight and separation of the phases can be gained by observing the bulk velocity u_{bulk} in Figure 4. The velocity shows a step increase in phase I and during the formation of the jet. After a plateau between $t = 100$ ms to 120 ms, a second step increase follows due to the rise in pressure. The gradual descent then follows in phase II, which is not dominated by high fluctuations in bulk velocity. While the glycerol and xanthan solutions align well, with xanthan showing a slightly lower bulk velocity during the entire time, water is of higher fluctuations and shows a lower bulk

velocity.

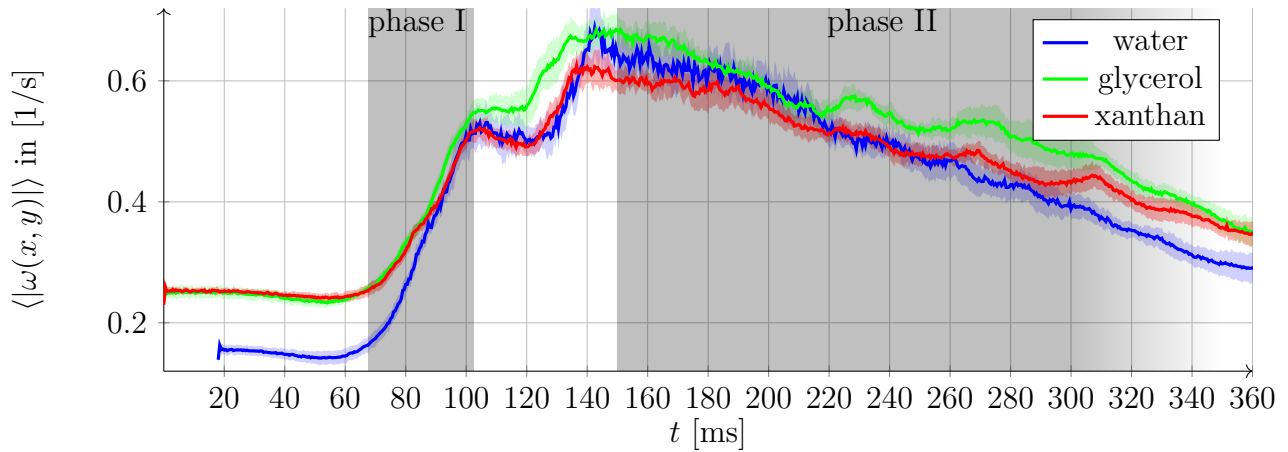


Figure 5. Mean absolute vorticity value of the FOV as a function of time.

The mean absolute vorticity of the field of view $|\omega|$ over the time t is an indicator for the definition of the end of phase I. In Figure 5 at $t = 103$ ms the vorticity has a clear edge in all jets. This correlates to the starting vortex exiting the field of view.

Moving to a comparison of the solutions, water has a lower vorticity at the start than the glycerol and xanthan solutions. During phase I the vorticity of the water jet rises steep to end up at the same level as the jets in glycerol and xanthan solutions. Comparing the xanthan to the glycerol jets, xanthan shows a decrease in the transitional phase between phase I and II, followed by a lower but parallel progress in phase II.

While water is of similar behavior as the viscosity-adapted BMF solutions with glycerol and xanthan, it can not be suggested to use water as the BMF. One obvious reason is the difference in Reynolds number, which is $Re_W \approx 60\,900$ for the water jet, while only $Re_G \approx 16\,900$ (i.e. a bare third) for the glycerol jet. In fluid mechanic laboratories adapting the Reynolds number might seem relatively easy, on the other hand MR is an interdisciplinary topic, which involves medical laboratories. The necessity motivated from the medical groups, to have a physiological pressure gradient, thus velocity and geometry, means that the Reynolds number can not be changed easily. Thus it is not recommended to use distilled water or fluids of similar viscosity like aqueous NaCl solutions as BMF if fluid dynamic aspects are relevant to the test case.

3.2. Phase I

In the initial formation of the jet – here named phase I – a starting vortex is a dominant feature. This vortex, if interacting with the walls of the atrium, is likely to be of great interest in characterizing MR.

The starting vortex was tracked using the Γ_1 -criterion introduced by Graftieaux et al. (2001). This is a criterion similar to the already introduced Q-criterion (see Figure 3b), which identifies vortex

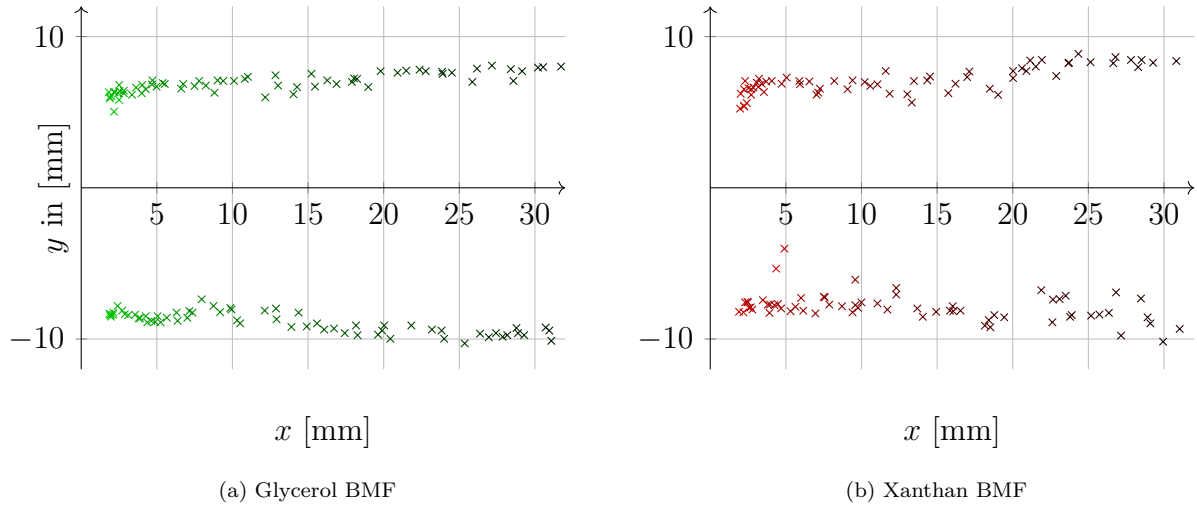


Figure 6. Starting vortex tracking based on Γ_1 , shading indicates time progression.

cores. In contrast to the Q-criterion, the Γ -criterion does not rely on velocity gradients but on the directional velocity information of the vector field. In Figure 6 the maximum Γ_1 of the field is used to track the starting vortex. The starting vortex forms near the orifice in the first time steps of both BMFs. The higher density of points suggests that the vortex gained circulation before starting to move in positive x direction. The diameter of the vortex, from one core to the other, widens in the progression through the atrium in both fluid cases.

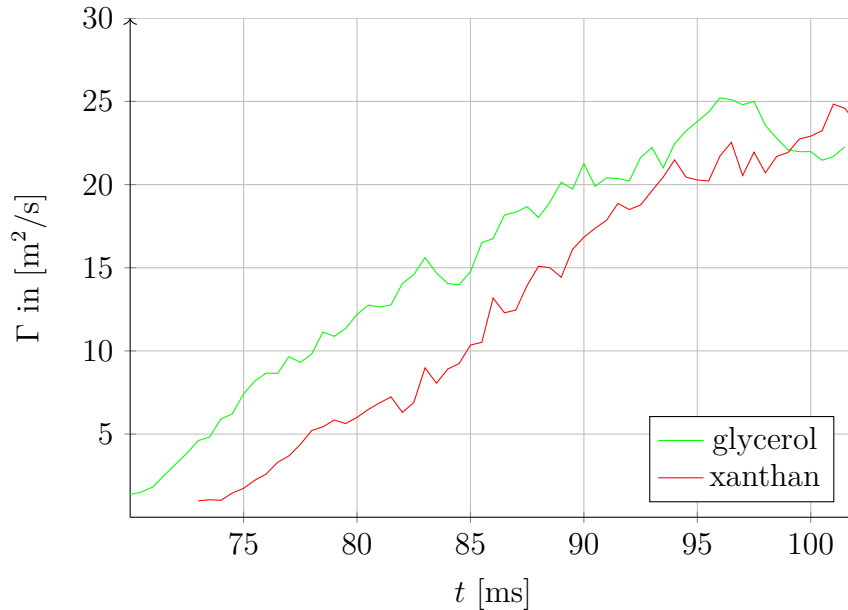


Figure 7. Circulation in Γ_2 marked regions.

The circulation of the vortex was calculated by summing the circulation of the the area marked as a vortex by the criterion of $|\Gamma_2| = 2/\pi$ (see Graftieaux et al., 2001). The starting vortex gains energy by the jet. In case of a "pinch-off"-event, the vortex would not gain more circulation and thus detach from the jet tracking the vortex (e.g. New & Yu (2015)). In Figure 7 the circulation

increases in both jets until they progress out of the field of view. A "pinch-off" process can thus not be observed, leading to the conclusion, that the vortex grows in strength over time, which might become relevant when the vortical structure interacts with the opposed wall.

On the other hand, Figure 7 is conclusive in terms of differences in the BMFs. In the glycerol solution, the vortex' circulation increases earlier than in NN-BMF xanthan solution. This aligns with the shear thinning behavior of the xanthan solution, which leads to a higher viscosity that needs to be overcome at the beginning of phase I.

3.3. Phase II

In phase II the pressure has reached the falling state of the systole. The jet is in a semi-converged condition.

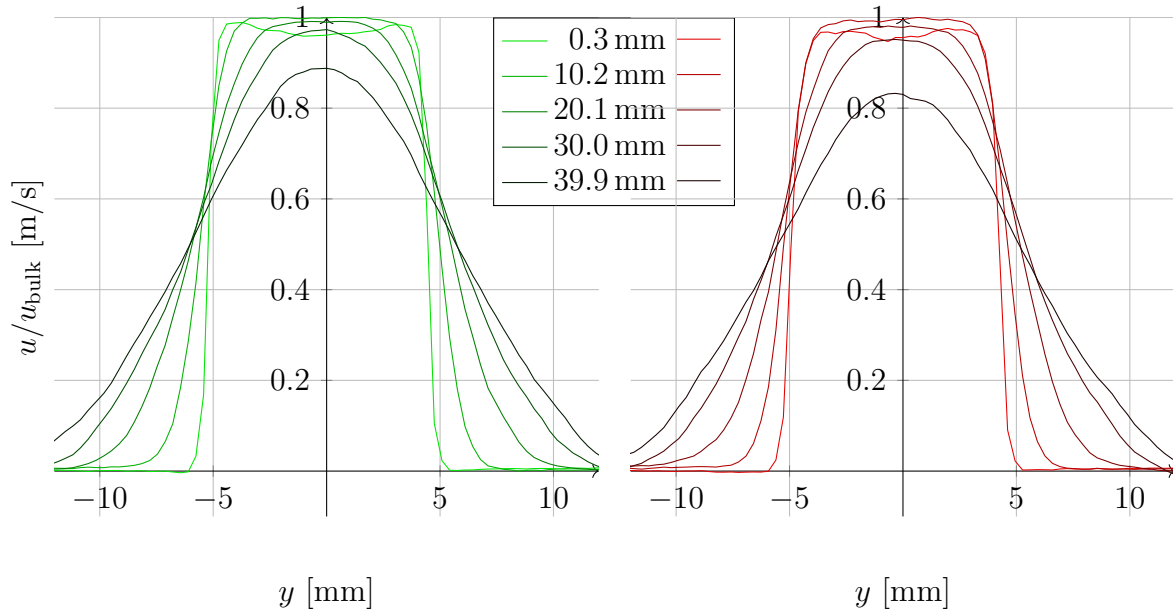


Figure 8. Velocity profiles at 5 different distances from the orifice (glycerol left, xanthan right).

The velocity profiles in Figure 8 at different x -locations show a striking consistency between the glycerol Newtonian and the xanthan NN BMF. Only close to the orifice, the block profile is mildly blunter in the xanthan jet.

An important value when observing NN fluids is the shear stress in the flow, as this dominates the local viscosity of the fluid. From the measurement-technical side, calculation of the shear stress has an upper limit, due to the restriction to resolve high gradients between the discrete interrogation windows. Nevertheless shear rates up to $\dot{\gamma} \approx 7000 \text{ s}^{-1}$ can be determined, which is more than sufficient for the fluids at hand, with plateau-like viscosities after 200 s^{-1} (see Figure 2).

While the similarities seen in Figure 9 are remarkable again, the shear rates of $|\dot{\gamma}| < 200 \text{ s}^{-1}$ play a role in both, the middle of the jet, as well as in the outer regions. Additionally, the formation

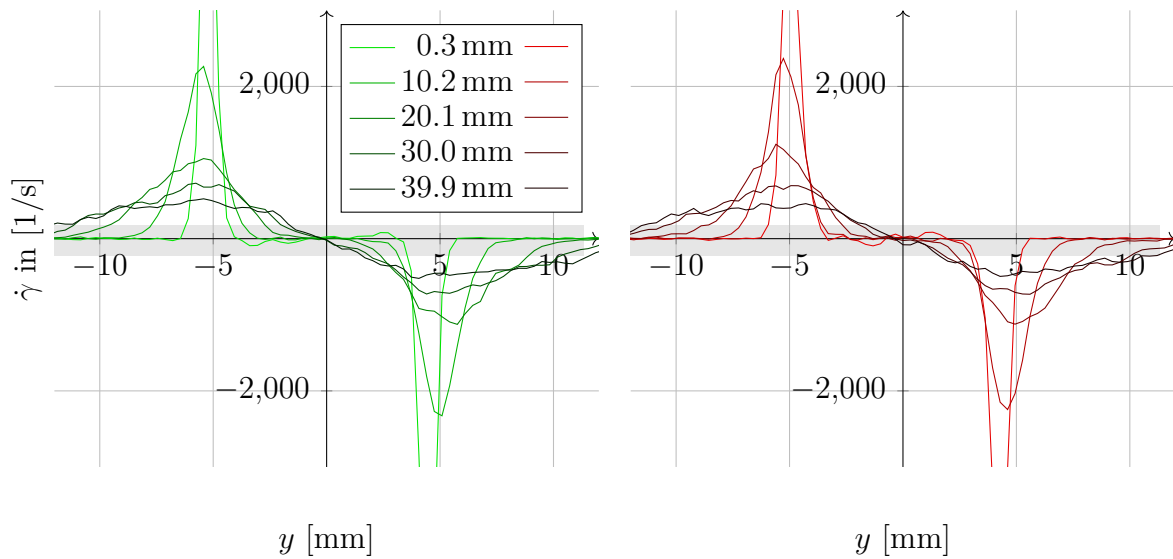


Figure 9. Shear rates at five different distances from the orifice. The gray box (■) indicates the shear stress region where the Non-Newtonian behavior becomes relevant (glycerol left, xanthan right).

and dissipation of instabilities – like the Kelvin-Helmholtz type at the orifice – observed by Leister, Karl, Neff, et al. (2025) might be affected by these factors as well.

4. Conclusions and Outlook

A new in-vitro hemodynamic setup focusing on mitral regurgitation was built and put into operation. The regurgitant jets of all three fluids, distilled water (**water**), water-glycerol (**glycerol**), water-glycerol-xanthan-starch (**xanthan**) through circular-shaped orifices were observed in the left heart setup. The MR jet was divided into two phases, a starting phase, where starting vortex and circulation dominate the flow field, and a second ejection phase, in which a semi-converged turbulent free jet is the dominating feature of the flow.

Water was discarded as a BMF, since the Reynolds number of the jets do not match. The xanthan-based NN-BMF showed a high similarity to the Newtonian BMF based on glycerol. Nevertheless, differences in mean vorticity, as well as starting-jet circulation were observed. This indicates that the NN-behavior of the fluid affects the flow even in a free jet, where shear rates of $\dot{\gamma} > 200 \text{ s}^{-1}$ occur. This contradicts the general assumption of irrelevance of NN-properties in macroscopic flow of high shear rates. It is thus recommended to utilize NN-fluids as BMF in the context of mitral regurgitation, since shear rates of $\dot{\gamma} > 200 \text{ s}^{-1}$ frequently occur. A relevant difference was shown with the circulation of the starting vortex.

For future work it is foreseen to implement more complex orifice geometries. Especially, in the light of the complex-shaped anatomy of the mitral valve, the observation of orifice shapes of much higher complexity is desirable, which calls for a 3D-3C resolving measurement setup.

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