

SUPPLEMENTAL MATERIAL

A. Pressure Driven Left Ventricular Flow

The basis for the flow rate within the ventricle was the ViVitro Systems FDA waveform as previously described in the Materials and Methods section. The pump waveforms prescribe the displacement of the pump piston over time. As the pump piston heads are of a fixed radius, the displacement waveform is interpreted as a time dependent volumetric change for any chamber connected to the pump (**Fig. S1(a)**). If we then take the derivative of the volume plot, we have a flow rate waveform that matches the volume/displacement waveform (**Fig. S1(b)**). We use the flow rate waveform to inform a pressure boundary condition at the inlet port using Poiseuille's Law to derive the following equation:

$$P_{i+1} = P_i + R_{\text{inlet}}(Q_{\text{pump}}(t) - Q_i)$$

where P_{i+1} is the updated pressure boundary condition, P_i is the current pressure measured at the inlet, R_{inlet} is the resistance of the inlet, $Q_{\text{pump}}(t)$ is the desired flow rate of the pump at time t , and Q_i is the current flow rate observed across the inlet. The resistance, R_{inlet} , describes a penalty term between the observed flow rate and the desired flow rate rather than a measured physical resistance, so we set its value to 666.61 (dyne/cm²)/(mL/s). In the pressure update equation, note that if the observed flow rate matches the prescribed flow rate the pressure will not change from the currently observed pressure to maintain the current flow rate. The base flow rate waveform describes a stroke volume of 1 mL, so we scale the flow rate waveform by the desired stroke volume to match any given cardiac output. The effectiveness of this pump boundary condition is demonstrated by the consistency of the stroke volumes measured in **Table 4**.

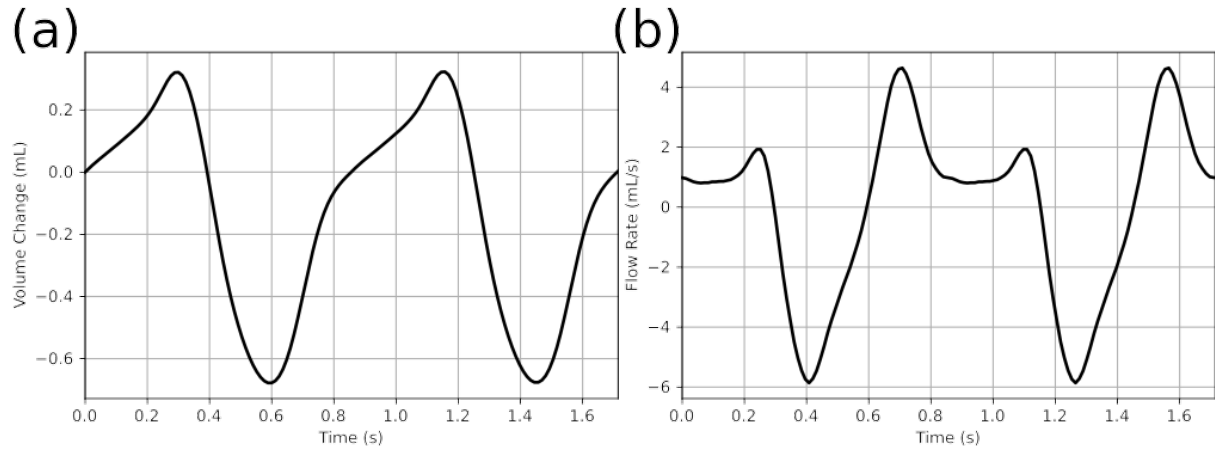


Figure S1: The pump displacement/volume change **(a)** with the corresponding flow rate **(b)**. The stroke volume of the volume change is 1 mL at 70 BPM. The flow rate and volume change waveforms are scaled by the appropriate stroke volume to produce the desired cardiac output at 70 BPM.

B. Bulk Flow Metric Cycle Convergence

As stated in the Results section, the first three of the eight cycles were discarded to allow for the time series data to approach periodic steady-state, and five cycles is the minimum required amount for the statistical power to exceed 0.8 for one-sided Mann-Whitney U Test, which it exceeds at approximately $p < 0.05$ assuming normal distributions [55]. The values we considered for cyclic convergence were the left ventricular pressure, the ventricular inflow flow rate, and ventricular outflow flow rate. Cross-cycle analysis was conducted by linear interpolation of 857 points for each cycle from the original data set to have matched relative time points for each of the eight cycles. This was done as the sampling rate for the pressure and flow rate data were 1000 Hz and the length of a cycle at 70 BPM is approximately 0.857 s. From the interpolated data, we compute the single standard deviation at each relative time point. With these values we compute the average standard deviation from the mean, and the average relative standard deviation, where the relative values were scaled by the mean values at each time point of the interpolated data.

The comparison was done for the normal healthy case (NHC) valve at 3.5 L/min for blood analogue and water with no cycles excluded and with the first three cycles excluded. The results are summarized in **Table S1**. For each variable, the mean standard deviation is reduced by at least 75%. We decided to compute the deviation metrics for the energy data as they are complex metrics reliant on the entire velocity field. The energy data is sampled at 100 Hz, so the interpolation was done for 86 time points for each cycle. We see approximately the same reduction in the mean standard deviation across the entire cycle when excluding the first three cycles, and most of the deviations are ~10% when only the last five cycles are considered.

In order to query time step convergence, we ran an additional simulation at 3.5 L/min with the NHC with a time step size of $0.5e-5$ s, which was half the original time step size. We report no statistically significant differences in the reported values for this case. Previous fluid grid resolution studies conducted by Lee *et al.* [38] suggest that there might be a $< 2\%$ difference in the bulk flow results with a refinement from the current Cartesian width of 0.55 mm to 0.44 mm, or a 20% refinement, and a ~10% difference within the energetic metrics under the same refinement.

Table S1: Mean difference reduction with cycle exclusion across the bulk flow metrics in the left ventricle (pressure, inflow and outflow flow rate) along with the energy metrics for the 3.5 L/min simulations with the NHC using blood analogue and water. **KE = kinetic energy, VED = viscous energy dissipation rate.**

Fluid	Blood Analogue		Water	
Cycles Excluded	0	3	0	3
Mean Pressure Deviation (mmHg)	0.2283	0.05464	0.3267	0.05707
Mean Relative Pressure Deviation (%)	4.352	0.8843	4.572	0.8281
Mean Inflow Deviation (mL/s)	5.723	0.8434	6.448	0.8680
Mean Relative Inflow Deviation (%)	11.15	1.560	13.19	1.634
Mean Outflow Deviation (mL/s)	5.711	0.8434	6.438	0.8684
Mean Relative Outflow Deviation (%)	8.847	1.246	9.631	1.314
Mean Planar KE Deviation (erg/mL)	3.868	1.073	4.410	1.455
Mean Relative Planar KE Deviation (%)	35.58	7.924	43.65	11.03
Mean Planar VED Deviation (erg/s/mL)	2.218	0.5446	4.421	0.9648
Mean Relative Planar VED Deviation (%)	28.42	5.630	41.76	7.169
Mean 3D KE Deviation (erg/mL)	1.088	0.2920	1.217	0.4851
Mean Relative 3D KE Deviation (%)	36.21	7.836	44.68	13.65
Mean 3D VED Deviation (erg/s/mL)	1.229	0.3177	2.058	0.5029
Mean Relative 3D VED Deviation (%)	32.59	6.685	39.96	7.797

C. Planar Vorticity and Streamlines

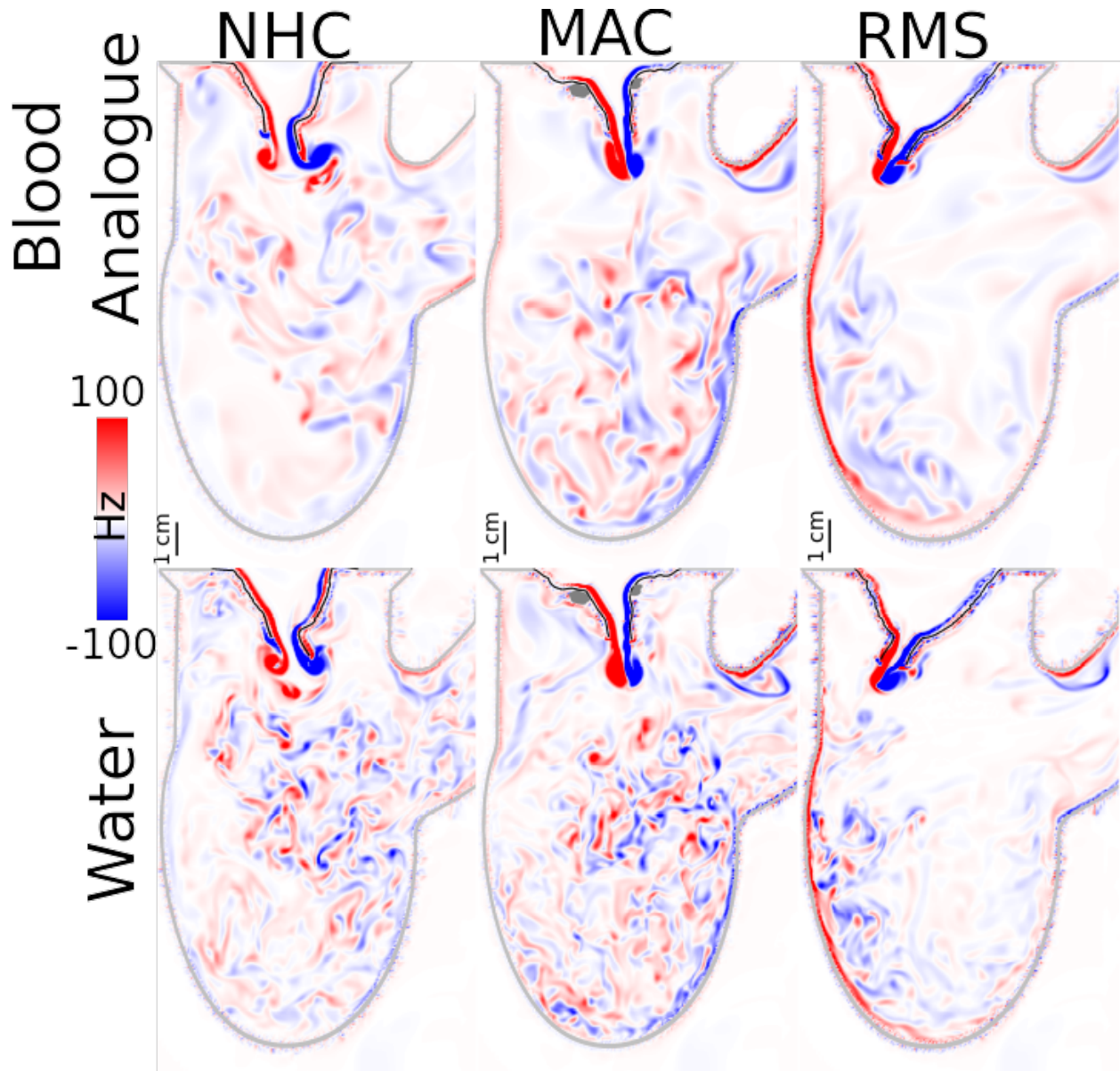


Figure S2: Snapshots of the planar vorticity during early diastole for the blood analogue and water simulations at 3.5 L/min.

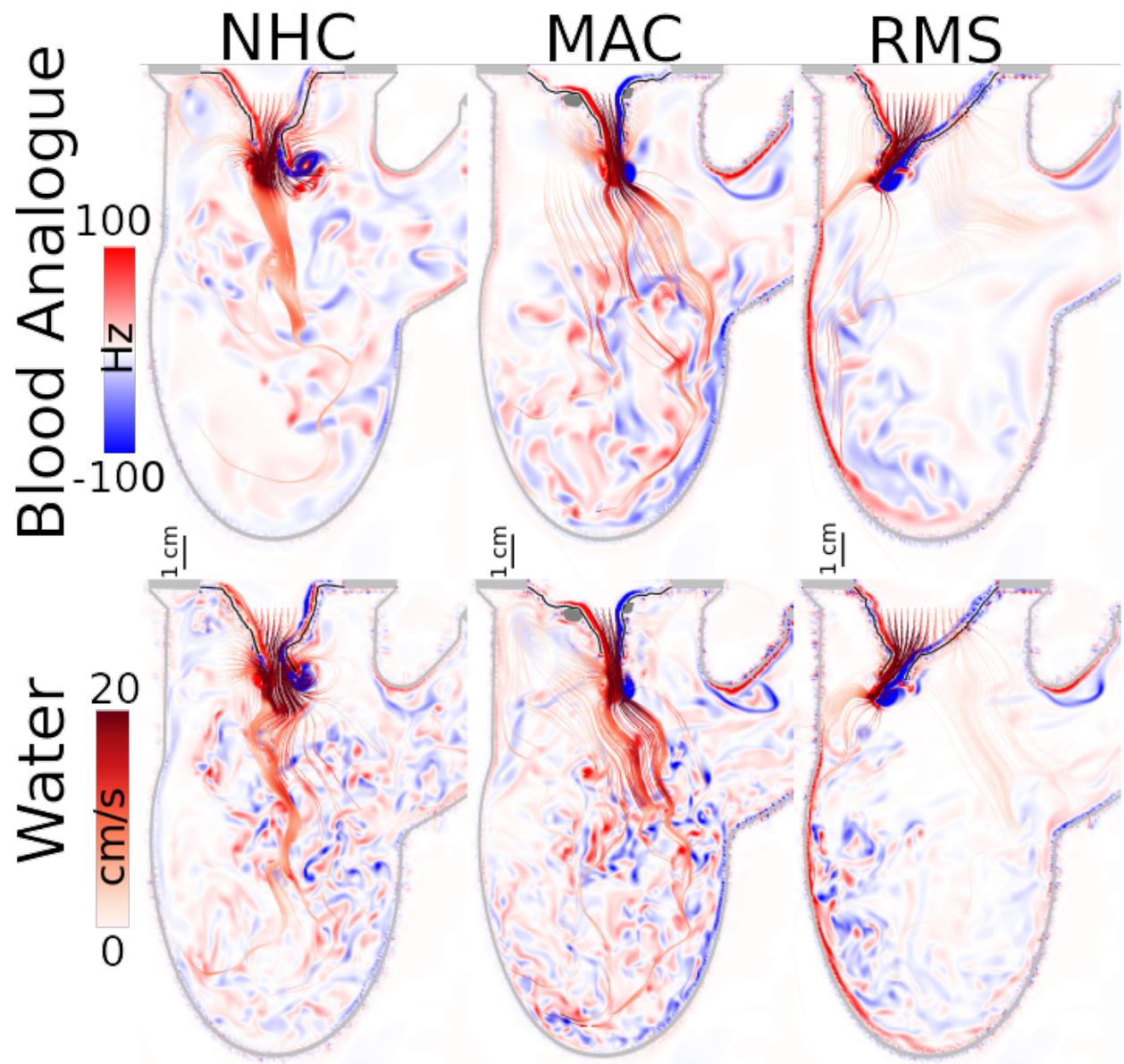


Figure S3: Snapshots of the planar vorticity with overlaid streamlines during early diastole for the blood analogue and water simulations at 3.5 L/min.

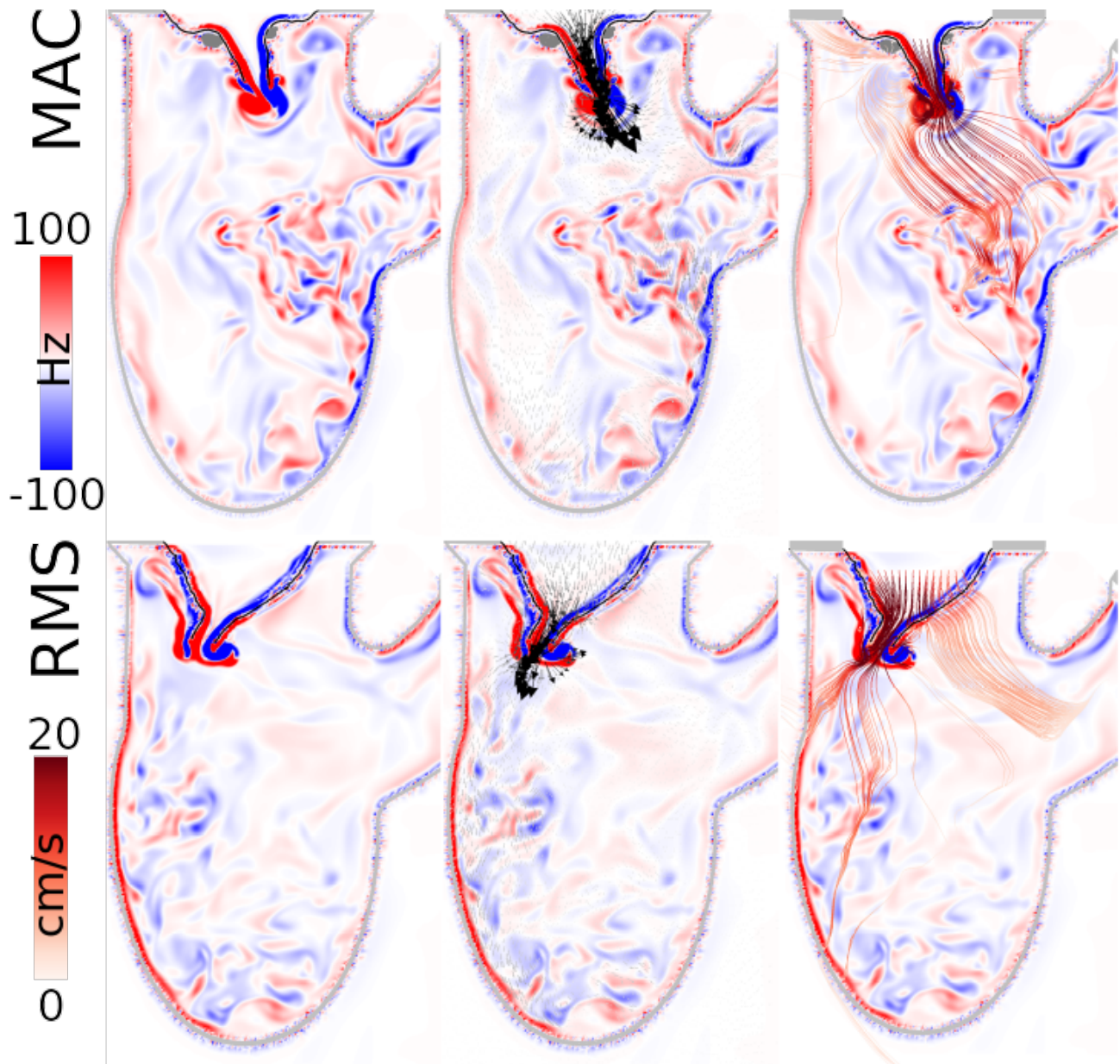


Figure S4: Snapshots of the planar vorticity (**left**), planar vorticity with velocity vectors (**middle**), and planar vorticity with overlaid streamlines (**right**) during early diastole for the simulations at 5.0 L/min.

D. Advecting Particles for 3.5 L/min with Water and 5.0 L/min

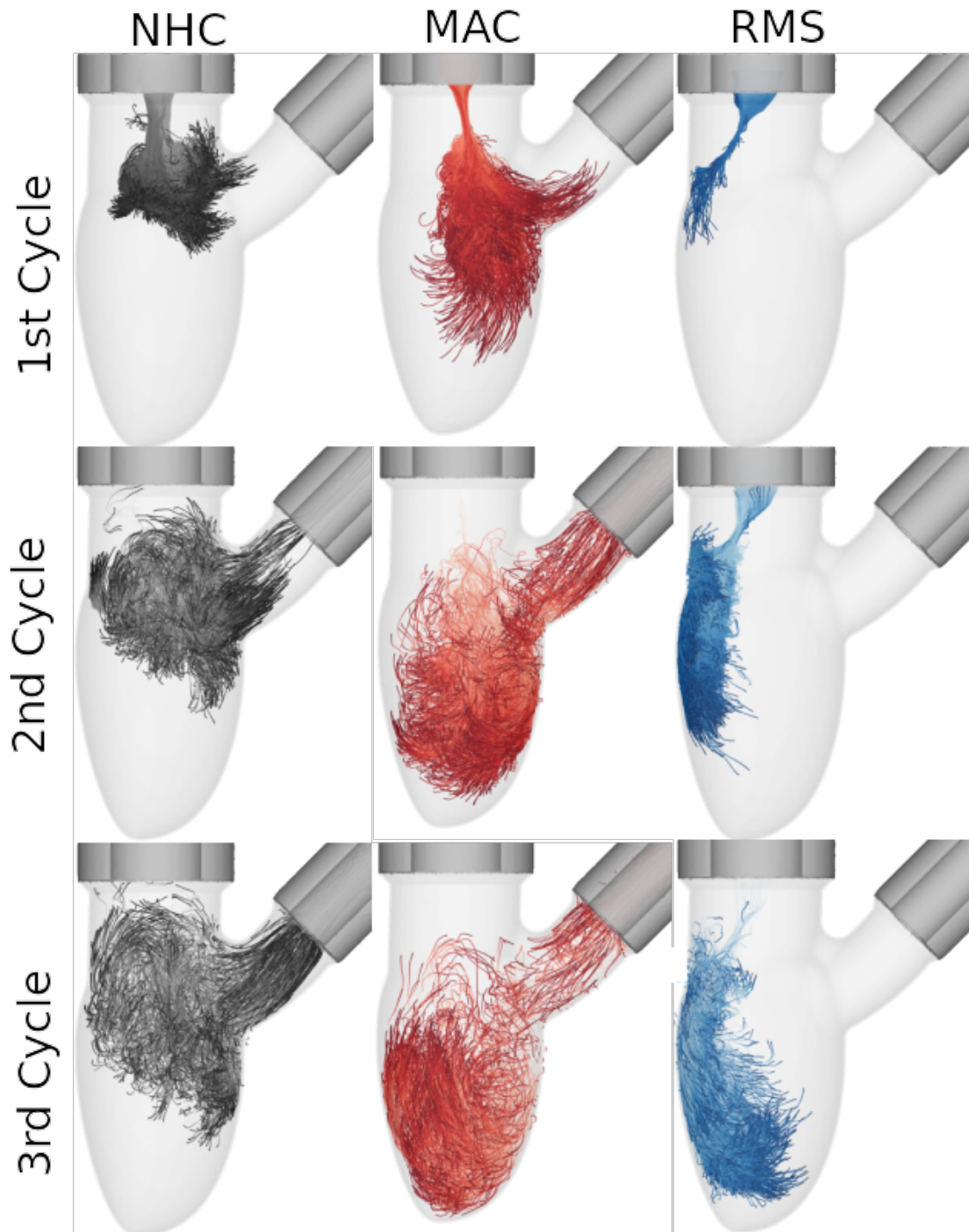


Figure S5: Particle traces one, two, and three cycles after particle release for the 3.5 L/min simulations with water. The particle tails correspond to the path each particle has taken over the previous cycle. The time point corresponds to the end of systole (see 1.4 seconds in **Fig. S1** for the corresponding pump phase).

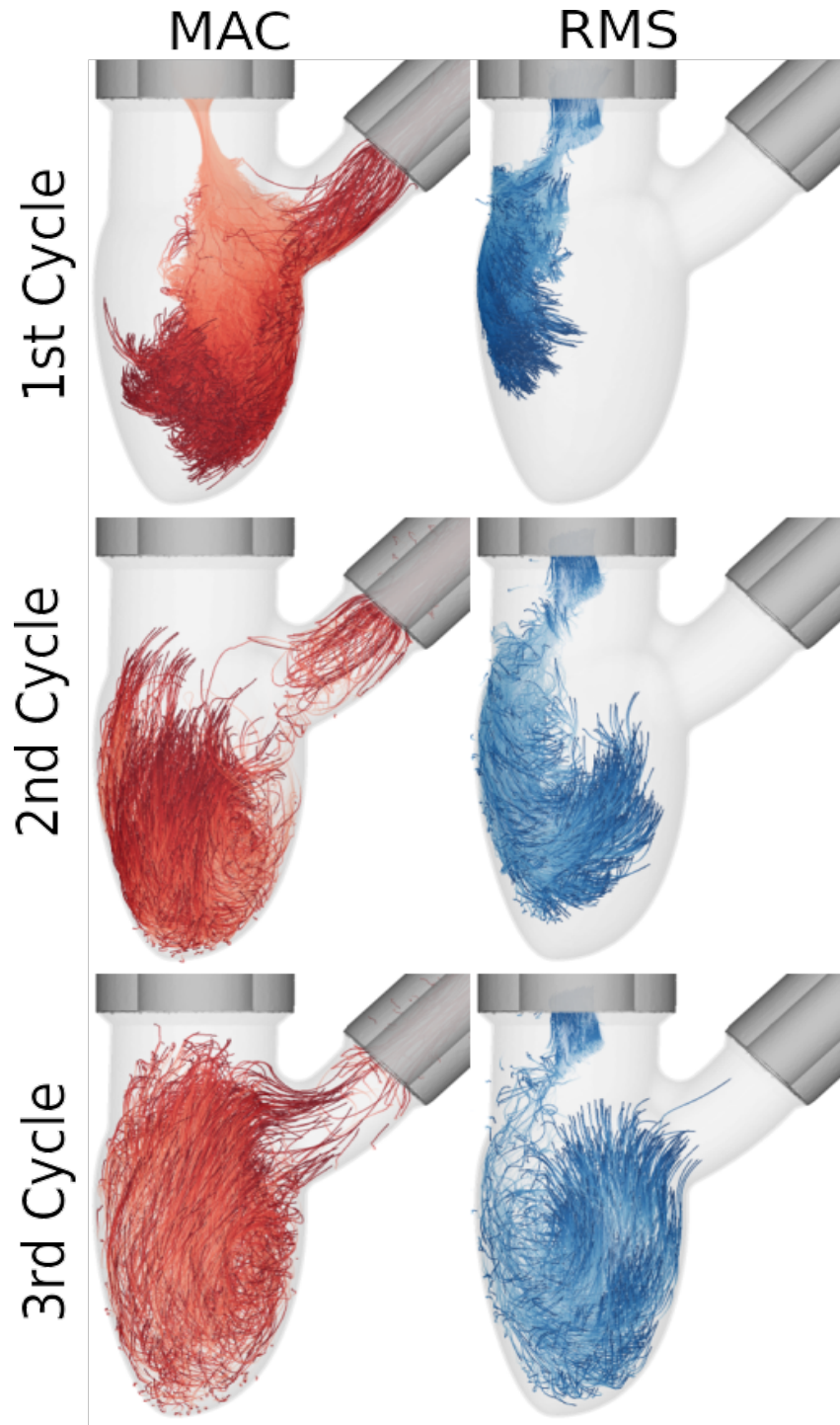


Figure S6: Particle traces one, two, and three cycles after particle release for the 5.0 L/min simulations. The particle tails correspond to the path each particle has taken over the previous cycle. The time point corresponds to the end of systole (see 1.4 seconds in **Fig. S1** for the corresponding pump phase).

E. In Vitro Flow Comparison

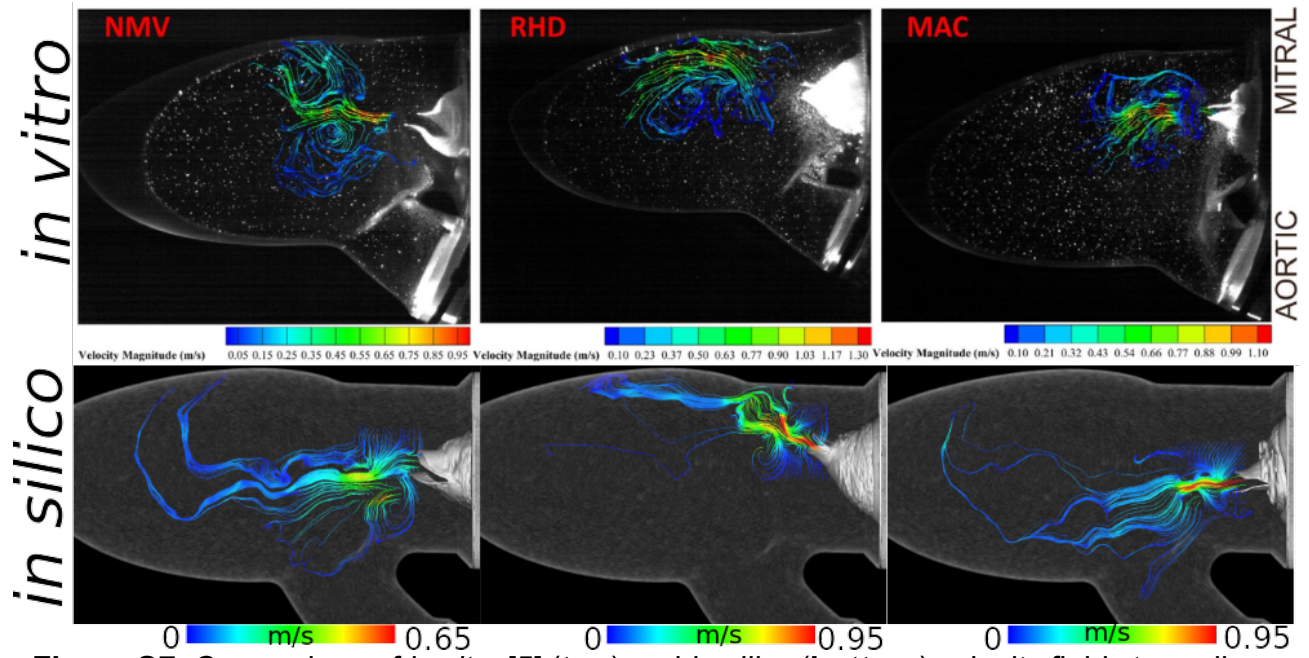


Figure S7: Comparison of *in vitro* [5] (**top**) and *in silico* (**bottom**) velocity field streamlines at 3.5 L/min with water. The velocity bounds were reduced *in silico* from the *in vitro* counterparts to better highlight the differences throughout the ventricle. The acronyms are from the *in vitro* paper where MAC is the same but NMV is normal mitral valve (NHC in this study) and RHD is rheumatic disease (RMS in this study).

F. Out-of-plane Velocity Magnitude

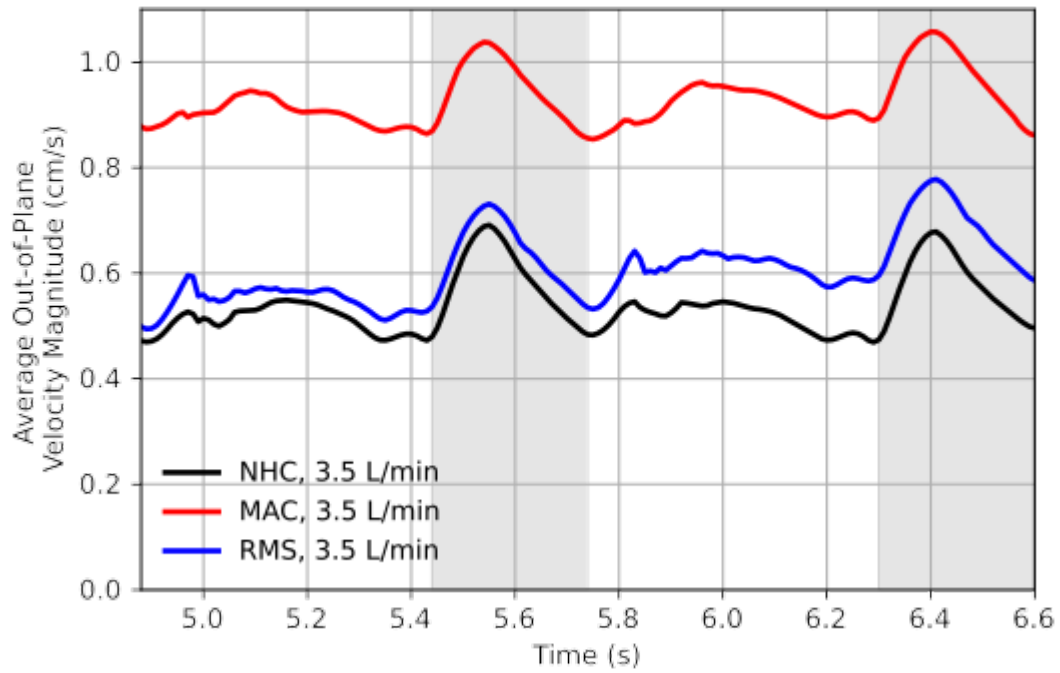


Figure S8: Average out-of-plane (defined by the plane that bisects the MV and AV) velocity magnitude within the entire ventricle across the final two cycles for the 3.5 L/min simulation with blood analogue.