# NUMERICAL INVESTIGATION OF HEMOLYSIS PHENOMENA IN THE FDA NOZZLE BENCHMARK: MIND THE EXTENSIONAL STRESSES

**Ali Maghouli[1,2], Annie-Claude Bayeul-Lainé[1], Sophie Simonet[1], Mohammad Haddadi[3], and Olivier Coutier-Delgosha[1,4]**

[1]ENSAM, Univ. Lille, CNRS, ONERA, Arts et Metiers Institute of Technology, Centrale Lille, UMR 9014 - LMFL - Laboratoire de Mécanique des Fluides de Lille - Kampé de Fériet, F-59000 Lille, France
[2]Medrik Dynamic Technology, Hopital Xavier Arnozan, Avenue du Haut Leveque, 33600 Pessac, France
[3]FineHeart, 28 Avenue Gustave Eiffel, Bâtiment C, 33600 Pessac, France
[4]Virginia Tech, Virginia Tech, Kevin T. Crofton, Dept. of Aerospace and Ocean Eng., Blacksburg, 460 Old Turner Street, VA 24061, USA

## SUMMARY

In recent years, the idea of using a pump as a left ventricle assist device is being well developed by several groups. Meanwhile, one of the challenges in this field is the occurrence of biological phenomena such as hemolysis. From an engineering point of view, a solution to this problem is to provide an accurate and efficient numerical method to predict the phenomenon. Hemolysis models are typically based on equivalent scalar stress and exposure time. This paper aims to study the impact of considering extensional stresses as the main reason for hemolysis in the FDA Nozzle benchmark. The idea comes from an experimental article. First of all, flow's hemodynamic was validated by comparing the results of normalized axial velocity in several sections with the experimental data. In this case, three different RANS models $k-\epsilon$, $k-\omega$ and $k-\omega\ SST$ were employed. As expected, it is clear that the $k-\omega\ SST$ is the most accurate model. In the next step, hemolysis simulations performed for different equivalent stresses. In this case, the impact of scaling up the extensional stresses on predicted hemolysis is studied by adding a coefficient $C_n$ to equivalent stress. It is concluded that by applying these new modifications the hemolysis index would be in a reliable range.

**Key words:** *Hemolysis, Power-Law Model, FDA Nozzle, Extensional Stress, Fluid Dynamics*

## 1 INTRODUCTION

The history of research on numerical modeling of the hemolysis phenomenon dates back to the last century. In 1990, the power-law model introduced by Giersiepen [1] was the first to predict this phenomenon. In the proposed equation (Equation 1), the hemolysis index is calculated as a function of flow field shear stress ($\tau\ (Pa)$) and time($t\ (s)$).

$$H = C\tau^{\alpha}t^{\beta} \tag{1}$$

$C$, $\alpha$, and $\beta$ are empirical dimensionless constants which are equal to $9.772\times10^{-7}$, $1.444$, and $0.2076$ respectively ([2]).

Over the last two decades, many pieces of research have been done to improve the performance of numerical methods in predicting hemolysis phenomena in blood pumps. In the meantime, the Eulerian approach has always been more popular due to its more straightforward structure for numerical discretization. In the Eulerian approach, the general form of the power-law model transforms to a standard transport equation:

$$\frac{\partial H_L}{\partial t} + v.\nabla H_L = C^{\frac{1}{\beta}}\tau^{\frac{\alpha}{\beta}} \tag{2}$$

Where $H_L = H^{\frac{1}{\beta}}$. In addition, $\tau$ is a scalar parameter that equivalent stress should replace. At this point, the main challenge is calculating the equivalent stress for use in the power-law model. As we know, the stress in the flow field is not a scalar parameter but a tensor one. So, the question is how to make this tensor parameter usable in the hemolysis model.

Many researchers have proposed different methods for calculating this equivalent stress. Most of these methods try to find a maximum value for the stress in different ways (see Table 1). One of the latest versions of equivalent stress is the one proposed by Faghih and Sharp [5]. In this paper, the authors try to study the impact of different types of stress on the hemolysis phenomena. They applied extensional, and shear stresses using two separate test cases. In conclusion, they found that hemolysis is more dependent on extensional stress components. Consequently, they provided new equivalent shear stress, which is based on a new parameter $C_n$.

Table 1: Equivalent stress based on different definitions .

| Definition | Source |
|---|---|
| $\tau_s = (\frac{1}{3}(\tau_{xx}^2 + \tau_{yy}^2 + \tau_{zz}^2 - \tau_{xx}\tau_{yy} - \tau_{xx}\tau_{zz} - \tau_{yy}\tau_{zz} + (\tau_{xy}^2 + \tau_{xz}^2 + \tau_{yz}^2))^{\frac{1}{2}}$ | [3] |
| $\tau_e = (\frac{1}{2}(\tau_{xx}^2 + \tau_{yy}^2 + \tau_{zz}^2) + (\tau_{xy}^2 + \tau_{xz}^2 + \tau_{yz}^2))^{\frac{1}{2}}$ | [4] |
| $\tau_n = (C_n^2(\tau_{xx}^2 + \tau_{yy}^2 + \tau_{zz}^2 - \tau_{xx}\tau_{yy} - \tau_{xx}\tau_{zz} - \tau_{yy}\tau_{zz}) + (\tau_{xy}^2 + \tau_{xz}^2 + \tau_{yz}^2))^{\frac{1}{2}}$ | [5] |

Where $C_n$ imposes the weighting between two types of stresses. Considering the experimental data, $\sqrt{3}C_n = 33.69$ is announced as the final value for this parameter. Of course, this value is obtained for human blood, which can be different for other species. The main objective of this work is to investigate the effect of parameter $C_n$ of $\tau_n$ on hemolysis simulation results. For this, simulations were first performed for three equivalent stresses $\tau_s$, $\tau_e$, and $\tau_n$, knowing that $\sqrt{3}C_n = 33.69$ is experimental based data for human blood. In the next step, the impact of parameter $C_n$ on the hemolysis index has been studied for four different species: Human, Bovine, Porcine, and Ovine.

### 1.1 FDA Nozzle Benchmark

The Nozzle benchmark project was the first part of the FDA (Food and Drug Administration of the United States) to evaluate CFD tools in the research and development of the LVADs. Initial outcomes of multi-laboratory experimental tests, which were based on the PIV (Particle Image Velocimetry) tools, published in 2011 [6]. Experimental data include velocities in different directions and pressure in several cross-sections of the flow. The second step of the project focused on the shear-inducer hemolysis [7]. This article investigates the impact of several fluid flow properties and blood properties on this phenomenon. Main goal of this paper was to generate some data to validate CFD-based predictions of this biological phenomenon.

The nozzle model contains regions of sudden and gradual changes in cross-sectional area, similar to many actual blood-contacting medical devices. Geometrical specifications of this test case are demonstrated in Figure 1. The nozzle is positioned at the test bench in two opposite directions: a) Sudden contraction (SC) at the inlet to the throat. b) gradual cone (GC) inlet configuration. These two configurations represent some critical regions that exist inside artificial cardiovascular devices.

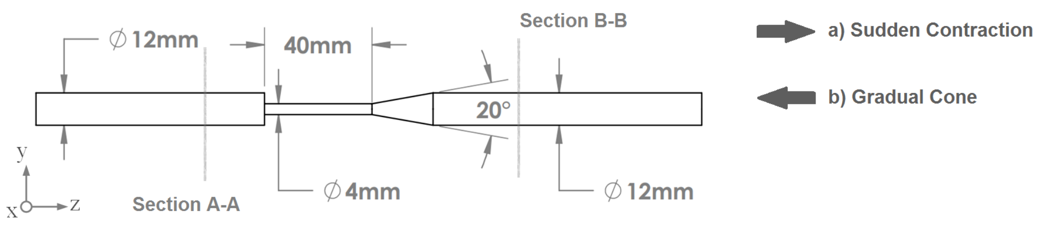


Figure 1: Dimensioning of the FDA Nozzle geometry and two opposite positions of it.

## 2 METHODOLOGY

In this paper, numerical simulations have been carried out using commercial solver ANSYS CFX v2020 R2 (ANSYS, Inc., Canonsburg, PA). This code has the potential to provide consistent and physical simulations of the complex flow. The Reynolds-averaged Navier-Stokes (RANS) in conjunction with the $k-\epsilon$, $k-\omega$ and $k-\omega\ SST$ turbulence models are used and compared for computations. In addition, the high-resolution scheme is used for the approximation of nonlinear convection terms in all transport equations. Blood flows in large arteries can be treated as a homogeneous Newtonian fluid, as it is an accepted assumption. Consequently, it is logical to consider the same hypothesis in the Nozzle. So, the dynamic viscosity and density have been adjusted as constant values 4 $cP$ and 1040 $\frac{kg}{m^3}$, respectively. In addition, Convergence criteria are specified as RMS residuals of $10^{-6}$.

Finally, the inlet and the outlet boundary conditions adjusted, focusing on the experimental conditions [6]. The outlet is set to have a static pressure of 0 $mmHg$. On the other side, a mass flow rate inlet condition is considered with Reynolds number: $Re_t = \frac{4\rho Q}{\pi d \mu}$. Where $\rho$, $Q$, $d_t$, and $\mu$ represents density, flow rate, throat diameter, and dynamic viscosity, respectively.

## 3 RESULTS AND CONCLUSIONS

### 3.1 Hemodynamic

The first step to predict reliable values for the hemolysis index is to validate the simulation considering hemodynamic of the model. In this case we have to verify our results based on experimental data of Hariharan [6]. To this, we performed hemodynamic simulation for both position (see Figure 1) in $Re = 6500$. Figure 2 and Figure 3 demonstrate normalized axial velocity ($\frac{V_z}{V_{z,inlet}}$) distribution in two different sections of the Nozzle (see Figure 1). Comparing different numerical method, we can conclude that the $k-\omega\ SST$ turbulent model results fits best with experimental data, as expected.

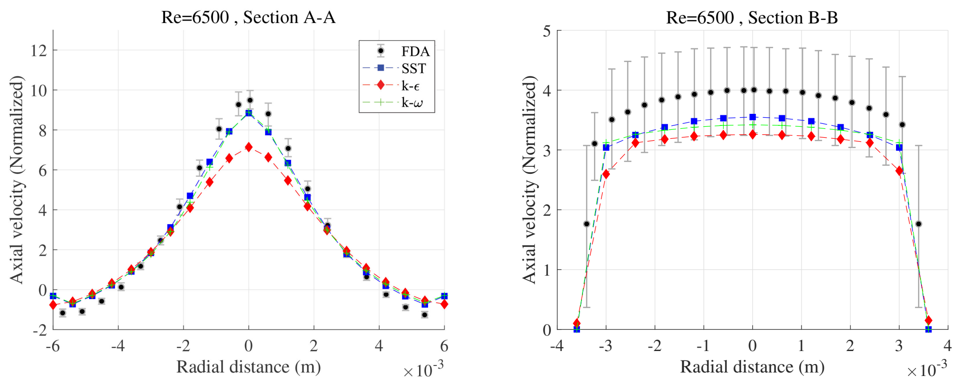


Figure 2: Hemodynamic validation of numerical simulation for Gradual cone position.

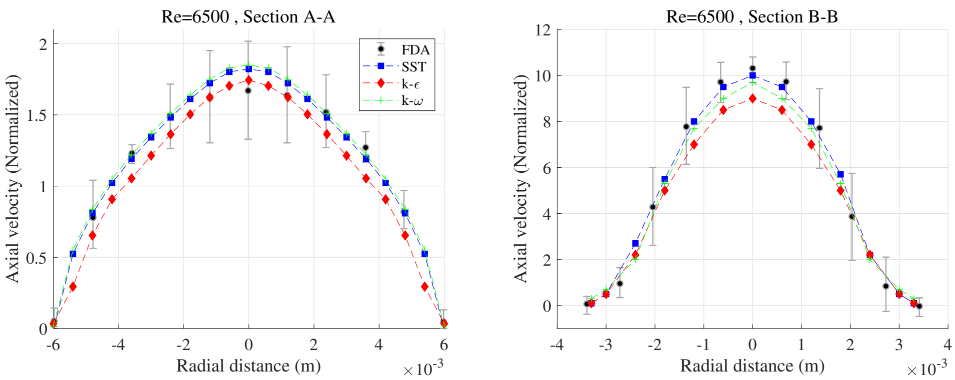


Figure 3: Hemodynamic validation of numerical results for Sudden contraction position.

### 3.2 Hemolysis

As mentioned in section 1, the main objective of this research is to find appropriate equivalent stress to be used in the Power-law model. And the idea is to test equivalent shear stress, which weighs extensional stress against shear stress—the idea belongs to an experiment-based article [5]. Hemolysis simulations were performed based on three equivalent stresses ($\tau_s$,$\tau_e$,$\tau_n$) for three operating conditions of the Nozzle ("Sudden contraction with 6 lpm", "Sudden contraction with 5 lpm", and Gradual cone with 6 lpm). To be consistent with experimental condition, model constants were selected as Bovine. Figure 4 demonstrates the Modified Hemolysis Index (MIH) outcomes for three different conditions. This represent a comparison between different numerical results and experimental data for all flow conditions. Considering this figure, we can conclude that in similar test cases implementation of $\tau_s$ and $\tau_e$ results in lower ranges of hemolysis in comparison with $\tau_n$. For example, for test condition "Sudden contraction with 6 lpm" both $\tau_s$ and $\tau_e$ cause an underestimation of MIH.

As a conclusion of this work, we can emphasize the importance of the extensional stresses on the accuracy and reliability of simulation outcomes of hemolysis simulations. The fact was found by an experimental procedure and now it is more clear as a consequence of a numerical study.

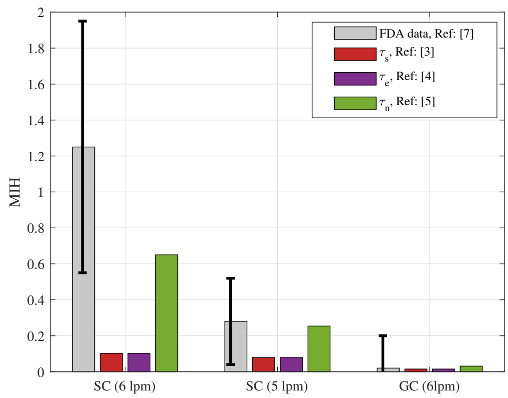


Figure 4: Hemolysis validation of numerical results for three operating conditions of the Nozzle.

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