Proceedings of the 11th World Congress on Mechanical, Chemical, and Material Engineering (MCM'25)
Paris, France - August, 2025
Paper No. HTFF 127
DOI: 10.11159/htff25.127

Effect of Micromagnetorotation on an MHD Blood Flow through an Idealised Stenosis

Kyriaki-Evangelia Aslani[1], Ioannis E. Sarris[2], Efstratios Tzirtzilakis[1]
[1]Department of Mechanical Engineering, University of the Peloponnese
M. Alexandrou 1, Koukouli Patra 26334, Greece
k.aslani@go.uop.gr; etzirtzilakis@uop.gr
[2]Department of Mechanical Engineering, University of West Attica
250 Thivon & P. Ralli, Egaleo Athens 12241, Greece
sarris@uniwa.gr

Abstract **-** This study examines the effect of micromagnetorotation (MMR) on MHD micropolar blood flow through an idealised stenosis. A numerical solver based on OpenFOAM was developed to analyse such flows, both with and without considering the effect of MMR behaviour. The solver was validated by comparing the numerical results with the analytical solutions of MHD micropolar Poiseuille blood flow with and without the effect of MMR. Here, emphasis is given to the impact of MMR on key characteristics of the flow, including the streamlines as well as the velocity and microrotation profiles both inside and outside the stenotic region. It was observed that the maximum velocity within the stenotic region decreases, while the vortex formed downstream of the stenosis is dampened. Considering the velocity and microrotation distributions within the stenotic region and downstream, it was found that including the MMR term can lead to a velocity reduction of up to 35% and a microrotation decrease of up to 99% when a magnetic field of 8T is applied. It is important to note that the impact of the Lorentz force alone (i.e., without acknowledging the MMR term) on the MHD micropolar blood flow through stenosis is minimal. In conclusion, the influence of MMR on MHD blood flow through stenosis is substantial. The findings of this study are expected to be valuable for bioengineering applications involving high-intensity applied magnetic fields on blood flows, such as magnetic hyperthermia and magnetic drug delivery.

Keywords**:** micropolar fluid; magnetohydrodynamics; ferrohydrodynamics; magnetisation; blood; stenosis.

1. Introduction

A micropolar fluid is defined as any fluid that contains small, rigid particles that move autonomously within the fluid. The theory of micropolar fluids was developed by Eringen (1966) and constitutes an extension of the Navier-Stokes equations used to describe Newtonian fluids [1]. Micropolar fluid theory differs from Navier-Stokes equations due to the presence of the antisymmetric part of the stress tensor, which is zero in the latter. The antisymmetric part of the stress tensor contributes to the equation for the conservation of angular momentum, which is not satisfied identically as in the Navier-Stokes equations. Thus, in micropolar fluids, another equation needs to be solved, i.e., the equation of the change of the internal angular momentum. Based on all this, a micropolar fluid is described not only by its axial velocity but also by its microrotation $\boldsymbol{\omega}$. Microrotation is defined as the total angular velocity of the particles in the internal microstructure of the fluid. Moreover, microrotation must differ from the vorticity of the fluid $\boldsymbol{\Omega} = \frac{1}{2}\,\nabla \times \boldsymbol{v}$, because if they coincide, the fluid can still be considered Newtonian, even when microrotation differs from zero. This rarely happens due to the introduction of an additional stress tensor related to the diffusion of microrotation, i.e., the couple stress tensor. Thus, microrotation coincides with vorticity only in specific cases [2]. The micropolar fluid theory has found application in the modelling of a variety of fluids, such as exotic lubricants, colloidal suspensions, liquid crystals, and blood [3].

An important class of micropolar fluids is the ferrofluids, which consist of ferromagnetic nanoparticles (such as magnetite) into a mother-liquid carrier (such as water or kerosene). Ferrofluids can have applications in both engineering and biomedicine, mainly because of their rheological properties that can be easily controlled by an externally applied magnetic field. A key characteristic of ferrofluids is the magnetic polarization of the particles (magnetisation) $\boldsymbol{M}$, which tends to 'relax' and align with the external magnetic field $\boldsymbol{H}$. This creates a magnetic moment $\boldsymbol{M} \times \boldsymbol{H}$ that contributes to the equation of change of the internal angular momentum and can significantly affect the microrotation of the ferromagnetic nanoparticles in the fluid. This phenomenon was extensively studied by Shliomis (1972) [4] and Rosensweig (1985) [5]. Later, Shizawa & Tanahashi (1986) [6] presented a complete mathematical model for ferrofluids using the micropolar fluid theory and Maxwell's equations, while introducing a new equation for the magnetisation change taking into account the existing

microrotation of the ferromagnetic nanoparticles (micromagnetorotation-MMR). The flow equations were derived using the methodology of irreversible thermodynamics, meaning that the dissipation function must always be positive.

Blood is also classified as a micropolar fluid, due to the existence of blood cells in the plasma. The experimental study of Ariman, Turk & Sylvester on steady and pulsatile blood flows showed a good agreement with the results that micropolar fluid theory predicts for these flows [7]. Moreover, some studies suggest that blood should be treated as a ferrofluid under an external applied magnetic field (for example when a magnetic resonance image-MRI scanner is used). Blood contains the haemoglobin molecule in the erythrocytes (red blood cells), which is an iron oxide and behaves like a magnetic particle, while blood plasma is the mother-liquid carrier [8]. Consequently, the applied magnetic field may influence the microrotation of erythrocytes due to the magnetisation of haemoglobin, subsequently impacting blood viscosity and velocity. There are experimental studies that provide statistically significant evidence of symptoms, such as vertigo, nausea, and metallic taste for magnetic field intensities of 1.5 and 4T associated with blood velocity reduction [9]. Furthermore, these observations cannot be attributed to the Lorentz force's impact on the blood flow, as the small electrical conductivity of blood prevents it from being influenced by the Lorentz force [8]. To the authors' knowledge, despite the evidence, the characterization of blood as a ferrofluid under the influence of an external magnetic field is not widely accepted; instead, blood is typically analyzed as a classical MHD Newtonian-or micropolar-fluid.

Recently, Aslani et al. [2, 10, 11] made several analytical studies on the impact of micromagnetorotation on micropolar magnetohydrodynamic (MHD) flows (such as blood) using the mathematical model of Shizawa and Tanahashi. These studies focused on establishing the differences arising by acknowledging and ignoring the MMR term, while the effect of the Lorentz force was always included. In summary, it was found that the MMR term has a strong braking effect both on velocity and microrotation. In the plane MHD micropolar Poiseuille flow, deceleration was up to 16%, whereas microrotation reduction was up to 99%. In the stability analysis of the same flow, it was shown that the MMR term has a strong stabilising effect, similar to the Lorentz force. Additionally, it was proven that the MMR reduces heat transfer by suppressing convection (temperature drop of up to 8.5%) due to the velocity reduction. Finally, it has been established that the micropolar nature of blood is more apparent than of a classical ferrofluid due to higher diffusion in erythrocytes microrotation, which is more intense in small vessels (such as arterioles). Moreover, it was shown that an externally applied magnetic field on a blood flow affects erythrocytes' microrotation through the MMR term, which in turn affects blood's velocity through vorticity, a phenomenon mentioned in many experimental studies that cannot be explained only by the effect of the Lorentz force.

Considering all the above, this study concerns the effect of micromagnetorotation on an MHD micropolar blood flow through an idealised stenosis. One value of stenotic region (66.67%) is considered. An applied magnetic field of 1, and 8T is used, while the hematocrit is kept constant at $\varphi = 45\ \%$. A numerical solver based on OpenFOAM was created to find solutions for micropolar MHD flows by acknowledging and ignoring the effect of MMR. The solver was validated by comparison with the analytical solutions of the classical MHD micropolar Poiseuille blood flow for various hematocrit and applied magnetic field values. Emphasis is given to the MMR effect on characteristic variables of the flow, i.e., the streamlines, and the velocity and microrotation profiles inside and outside the stenotic region. For comparison, all these variables were derived for the Newtonian blood flow through the stenosis, the corresponding micropolar blood flow through the stenosis, and the MHD blood flow through stenosis by ignoring and considering the effect of the MMR term.

2. Mathematical Formulation

As mentioned above, here, the effect of micromagnetorotation on blood flow in stenosis under the influence of an applied magnetic field is studied. Consider a cylinder of radius $R = 0.0015\ m$ (diameter $D = 2R = 0.003\ m$) and length $l = 0.06\ m$. An idealised stenosis was created on this cylinder, as depicted in Fig.1 using the formula of Hogan & Henriksen (1989) [12]. A magnetic field H_0 is applied transverse to the direction of the blood flow. The governing differential equations of the flow are shown below (Equations 2-9). The 3D cylindrical coordinate system *(r, θ, z)* is utilized, where z is the axial coordinate, r is the radial coordinate, and θ is the azimuthal angle. The assumptions of the flow being axisymmetric, i.e. $\frac{\partial}{\partial \theta} = 0$ and the radial and azimuthal velocity components being zero, i.e. $v_r = v_\theta = 0$ are applied. Therefore, the components of the velocity and microrotation are defined as $\boldsymbol{v} = 0, 0, v_z(r)$ and $\boldsymbol{\omega} = 0, \omega_\theta(r), 0$, respectively. No-slip boundary conditions are imposed for the velocity and Condiff-Dahler conditions are imposed for microrotation, i.e.:

$$v_z(R) = 0 \text{ and } \omega_\theta(R) = 0 \tag{1}$$

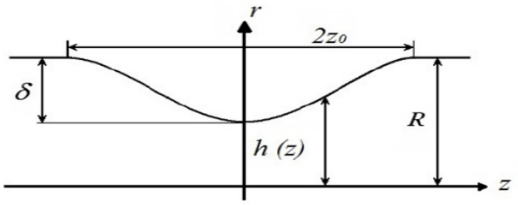


Fig. 1: Schematic representation of the geometry of the stenosis.

For the examination of the stenosis, one value of the hematocrit was used, i.e., $\varphi = 45\ \%$. Considering blood's physical properties, they were derived from the papers of Ariman et al. (1974) and Pai et al. (1996) [7, 12] and they are tabulated in Table 1.

Table 1: Blood's physical properties.

Physical properties	**Value**
volume fraction $\varphi\,(\%)$	45
dynamical viscosity $\mu\,(Pa\cdot s)$	$4\cdot 10^{-3}$
rotational viscosity $\mu_r\,(Pa\cdot s)$	$2.7\cdot 10^{-3}$
microinertia coefficient $j\,(m^2)$	$7.5\cdot 10^{-10}$
spin viscosity $\gamma\,(\frac{kg\cdot m}{sec})$	$3\cdot 10^{-12}$
fluid density $\rho_f(kg\cdot m^{-3})$	1050
saturation magnetisation $M_s\,(A\cdot m^{-1})$	100
electrical conductivity $\sigma\,(S\cdot m^{-1})$	0.067
magnetisation relaxation time $\tau\,(sec)$	0.001

As mentioned in the Introduction, three values of the applied magnetic field's intensity were used, which are frequently found in experimental and numerical studies associated with biomedical and engineering applications, i.e. $H_0 = 795{,}774.72\ \frac{A}{m}$, and $H_0 = 6{,}366{,}197.72\frac{A}{m}$. Reminding that $H_0 = \frac{B_0}{\mu_0}$ which leads to $H_0 = 795{,}774.72\ \frac{A}{m}$ corresponding to $1\ T$, and $H_0 = 6{,}366{,}197.72\frac{A}{m}$ corresponding to $8\ T$.

For the simulation of the blood flow, a suitable pressure gradient G or an inlet velocity must be specified. For the numerical simulations to resemble a realistic human blood flow, no value for the inlet velocity was applied [12]. Instead, a maximum velocity value was selected according to the size of the blood vessel, and the corresponding pressure gradient was calculated using Poiseuille law $v_{zmax} = \frac{(2R)^2}{8\mu}G$. Considering that $v_{zmax} = 0.1\ {}^{m}/_{sec}$, it was derived that $G = 355.556\ \frac{Pa}{m}$. Using that $l = 0.06$ and $\frac{P_2 - P_1}{l} = G$, and that the outlet pressure $P_2 = 0\ Pa$, the inlet pressure was calculated as $P_1 = 21.336\ Pa$.

The governing equations for describing an MHD micropolar flow with ferromagnetic particles (such as blood) are:

$$\rho\frac{D\boldsymbol{v}}{Dt} = -\nabla p + \mu\nabla^2\boldsymbol{v} + 2\mu_r\nabla\times(\boldsymbol{\omega}-\boldsymbol{\Omega}) + (\boldsymbol{M}\cdot\nabla)\boldsymbol{H} + \mu_0(\boldsymbol{j}\times\boldsymbol{H}) \tag{2}$$

$$\rho j\frac{D\boldsymbol{\omega}}{Dt} = 4\mu_r(\boldsymbol{\Omega}-\boldsymbol{\omega}) + \gamma\nabla^2\boldsymbol{\omega} + \boldsymbol{M}\times\boldsymbol{H} \tag{3}$$

$$\nabla\cdot\boldsymbol{B} = 0 \tag{4}$$

$$\nabla\times\boldsymbol{H} = \boldsymbol{j} \tag{5}$$

$$\boldsymbol{j} = \sigma(\boldsymbol{v}\times\boldsymbol{B}) \tag{6}$$

$$\boldsymbol{M} = \frac{M_0}{H}[\boldsymbol{H} - \tau(\boldsymbol{H}\times\boldsymbol{\omega})] \tag{7}$$

where ρ is fluid's density, t is time, p is pressure, μ is the dynamic viscosity, μ_r is the rotational or dynamic viscosity, $\boldsymbol{M}$ is the magnetisation vector, $\boldsymbol{H}$ is the applied magnetic field from the magnetic flux density $\boldsymbol{B} = \mu_0\boldsymbol{H} + \boldsymbol{M}$ with μ_0 being the

magnetic permeability of free space ($4\pi \cdot 10^{-7}\ {}^{H}/_{m}$), $\boldsymbol{j}$ is the current density vector, j is the microinertia coefficient, γ is spin viscosity, σ is the electrical conductivity, M_0 is the equilibrium magnetisation and τ is the magnetisation relaxation time. The first relation represents the conservation of linear momentum, while the second relation represents the equation of change of the internal angular momentum, which appears in micropolar fluids and is derived from the antisymmetric part of the stress tensor that contributes to the conservation of the total angular momentum. In the first equation, the term $2\mu_r \boldsymbol{\nabla} \times (\boldsymbol{\omega} - \boldsymbol{\Omega})$ represents force due to microrotation-vorticity difference that characterises micropolar fluids, the term $(\boldsymbol{M} \cdot \boldsymbol{\nabla})\boldsymbol{H}$ is the magnetic body force, and the term $\mu_0(\boldsymbol{j} \times \boldsymbol{H})$ is the Lorentz force. In the second equation, the term $4\mu_r(\boldsymbol{\Omega} - \boldsymbol{\omega})$ represents the internal torque due to the microrotation-vorticity difference, the term $\gamma \boldsymbol{\nabla}^2 \boldsymbol{\omega}$ is the microrotation diffusion and the last term $\boldsymbol{M} \times \boldsymbol{H}$ is the magnetic torque generated due to the magnetisation of the particles (micromagnetorotation). Equations (4), (5) and (6) represent Gauss law for magnetic monopoles, Ampere's law and Ohm's law, respectively (Maxwell's equations). Equation (7) is the constitutive magnetisation equation from the mathematical model of Shizawa and Tanahashi [6]. The vorticity of the fluid remains $\boldsymbol{\Omega} = \frac{1}{2} \boldsymbol{\nabla} \times \boldsymbol{v}$ as in Newtonian fluids. It is evident from Equations (2)-(3) that when $\mu_r = 0$ or when $\boldsymbol{\omega} = \boldsymbol{\Omega}$, the classical Newtonian hydrodynamic equations are retrieved. Moreover, from Equation (7) it can be seen that when no microrotation exists, i.e., when $\boldsymbol{\omega} = 0$, the magnetisation attains its equilibrium value M_0 parallel to the applied magnetic field $\boldsymbol{H}$. Provided that the blood flow is viscous and incompressible, the mass conservation law should be included:

$$\boldsymbol{\nabla} \cdot \boldsymbol{v} = 0 \tag{8}$$

$$\boldsymbol{\nabla} \cdot \boldsymbol{\Omega} = 0 \tag{9}$$

2.1. Solution Algorithm

Transient solvers for the numerical solution of Equations (2)-(9) were created based on the open-source library toolbox OpenFOAM. The popular transient solver icoFoam was used for the solution of the Newtonian blood flow through stenosis. This solver includes the PISO (Pressure Implicit with Splitting of Operators) algorithm. Considering the simple micropolar blood flow through stenosis without an externally applied magnetic field, micropolarFoam was utilised. This solver was first developed by Manolis and Koutsoukos (2018) [12] by altering the icoFoam solver to include the force term due to the microrotation-vorticity difference and the equation of change of the internal angular momentum. For the solution of the MHD micropolar blood flow through stenosis without the effect of micromagnetorotation, the solver epotMicropolarFoam was created for the first time. This solver is a modification of the transient solver epotFoam [12], which is used for incompressible, laminar flows of conducting fluids under the influence of a magnetic field. EpotMicropolarFoam uses the low magnetic Reynolds number approximation, where the magnetic induction equation is ignored and an electric potential formulation is employed. Finally, for the solution of the MHD micropolar blood flow through stenosis including the effect of MMR, epotMMRFoam was created. For this solver, the epotMicropolarFoam was modified to include the magnetic torque $\boldsymbol{M} \times \boldsymbol{H}$ in the internal angular momentum equation as a source term and the constitutive magnetisation equation (Equation 7).

All the equations in all solvers were discretised using the finite volume method (FVM). In detail, the second-order upwind Euler scheme was used for the discretisation of the transient terms, the Gauss linear corrected scheme was utilised for the diffusion terms, the second-order unbounded scheme was used for the convection terms, and the Taylor-like series expansion was applied to source terms. For the solution of the algebraic equations obtained by discretization, an explicit under-relaxation method and the preconditioned conjugate gradient method were utilised. More information about the solvers and the iteration methods used in this study can be found in Chapter 3 of the thesis of Aslani (2024) [12].

2.2. Code Validation

After the creation of the new OpenFOAM transient solvers for MHD micropolar flows with ferromagnetic particles, the first step was to validate them with the analytical results obtained from [2] regarding the plane MHD micropolar Poiseuille blood flow. Both analytical and numerical results were drawn using different values for the hematocrit, the height of the channel and the intensity of the applied magnetic field. The values that were chosen for blood's physical properties were derived from other numerical and experimental studies regarding various biomedical applications [8]. For the sake of comparison, the numerical results of the Newtonian Poiseuille blood flow, the simple micropolar Poiseuille blood flow and the MHD micropolar Poiseuille blood flow, where the effect of MMR is acknowledged and ignored, are also compared to the corresponding analytical results.

The results showed that the numerical results are in very good agreement with the analytical ones for all Poiseuille blood flow cases. The error does not exceed the value of 2% for both velocity and microrotation. This validates that the epotMicropolarFoam and the epotMMRFoam solvers, which were created for the first time, give accurate results with small errors, even for larger hematocrit values and high magnetic field intensities. More information for the validation of the solvers can be found in Chapter 5.1 of the thesis of Aslani (2024) [12].

2.3. Geometry and Mesh Refinement

As mentioned above, the geometry of the problem is an artery with an idealised stenotic region. For this study, one value was considered for the stenosis diameter D_s being the one-third of the nominal artery diameter D, i.e., $D_s = 0.001\ m$, yielding to stenosis of $66.67\ \%$. The discretisation of the computational domain was achieved with the use of a hexagonal O-grid mesh, which was created using the open-source platform Salome. In Fig. 2 the computational mesh is illustrated as seen from Paraview.

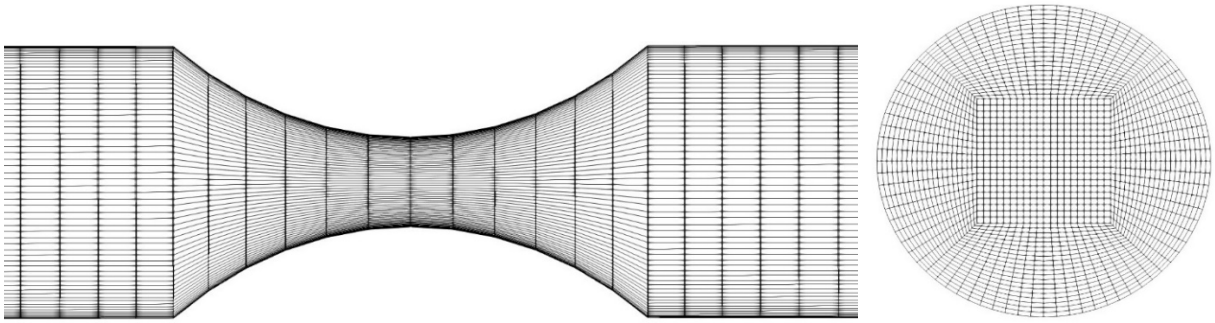


Fig. 2: Computational mesh for the stenosis as depicted in Paraview.

The independence of the mesh was tested by comparing the maximum velocity, vorticity and microrotation values for all flow cases (Newtonian, micropolar and MHD by ignoring and considering the effect of the MMR term), both in the stenotic region and downstream the stenosis, using different cell density. The time step of every simulation was selected such as the Courant number to be smaller than one. The time that the simulation ends was selected such as the flow to reach the steady state and to be fully developed. For more information about the simulation, one can refer to Chapter 5.3 of the thesis of Aslani (2024) [12].

3. Results

In this subsection, numerical results for the MHD blood flow through a stenosis are discussed. Emphasis is given on the effect of MMR on characteristic variables of the flow, i.e., the streamlines, and the velocity and microrotation profiles inside and outside the stenotic region. All variables are plotted for the Newtonian blood flow through stenosis, the corresponding micropolar blood flow through stenosis, and the MHD blood flow through stenosis by ignoring and considering the effect of the MMR term. In this manner, Fig. 3 shows the streamlines of the stenotic blood flow using an applied magnetic field of $1\ T$, and $8\ T$. It is evident that the velocity of the blood increases at the stenosis, while it reaches its maximum value at the position where the stenosis is most tight; then it returns to its initial inlet value at the downstream positions away from the stenosis. A vortex is created inside the stenotic area. The streamlines for the simple micropolar blood flow and the MHD micropolar blood flow without acknowledging the MMR term do not exhibit noticeable differences, for all considered values of the applied magnetic field. On the other hand, when the MMR term is considered, the value of the maximum velocity inside the stenotic region decreases, while the vortex downstream stenotic region is slightly dampened.

Fig.4 illustrates the velocity distributions inside the stenotic region and outside the latter for the Newtonian blood flow, the corresponding micropolar blood flow, and the MHD micropolar blood flow, by ignoring and considering the effect of the MMR term (acronyms MHD and MMR, respectively), using an applied magnetic field of $1\ T$, and $8\ T$. As expected from the results for the streamlines, the velocity inside the stenosis is almost increased by two times compared to the velocity downstream of the stenosis. The velocity of the micropolar blood flow in the stenosis is slightly smaller than the Newtonian one. The maximum velocity difference appears at the centre of the vessel and is calculated as $\Delta v_{zmax} = 2.72\ \%$ in the stenotic region and $\Delta v_{zmax} = 4.19\ \%$ downstream the latter.

Considering the application of the magnetic field on the blood flow when the MMR term is ignored, it does not affect the velocity noticeably both inside the stenotic region and outside the latter, due to the small effect of the Lorentz force on

the flow. As expected, when the MMR term is considered, the velocity decreases both inside the stenotic region and outside the latter.

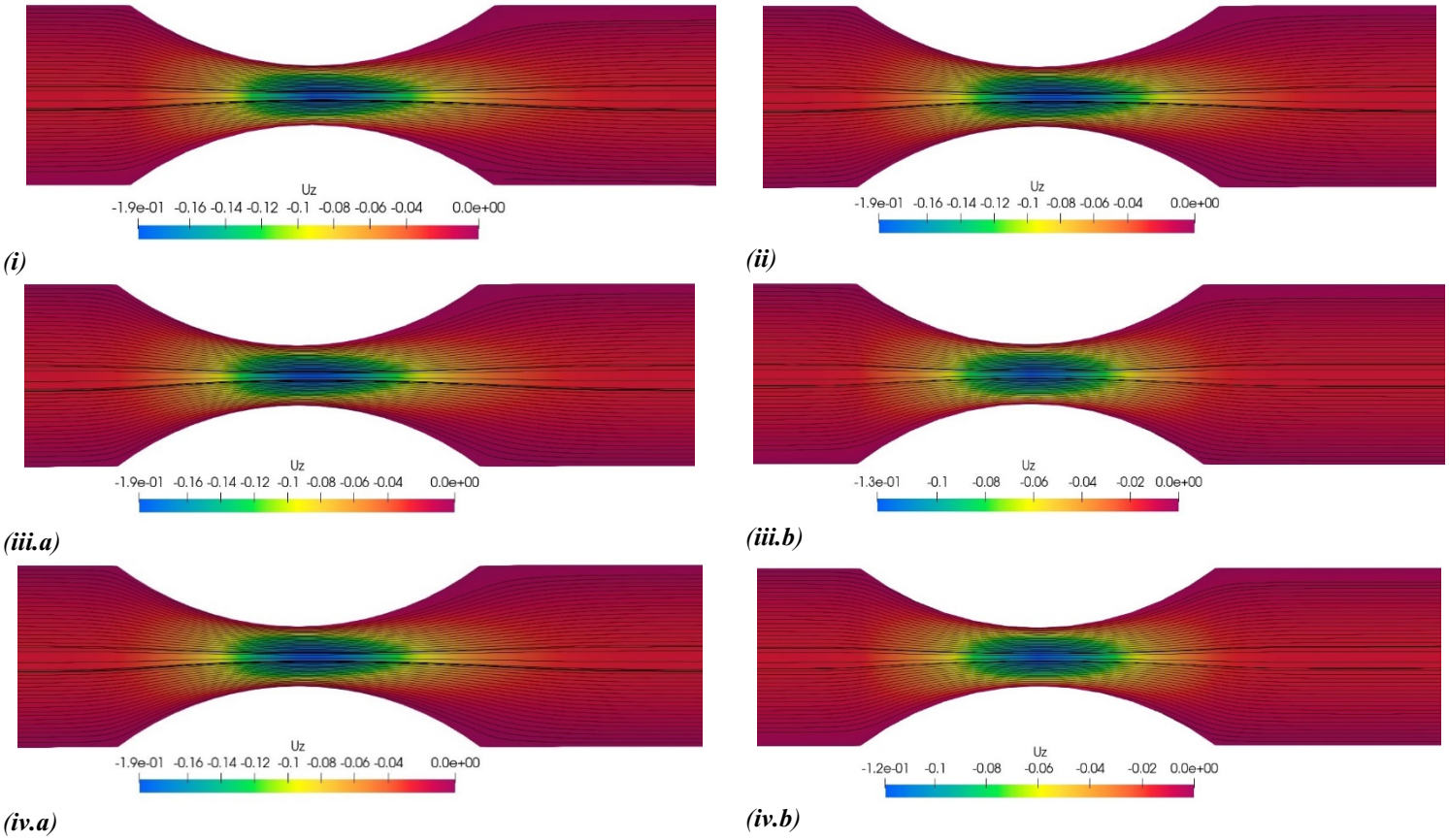
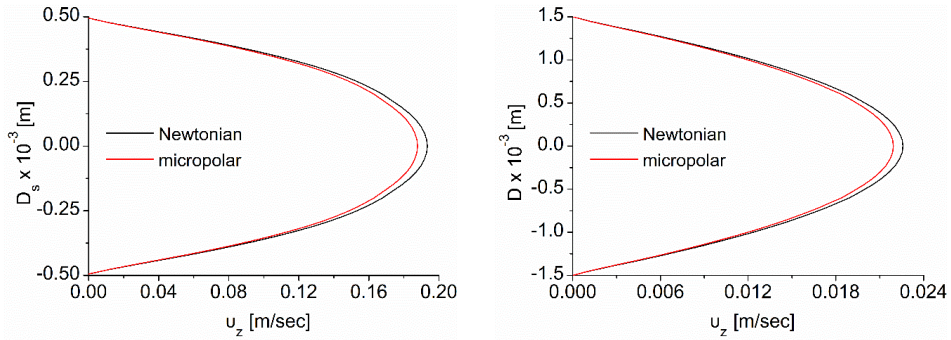


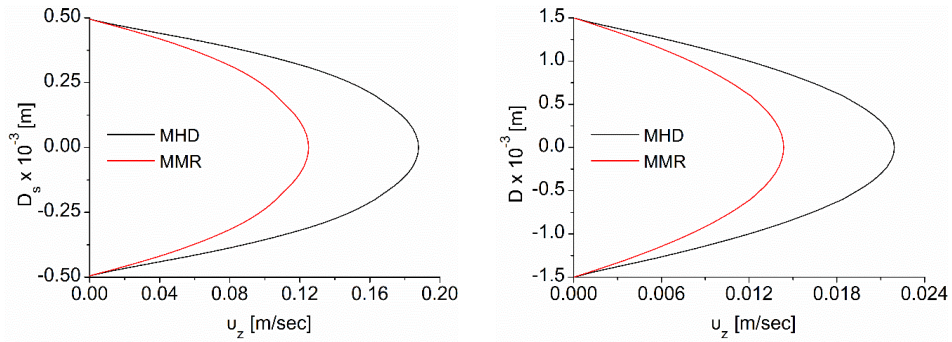
Fig. 3: Streamlines for the blood flow through stenosis using (i) Newtonian modelling, (ii) micropolar modelling without magnetic field, (iii) MHD micropolar modelling with a magnetic field of 1 T and (iv) 8 T by (a) ignoring MMR and (b) considering MMR.

The maximum velocity differences in the stenotic region are $\Delta v_{zmax} = 31.32\ \%$, and $\Delta v_{zmax} = 33.55\ \%$ forthe applied magnetic field of 1 T, and 8 T, respectively. Similarly, the maximum velocity differences downstream the stenotic region are $\Delta v_{zmax} = 32.94\ \%$, and $\Delta v_{zmax} = 35.02\ \%$ for an applied magnetic field of 1 T, and 8 T, respectively. These results confirm that the influence of the MMR term is decreased when the characteristic diameter of the vessel decreases, which implies that there is greater resistance between the erythrocytes.

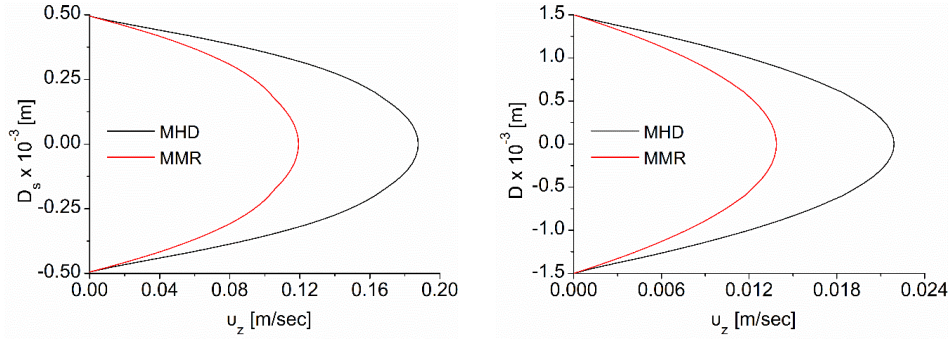
Fig. 5 shows the microrotation distributions inside the stenotic region and outside the latter for the micropolar blood flow without an applied magnetic field and the MHD micropolar blood flow by ignoring and considering the effect of the MMR term (acronyms MHD and MMR, respectively), using an applied magnetic field of 1 T, and 8 T.



(a)



(b)



(c)

Fig. 4: Velocity profiles for the centre of the stenotic region (left) and downstream the stenotic region (right) for (a) the Newtonian and micropolar blood flow without an applied magnetic field and (b) the micropolar blood flow by ignoring and considering the MMR term with a magnetic field of 1 T, and (c) 8 T. D_s refers to the diameter of the stenosis and D refers to the diameter of the artery.

The microrotation inside the stenotic region is by 96 % greater compared to the latter downstream of the stenotic region, due to the big stenosis (over 50 %). The application of the external magnetic field without the consideration of the MMR term does not bring any noticeable differences (micropolar and MHD profiles – black and red lines- coincide). On the other hand, when the effect of MMR is included (blue line), microrotation decreases vastly both in the stenotic region and outside the latter. The maximum velocity differences in the stenotic region are $\Delta\omega_{\theta max} = 92.91\ \%$, and $\Delta\omega_{\theta max} = 99.03\ \%$ for applied magnetic field of 1 T, and 8 T, respectively. Similarly, the maximum microrotation differences downstream the stenotic region are $\Delta\omega_{\theta max} = 93.18\ \%$, and $\Delta\omega_{\theta max} = 99.08\ \%$ for applied magnetic field of 1 T, and 8 T, respectively. It is evident that the MMR term affects microrotation greatly reaching a 99 % reduction for an applied magnetic field of 8 T. This physically means that the internal rotation of the erythrocytes almost "freezes", due to the alignment of the latter with the applied magnetic field, which leads to a considerable velocity reduction.

4. Conclusions

This study concerns the effect of MMR on an MHD micropolar blood flow through an idealised stenosis. A numerical solver based on OpenFOAM was created to find solutions for micropolar MHD flows by acknowledging and ignoring the effect of MMR. The solver was validated by comparison with the analytical solutions of the classical MHD micropolar Poiseuille blood flow for various hematocrit, channel sizes, and applied magnetic field values. Emphasis is given to the MMR effect on characteristic variables of the flow, i.e., the streamlines, and the velocity and microrotation profiles inside and outside the stenotic region. For the streamlines, it was found that the value of the maximum velocity inside the stenotic region decreases, while the vortex downstream of the stenotic region is dampened. Considering the velocity and microrotation distributions inside the stenotic region and downstream the latter, it was found that the consideration of the MMR term leads to a velocity reduction of up to 35% and microrotation reduction up to 99% for an applied magnetic field of 8 T. It should be mentioned that the effect of the Lorentz force alone (i.e. without the consideration of the MMR term) on the MHD micropolar blood flow through stenosis is barely noticeable. In conclusion, the effect of micromagnetorotation on an MHD blood flow through stenosis is significant. The results of this study are anticipated to have great value for bioengineering applications that involve applied magnetic fields on blood flows (such as magnetic hyperthermia and magnetic drug delivery).

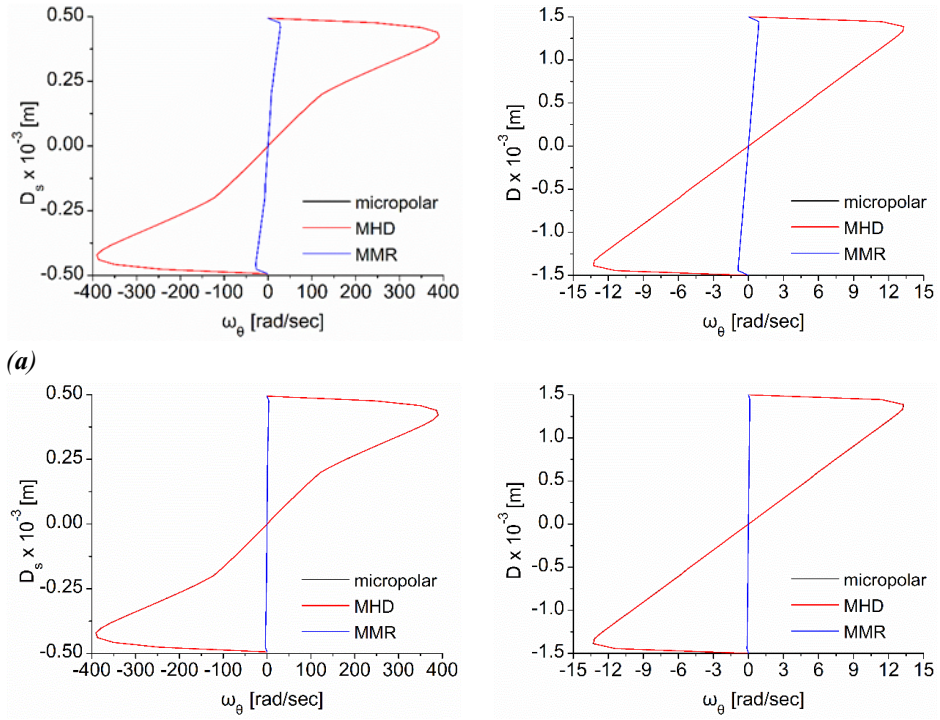


Fig. 5: Microrotation profiles for the centre of the stenotic region (left) and downstream the stenotic region (right) for the micropolar blood flow without an applied magnetic field and the micropolar blood flow by ignoring and considering the MMR term with a magnetic field of (a) 1 *T*, and (b) 8 *T*. D_s refers to the diameter of the stenosis and D refers to the diameter of the artery.

Acknowledgements

This research was funded by the Action "Flagship actions in interdisciplinary scientific fields with a special focus on the productive fabric," implemented through the National Recovery and Resilience Fund Greece 2.0 and funded by the European Union–NextGenerationEU (Project ID: TAEDR-0535983).

References

[1] A. C. Eringen, "Theory of micropolar fluids," *J. Math. Mech.,* vol. 16, no. 1, pp. 1-18, 1966.
[2] K.-E. Aslani, E. Tzirtzilakis, and I. E. Sarris, "On the mechanics of conducting micropolar fluids with magnetic particles: Vorticity–microrotation difference," *Phys. Fluids,* vol. 36, no. 10, 2024.
[3] G. Lukaszewicz, *Micropolar fluids: theory and applications*. Springer Science & Business Media, 1999.
[4] M. Shliomis, "Effective viscosity of magnetic suspensions," *Zh. Eksp. Teor. Fiz,* vol. 61, no. 2411, pp. s1971d-1971, 1971.
[5] R. E. Rosensweig, *Ferrohydrodynamics*. London: Cambridge University Press, 1985.
[6] K. Shizawa and T. Tanahashi, "New constitutive equations for conducting magnetic fluids with internal rotation: Thermodynamical discussions," *Bull. JSME,* vol. 29, no. 255, pp. 2878-2884, 1986.
[7] T. Ariman, M. Turk, and N. Sylvester, "On steady and pulsatile flow of blood," *J. Appl. Mech.,* vol. 41, no. 1, pp. 1-7, 1974.
[8] E. Tzirtzilakis, "A mathematical model for blood flow in magnetic field," *Phys. Fluids,* vol. 17, no. 7, p. 077103, 2005.
[9] J. F. Schenck, C. Dumoulin, R. Redington, H. Kressel, R. Elliott, and I. McDougall, "Human exposure to 4.0-Tesla magnetic fields in a whole-body scanner," *Med. Phys.,* vol. 19, no. 4, pp. 1089-1098, 1992.
[10] K.-E. Aslani and I. E. Sarris, "Effect of micromagnetorotation on magnetohydrodynamic Poiseuille micropolar flow: Analytical solutions and stability analysis," *J. Fluid Mech.,* vol. 920, p. A25, 2021.
[11] K.-E. Aslani and I. E. Sarris, "Effect of micromagnetorotation on the heat transfer of micropolar Hartmann flow," *Therm. Sci. Eng. Prog.,* vol. 26, p. 101129, 2021/12/01/ 2021.
[12] K.-E. Aslani, "Biological flows under the effect of strong magnetic fields with micropolar theory," Ph.D. dissertation, Dept. Mech. Eng., Uni. of West Attica, Athens, Greece.