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Effect of static magnetic field on the hemodynamic properties of blood flow containing magnetic substances

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Abstract

The use of magnetic fields in targeted drug delivery, especially for treatment of cancers and tumoral regions, is one of the significant techniques in the field of modern methods of treatment. Considering that many vital biological tissues have been located deep in the body, then for targeted drug delivery and effective treatment in these tissues, it is required to bring therapeutic agent to the desired location and focus on that location. The purpose of this study is to evaluate the static magnetic field interaction with hemodynamic properties of blood flow containing a magnetic carrier substance as a bio magnetic fluid. The finite element method (FEM) is used for 2D numerical simulation of magnet with different tip shapes and evaluating of external static magnetic field and its effects on the blood flow with aforementioned properties. The results show that the static magnetic fields generated from magnets with different tip shapes have different effects on the distribution of the fluid velocity field. Furthermore it can be concluded that when magnetic field flux density is concentrated around the magnet tip, the intensity of these hemodynamic effects become more concentrated within the fluid and the location of the magnet tip on the tissue, though the hemodynamic variables have been changed.

Keywords: Static Magnetic Field, Hemodynamic Properties, Blood Flow, Magnetic Particles

1. Introduction

The main objective of targeted drug delivery to damaged tissues and organs is to deliver a specific dose of drug at the certain time to the location of diseased tissues in the body of living creatures. Overspreading of drug over the circulatory system is one of difficulties in non-targeted treatment. This problem causes various defects such as large amount of drug dissipation, low drug concentration in the diseased tissues, and increasing side effects in healthy tissues [1, 2].

To achieve the new targets for pharmaceutical, nanotechnology's role is undeniable. Several recent progresses in nanotechnology have shown that nanoparticles have a plentiful potential as drug carriers. Among the different nanostructure, magnetic nanoparticles (MNPs) have shown useful properties and applications. Magnetic nanoparticles present a wide variety of attributes, which make them highly promising carriers for drug delivery. Among MNPs, Iron oxide (Fe_3O_4) is the only magnetic material approved by Food and Drug Administration (FDA) [3].

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On the other hand, according to the compatibility interaction of static magnetic fields with living bodies, new methods of treatment is developed because it permits to put together the magnetic nanoparticles abilities and static magnetic field properties. All living organisms on the Earth planet, at any time are surrounded by the Earth's static magnetic field. The natural static magnetic field of the Earth is ~ 50 μT and, according to the geographic location, varies from ~ 30 to 70 μT [4]. A Ferro fluid is composed of many magnetizable nano-particles and is essentially super-paramagnetic. In the presence of an external field, ferro fluids are magnetized strongly and then lose their magnetization once the external field is removed due to rapid random particle reorientation [5].The investigation in magnetic drug targeting and consequently in bio magnetic fluid dynamic (BFD) is an interesting topic because of abundant proposed applications in medical sciences and bioengineering. Other application of magnetic-based manipulation includes cell separation, magnetic wound treatment, cancer tumor treatment, reduction of bleeding during surgeries or provocations of occlusion of the feeding vessels of cancer tumors and development of magnetic tracer [6]. Design of an effective magnetic drug targeting (MDT) system requires study in three main fields: synthesis of composite drug and magnetic particles, real-time imaging of magnetic particles as they are transported through the blood stream, and development of technology to direct the particles through the circulatory system and keep them at the correct location [2].The purpose of this study is to evaluate and understand more about third field. Due of their strength, low cost, and availability, permanent magnets are usually one of the first choices in several MDT studies for the purpose of particle steering and capture [1].

In addition to what was mentioned about MDT, since vessels containing blood, are important way to implement this process, so blood behavior in magnetic fields is also a very important issue. Red blood cells have the characteristics of a paramagnetic fluid when deoxygenated (in veins) and diamagnetic when oxygenated (in arteries) [7]. The response of blood on static magnetic fields depends on a different of factors such as: oxygen content, pH, temperature gradients, etc [8]. Since about 40 years ago until now, variety studies and researches have been done for better recognizing of magnetic fluid dynamics, properties and applications of magnetic field in the medical sciences and investigate of different material and process as pharmaceutical carrier. Some of these researches are: Senyei et al. who were the first to contain nanometer-sized iron oxide particles in a well-localized region of flow using a permanent magnet [2]. Scott Goodwin et al. studied the initial part of the basis for a preclinical rationale for magnetic targeted delivery using magnetic targeted carriers (MTCs). Their experiments show that MTCs can be accurately targeted in a relevant animal model to regions of interest in the liver and lung, with sufficient specificity to allow multiple administrations to different sites within the same region of interest [9]. Yousef Haik et al. discussed separation of red blood cells from the whole blood using a magnetic technique [7]. Curtism M. Oldenburg et al. presented the equations and methods used to simulate the flow of ferro fluids in porous media and to compare simulation results against laboratory measurements. In addition, an example simulation of a potential application of ferro fluids for barrier verification presented along with a brief discussion of the length scale over which ferro fluid flow can be practically induced [10]. Ovidiu Rotariu and Norval J.C. Strachan developed computer simulations to evaluate the focusing of MPs of nano - and micrometer size within tumors situated deep inside the human body. The results showed that for large external magnets, magnetite particles of diameter 1mm can be captured with high efficiency within the tumor capillaries and arterioles [11]. E. E. Tzirtzilakis proposed a mathematical model of BFD. He studied an application where the 3D blood flow in a rectangular duct under the action of a magnetic field numerically studied. Results showed generally, the magnetic field reduces the flow rate [12]. E. E. Tzirtzilakis in another work investigated the BFD flow in a channel with symmetric stenosis numerically. The results concerning the velocity field indicate that the symmetry of the flow downstream the stenosis breaks and vortex close to the magnetic field source is enlarged [6]. Misael O. Avile´s et al. examined the feasibility of using a ferromagnetic wire implanted under the skin next to the carotid artery to assist in the collection of magnetic drug carrier particles (MDCPs) using an external magnetic field theoretically [13]. Ranjan Ganguly et al. visualized and characterized the magnetically induced accumulation of ferro fluid at a target location and its subsequent dispersion in a steady host fluid flow [14]. Saud A. Khashan and Yousef Haik investigated the isothermal bio-

magnetic fluid downstream an eccentric stenotic orifice numerically using a finite volume method. They founded out that based on the location of the magnetic field, the reattachment point downstream the stenotic orifice changes. They understudied also that the shear stress will be affected based on the magnetic field location [15]. J.C. Misra and G.C. Shit examined various aspects of blood flow in different segments of the circulatory system in a situation where the system has been subjected to an external magnetic field. Their study showed that the instantaneous flow characteristics are significantly affected by the magnetic parameter and unsteadiness parameter as well as by the radius phase angle [16]. Usha K. Veeramachaneni and R. Lloyd Carroll presented a model for predicting the motion of magnetizable particles in a gradient magnetic field, considering the effects of fluidic forces on particles in the micro system; they observed that the particle magnetization dependent on the applied magnetic field and the value of the magnetization should increase as the particle moves to higher magnetic field [17]. I. Hoke et al. using Comsol software, determined the optimal geometry that assures the necessary magnetic field properties to let magnetic nanoparticles overcome the blood brain barrier of a laboratory mouse [18]. R. Probst et al. considered an initial ferro fluid control problem: the precise manipulation of a single drop of ferro fluid by four external electromagnets; they sensed the location of the droplet by a camera and imaging software and then correctly actuated the electromagnets at each time to move it from where it is to closer to where it should be [19]. Sachin Shaw and P.V.S.N. Murthy observed the significant effect of the glycocalyx layer on the magnetic targeting of the carrier particle in impermeable micro vessel with two-phase fluid model [20]. Pengtao Yue et al. studied the motion of spherical superparamagnetic clusters of roughly 10^{-7} m carried in a flow through a channel of radius 10^{-4} m, guided by a magnetic field [21]. Alexanderu Mihail Morega et al. presented the mathematical model and numerical simulations results for complex magnetic field arterial blood flow-structural coupled models specific to magnetic drug targeting (MDT). In their approach, the computational domains are generated using image-based reconstruction techniques, providing more realistic description [22]. A. Nacev et al. carried out a detailed analysis to better understand and quantify the behavior of magnetizable particles in-vivo. They found that there are three types of behaviors (velocity dominated, magnetic dominated, and boundary layer formation) uniquely identified by three essential non-dimensional numbers (the magnetic-Richardson, mass Péclet, and Renkin numbers) [5]. Arash Komaee and Benjamin Shapiro considerd the design, implementation, and verification of a class of feedback control policies for steering a magnetic particle (or a drop of a ferromagnetic fluid) along arbitrary trajectories in a controlled magnetic field [23]. A. Sarwar et al. increased magnetic forces at a depth. To make the problem concrete, they chose to optimize force at 10cm for Halbach designs with a total volume of 2000 cm^3, 36elements, and each element having a remanence of <1T. Methods based on semi-definite quadratic programming have been presented to optimize Halbach arrays to maximize push and pull forces on magnetic particles at depth [24]. Erica M. Cherry and John K. Eaton presented a model for magnetic particle motion in blood and use it to optimize injection conditions for particles that will be steered through the bloodstream. The overall objective of this study was to examine the dynamics of externally-forced magnetic particles by simulating their flow through a straight artery section, accounting for complex fluid properties, all forces on the particles, and dispersion [1]. Erica M. Cherry and John K. Eaton in another work developed a model for the motion of magnetic particle-laden blood flow that accounts for all non-trivial forces. They explored the parameter space of MDT, determining how parameters such as particle cluster concentration and applied magnetic field gradient affect the success with which magnetic particles can be controlled [2]. Helene Rahn et al. presented a method which enables a 3-dimensional quantitative analysis of the nanoparticle content. This method was a calibration procedure which has been applied to a polychromatic X-ray micro computed tomography instrument [25]. Türk et al. investigated numerically the effect of irregular stenosis occurring through the channel for the two-dimensional bio magnetic fluid flow under the effect of an external magnetic field. They also investigated the link between the variations of the flow in channels with constrictions, and the location and the intensity of the externally applied magnetic field. The finite element method employed to solve the problem in irregularly stenosed channels for the first time [26].

In most of investigations that mentioned above, the emphasis has been more on dynamic of ferro carrier concept but there is no extensive research of

how can be produce effective magnetic field to affect on magnetic drug carrier in blood circulation. In our study has been tried to understand how can be generate operative and effective magnetic field by changing in geometry of magnet. We use the finite element method (FEM) for 2D numerical simulation of magnet with different tip shapes. Then we evaluate the external static magnetic field and its hemodynamic effects on blood flow containing a magnetic carrier substance as a bio magnetic fluid. FEM is a well-developed and widely used reliable numerical approach handling complicated geometries which usually achieve high accuracy with a course mesh compared with finite difference Method (FDM) and finite volume method (FVM) [26].

2. Problem Statement (Material and Method)

In order to bring the magnetic Nano-carriers containing pharmacological agents to the specific location in the body via cardiovascular system, we should take appropriate action to manipulate and maintaining against vascular blood flow. In this direction, some important points should be taken into account:

- The concentration region corresponding to the magnetic carriers is better to be narrowing as much as possible.
- The external magnetic field should be able to manipulate the magnetic carries to this region of action.
- The external magnetic field should be able to maintain the magnetic carriers in the concentration region.

Therefore, the first step in manipulation of magnetic carries is to create an appropriate magnetic field. One of the most important parameters in creating such magnetic field is geometrical aspect of the magnet core which is the main subject of the current work.

In this research, we study a model consisting of three parts including a permanent magnet, delimiter tissue, and a blood vessel containing blood flow and ferro magnetic fluid. This arrangement is shown in Figure 1. To investigate the influence of magnet tip shape on magnetic field distribution and its effects on fluid hemodynamics, we consider five models; one of them is considered as the reference model and four states are the comparison models. The dimensions of the reference model are according to Ref. [27]. Figure 2 shows both reference and comparison models with their dimensions. Ovidiu Rotariu and Norval J.C. Strachan using computer simulation found that cylindrical magnet and magnetic circuit with parabolic shaped confocal poles (MCPSCP) produces magnetic fields with higher magnitudes ($>10^5$ Am^{-1}) at distances > 5 cm inside the body which are essential for the efficient targeting of MPs; the needle magnet facilitates a high local concentration of magnetic field lines that is useful for accurate magnetic targeting in small areas [11]. Although at this pointed research, the relationship between magnet shape and magnetic field distribution was studied, the effects of magnetic field on fluid hemodynamics have not been reposted yet.

3. Governing Equations and Numerical Model

To study the interaction of static magnetic field on blood flow containing magnetic fluid, two general categories of physical equations such as Maxwell's equations and the Navier-Stokes equations are used.

3.1. *Equations of magnetic section*

The main Maxwell-Ampere's law equation to express magnetic field and Gauss' law equation to state magnetic flux density, respectively, as follows [5, 22, 27]:

$$\nabla \times \boldsymbol{H} = \boldsymbol{J} \quad (1)$$

$$\nabla \cdot \boldsymbol{B} = 0 \quad (2)$$

where $\boldsymbol{H}$ is the magnetic field vector (A/m), $\boldsymbol{J}$ is the current density vector (A/m^2), $\boldsymbol{B}$ is the magnetic flux density vector (T). Because the source of the magnetic field is a permanent magnet, static magnetic field, then in equation (1), the value of the electric current density vector is zero vector and the equation changes as follows [22]:

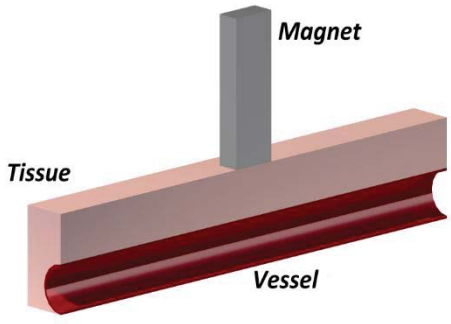
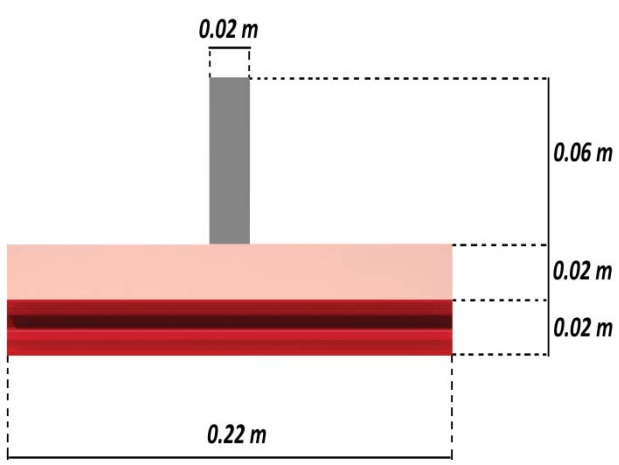
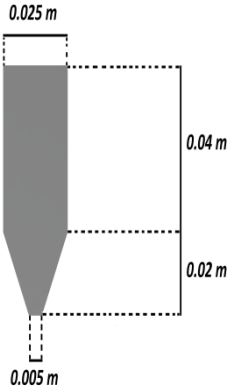


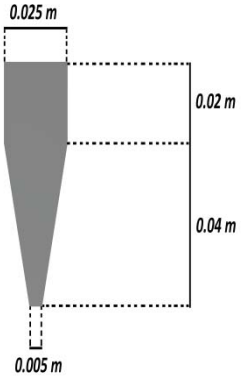
Figure 1: Arrangement of model geometry



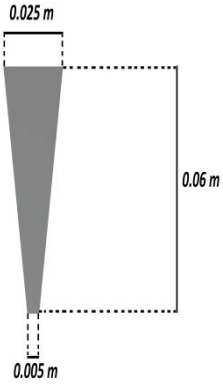
Magnet 1



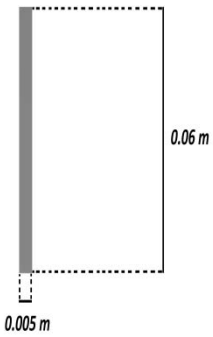
Magnet 2



Magnet 3



Magnet 4



Magnet 5

Figure 2: reference and comparison models; Magnet 1 is used as reference model; Magnets 2, 3, 4, and 5 are used for comparison purposes

$$\nabla \times \boldsymbol{H} = 0 \tag{3}$$

To express the relationship between the magnetic field and the magnetic field density and adapt this relation to different parts of the model, the following equations will be obtained as [22, 27]:

$$\boldsymbol{B} = \mu_0\mu_r\boldsymbol{H} + \boldsymbol{B}_{rem} \quad (4)$$

$$\boldsymbol{B} = \mu_0[\boldsymbol{H} + \boldsymbol{M}(\boldsymbol{H})] \quad (5)$$

$$\boldsymbol{B} = \mu_0\boldsymbol{H} \quad (6)$$

Here equation (4) is related to permanent magnet, equation (5) corresponds to blood flow containing magnetic fluid, and equation (6) is associated with tissue and air environments. In equations (4) to (6), μ_0 is the magnetic permeability of vacuum ($4\pi\times 10^{-7}$, N/A^2), μ_r is the relative magnetic permeability of the permanent magnet, $\boldsymbol{B}_{rem}$ is the remanente magnetic flux density (A/m) and $\boldsymbol{M}$ is the magnetization vector in the blood stream as a function of $\boldsymbol{H}$. In addition to the above equations, two other equations involving the magnetic vector potential, $\boldsymbol{A}$, and its relation to the magnetic flux density vector is defined as follows [22, 27]:

$$\nabla\cdot\boldsymbol{A} = 0 \quad (7)$$

$$\nabla\times\boldsymbol{A} = \boldsymbol{B} \quad (8)$$

According to Equations (2) to (8), the final equation is presented as follows [27]:

$$\nabla\times\left(\frac{1}{\mu_0}\nabla\times\boldsymbol{A} - \boldsymbol{M}\right) = 0 \quad (9)$$

In Equation (9), several assumptions are very important: Firstly, problem is assumed two dimensional; Secondly, the magnetic vector potential is considered as a non-zero vector component with the form of $\boldsymbol{A}$ = (0, 0,A_z); thirdly, the value of μ_r is assumed 1, and lastly, $\boldsymbol{M}$ as the magnetization vector is defined with form of $\boldsymbol{M}$ = (M_x, M_y, 0) whose components will be as follows [10, 27]:

$$\begin{cases} M_x = \alpha\tan^{-1}\left(\dfrac{\beta}{\mu_0}\dfrac{\partial A_z}{\partial y}\right) \\ M_y = \alpha\tan^{-1}\left(\dfrac{\beta}{\mu_0}\dfrac{\partial A_z}{\partial x}\right) \end{cases} \quad (10)$$

where both α (A/m) and β (m/A) are material parameters. Equation (10) can be defined in linearized form as the following too [27]:

$$\begin{cases} M_x = \dfrac{X}{\mu_0}\dfrac{\partial A_z}{\partial y} \\ M_y = \dfrac{-X}{\mu_0}\dfrac{\partial A_z}{\partial x} \end{cases} \quad (11)$$

where $\boldsymbol{X} = \alpha\beta$ is the magnetic susceptibility.

3.2. *Equations of fluidic section*

The time-dependent Navier-Stokes equation, which represents balance of mass and momentum for a non-compressible fluid Together with the continuity equation are two governing equations of fluid flow physics, which is defined as follows [12, 22, 27]:

$$\rho\frac{\partial\boldsymbol{U}}{\partial t} - \nabla\cdot\eta(\nabla\boldsymbol{U} + (\nabla\boldsymbol{U})^T) + \rho\boldsymbol{U}\cdot\nabla\boldsymbol{U} + \nabla\boldsymbol{P} = \boldsymbol{F} \quad (12)$$

$$\nabla\cdot\boldsymbol{U} = 0 \quad (13)$$

where ρ is the fluid density (Kg/m^3), $\boldsymbol{U}$ is the velocity (m/s), η is the dynamic viscosity (Kg/m.s), P is the pressure (N/m^2) and $\boldsymbol{F}$ is a volume force (N/m^3). According to equations (4) to (8) and equation (11), the force acted on blood flow containing magnetic fluid is expressed as follows [27]:

$$\begin{cases} F_x = \dfrac{X}{{\mu_0\mu_r}^2}\left(\dfrac{\partial A_z}{\partial x}\dfrac{\partial^2 A_z}{\partial x^2} + \dfrac{\partial A_z}{\partial y}\dfrac{\partial^2 A_z}{\partial x\partial y}\right) \\ F_y = \dfrac{X}{{\mu_0\mu_r}^2}\left(\dfrac{\partial A_z}{\partial x}\dfrac{\partial^2 A_z}{\partial x\partial y} + \dfrac{\partial A_z}{\partial y}\dfrac{\partial^2 A_z}{\partial y^2}\right) \end{cases} \quad (14)$$

In Equation (14), it is considered several assumptions: Firstly, because the problem is two dimensional, volumetric force consists of two components, $\boldsymbol{F}$ = (F_x, F_y, 0); secondly, here only the external magnetic field is considered and other forces, such as the interaction of magnetic nanoparticles composed of magnetic fluid, the force of gravity and the force due to diffusion are neglected [5, 27].

3.3. *Initial and boundary conditions*

To solve the governing equations, COMSOL Multiphysics 4.3b software is used. No-slip conditions are assumed for the vessel walls. Flow is considered laminar, incompressible and pulsatile. Because blood flow is a complex non-Newtonian, blood viscosity is assumed with form of power law equation, which is presented as follow [28]:

$$\mu = m(\dot{\gamma})^{n-1} \quad (15)$$

where m and n are Power law model parameters, $\dot{\gamma}$ is the shear rate (1/s) and μ is the blood viscosity (Pa.s). The inlet velocity is defined as following [27]:

$$\boldsymbol{V}_{in} = 2\boldsymbol{U}_{max}s(1-s)(\sin(\omega t) + \sqrt{\sin(\omega t)^2}) \quad (16)$$

where the boundary segment length parameter s is between 0 and 1, $\boldsymbol{U_{max}}$ is the maximum velocity of blood flow. Figure 3 shows the diagram of the main part of inlet velocity function ($\boldsymbol{V_{in}} = \boldsymbol{U_{max}}(\sin(\omega t) + \sqrt{\sin(\omega t)^2})$) for and interval of 4 seconds.

Outlet pressure condition is defined to be zero. Other considered assumptions and parameters for modeling and simulation are listed in Table 1.

4. Results and Discussion

To verify the used method in this study, present numerical procedure is compared to the result of Ref. [27]. The graphical comparison can be seen in Figure 4. Also, comparison between percentage of absolute differences of maximum magnetic flux density norm and maximum velocity magnitude have been shown in Table 2. As can be seen in Table 2, the percentage of absolute difference of the results is reasonable and acceptable. According to Table 2, there is a negligible difference between result of present work and Ref. [27] in figures A and C. It is because of the difference in the definition of viscosity. In Ref. [27] was not pointed to Newtonain or non-Newtonan blood then we assume Power law as a non-Newtonian, blood viscosity model. The results show that how geometry of the core tip of the magnets affects the distribution of the corresponding magnetic field and therefore changes the hemodynamics of the blood flow.

Figure 5 shows the distribution of the static magnetic flux density generated by five permanent magnets with different tip shapes. It is apparent that the concentration of the magnetic flux density increases from Figure 5-a (magnet 1) to Figure 5-e (magnet 5). In other words, the spread and dispersion of the magnetic flux density decreases with

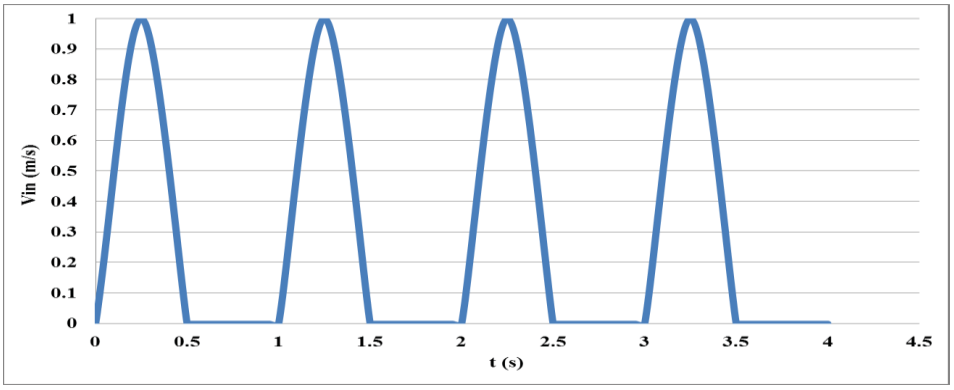


Figure 3: Plot of the main part of inlet velocity function

Table 1: Assumptions and parameters used in the simulation

Parameters or Assumptions	Value	Unit	Explanation
U_{max}	50	cm/s	maximum velocity of blood flow
ω	2π	Rad/s	Pulse angular velocity*
f	1	1/s	Heart-beat rate
ρ	1060	Kg/m^3	Density of blood
m	0.035	-	Power law model parameter
n	0.6	-	Power law model parameter
B_{rem}	1	T	magnet remanent flux density
α	10^{-4}	A/m	Ferrofluid magnetization-curve parameter
β	3.10^{-5}	m/A	Ferrofluid magnetization-curve parameter
$\mu_{r,\ mag}$	1	-	Magnet Relative permeability
$\mu_{r,\ ferro\text{-}fluid}$	1.3	-	Ferrofluid Relative permeability

*Pulse angular velocity: $\omega = 2\pi . f$

Results of Ref. [27] Results of Present study

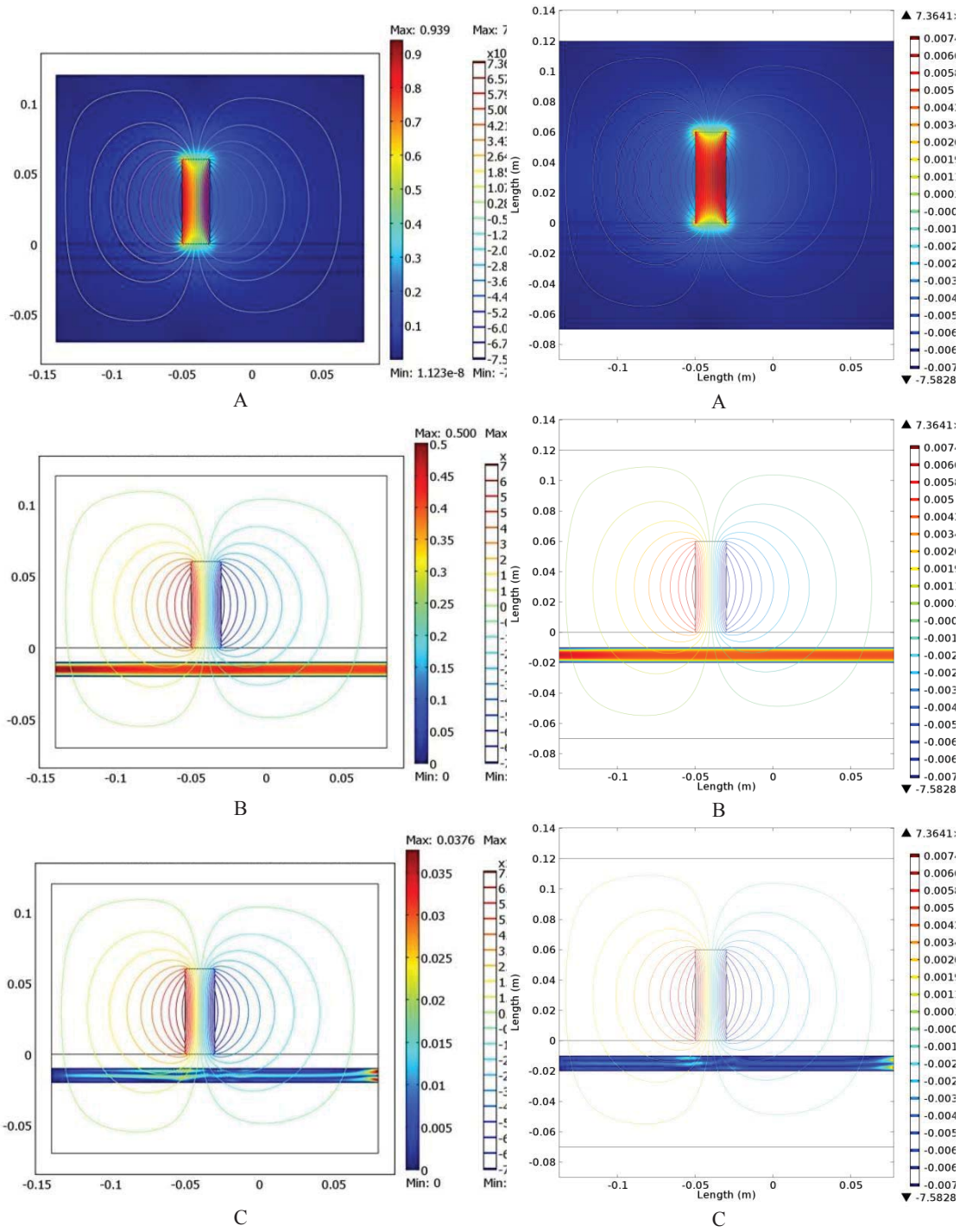


Figure 4: Graphical comparison of present producer with results of Ref. [27]; (A) At time 2 sec, (B) At time 0.25 sec, (C) At time 1 sec. In all cases, Surface is for Magnetic flux density norm (T), and Contours are Magnetic vector potential at Z component (Wb/m)

Table 2: comparison in absolute differences of maximum magnetic flux density norm and maximum velocity magnitude in verification trial

Maximum of:	Figure	Results of Ref. [27]	Results of Present study	absolute results differences
Magnetic flux density norm	A	0.939 T	0.890 T	5.21 %
Velocity magnitude	B	0.5 m/s	0.5 m/s	0 %
Velocity magnitude	C	0.037 m/s	0.035 m/s	5.4 %

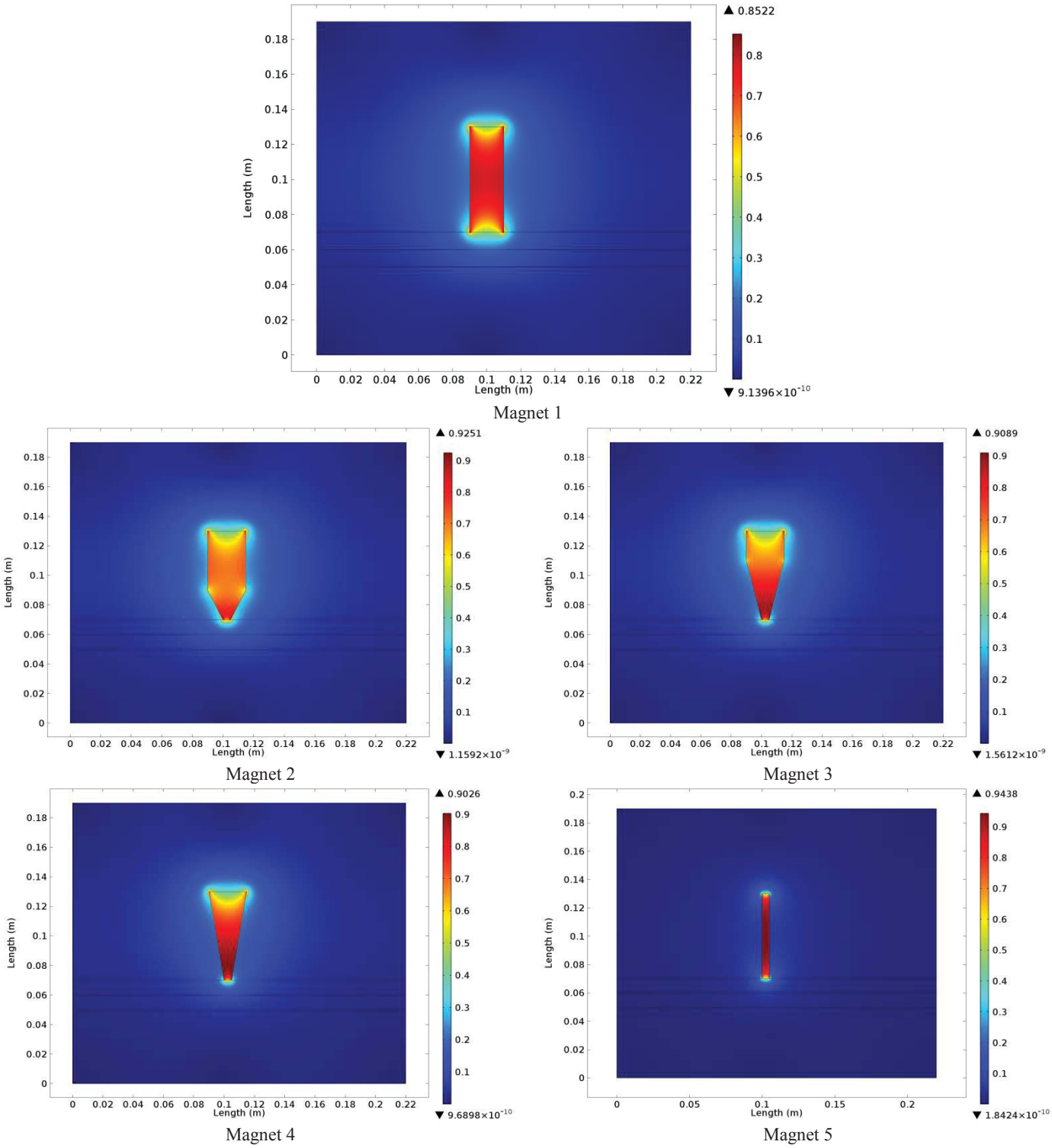


Figure 5: Magnetic flux density around the magnet (T)

narrowing the tip shape of the magnets. The maximum value of the magnetic flux density in the body of magnet 5 is greater than those corresponding to magnets 1 to 4. These maximum values for magnet 1 to magnet 5 are 0.8522 T, 0.9251 T, 0.9089 T, 0.9026 T and 0.9438 T respectively. Therefore, the concentration of magnetic flux density as well as its

maximum value in the body of magnet 5 increases in comparison to those of other magnets (magnets 1 to 4).

Figure 6 depicts the magnetic flux density changes along the centerline of the vessel for each magnet. As it is seen, the maximum produced magnetic flux density corresponds to the magnet 1 while the minimum value corresponds to the magnet 5. In fact, the trend of changes in this value is descending from magnet 1 to magnet 5; these values are 0.131 T, 0.110 T, 0.087 T, 0.073 T, and 0.038 T for magnets 1 to 5 respectively. Considering the diagrams of both Figures 5 and 6, it is concluded that although the maximum value of magnetic flux density produced by magnet 5 along the centerline of the vessel is less than other cases, the distribution of the magnetic flux density just below the tip of magnet 5 is much concentrated in comparison to other cases. This configuration is appropriate to maintain the magnetic carrier which is an important outcome of this study.

Figure 7 shows the distribution of the velocity field of the blood flow. It is transparent that the velocity distribution is symmetric in the absence of magnetic field. However, the velocity distribution is changed to an asymmetric regime by applying external magnetic field. It is notable that some vortices have appeared due to external magnetic field; see Figure 7. Although the velocity of blood in case 5 is less than other cases, the rate of vortex construction in case 5 is more than other cases. When a vortex is composed, it means that the velocity of blood flow has decreased and therefore the magnetic carrier has enough time to transfer the pharmacological agent. However, special consideration must be taken into account to avoid probable side effects on blood flow.

Figure 8 depicts the diagram of blood velocity changes along the center line of the vessel. In Figure 8-a, these changed has been plotted in presence and absence of external magnetic field. It could be observed that the velocity changes dramatically decreases upon exerting the external magnetic field. Figure 8-b, indicates the changes of blood flow velocity along the center line of the vessel in presence of external magnetic field. As could be seen in this figure, the exerting of external magnetic field results in constructing peaks and valleys in the velocity diagram just below the place of magnet tip. These peaks and valleys in velocity diagram are symptoms of existing some vortices in flow. Moreover, it is obvious that the changes in peaks and valleys are directly depending on the tips of magnets and their corresponding geometry.

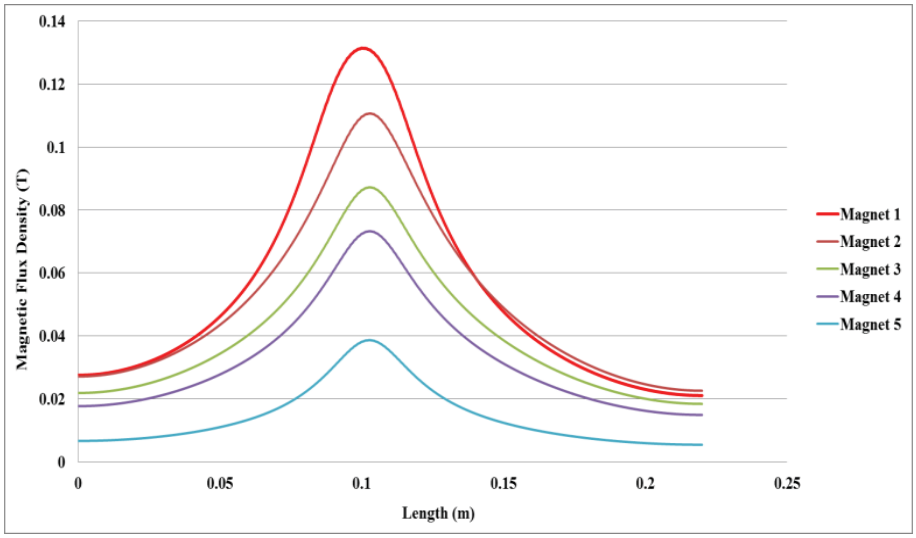


Figure 6: Magnetic flux density on the centerline of the vessel (T)

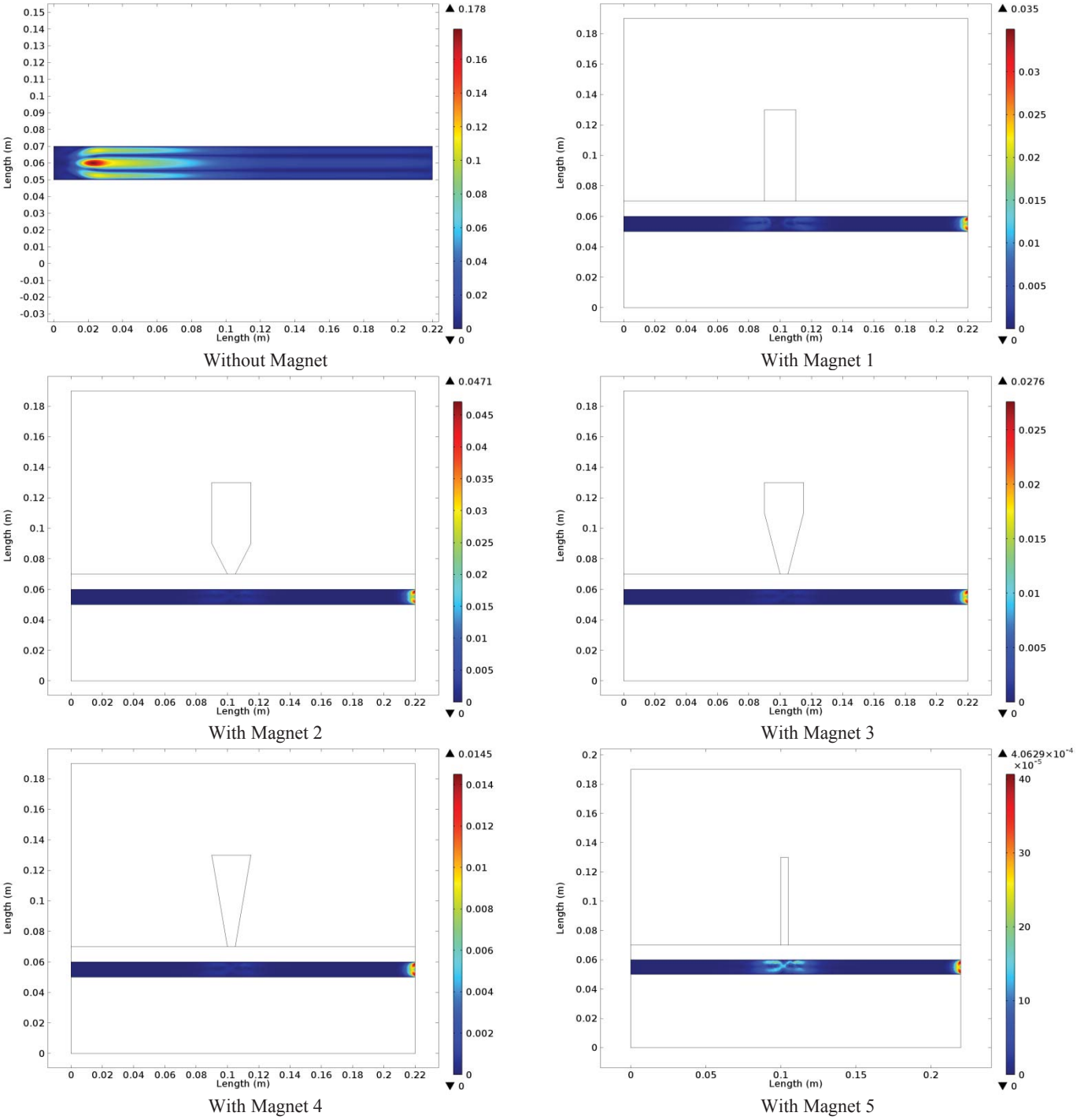


Figure 7: fluid velocity distribution (m/s)

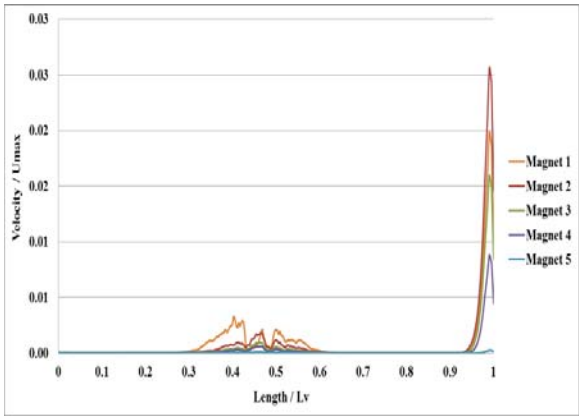
The diagram of flow pressure changes along the center line of the vessel in presence and absence of external magnetic field is demonstrated in the Figure 9. It is worthy to note that by applying external magnetic field, the pressure substantially increases just below the place of magnet tip. With a little investigation in this figure, it is perceived that the diagram of the blood flow pressure in this case much more resembles that of the case without external mantic field. Moreover the maximum value of the diagram for the magnet 5 is less than that of other cases. A consequence of this discussion is that the effect of tip geometry of magnet 5 on blood flow more natural than other cases.

Figure 10 depicts the diagram of wall shear stress for the upper side of the vessel (upper line in 2D view). It is observed that in the case of exerting external magnetic field, the wall shear stress just below the place of magnet tip is less than that of magnetic field free case. According to the previous discussion, the blood flow velocity dramatically decreases upon exerting the external magnetic field; this phenomenon could be the main reason of decreasing the wall shear stress in the presence of external magnetic field. It is worthy to note that in the case of exerting the external magnetic field, the wall shear stress described above experiences a vibrating behavior, but it has a nearly steady behavior in the magnetic field free case.

In this study we tried to understand how can be generate operative and effective magnetic field. Our emphasis was more on geometry of magnet concept but in another work for studding extensively, It seems appropriate to investigate of more parameter and condition like: Reynolds and Womersley number effect, different viscosity model and etc.

5. Conclusion

The purpose of this study was to evaluate the static magnetic field interaction with hemodynamic properties of blood flow containing a magnetic carrier substance as a bio magnetic fluid. In this study we used the finite element method (FEM) for 2D numerical simulation of magnets with different tip shapes and evaluating of external static magnetic field and its hemodynamic effects on blood flow density of the magnetic flux intensity at the tip of the magnet, but also concentrates the distribution of magnetic flux density around the magnet. Moreover, the results showed that existence of static magnetic field on the blood flow containing magnetic carrier containing a magnetic carrier substance as a bio magnetic fluid. It was found out that in the contact region between the magnet and the tissue, usage of magnets with sharper tips, not only increase the



a

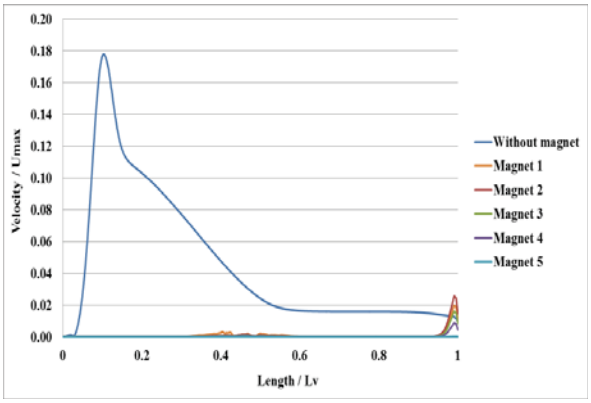


Figure 8: Changes in velocity on the centerline of the vessel; a: effect of different magnets; b: Comparison of the magnet-less case and with-magnet cases

b

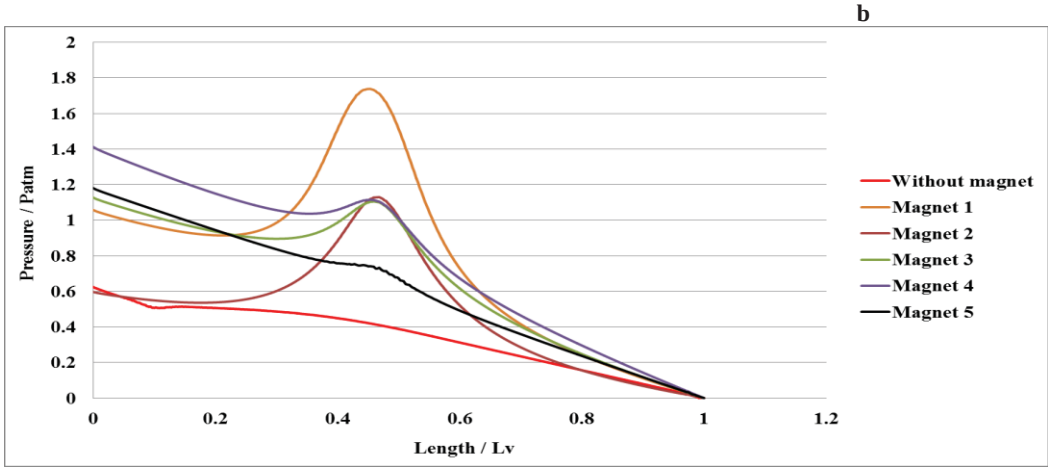
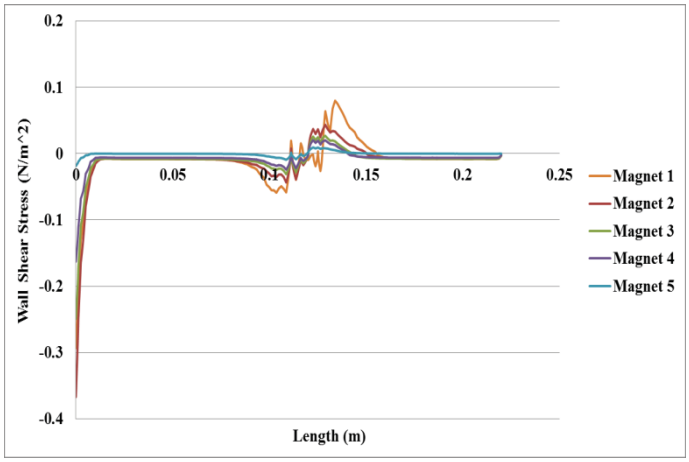
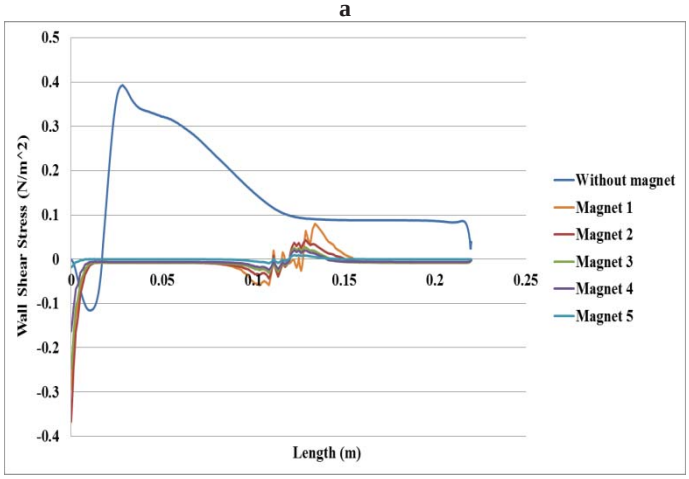


Figure 9: Changes in Pressure on the centerline of the vessel



a



b

Figure 10: distribution of shear stress on the upper vessel wall (N/m^2); a: effect of different magnets; b: Comparison of the magnet-less case and with-magnet cases

substances and also changing the shape of the magnet tips would result in considerable changes in the distribution of fluid flow velocity. One of the most important results is that the mutating changes in velocity and pressure of the flow in the centerline of blood vessel have decreased with concentrating of magnetic flux density distribution. Finally, the wall shear stress values have also decreased, just under the position of the magnet tip on the tissue. This effect becomes more intensive when the tip of magnet become sharper.

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