Optimization of the Performance of a Biomedical Micro-Pump

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ABSTRACT
This paper discusses the optimization of a micro-pump composed by deformable polymeric membrane in contact with reservoir and examines the effect of the materials property at the performance and the functionality of the system. The Neo Hookean hyperelastic material model is used to simulate the deformation of polydimethylsiloxane (PDMS) elastomer and compared with Poly methyl methacrylate (PMMA). The results of simulation by finite element are presented and discussed. In second steps we study the power to inject by active membrane a Newtonian and a non Newtonian fluid in microcanalization, the power law is used to model the variation of the blood viscosity and precise the maximum value of flow rate at minimum applied pressure and control the fluid transportation. This type of micropump appears to be suitable for biomedical applications and demonstrate the versatile use of active membrane as moving parts to inject the fluids us blood or glucose.

1. INTRODUCTION

The MEMS technology allow to improve the performance of the systems which use it, to increase their speed, their reliability and their potential of integration, but also to reduce their energy consumption and their dimension, this phenomenon of miniaturization allows to multiply the features integrated into the systems and to answer new needs in the biomedical industry such as the development of micro-pumps for various applications, and making possible parallel processing that leads to a production in large quantities at low cost.

The development of the biomedical micro electromechanical system technology require a better understanding of fluid flows in micrometer scale, which gave birth to a new discipline called micro-fluidics.

The use of polymeric materials such as PDMS(Polydimethylsiloxane) and PMMA(Poly methyl methacrylate) for the manufacture of microfluidic devices is a promising way made it possible to fabricate small size and high performance biomedical devices. The polymeric materials having particular physical and chemical properties are good candidates for the development of new generation actuator. With greater complexity, shorter manufacturing, lower cost. They are much cheaper than silicon. Currently the technology related to PDMS and PMMA is the most widespread in the field of microfluidics.

The PDMS finds its use in a wide range of microfluidic applications due to its flexibility and low cost [1]. The PMMA attracts growing interest in microfluidics research community

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due to its low cost, high transparency, good mechanical and chemical properties [2].

Modeling and simulation of fluid-structure system are the key stages of the designation and optimization of microfluidic applications. This work is related to the description of mechanical behavior of the elastomer and the displacement of the Newtonian and non-Newtonian fluids generated by the deformed membrane which are investigated by the coupling between the mechanical model and the fluid model.

2. STRUCTURE FLUID MODELLING

The system is composed of a circular polymeric membrane of 400µm in radius and a thickness of 20 µm (Figure1). The materials used for the membrane are PDMS Silgel 612 [3] and PMMA reinforced shock by polybutadiene elastomer particles [4]. The membrane is contacted with a reservoir of 200µm of height. The ejection of the liquid is done by applying pressure on the membrane thus causing its deformation. The displacement of the membrane can drain liquid from the reservoir through a micocanalization. The cylindrical geometry canalization have a length of 30mm and width of 40µm (Figure 2).

Before simulate the coupling process, it is important to simulate the behavior of the deformable membrane. This step is important in order to predict fluid flow to drain in the microcanalization. On the other hand the test for the entire system shall provide a maximum amount of fluid at minimum pressure applied on the membrane, while ensuring the rigidity of the system.

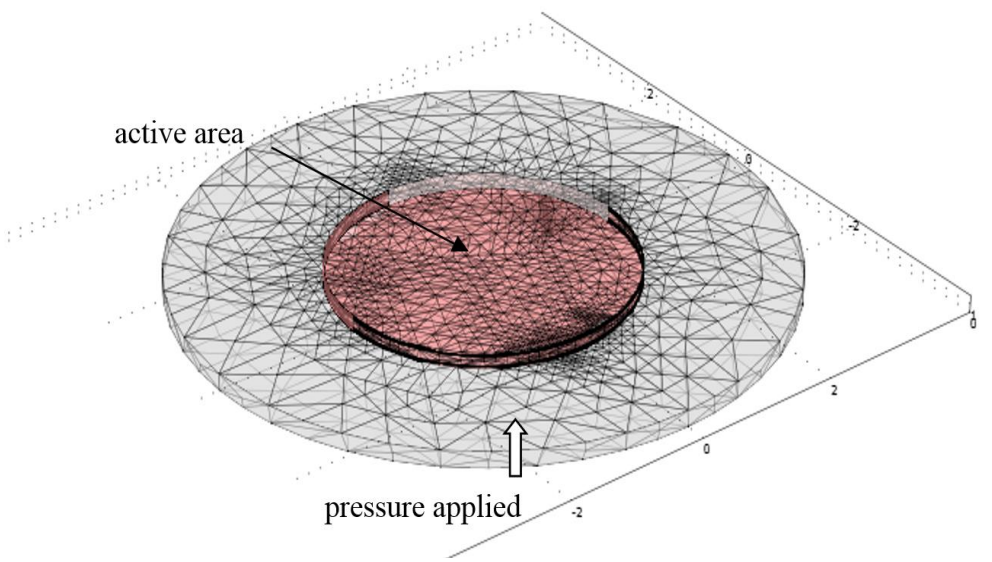

Figure 1: Three-dimensional mesh of the elastomeric membrane

2.1. Membrane Behavior

Our model is constituted by the equation of the mechanical behavior of the membrane as described by the model to the large deformation [5, 6].

The strain tensor at large strain contains a quadratic term, meaning that analysis in large deformation is non-linear:

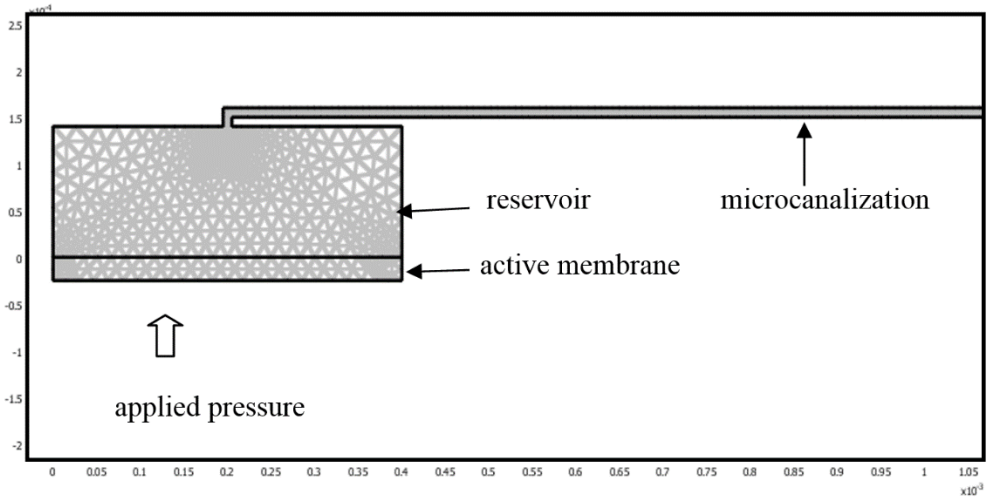

Figure 2: Two-dimensional mesh structure-fluid system

$$\bar{\bar{E}} = \frac{1}{2}\left(\left(\frac{\partial \vec{u}}{\partial \vec{X}}\right)^T + \left(\frac{\partial \vec{u}}{\partial \vec{X}}\right) + \left(\frac{\partial \vec{u}}{\partial \vec{X}}\right)^T \cdot \left(\frac{\partial \vec{u}}{\partial \vec{X}}\right)\right) \tag{1}$$

For a hyperelastic material, the stresses S tensor is calculated by means of an energy function W.

$$\bar{\bar{S}} = 2\sum_{i=1}^{2} \frac{\partial W}{\partial I_i} \frac{\partial I_i}{\partial \bar{\bar{C}}} \tag{2}$$

The relationship between the deformation and stress is given by the following equation:

$$\bar{\bar{\sigma}} = 2J^{-1}\bar{\bar{F}}\left[\sum_{i=1}^{3} \frac{\partial w}{\partial I_i} \frac{\partial I_i}{\partial \bar{\bar{C}}}\right]\bar{\bar{F}}^T \tag{3}$$

In our calculation we use the New Hookeen model of strain density energy [7]:

$$W = \frac{1}{2}G(I_1 - 3) + \frac{1}{2}K(J-3)^2 \tag{4}$$

When G is the shear modulus and K is the bulk modulus of the hyper elastic material. I_1 is the strain invariant: I_1=trC and J=det(F), F is the deformation tensor. C is Cauchy – Green tensor: $C=F^T F$.

2.2. Flow Modelling

We interest to model the flow of two fluids, the blood and glucose. The glucose follows the Newtonian model, the fluid can be considered incompressible therefore its viscosity does not change (η=cst), regardless of the strike force. The equation of fluid dynamics that describes the movement of this fluid is the Navier-Stokes equation:

$$\rho \frac{\partial \vec{v}}{\partial t} + \rho(\vec{v}.\overrightarrow{grad})\vec{v} = \rho\vec{f} - \overrightarrow{grad}p + \eta\Delta\vec{v} \quad (5)$$

With ρ is the density of the fluid and is the velocity of a fluid element, represents the set of volume forces applied to the fluid, p the pressure and η the viscosity of the fluid.

However the behavior of non-Newtonian fluids is not as predictable because their viscosity is dependent on the applied force. The viscosity of certain fluids decreases when subjected to a force, these are shear-thinning fluids (pseudoplastic) such us blood.

The blood is non Newtonian fluids namely biological fluids, they are rheologically complex due to their multicomponent nature [8]. For our numerical calculations, we consider rheological power law model to describe the variation of the blood viscosity with the shear rate, the law are written us [9]:

$$\eta = k\dot{\gamma}^{n-1} \quad (6)$$

Where k is referred as the consistency index and n is the power law exponent, $\dot{\gamma}$ is the shear strain rate. When the magnitude of $n < 1$ the fluid is shear thinning, Or where $n = 1$ corresponds to the Newtonian behavior.

2.3. Testing Mesh

In order to increase the accuracy of the calculations in sensitive areas as shown in Figure.1. We chose a non-uniform tetrahedral mesh for the membrane. This mesh allows faster convergence and reduced computational time. We were careful to use a sufficiently fine mesh to minimize the error in the modelling. To maximize the strain response of the membrane, the finite element simulation results for different mesh element values and computation times are summarized in Table 1. Note that the relative error is less than 8%. We choose the small number of elements to optimize the computing time.

Table 1: The mesh test for deformation of the PDMS membrane

Element number	**Maximum of deformation (µm)**	**computational time (s)**	**The relative error**
1481	28,13	1,015	7,36%
3754	29,22	2,156	6,73%
10611	30,89	6,469	1,02%
14214	31,21	29,20	0,00%

2.4. Boundary Conditions

For our study is close to a real system it is important to impose boundary conditions to the membrane simulation:

- We blocked the nodes in the sides along the segment in three directions the active area to free deformation
- The default initial conditions are zero initial displacement.
- A pressure is applied under the membrane, and residual stresses are neglected.

To model the fluid structure system, we decided to undertake a two-dimensional modeling to reduce the computation time. We model a cross section of the system and infinite in the third direction. The boundary conditions express:

- The fluid velocity is zero at the channel wall and all surfaces).
- On the membrane-fluid interface at the same speed.
- The pressure applied at the outlet of the pipe is zero.
- A fluid pressure is applied to the membrane.
- Pressure a function of time is applied to the membrane (equation 7):

$$p = Ae^{B/2t} \tag{7}$$

Where A=140 and B=1

3. RESULTS AND DISCUSSION

Figure 3 represent the three-dimensional deformation of the PDMS and PMMA membrane at a pressure of 30 kPa. We observe an important change in the center of the PDMS membrane under a pressure compared with the PMMA membrane.

To compare the two strain amplitudes of materials, we have plotted the curves of maximum vertical displacement as a function of the pressure and as a function of thickness for the PDMS and PMMA. The results obtained are shown in Figures 4 and 5 using the hyper elastic model and linear model. Particularly, the results show that the PDMS membrane has a large deformation compared to PMMA membrane at same conditions.

The purpose of this part concerning specifically the flow control

Fluid in the microcalization (velocity profile and flow calculation), Indeed, the results of the simulation with the blood and glucose are presented in Figures 6, 7, 8.

Figures 6, 7 and 8 respectively show the velocity field and the velocity profile in the microcanalization at 0,015 seconds. The results show that the speed is important at the entrance and in the micocanalization.

The velocity profile of the two fluids shows that we are in the presence of a laminar flow, as demonstrated in Figures 7 and 8. The value of the average speed at t = 0.0015 equal to 0.054 m / s for glucose and 0018 m / s for the blood. And the flow rate obtained equal to 468 nl / s for glucose and 155.9nl / s for blood

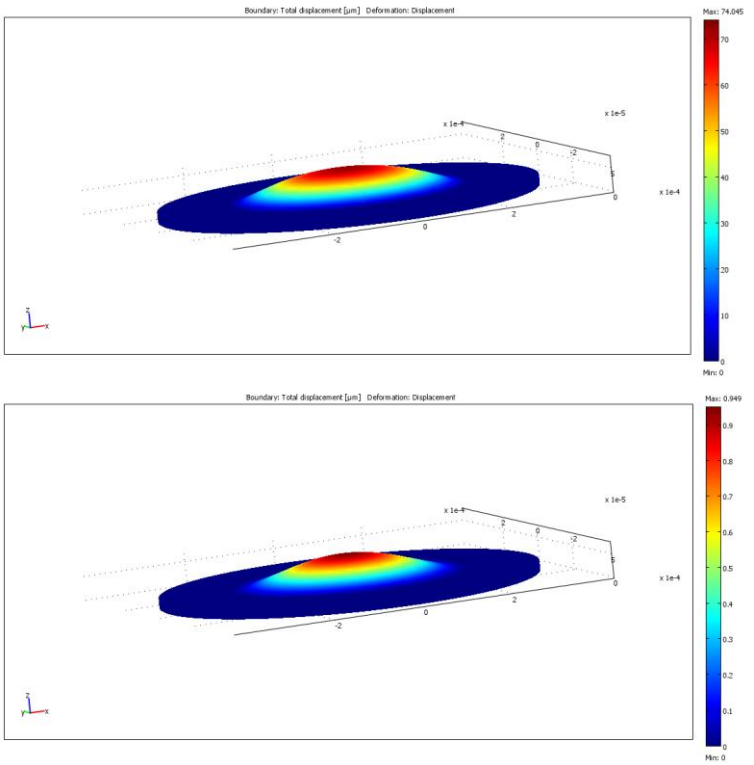

Figure 3: Deformation of PDMS and PMMA membrane induced by applied pressure of 30kpa

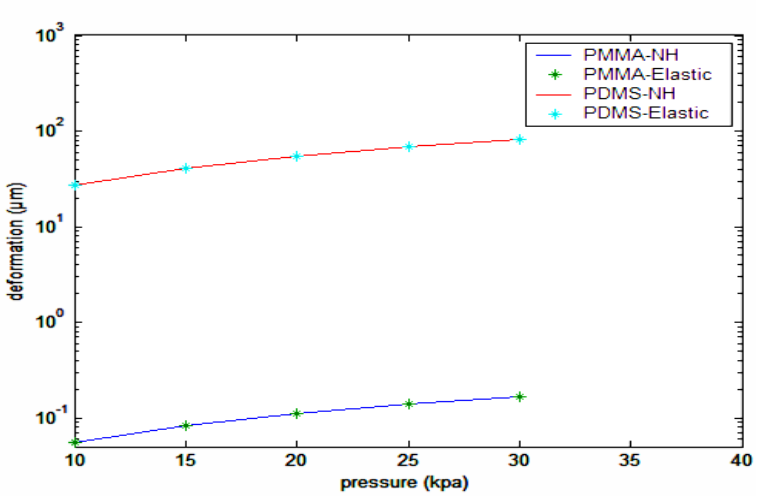

Figure 4: Maximum of Vertical Displacement on Surface of PDMS and PMMA Membrane at Different Pressure.

Indeed the Reynolds number is proportional to the characteristic dimension of the system. In microfluidics, so we still very low Reynolds numbers (Reynolds number calculated average of 2.4). This means that the flows are always laminar.

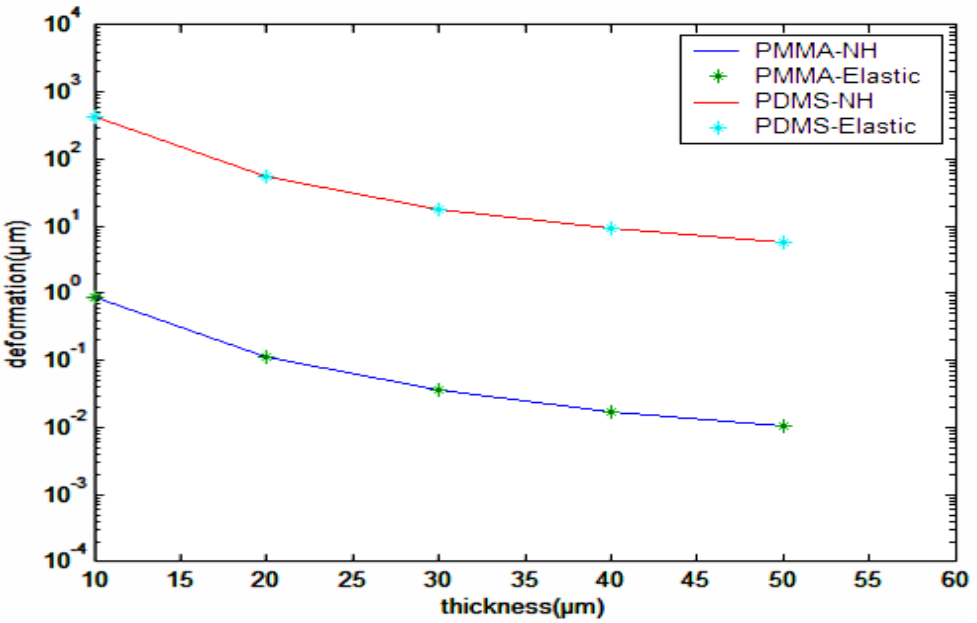

Figure 5 Maximum of Vertical Displacement on Surface of PDMS and PMMA Membrane at Different Thickness.

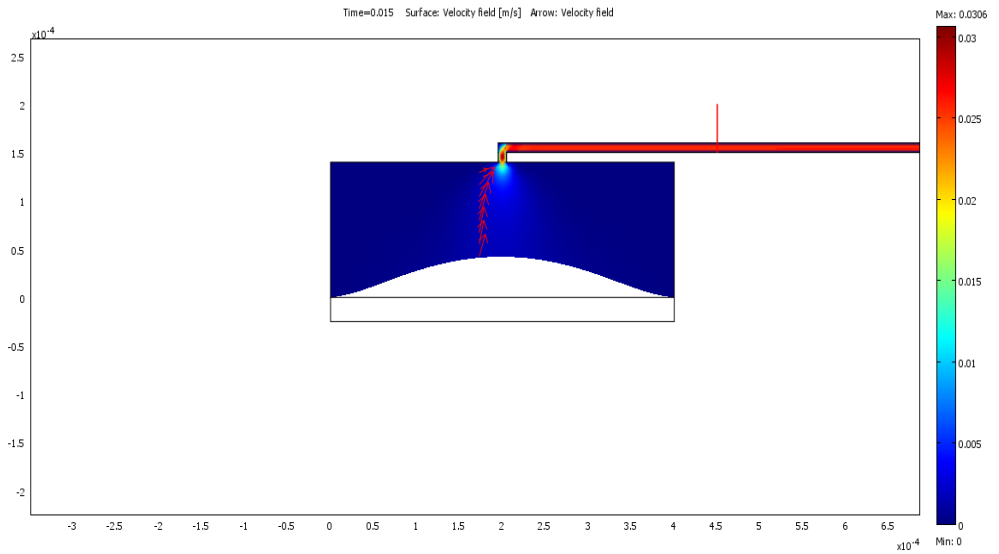

Figure 6: Velocity surface of blood at 0.015s

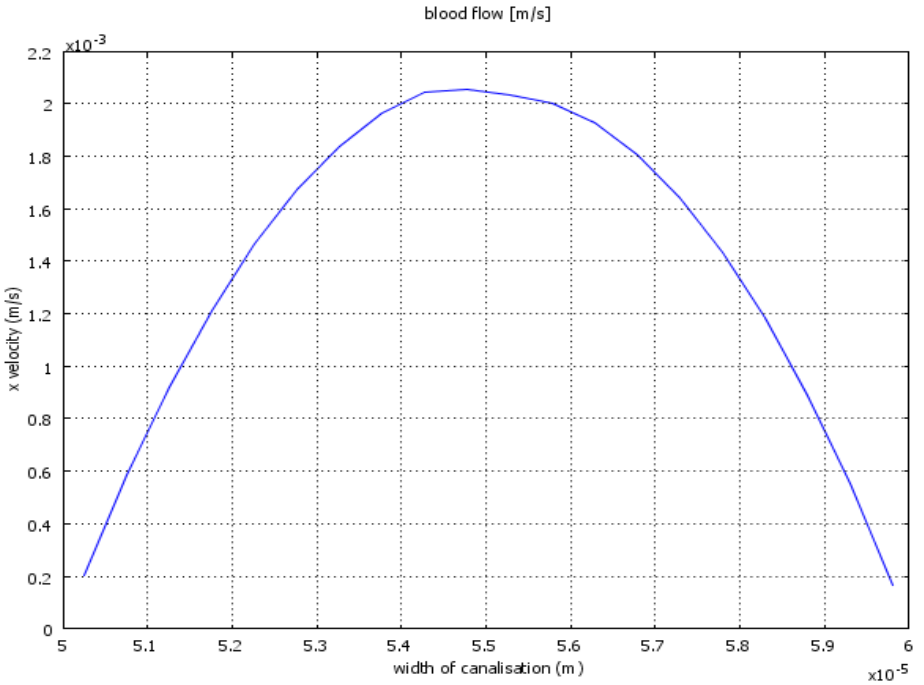

Figure 7: Velocity profile of the blood at 0.015s

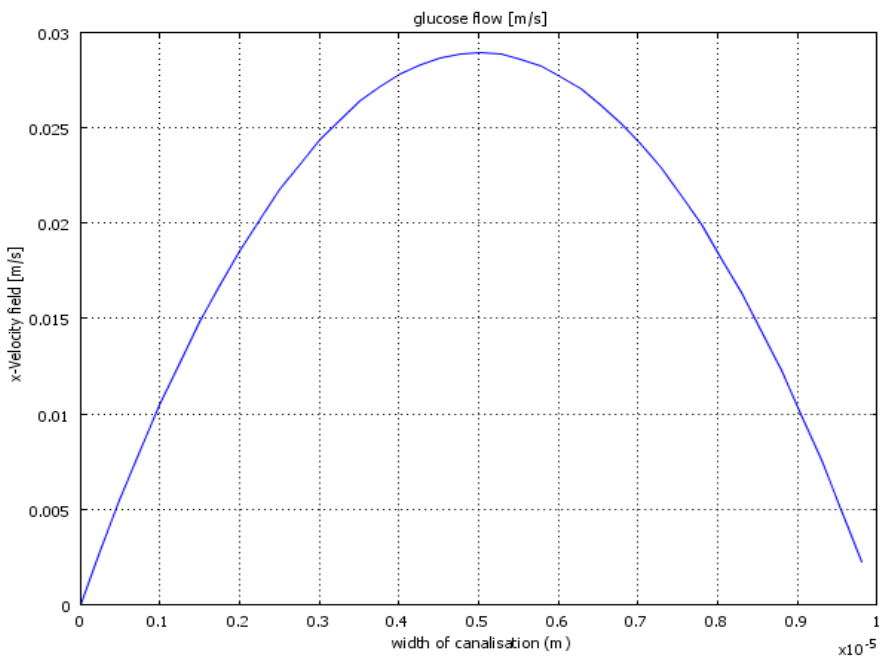

Figure 8: Velocity profile of the glucose at 0.015s

4. CONCLUSION

The controlled flow through microcanalization injected by the mean of polymeric deformable membrane has various applications such as drug delivery. At 1s we can inject 468 nl of glucose or 155.9nl of blood.

In conclusion, this type of micropump appears to be suitable for biomedical applications and demonstrate the versatile use of active membrane as moving parts to inject the fluids. The PDMS material is good candidate for micro fluidic devices because of its flexibility and its rapid prototyping and its biocompatibility compared to the PMMA material.

Despite the advantages of rapid prototyping of PDMS, this material suffers from a serious drawback in which it swells in most organic solvents [10]. This swelling of the device based on PDMS nullifies the flow in the microchannels. PMMA is an elastomer with little deformation as compared to the PDMS. This property can be used for the construction of micro-pipes when the rigidity is required.

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