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Modeling of fast-response magnetic-sensitive hydrogel for dynamic control of microfluidic flow

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Abstract

A magnetic-sensitive hydrogel-based microfluidic system is designed via a magneto-chemo-hydro-mechanical model for replicating various physiological and pathological conditions in the human body, by which the desired flow patterns can be generated in real time due to the fast-response deformation of the magnetic hydrogel. In the model, the fluid-structure interaction is characterized between the deformable magnetic hydrogel and surrounding fluid flow through the fully coupled arbitrary Lagrangian-Eulerian (ALE) method. Moreover, the physicochemical mechanisms are characterized including hydrogel magnetization, fluid diffusion, fluid flow, and hydrogel large deformation. After validation of the present model with both the finite difference and experimental results in open literatures, the transient behavior of the magnetic hydrogel is investigated, and the results show that the response time for the magnetic hydrogel is improved significantly in a uniform magnetic field compared with hydrogel without magnetic effect. Furthermore, various patterns of pulsatile flow are generated for mimicking the cell physiological microenvironment experienced by bone marrow stromal cells, and also for the pathological condition at femoral artery during diastole and systole, respectively. Therefore, the present magnetic-sensitive hydrogel-based microfluidic system via the multiphysics model may provide a relevant humanized manipulation platform to investigate cell behavior and function through microfluidic chips.

1. Introduction

Hemodynamic flow is one of the key factors to physically stimulate the cells in a microphysiological system.¹ Its respective regulation and dysregulation are highly associated with the tissue health and disease states.² For example, under normal physiological condition, the vascular endothelial cells are stimulated to release bioactive molecules that maintain the stress of the blood vessel within a certain range, and inhibit the adhesion of blood platelet on intima and the proliferation of smooth muscle cells.³ Under the condition of pathological flow pattern, hypertension activates signal pathways of smooth muscle cells and accelerates the inflammatory reactions.³ The low fluid shear stress increases the permeability of the vascular endothelial cells, allowing the cascade of intimal hyperplasia and resulting in the formulation of atheroma, which is the precursor of atherosclerosis.¹

For the influence of the fluid shear stress or flow pattern on the cellular behavior, several *in vitro* methods were developed to mimic the *in vivo* physiological/pathological condition, such as the cone-and-plate and the parallel plate models.⁴ However, their robustness and applicability are significantly weakened, due to high reagent consumption and limited reproducibility.¹ Recently the microfluidic or so-called lab-on-a-chip technology ^{has been} is of great interest in biomedical and biotechnological fields including cancer cell analysis,⁵ cell culture,⁶ and drug administering.⁷ In particular, the microphysiological system was successfully implemented in microfluidics, providing a more relevant humanized platform to probe the function of the individual organ and evaluate the safety and efficacy for various candidate therapeutics.⁸ Despite the importance of the flow pattern in microphysiological system, it is still challenging for the existing flow controller,

such as the flexible membranes,⁹ or external pressure pump,¹⁰ to precisely handle and control the microfluidic flow, due to the sophisticated system configuration and high cost of the supporting equipment.

As an alternative option, the stimuli-sensitive hydrogel ^{can be} **is** employed as the microfluidic flow controller due to its simple system configuration, since it eliminates the necessity for integrated electronics, controls, and power source, which are required for a conventional microfluidic device.¹¹ So far numerous stimuli-responsive hydrogels have been used in microfluidics, where the pH-sensitive hydrogel may be the most common one as the sensor or actuator for flow control.¹² However, the hydrogel is required to contact the surrounding solution with the specific pH,^{13, 14} which may limit the development of the microfluidic system in analytical instrument at remote locations.¹⁵ In addition, the time response of the single pH-sensitive hydrogel was relatively slow.^{11, 12, 16} Although two strategies were proposed to improve the response rate, such as utilization of the macroporous hydrogel or multiple smaller hydrogel disks, the hydrogel suffers from low mechanical stiffness,¹⁶⁻¹⁸ which makes it inappropriate for dynamic control of the microfluidic flow with a wide variety of rates, frequencies, and patterns required by the microphysiological systems. As such, it is desirable to develop a novel environmentally sensitive hydrogel-based microfluidic system with the fast responsiveness and excellent controllability.¹⁵

Recently much effort was made to investigate the magnetic-sensitive hydrogel as microfluidic controller.^{8, 19-21} An important advantage of the magnetic hydrogel over pH actuated controller is that the hydrogel behavior can be controlled from outside the

device,²² avoiding direct contact with the external magnetic stimulus. Furthermore, the magnetic hydrogel responds fast to the external magnetic stimulus, which is crucial for a microfluidic flow manipulation.¹⁵ The magnetic hydrogel undergoes a fast and reversible deformation in response to the external magnetic field, and it is thus potential in dynamic microfluidic flow control in real time. To the best of our knowledge, there is no prior theoretical or computational model to uncover the responsive mechanism or provide useful guidelines for harnessing the unique properties of the magnetic hydrogel. From the open literature, it is known that most of the reported models focused on the magnetic elastomers that were immersed in an air environment, and assumed an incompressible elastomer with constant volume.^{23, 24} However, the model developed in this paper focuses on the deformable hydrogel placed in a moving fluid environment, where the hydrogel absorbs large amounts of fluid, which is characterized by a chemical field, and significant volume change described by a mechanical field. Recently, the behavior of fast responsiveness attracts considerable attention in hydrogel study. For example, based on continuum mechanics and thermodynamics, Zheng *et al.*²⁵ theoretically studied the fast homogeneous diffusion of hydrogel under three different stimuli: the chemical potential, the temperature, and the hydraulic pressure respectively. It was shown that at the beginning of the hydrogel swelling, the higher hydraulic pressure leads to a larger diffusion rate, while the hydraulic pressure shows insignificant effect on the diffusion rate when an equilibrium state is reached. In another work, Zheng and Liu,²⁶ investigated the phase transition of temperature-sensitive hydrogel under mechanical constraint and it was found that phase transition may also be used to improve the diffusion rate. In terms of fast responsiveness of magnetic hydrogel subject to externally applied magnetic field, the

literature search reveals so far no study was made in theoretical aspect to quantify the peculiar property of magnetic hydrogel. Herein, a magneto-chemo-hydro-mechanical model is developed to characterize the fluid-structure interaction between the deformable magnetic hydrogel and moving fluid via the fully coupled arbitrary Lagrangian-Eulerian (ALE) method. The multiphysics model is examined by both the finite difference result and experimental data in open literature. After validation, a transient simulation is conducted to quantify the response rate of the magnetic hydrogel, and the magnetic response behavior of the magnetic hydrogel is subsequently investigated in detail. Furthermore, several desired patterns of pulsatile flows are achieved for simulating the cell physiological microenvironment at bone marrow stroma, and for the pathological conditions at femoral artery during diastole and systole.

2. Model formulation

2.1. Governing equations

Without free current, the distributive magnetostatic field is described by the Maxwell equations²⁷

$$\nabla_{\mathbf{X}} \times \mathbf{H} = 0, \quad \nabla_{\mathbf{X}} \cdot \mathbf{B} = 0 \quad (1)$$

where $\nabla_{\mathbf{X}} \times (\bullet)$ and $\nabla_{\mathbf{X}} \cdot (\bullet)$ are the material curl and divergence operators with respect to Lagrangian coordinate $\mathbf{X}$ respectively, $\mathbf{B}$ and $\mathbf{H}$ the magnetic induction and intensity respectively with respect to $\mathbf{X}$, and they are interacted by

$$\mathbf{B} = \mu_0 J \mathbf{C}^{-1} (\mathbf{H} + \mathbf{M}) \quad (2)$$

where μ_0 is the magnetic permeability at vacuum, $\mathbf{M}$ the magnetization vector, $\mathbf{C} = \mathbf{F}^T \mathbf{F}$ the right Cauchy-Green tensor, here $\mathbf{F} = \nabla_{\mathbf{x}} \mathbf{x}$ the deformation gradient, $J = \det(\mathbf{F})$ the determinant with the standard convention $J > 0$.

Over the interface between the magnetic hydrogel and fluid medium, the magnetic induction $\mathbf{B}$ and intensity $\mathbf{H}$ satisfy the jump conditions

$$\mathbf{N} \cdot [[\mathbf{B}]] = 0, \quad \mathbf{N} \times [[\mathbf{H}]] = 0 \quad (3)$$

where $\mathbf{N}$ is the Lagrangian unit outward normal, and the double square bracket a quantity jump across the material surface from its inside to the outside.

Based on the conservation of mass,²⁸ the chemical field with respect to the fluid diffusion is written in the local form of

$$\dot{C}_s = -\nabla_{\mathbf{x}} \cdot \mathbf{J}_s + R_s \quad (4)$$

where $C_s(\mathbf{X}, t)$, $\mathbf{J}_s$, and R_s denote the fluid concentration, flux, and generation respectively. Over the hydrogel-fluid interface, the chemical boundary condition is given by

$$\mathbf{J}_s \cdot \mathbf{N} = -i \quad (5)$$

where i is the surface flux rate.

By the conservation of momentum,²⁸ the mechanical field with the large deformation of magnetic hydrogel is expressed as

$$\rho_0 \dot{\mathbf{V}}_h = \nabla_{\mathbf{x}} \cdot \mathbf{P} + \mathbf{f}_b \quad (6)$$

where ρ_0 is the nominal mass density of the polymer network, $\mathbf{V}_h$ the deformation velocity, $\mathbf{P}$ the first Piola-Kirchhoff stress, and $\mathbf{f}_b$ the external force density. Across the hydrogel-fluid interface, the stress $\mathbf{P}$ satisfies the following mechanical boundary condition ²⁹

$$\mathbf{P} \cdot \mathbf{N} = \mathbf{T}_a + \mathbf{T}_m \quad (7)$$

where $\mathbf{T}_a$ and $\mathbf{T}_m$ denote the mechanical and magnetic tractions exerted on the hydrogel boundary respectively.

To characterize the hydrodynamic flow within microchannel in the microfluidic system, the incompressible Navier-Stokes equations are employed, namely ³⁰

$$\nabla \cdot \mathbf{v}_s = 0 \quad (8)$$

$$\rho_s \frac{\partial \mathbf{v}_s}{\partial t} + \rho_s (\mathbf{v}_s \cdot \nabla) \mathbf{v}_s = \nabla \cdot \boldsymbol{\sigma}_s + \rho_s \mathbf{g} \quad (9)$$

where $\nabla \cdot (\bullet)$ and $\nabla(\bullet)$ denote the spatial divergence and gradient operator respectively with respect to the Eulerian coordinate $\mathbf{x}$. ρ_s , $\mathbf{v}_s$ and $\mathbf{g}$ represent the fluid mass density, velocity, and gravitational acceleration respectively. $\boldsymbol{\sigma}_s$ is the fluid stress tensor defined as $\boldsymbol{\sigma}_s = -p\mathbf{I} + \phi[\nabla\mathbf{v}_s + (\nabla\mathbf{v}_s)^T]$, ³⁰ where p and ϕ are the fluid pressure and dynamic viscosity respectively. Across the hydrogel-fluid interface, the following kinetic and dynamic conditions are used ³⁰

$$\mathbf{v}_s = \dot{\mathbf{u}}, \quad \mathbf{n} \cdot \boldsymbol{\sigma} = \mathbf{n} \cdot (\boldsymbol{\sigma}_s + \boldsymbol{\sigma}_m) \quad (10)$$

where $\mathbf{u}$ is the hydrogel displacement, $\boldsymbol{\sigma}$ the Cauchy stress associated with the stress $\mathbf{P}$ via $\boldsymbol{\sigma} = J^{-1} \mathbf{P} \mathbf{F}^T$, and $\boldsymbol{\sigma}_m$ the Maxwell stress tensor in Eulerian form associated with corresponding stress $\mathbf{P}_m$ by $\boldsymbol{\sigma}_m = J^{-1} \mathbf{P}_m \mathbf{F}^T$.

2.2. Constitutive equations

Based on thermodynamics,²⁸ it is known that the increase in free energy of the hydrogel system is less than or equal to the external power plus the free energy due to fluid transport and hydrogel magnetization, namely

$$\begin{aligned} \frac{d}{dt} \left\{ \int_{V_0} \left(F + \frac{1}{2} \rho_0 |\mathbf{V}_h|^2 \right) dv \right\} \leq & \int_{S_0} \mathbf{P} \mathbf{N} \cdot \mathbf{V}_h ds + \int_{V_0} \mathbf{f}_b \cdot \mathbf{V}_h dv - \int_{S_0} \mu_s \mathbf{J}_s \cdot \mathbf{N} ds + \int_{V_0} \mu_s R_s dv \\ & - \int_{V_0} [-(\mathbf{M} \cdot \mathbf{B} \cdot \mathbf{F}^{-T}) : \dot{\mathbf{F}} + (\mathbf{F}^{-T} \cdot \mathbf{M} \otimes \mathbf{B}) : \dot{\mathbf{F}} + \mathbf{M} \cdot \dot{\mathbf{B}}] dv \end{aligned} \quad (11)$$

where μ_s is the chemical potential and F the free energy density.

By Equations (4) and (6), and the divergence theorem, the local form of the free energy imbalance (11) is written as

$$\dot{F} \leq \mathbf{P} : \dot{\mathbf{F}} + (\mu_s \dot{C}_s - \mathbf{J}_s \cdot \nabla_{\mathbf{x}} \mu_s) - [-(\mathbf{M} \cdot \mathbf{B} \cdot \mathbf{F}^{-T}) : \dot{\mathbf{F}} + (\mathbf{F}^{-T} \cdot \mathbf{M} \otimes \mathbf{B}) : \dot{\mathbf{F}} + \mathbf{M} \cdot \dot{\mathbf{B}}] \quad (12)$$

If the free energy density F is a function of state relating to the three independent variables, i.e. the fluid concentration C_s , magnetic induction $\mathbf{B}$, and deformation gradient $\mathbf{F}$, we have

$$(\mathbf{P} + \mathbf{M} \cdot \mathbf{B} \cdot \mathbf{F}^{-T} - \mathbf{F}^{-T} \cdot \mathbf{M} \otimes \mathbf{B} - \frac{\partial F}{\partial \mathbf{F}}) : \dot{\mathbf{F}} - \left(\frac{\partial F}{\partial \mathbf{B}} + \mathbf{M} \right) \cdot \dot{\mathbf{B}} + \left(\mu_s - \frac{\partial F}{\partial C_s} \right) \dot{C}_s - \mathbf{J}_s \cdot \nabla_{\mathbf{x}} \mu_s \geq 0 \quad (13)$$

The coefficient in any bracket vanishes due to arbitrariness of the variables.³¹ To ensure the inequality held, the fluid flux $\mathbf{J}_s$ is associated with the chemical potential gradient $\nabla_{\mathbf{x}} \mu_s$ through³²

$$\mathbf{J}_s = -\mathbf{L} \nabla_{\mathbf{x}} \mu_s \quad (14)$$

where the mobility $\mathbf{L}$ is a positive-definite tensor related to the diffusion coefficient D_s via $\mathbf{L} = C_s D_s \mathbf{C}^{-1} / (k_B T)$. Consequently, we have the following constitutive relations

$$\mathbf{P} = \frac{\partial F}{\partial \mathbf{F}} - \mathbf{M} \cdot \mathbf{B} \cdot \mathbf{F}^{-T} + \mathbf{F}^{-T} \cdot \mathbf{M} \otimes \mathbf{B}, \quad \mathbf{M} = -\frac{\partial F}{\partial \mathbf{B}}, \quad \mu_s = \frac{\partial F}{\partial C_s} \quad (15)$$

If the magnetic intensity $\mathbf{H}$ is adopted as the independent variable, the augmented energy of F with partial Legendre transformation is required via²⁷

$$\Omega(\mathbf{F}, \mathbf{H}, C_s) = F(\mathbf{F}, \mathbf{B}, C_s) + (\mathbf{C} : \mathbf{B} \otimes \mathbf{B}) / (2\mu_0 J) - \mathbf{H} \cdot \mathbf{B} \quad (16)$$

then the constitutive relation (15) is rewritten as

$$\mathbf{P} = \frac{\partial \Omega}{\partial \mathbf{F}}, \quad \mathbf{B} = -\frac{\partial \Omega}{\partial \mathbf{H}}, \quad \mu_s = \frac{\partial \Omega}{\partial C_s} \quad (17)$$

2.3. Free energy density

In the present hydrogel system, the free energy density Ω is contributed by three components, stretching of the polymer Ω_{ela} , mixing of the polymer and fluid Ω_{mix} , and the hydrogel magnetization Ω_{mag} . If the linear magnetic behavior and the molecular incompressibility condition $1 + v_s C_s = \det(\mathbf{F})$ ³³ are included, the free energy density Ω is expressed as^{24, 34, 35}

$$\begin{aligned} \Omega = & \frac{1}{2} N k_B T \{ \text{tr}(\mathbf{F}^T \mathbf{F}) - 3 - 2 \ln[\det(\mathbf{F})] \} - \frac{J \mu_m}{2} [(\mathbf{F}^{-T} \mathbf{H}) \cdot (\mathbf{F}^{-T} \mathbf{H})] \\ & + \frac{K_B T}{v_s} [v_s C_s \ln(\frac{v_s C_s}{1 + v_s C_s}) + \frac{\chi_H v_s C_s}{1 + v_s C_s}] + \Pi [1 + v_s C_s - \det(\mathbf{F})] \end{aligned} \quad (18)$$

where N denotes the effective number of polymer chains per unit reference volume, k_B the Boltzmann constant, T the absolute temperature, v_s the volume per fluid species, χ_H the Flory-Huggins parameter, Π the Lagrange multiplier that is also interpreted as osmotic pressure,^{32, 33} and μ_m the magnetic permeability that is associated with the relative permeability μ_{mr} via $\mu_m = \mu_0 \mu_{mr}$.

From the constitutive relation (17) and the free energy density (18), the stress $\mathbf{P}$, the magnetic induction $\mathbf{B}$, and the chemical potential μ_s are expressed respectively as

$$\mathbf{P} = N k_B T (\mathbf{F} - \mathbf{F}^{-T}) - \Pi \mathbf{F}^{-T} + J \mu_m \mathbf{F}^{-T} (\mathbf{H} \otimes \mathbf{H}) \mathbf{C}^{-1} - \frac{1}{2} J \mu_m \mathbf{F}^{-T} [(\mathbf{C}^{-1} \mathbf{H}) \cdot \mathbf{H}] \quad (19)$$

$$\mathbf{B} = J \mu_m \mathbf{C}^{-1} \mathbf{H} \quad (20)$$

$$\mu_s = k_B T \left[\ln \left(1 - \frac{1}{1 + \nu_s C_s} \right) + \frac{1}{1 + \nu_s C_s} + \frac{\chi_H}{(1 + \nu_s C_s)^2} \right] + \Pi \nu_s \quad (21)$$

Based on Equation (1), the magnetic intensity vector $\mathbf{H}$ relates to the magnetic scalar potential ψ by $\mathbf{H} = -\nabla_{\mathbf{x}}\psi$. By Equations (1) and (20), the magnetic potentials ψ in both the magnetic hydrogel and the exterior fluid satisfy

$$\nabla_{\mathbf{x}} \cdot (J \mu_m \mathbf{C}^{-1} \nabla_{\mathbf{x}} \psi) = 0 \quad (22)$$

So far the multiphysics model has been formulated for the magnetic hydrogel placed in the fluid subject to the coupled magneto-chemo-hydro-mechanical fields. It consists of the magnetostatic equation (22), the mass conservation equation (4), the mechanical balance equation (6), and the incompressible Navier-Stokes equations (8) and (9). In addition, the free energy density is given by Equation (18), and the constitutive relations are described by Equations (14), (19), (20), and (21). The model is implemented through the finite element package, COMSOL Multiphysics 5.3. As is known, the hydrogel deformation is conveniently formulated by a Lagrangian description and a material frame, whereas the fluid motion by a Eulerian description and a spatial frame. To connect the interface between the spatial frame of the fluid and material frame of the hydrogel, the fully coupled ALE method is adopted, which calculates the new mesh coordinates according to the movement of the hydrogel boundary.³⁶

3. Results and discussion

3.1. Validations of the multiphysics model

In order to examine the model, three comparisons are made, including (i) quantitative validation with the finite difference result for transient swelling of a constrained hydrogel

slab,³³ (ii) quantitative validation with the experimental data in open literature for equilibrium deformation of magnetic hydrogel,³⁷ and (iii) qualitative validation with experimental work for flow pattern generation for the real cell physiological microenvironment experienced by the bone marrow stromal cells.³⁸

3.1.1. Validation for swelling of a hydrogel slab

The first comparison is made with the finite difference solution by Bouklas *et al.* for one-dimensional (1D) transient swelling of a hydrogel slab.³³ The hydrogel is placed in a rigid container and its deformations are constrained in X_1 and X_3 directions, in which the top surface of the hydrogel is in contact with the fluid, while its bottom surface is fixed and no fluid is allowed to diffuse, as shown in Figure 1 (a). In this validation, the model developed is reduced to a two-dimensional (2D) domain and then compared with that obtained by the 1D finite difference method.³³ Based on the field equations (4) and (6), and the constitutive equations (14), (19), and (21), the stretch λ_2 along X_2 direction is obtained via solving the nonlinear diffusion equation

$$\lambda_0^2 \frac{\partial \lambda_2}{\partial t} = D_s \frac{\partial}{\partial X_2} \left\{ \left[\frac{1}{\lambda_0^2 \lambda_2^4} - \frac{2\chi_H(\lambda_0^2 \lambda_2 - 1)}{\lambda_0^4 \lambda_2^5} + N\nu_s \frac{(\lambda_0^2 \lambda_2 - 1)(\lambda_2^2 + 1)}{\lambda_0^2 \lambda_2^4} \right] \frac{\partial \lambda_2}{\partial X_2} \right\},^{33} \text{ where } \lambda_0$$

is the initial stretch along X_2 direction. The boundary conditions are given as follows:

the hydrogel displacement $\mathbf{u}|_{X_2=0}=0$, the fluid flux at the bottom surface $\mathbf{J}_s|_{X_2=0}=0$, the chemical potential at top surface $\mu_s|_{X_2=L_H}=0$, the hydrogel displacement along X_1 direction $u_1|_{X_1=-L_0, L_0}=0$ (Here $2L_0$ is the hydrogel length), and the fluid flux at lateral surfaces $\mathbf{J}_s|_{X_1=-L_0, L_0}=0$, respectively. The initial stretch along X_2 direction is set as

$\lambda_0 = 1.4$, and the initial chemical potential $\mu_{s0}(\mathbf{X}, 0) = -0.035k_B T$ is determined by Equations (19) and (21). The input parameters include $L_H = 2L_0 = 1 \text{ m}$, $D_s = 8 \times 10^{-9} \text{ m}^2 \text{ s}^{-1}$, $v_s = 10^{-28} \text{ m}^3$, $G = 4 \times 10^4 \text{ Pa}$, and $\chi_H = 0.4$.³³ For simplification, the time t is normalized as the non-dimensional time $\bar{t} = tD/L_H^2$, chemical potential μ_s as the non-dimensional chemical potential $\bar{\mu}_s = \mu_s/k_B T$, and the stress $\mathbf{P}$ as the non-dimensional stress $\bar{\mathbf{P}} = \mathbf{P}v_s/k_B T$. Figures 1(b)-(d) are plotted to compare the present 2D numerical results with 1D finite difference solutions³³ for distributions of the non-dimensional chemical potential $\bar{\mu}_s$, the stretch λ_2 , and the non-dimensional compressive stress $\bar{P}_{x1}$, against the non-dimensional coordinate X_2/L_H , at different non-dimensional times $\bar{t} = 0.1, 10, 50, 200, 1000$, and t^* , respectively, where t^* represents a time for evolution of the swelling fully developed.³³

It is found from Figures 1(b)-(d) that the chemical potential $\bar{\mu}_s$, the stretch λ_2 , and the compressive stress $\bar{P}_{x1}$ have the similar profile patterns, where the sharp gradients of $\bar{\mu}_s$, λ_2 , and $\bar{P}_{x1}$ are developed near the upper surface $X_2 = L_H$ at the early stage of swelling (e.g. $\bar{t} \leq 50$), then the gradients become smaller gradually (e.g. $\bar{t} \leq 1000$), and finally they vanish to reach the equilibrium state (e.g. $\bar{t} = t^*$). Moreover, it is demonstrated that the increasing rate of the hydrogel stretch λ_2 decreases with non-dimensional time $\bar{t}$, as illustrated in Figure 1(c). It is because that the gradient of chemical potential $\bar{\mu}_s$ is large at the surface $X_2 = L_H$, and the compressive stress $\bar{P}_{x1}$ small in the interior part of the hydrogel during the initial swelling process, which both

contribute to the fluid absorption and diffusion.³⁹ However, when more and more fluid enters into the hydrogel, the gradient of $\bar{\mu}_s$ becomes smaller and the compressive force $\bar{P}_{x1}$ larger, which both prevent the fluid penetrating into the hydrogel.

3.1.2. Validation for equilibrium deformation of magnetic hydrogel

The second comparison is made with the experimental data,³⁷ for equilibrium deformation of the magnetic hydrogel, where the spherical magnetic polydimethylsiloxane (PDMS) hydrogel was synthesized and then swollen to equilibrium in cyclohexane. After that, the hydrogel was placed in air and exposed to a uniform magnetic field generated by a large electromagnet, and the hydrogel shape change was recorded with increasing magnetic field, as illustrated in Figure 2(a). The input parameters include: $L_R = 24 \text{ mm}$, $L_H = 40 \text{ mm}$, $R_0 = 3.225 \text{ mm}$, $G = Nk_B T = 1.5 \times 10^4 \text{ Pa}$, $\mu_0 = 4\pi \times 10^{-7} \text{ N A}^{-2}$ and $\mu_{mr} = 1.12$. As seen from Figure 2 (b), the magnetic hydrogel elongates along the field direction and contracts in the transverse direction, and the deformation λ increases with the external magnetic field H_0 . This is because the magnitude of the magnetic force F_m is associated with the local magnetic field via $F_m = \mu_0(\mu_{mr} - 1)(H_t^2 + \mu_{mr}H_n^2)/2$,⁴⁰ where H_t and H_n are the tangential and normal components of the magnetic fields over the hydrogel-surrounding interface respectively. It is known that magnetic field is uniformly distributed inside the hydrogel, and the tangential magnetic field $H_t = 0$ at the top and bottom apexes of the hydrogel. In addition, the relative magnetic permeability of the hydrogel μ_{mr} is higher than 1, such that the force F_m maximizes at the top and bottom apexes of hydrogel and thus makes the hydrogel deformed to the prolate spheroid.⁴⁰

3.1.3. Validation for flow pattern for various physiological conditions

For further validation of the present work, the predicted flow pattern in Figure 3 generated by the present magnetic-sensitive hydrogel-based microfluidic system (as illustrated in Figure 4(a)) is qualitatively compared with the experimental measurement.³⁸ As observed from Figure 3, the reversible variation of the flow rate w is shown with the time t at different inlet pressures p_i of 0.5 and 1 kPa, in response to the pulsatile magnetic field with frequencies f_H of 0.015, 0.044, and 0.074 Hz respectively, where the hydrogel radius $R_0 = 200 \mu\text{m}$ and external magnetic field $H_0 = 200 \text{ kA m}^{-1}$. It is seen that the simulated flow patterns by the present model are qualitatively consistent with those in experimental observation,³⁸ where the real mechanical microenvironment experienced by the bone marrow stromal cells were generated by a programmable microprocessor to open and close a pinch valve that is connected to the parallel-plate flow chamber. Therefore, it is indicated that the magnetic-sensitive hydrogel-based microfluidic system can be precisely controlled for formation of the physiological condition on demand within bone tissue. Apart from that, it is observed that the amplitude of the flow rate w can be altered by the inlet pressure p_i , and the frequency tuned by the external magnetic field frequency f_H . Moreover, no hysteresis in the frequency is found between the applied magnetic field and the resulting flow rate w , probably due to the fast-response deformation and excellent controllability of the magnetic hydrogel.

From the above comparisons, it is found that good agreements are achieved between the present model and the finite difference solution, and between the model and experimental findings. It is thus concluded that the validated model can well capture the

performance of the magnetic hydrogel, and then be used to undertake the following studies.

3.2. Fast-response behavior of magnetic-sensitive hydrogel

As well known, the stimuli-sensitive hydrogel is widely used for self-regulated flow control in microfluidic chips,¹² where its performance is generally associated with the response rate of the hydrogel. As discussed previously, fast responsiveness is a remarkable feature of the magnetic hydrogel,⁴¹ whereas no effort was made to quantitatively characterize the behavior with the magneto-chemo-hydro-mechanical coupled field. To the best of our knowledge, it is the first theoretical study to reveal the property of the magnetic hydrogel. Here a transient simulation is carried out for the magnetic hydrogel placed in the surrounding fluid, with the effect of externally applied magnetic field. As demonstrated in Figure 4(a), the cylindrical magnetic hydrogel with outer radius R_o is placed in the microchannel of the microfluidic system, where the displacements of the top and bottom surfaces of the hydrogel are restricted by the glass channel, and the inner part of the magnetic hydrogel is coated on a non-magnetic fixed rod with the radius $R_i = 0.4R_o$. As such, it is reasonable to assume that the hydrogel deforms only in the radial direction, and a 2D transient simulation is thus conducted here to investigate the time-dependent responsive behavior of the hydrogel. The fluid is delivered to the microchannel with a constant inlet pressure p_i . Initially the hydrogel is at its dry state, and it then swells gradually to regulate the fluid flow. To improve the hydrogel response rate, a uniform magnetic field is imposed in the surrounding air.

Figures 4(b)-(c) demonstrate the effect of the external magnetic field H_0 on variation of the average fluid flow rate w at the outlet of microchannel against the time t , where $R_0 = 200 \mu\text{m}$ and $R_0 = 250 \mu\text{m}$ respectively. The inputs of the parameters required for the numerical simulation include: $L_x = 6 \text{ mm}$, $L_y = 4 \text{ mm}$, $L_c = 0.8 \text{ mm}$, $D_s = 8 \times 10^{-9} \text{ m}^2 \text{ s}^{-1}$, $\chi_H = 0.6$, $\varphi = 0.001 \text{ Pa s}$, $\rho_s = 1000 \text{ kg m}^{-3}$, $G = 10^5 \text{ Pa}$, and $\mu_{mr} = 3$. It is observed from the figures that, the flow rate w decreases with the increase of time t , where the rate w further decreases with increasing magnetic field H_0 . As shown in Figure 4(b), the flow rate w decreases drastically from about 4.5 to 1.1 mL min^{-1} , when subjected to the magnetic field H_0 of 150 kA m^{-1} , if the time t is less than 50 s . Subsequently, the rate w decreases slowly with a minimal variation from 1.1 to 0.4 mL min^{-1} during the following 150 s . After that, it approximates the steady state with $w = 0.25 \text{ mL min}^{-1}$. It is also found from Figure 4(b) that, the rate w at steady state is nonzero if magnetic field H_0 is 0 or 150 kA m^{-1} , whereas it becomes zero when $H_0 = 300 \text{ kA m}^{-1}$, where the magnetic hydrogel contacts the microchannel wall and fully blocks the fluid stream. Therefore, the application of the external magnetic field accelerates the hydrogel swelling. Furthermore, it is observed that variation of the flow rate w is inversely related to evolution of the hydrogel swelling, where the hydrogel swells sharply at the beginning, and then slowly until the equilibrium state is reached, as shown in the contours for hydrogel deformation under the magnetic field H_0 of 150 kA m^{-1} . As such, the flow manipulation may be realized by controlling the hydrogel deformation under the external magnetic field. As also demonstrated from Figure 4(c), it

is clearly known that the times required are different for the magnetic hydrogels to reach the closed state. For example, it takes about 15 s for the hydrogel under H_0 of 300 kA m⁻¹, while about 45 s and 70 s under H_0 of 150 kA m⁻¹ and 0 respectively. Thereby, the response times for the magnetic hydrogels are improved by 78% and 35% respectively, when subjected to the magnetic fields H_0 of 300 and 150 kA m⁻¹, compared with hydrogel without magnetic effect. This finding is qualitatively in line with the experimental result,¹³ in which the hydrogel rapidly deformed when the external magnetic field is applied. It is worth noting that the magnetic hydrogel with fast responsiveness may be potential as the flow controller in microfluidic chips and will be discussed in the section 3.4.

3.3. Magnetic response behavior of magnetic-sensitive hydrogel

For further understanding the responsive behavior of magnetic-sensitive hydrogel, it is crucial to characterize the magnetic field distribution. As a key physical parameter in magnetostatics, magnetic permeability is used to characterize the magnetizability of the magnetic hydrogel, such that different spatial distribution of magnetic field occurs in hydrogels with different permeabilities. Hence, the magnetic field distribution is investigated over a computational domain covering the surrounding air and fluid, the hydrogel, as well as the fixed rod, when subjected to varying relative magnetic permeabilities μ_{mr} , i.e. 1.5, 3, and 4.5. In the present simulation, the outer radius of the hydrogel is $R_o = 200 \mu\text{m}$, the imposed magnetic field $H_0 = 200 \text{ kA m}^{-1}$. The other inputs of the parameters follow those in Section 3.2.

Figure 5 visualizes the influence of the relative magnetic permeability μ_{mr} on variation of the non-dimensional magnetic field H_y / H_0 at different times, 0 and 40s, along a vertical line located at $X = 2$ mm across the surrounding air and fluid, the hydrogel, and the non-magnetic rod. As observed from Figure 5, the non-dimensional magnetic field H_y / H_0 is around 1 if the non-dimensional distance y / L_y is less than 0.2, demonstrating that the original magnetic field applied is not disturbed far away from the hydrogel-surrounding interface. However, when the distance y / L_y increases continuously, the magnetic field H_y / H_0 increases slightly and then dramatically until it reaches a maximum. After that, a sharp decrease is found in the field H_y / H_0 , and then it decreases further until it reaches a minimum, where the field H_y / H_0 follows a step function with a sharp increase at the hydrogel-rod interface with a small range of the normalized distance y / L_y . The profile of the magnetic field distribution may be understood via the following two reasons, (i) the different relative magnetic permeabilities among the surrounding, the hydrogel, and the rod, and (ii) the boundary condition for the magnetic field in Equation (3), showing that the normal component of the magnetic field is discontinuous across the material boundary while the tangential one is continuous across the material boundary. The figures also demonstrate that the original magnetic field is more perturbed with the increase of the relative magnetic permeability μ_{mr} , due to the increasing magnetizability of the magnetic hydrogel. It is also seen from the figures, the hydrogels extend along the field direction with increasing time. In addition, a higher relative magnetic permeability μ_{mr} contributes to a larger hydrogel

deformation. Furthermore, it is found that the magnetic field is distributed uniformly within the fixed rod, due to the constant demagnetizing factor for a spherical shape.²⁹

3.4. Flow pattern generation for various physiological/pathological conditions

For accurate representation of human physiology, it is essential to simulate physiological and pathological conditions in microfluidic chips. In Section 3.1.3, various patterns of pulsatile flows are generated by the magnetic-sensitive hydrogel-based microfluidic system to mimic the real cell physiological microenvironment experienced by the bone marrow stromal cells. In the present work, the pathological conditions are studied in detail at femoral artery during diastole and systole.

According to the Doppler ultrasound measurement, the femoral blood flow patterns may be categorized as the triphasic, biphasic, sharp monophasic, and poor monophasic waveforms, based on different levels of the peripheral arterial disease progression.⁴² It is reported that the poor monophasic waveform is highly associated with atherosclerosis, and the sharp monophasic waveform associated with the stenosis at femoral arteries.⁴²⁻⁴⁴ Therefore, it is crucial to implement such flow patterns in the microphysiological system to study the pathophysiology at the femoral artery, for deeper understanding of the fundamental mechanism underlying real events that occur in human body. However, it is challenging to generate the complicated flow patterns with the conventional microfluidic chips.²⁰ A major advantage of the presently developed magnetic-sensitive hydrogel-based microfluidic system over the conventional microfluidic chips is to generate the user-defined flow patterns. For example, when the externally applied magnetic field H_0 is prescribed to follow the pulsatile profile, as shown in Figure 6(a), the poor monophasic

waveform is achieved with a blunted systole rise and a slow diastolic fall, as shown in Figure 6(b). When the magnetic field H_0 is prescribed to follow the profile, as demonstrated in Figure 7(a), the sharp monophasic flow is achieved with a sharp systole rise, and a slow diastolic fall, as well as a pause phase, as shown in Figure 7(b). Furthermore, it is noted that the waveforms simulated are consistent qualitatively with the real blood flow pattern measured at the femoral artery in pathological condition,⁴² indicating the strong capability of the novel magnetic-sensitive hydrogel-based microfluidic device in manipulating the microfluidic flows for various microphysiological systems.

4. Conclusions

The magnetic-sensitive hydrogel-based microfluidic system is numerically designed and investigated via the magneto-chemo-hydro-mechanical model, where the fluid-structure interaction of the deformable magnetic hydrogel with the surrounding fluid is formulated by the fully coupled ALE method. In the present model, the physicochemical responsive mechanisms are characterized including hydrogel magnetization, fluid diffusion, fluid flow, and hydrogel large deformation. The model is examined by both the finite difference solution³³ and experimental data,^{37, 38} where satisfactory agreements are obtained. Moreover, the transient simulation results show that the response times for the present magnetic hydrogels are improved by 78% and 35%, when subjected to the uniform magnetic fields of 300 and 150 kA m⁻¹ respectively, compared with the hydrogel without external magnetic stimulation. Furthermore, the magnetic response behavior of the magnetic hydrogel is studied, showing that the sharp variations of magnetic field occur at the hydrogel-surrounding and hydrogel-rod interfaces, and a higher relative

magnetic permeability contributes to a larger hydrogel deformation. After that, several patterns of pulsatile flows are generated for mimicking the real physiological cell mechanical microenvironment experienced by bone marrow stromal cells, and for the pathological condition, such as poor and sharp monophasic waveforms, at femoral artery during diastole and systole. Hence, the present multiphysics model is a useful tool: (i) to achieve the kinetic behavior of the magnetic hydrogel with the magneto-chemo-hydro-mechanical coupled effect for optimized design of the fast-response magnetic hydrogel, and (ii) to elucidate the correlation of the flow rate with the varying inlet pressure, external magnetic field intensity, and frequency for generation of the user-defined fluid flow patterns. Therefore, the magnetic-sensitive hydrogel-based microfluidic system with tunable flow patterns is particularly suitable for investigation of cell behavior and function on microfluidic chips by providing the real physiological/pathological microenvironment.

Conflicts of interest

There are no conflicts to declare

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References

1. C. Zheng, X. Zhang, C. Li, Y. Pang and Y. Huang, *Anal. Chem.*, 2017, **89**, 3710-3715.
2. J. Shemesh, I. Jalilian, A. Shi, G. Heng Yeoh, M. L. Knothe Tate and M. Ebrahimi Warkiani, *Lab Chip*, 2015, **15**, 4114-4127.
3. W. Zheng, R. Huang, B. Jiang, Y. Zhao, W. Zhang and X. Jiang, *Small*, 2016, **12**, 2022-2034.
4. J. A. Frangos, S. G. Eskin, L. V. McIntire and C. L. Ives, *Science*, 1985, **227**, 1477.
5. M. Chung, J. Ahn, K. Son, S. Kim and N. L. Jeon, *Adv. Healthc. Mater.*, 2017, **6**, 1700196.
6. T. Osaki, Y. Shin, V. Sivathanu, M. Campisi and R. D. Kamm, *Adv. Healthc. Mater.*, 2018, **7**, 1700489.
7. N. T. Nguyen, S. A. Shaegh, N. Kashaninejad and D. T. Phan, *Adv Drug Deliv Rev*, 2013, **65**, 1403-1419.
8. G. A. Truskey, H. E. Achneck, N. Bursac, H. F. Chan, C. S. Cheng, C. Fernandez, S. Hong, Y. Jung, T. Koves, W. E. Kraus, K. Leong, L. Madden, W. M. Reichert and X. Zhao, *Stem. Cell. Res. Ther.*, 2013, **4**, S10.
9. H. Chen, J. Cornwell, H. Zhang, T. Lim, R. Resurreccion, T. Port, G. Rosengarten and R. E. Nordon, *Lab Chip*, 2013, **13**, 2999-3007.
10. R. Estrada, G. A. Giridharan, M.-D. Nguyen, T. J. Roussel, M. Shakeri, V. Parichehreh, S. D. Prabhu and P. Sethu, *Anal. Chem.*, 2011, **83**, 3170-3177.
11. J. C. Kurnia, E. Birgersson and A. S. Mujumdar, *J. Funct. Biomater.*, 2011, **2**, 195-212.
12. D. J. Beebe, J. S. Moore, J. M. Bauer, Q. Yu, R. H. Liu, C. Devadoss and B.-H. Jo, *Nature*, 2000, **404**, 588-590.
13. W. Gao, L. Wang, X. Wang and H. Liu, *ACS Appl. Mater. Interfaces*, 2016, **8**, 14182-14189.
14. N. S. Satarkar, W. Zhang, R. E. Eitel and J. Z. Hilt, *Lab Chip*, 2009, **9**, 1773-1779.
15. J. ter Schiphorst, S. Coleman, J. E. Stumpel, A. Ben Azouz, D. Diamond and A. P. H. J. Schenning, *Chem. Mater.*, 2015, **27**, 5925-5931.

16. Y. Qiu and K. Park, *Adv. Drug Del. Rev.*, 2001, **53**, 321-339.
17. E. Birgersson, H. Li and S. Wu, *J. Mech. Phys. Solids*, 2008, **56**, 444-466.
18. E. Lee, D. Kim, H. Kim and J. Yoon, *Sci. Rep.*, 2015, **5**, 15124.
19. O. Philippova, A. Barabanova, V. Molchanov and A. Khokhlov, *Eur. Polym. J.*, 2011, **47**, 542-559.
20. S. Hong, Y. Jung, R. Yen, H. F. Chan, K. W. Leong, G. A. Truskey and X. Zhao, *Lab Chip*, 2014, **14**, 514-521.
21. R. Byrne, F. Benito-Lopez and D. Diamond, *Mater. Today*, 2010, **13**, 16-23.
22. L. Rödling, E. M. Volz, A. Raic, K. Brändle, M. Franzreb and C. Lee - Thedieck, *Adv. Healthc. Mater.*, 2018, **7**, 1701403.
23. R. Zhao, Y. Kim, S. A. Chester, P. Sharma and X. Zhao, *J. Mech. Phys. Solids*, 2019, **124**, 244-263.
24. J.-P. Pelteret, D. Davydov, A. McBride, D. K. Vu and P. Steinmann, *Int. J. Numer. Meth. Eng.*, 2016, **108**, 1307–1342.
25. S. Zheng, Z. Li and Z. Liu, *International Journal of Mechanical Sciences*, 2018, **137**, 263-270.
26. Z. Shoujing and L. Zishun, *J. Appl. Mech.*, 2017, **85**, 021002.
27. B. Nedjar, *J. Mech. Phys. Solids*, 2015, **84**, 293-312.
28. M. E. Gurtin, E. Fried and L. Anand, *The Mechanics and Thermodynamics of Continua*, Cambridge University Press, Cambridge, U.K., 2010.
29. Q. Liu, H. Li and K. Y. Lam, *J. Microelectromech. Syst.*, 2018, **27**, 127-136.
30. Y. Zhang, Z. Liu, S. Swaddiwudhipong, H. Miao, Z. Ding and Z. Yang, *J. Funct. Biomater.*, 2012, **3**, 464-479.
31. Q. Liu, H. Li and K. Y. Lam, *J. Phys. Chem. B*, 2017, **121**, 5633-5646.
32. W. Hong, X. Zhao, J. Zhou and Z. Suo, *J. Mech. Phys. Solids*, 2008, **56**, 1779-1793.
33. N. Bouklas and R. Huang, *Soft Matter*, 2012, **8**, 8194-8203.
34. P. J. Flory, *Principles of Polymer Chemistry*, Cornell University, New York, 1953.

35. P. J. Flory and J. R. Jr., *J. Chem. Phys*, 1943, **11**, 521-526.
36. X. Xu, Z. Li and A. Nehorai, *Biomicrofluidics*, 2013, **7**, 054108.
37. F. Genoveva and Z. Miklos, *J. Phys.: Condens. Matter*, 2010, **22**, 276001.
38. L. A. Sharp, Y. W. Lee and A. S. Goldstein, *Ann. Biomed. Eng.*, 2009, **37**, 445-453.
39. Y. Liu, H. Zhang, J. Zhang and Y. Zheng, *Int. J. Solids Struct.*, 2016, **80**, 246-260.
40. M. S. Korlie, A. Mukherjee, B. G. Nita, J. G. Stevens, A. D. Trubatch and P. Yecko, *J. Phys. Condens. Matter*, 2008, **20**, 204143.
41. N. S. Satarkar and J. Zach Hilt, *Acta Biomater.*, 2008, **4**, 11-16.
42. S. Spronk, P. T. den Hoed, L. C. W. de Jonge, L. C. van Dijk and P. M. T. Pattynama, *J. Vasc. Surg.*, 2005, **42**, 236-242.
43. R. Scissons and A. Comerota, *J. Diagn. Med. Sonogr.*, 2009, **25**, 185-194.
44. A. Javadzadegan, A. Lotfi, A. Simmons and T. Barber, *Comput. Methods Biomech. Biomed. Eng.*, 2016, **19**, 1143-1153.

List of Figures

- Figure 1.** Schematic of the constrained hydrogel slab in transient swelling (a). Comparison of the present theoretical simulation (solid lines) with the finite difference solutions ³³ (circles) for variation of the non-dimensional chemical potential $\bar{\mu}_s$ (b), the stretch λ_2 (c), and the non-dimensional stress $\bar{P}_{x1}$ (d), against the non-dimensional coordinate X_2 / L_H 29
- Figure 2.** Schematic of the magnetic hydrogel placed in surrounding air (a). Comparison of the present theoretical simulation (solid lines) and the experimental results ³⁷ (square bricks) for the magnetic field-dependent deformation of the magnetic hydrogel at equilibrium state (b). 30
- Figure 3.** The flow pattern analysis at various conditions. The pulsatile flow patterns at the frequencies f_H of (a) 0.015 Hz, (b) 0.044 Hz, and (c) 0.074 Hz with the inlet pressure p_i of 0.5 kPa. The pulsatile flow patterns at the frequencies f_H of (d) 0.015 Hz, (e) 0.044 Hz, and (f) 0.074 Hz with the inlet pressure p_i of 1 kPa. 31
- Figure 4.** Schematic of the magnetic hydrogel as microfluidic flow controller (a). Influence of the magnetic field H_0 on variation of fluid flow rate w against the time t when subjected to different hydrogel radii: $R_o = 200\mu\text{m}$ (b), and $R_o = 250\mu\text{m}$ (c), where the contours represent the hydrogel displacement under the field H_0 of 150 kA m^{-1} 32
- Figure 5.** Influence of relative magnetic permeability μ_{mr} on variation of non-dimensional magnetic field H_y / H_0 along a vertical line located at $X = 2 \text{ mm}$ across the surrounding air and fluid, the hydrogel, and the non-magnetic rod when time $t = 0$ (a), and 40 s (b)..... 33
- Figure 6.** (a) Prescribed external magnetic field H_0 as a function of time t . (b) Generation of the pathological flow pattern for replication of the typical poor monophasic blood waveform at the common femoral artery. 34

Figure 7. (a) Prescribed external magnetic field H_0 as a function of time t . (b)
Generation of the pathophysiological flow pattern for replication of the typical
sharp monophasic blood waveform at the common femoral artery. 34

Figure 1. Schematic of the constrained hydrogel slab in transient swelling (a). Comparison of the present theoretical simulation (solid lines) with the finite difference solutions³³ (circles) for variation of the non-dimensional chemical potential $\bar{\mu}_s$ (b), the stretch λ_2 (c), and the non-dimensional stress $\bar{P}_{X1}$ (d), against the non-dimensional coordinate X_2/L_H .

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Figure 5. Influence of relative magnetic permeability μ_{mr} on variation of non-dimensional magnetic field H_y / H_0 along a vertical line located at $X = 2$ mm across the surrounding air and fluid, the hydrogel, and the non-magnetic rod when time $t = 0$ (a), and 40 s (b).

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Figure 7. (a) Prescribed external magnetic field H_0 as a function of time t . (b) Generation of the pathophysiological flow pattern for replication of the typical sharp monophasic blood waveform at the common femoral artery.

Table of Contents Entry

A magnetic-sensitive hydrogel-based microfluidic system is designed numerically for replicating various physiological and pathological conditions in the human body