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Preliminary Design of a SIMO Fuzzy Controller for Steering Microparticles inside Blood Vessels by Using a Magnetic Resonance Imaging System

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***Abstract*—In this paper, a Single-Input-Multiple-Output (SIMO) fuzzy controller is designed to drive an upgraded clinical real-time Magnetic Resonance Imaging (MRI) system to provide steering forces for an aggregation of ferromagnetic microparticles in the human cardiovascular system according to a pre-set pathway. This kind of endovascular navigation is considered as an important procedure of the catheter-based method for medical treatments against diseases such as some particular types of cancers. The validity of the fuzzy controller has been tested by preliminary simulation results.**

I. Introduction

THE Magnetic Resonance Submarine (MR-Sub) project intends to propel and guide ferromagnetic microparticles properly in human cardiovascular system to deliver drugs through blood vessels from the injection point to the target point for medical interventions such as cancer treatment.

Previous work has already proved that an upgraded clinical Magnetic Resonance Imaging (MRI) platform is capable to provide micro devices with adequate magnetic fields and gradients for endovascular propulsion [1, 2, 3]. Being equipped with a real-time tracking unit, the positioning unit embedded inside the system is able to feedback the coordinates of a tiny microparticle from a three-dimensional (3-D) Magnetic Resonance (MR) image, which would then allow us to perform a closed-loop control along complex blood vessel pathways [4].

A Proportional-Integral-Derivative (PID) controller was chosen to achieve the real-time navigation of a ferromagnetic bead along a pre-defined trajectory [5, 6]. *In-vitro* and *in-vivo* experimental results have shown that real-time PID control is feasible [4, 5, 6]. However, the difficulties of control due to wide range of vessel diameters as well as time-varying environment parameters were also mentioned as probable constraints to any *in-vivo* experiment attempts while navigating microparticles using a PID controller [4, 6].

In this paper, the authors try to find a new approach to navigate microparticles by developing a SIMO fuzzy control system to overcome the shortcomings of traditional control methods. Through the tests based on a fluid dynamic model, preliminary simulation results are given and illustrated.

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II. Modeling

A. MRI machine and sequence

For our upgraded MRI system, the 3-D navigation-capable propulsion pulse sequence for a 2-D control experiment was developed [3]. Fig.1 is a much simplified figure to show how the sequence is processed. In Fig.1, there are some important factors related to the propulsion controller, i.e., *t_track* for the tracking unit of the MRI system to take images and return the current coordinates of a microparticle; the maximum magnetic gradient ∇B to be reached in the next propulsion period and the time *t_maintain* to maintain that maximum gradient; the rising time *t_rise* and the falling time *t_fall* allowing the gradient to go to its maximum in the next prolpulsion period and to drop back to 0 afterwards.

Taking the magnetic gradient $\vec{\nabla B}$, the magnetic force induced by a MRI gradient coil is

$$\vec{F}_{mag} = V_f \cdot (\vec{M} \cdot \vec{\nabla})\vec{B} \quad (1)$$

where $\vec{F}_{mag}$ is the magnetic force (N), V_f is the volume of the ferromagnetic entity (m^3), $\vec{M}$ is the magnetization of the material (A/m), $\vec{\nabla B}$ is the induced magnetic gradient (T/m).

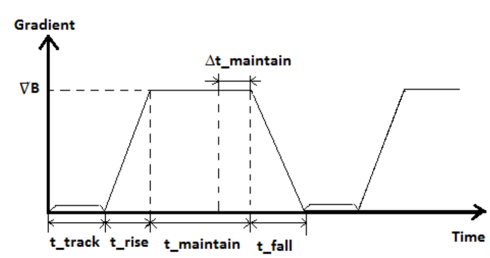


Fig. 1. Overview of a real-time MRI propulsion sequence

B. Motion of Microparticles under Navigation

A coordinate system was set up in our approach as depicted in Fig.2. For modeling inside a blood vessel, the first step is to calculate the Reynolds number to determine the flow type.

In the direction parallel to the x-axis, we assume that the flow is all in one direction and could not be affected by the microparticle due to its small size. We have

$$R_{e\_p} = UD/\nu = UD\rho/\mu \tag{2}$$

where R_{e_p} is the Reynolds number for the flow parallel to the x-axis, U is the average flow velocity in that direction (m/s), D is the diameter of the pipe (m), ν and μ represent the kinematic viscosity and the dynamic viscosity respectively, ρ is the flow density (kg/m^3). In our case, $R_{e\_v} \approx 104.5$ (kg/Pa·m·s^2), thus in the direction along the x-axis, it could be treated as poisseuille flow [7], which means that a numerical relation could always be discovered between the local velocity of a certain point inside the blood vessel and the average velocity. That could be described as

$$u_l = [1-(l/L)^2]\times u_{\max} = 2u_{avg}[1-(l/L)^2] \tag{3}$$

where u_l is a local velocity at a certain point inside the blood vessel, $u_{\max}$ and u_{avg} are the maximum velocity and average velocity of the flow, l stands for the distance from the certain point to the middle of the vessel, and L stands for the radius of the vessel.

In the direction vertical to the x-axis, we have

$$R_{e\_v} = v_y \times 2r \times \rho/\mu \le (F_{mag}/6\pi\mu r 2r)\times 2r\rho/\mu$$
$$= 4M\nabla B\rho r^2/9\mu^2 \tag{4}$$

where R_{e_v} is the Reynolds number for the flow vertical to x-axis, v_y is the steering velocity of the particle (m/s), r is the radius of the particle (m). Here the maximum Reynolds number is $R_{e\_v} \approx 0.1283$ (kg/Pa·m·s^2). Hence, the Stokes Law could be applied to calculate the drag force in the direction vertical to the x-axis.

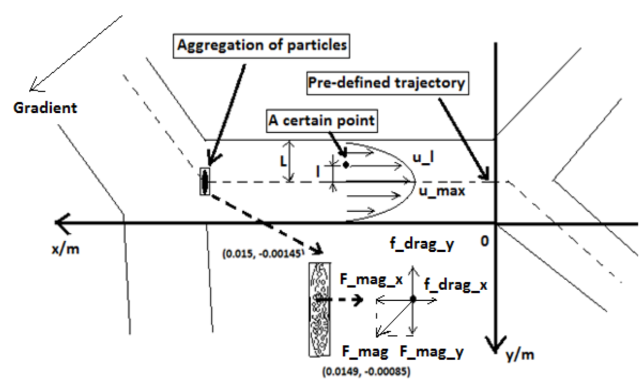


Fig. 2. Intravascular modelling, poisseuille flow and aggregation

According to the results from equations (1), (2), (4), a mathematical model was established in y-axis from the simple Newton's Law to analyze the forces applied on a single particle while under control. We ignore gravity force as it's negligible comparing to other forces:

$$\vec{F}_{mag\_y} - \vec{f}_{drag\_y} = V_f \cdot (\vec{M}\cdot\vec{\nabla})\vec{B} - 6\pi\mu r\frac{d\vec{s}}{dt} = m\frac{d^2\vec{s}}{dt^2} \tag{5}$$

where m represents the mass of the ferromagnetic particle, $\vec{s}$ is the distance that the particle has travelled along the y-axis.

Another important assumption associated in our model is that the microparticles form an ellipse-shaped aggregation with a uniform distribution. The aggregation is close to a needle which is always parallel to the main magnetic field of the MRI machine [1].

III. CONTROLLER

For navigating a single microparticle, the input would be the difference of its x-coordinate and the x-coordinate of a given target waypoint. Obtaining the velocity along x-axis of the particle from the derivative of the input, the controller gives the values of ∇B and $\Delta t\_maintain$ shown in Fig.1 as the two operational outputs to drive the MRI machine in the next propulsion period according to the current position and velocity of the microparticle. Fig.3 is a block diagram that shows the structure of the controller.

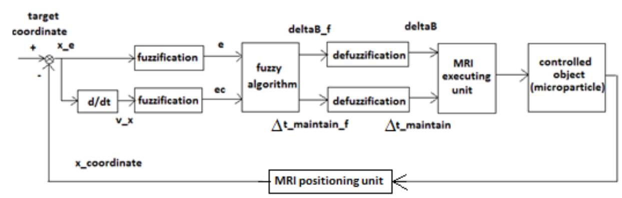
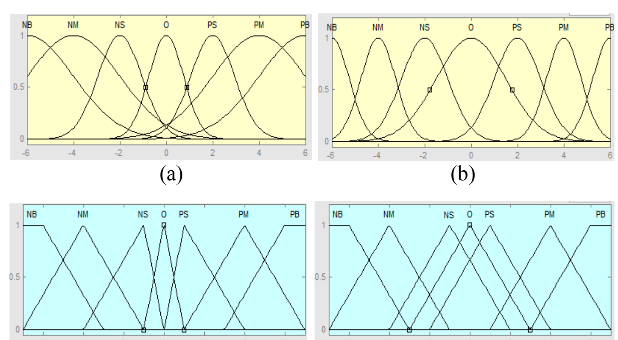


Fig. 3. SIMO fuzzy controller block diagram

The fuzzy sets for the inputs and outputs of the controller are defined as {Negative Big (NB), Negative Medium (NM), Negative Small (NS), Zero (O), Positive Small (PS), Positive Medium (PM), Positive Big (PB)}. The membership function definitions for the two inputs over a field of {-6, -5, -4, -3, -2, -1, 0, 1, 2, 3, 4, 5, 6} are given by Fig.4(a)(b) while those for the two outputs over a field of {-7, -6, -5, -4, -3, -2, -1, 0, 1, 2, 3, 4, 5, 6, 7} are shown in Fig.4(c)(d).



(a) (b)

(c) (d)

Fig.4. Membership functions for inputs and outputs. (a) Functions for x-coordinate error input Δx -E; (b) Functions for its velocity along x-axis v_x -EC; (c) Functions to output magnetic gradient ∇B applied in the next propulsion period -G; (d) Functions to output the gradient maintaining time $\Delta t\_maintain$ in the next propulsion period –T

Since we do not have any coupling between the two outputs, two rule sets R1 and R2 are given described in table I and table II, depending on operators' experience, e.g., the first block in table I indicates that when E and EC are both NB, the outputting gradient will be O.

TABLE I
RULE SET R1 FOR GRADIENT

ec / e	NB	NM	NS	O	PS	PM	PB
NB	O	O	O	O	NS	NS	NM
NM	PS	PS	O	O	O	O	NS
NS	PM	PM	PS	PS	O	O	NS
O	PB	PM	PM	PM	PM	PS	O
PS	PB	PM	PM	PM	PM	PS	O
PM	PB	PB	PM	PM	PS	PS	O
PB	PB	PB	PM	PS	PS	O	O

TABLE II
RULE SET R2 FOR MAINTAINING TIME

ec / e	NB	NM	NS	O	PS	PM	PB
NB	NB	NB	NB	NB	NM	NS	O
NM	NB	NB	NB	NB	NM	NM	NM
NS	O	NM	NB	NB	NB	NB	NB
O	PS	O	NM	NB	NB	NB	NB
PS	PM	PS	NM	NM	NB	NB	NB
PM	PB	PM	NM	NM	NB	NB	NB
PB	PB	PB	NM	NM	NB	NB	NB

The operations for fuzzy operators AND and OR between antecedents are defined as follows: AND simply uses the minimum weight of all the antecedents, while OR uses the maximum value. To obtain the result of a rule, the common "max-min" inference method is used.

The control sets were calculated as follows:

$$R=(E\times EC)\rightarrow G \tag{6}$$

$$\begin{aligned} G &= E\times EC\circ R=(E\times EC)\circ \cup_{i=1}^{n} R_i \\ &= \cup_{i=1}^{n}\{(E\times EC)\circ[(E\times EC)\rightarrow G_i]\} \\ &= \cup_{i=1}^{n}\{[E\circ(E_i\rightarrow G_i)]\cap[EC\circ(EC_i\rightarrow G_i)]\} \\ &= \cup_{i=1}^{n}\{G_{iE}\cap G_{iEC}\}=\cup_{i=1}^{n} G_i \end{aligned} \tag{7}$$

The defuzzification uses the "centroid" method, i.e., to take the weighted arithmetic mean of its membership function as the output to the executing unit of the MRI system, e.g., for the output ∇B, the defuzzification applies

$$g_0=\sum_{i=1}^{n} g_i\mu_g(g_i)/\sum_{i=1}^{n}\mu_g(g_i) \tag{8}$$

where $\mu_g(g_i)$ denotes the membership functions which are shown in Fig.4(c).

IV. SIMULATION RESULTS

Fig.5 shows the input-output surfaces of the controller.

A computer-aided platform was set up to simulate the motion of microparticles when under navigation with MATLAB/SIMULINK/C++ hybrid programming. All the simulation results were obtained based on parameters specified in table III.

Fig.6 plots the different simulated trajectories for microparticles starting from different initial positions. Fig.7 shows the magnetic gradients applied to a corresponding particle during its travel. The 'escape time point' indicates the time at which the x-coordinate of the particle reaches 0.

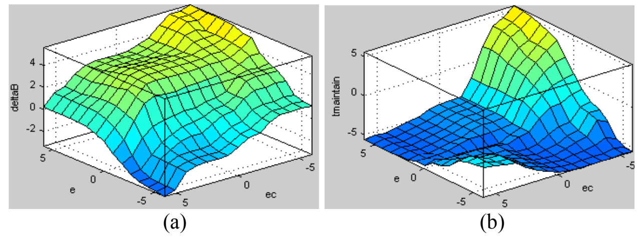


Fig. 5. Input-output surfaces for the SIMO fuzzy controller. (a) Surface for (E×EC)->G; (b) Surface for (E×EC)->T

TABLE III
SIMULATION PARAMETERS

Microparticle		Blood vessel		MRI sequence	
Name	*Value*	*Name*	*Value*	*Name*	*Value*
Diameter (μm)	58	Length (m)	0.015	Tracking time(ms)	30
Mass (kg $\times 10^{-10}$)	4.56	Diameter (m)	0.0023	Acceleration (T/m·s)	40
Magnetic weight percentage(%)	45	Avg flow velocity (m/s)	0.15	Reference maintaining time(ms)	60
Saturation Magnetization (A/m)	401440	Dynamic viscosity (Pa·s)	0.0035	Main magnetic field(T)	1.5
Microparticle density (kg/m^3)	4460	Flow density (kg/m^3)	1060	Maximum magnetic gradient(T/m)	0.4

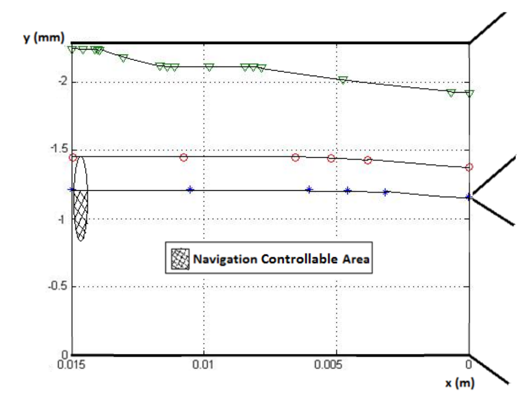


Fig. 6. Simulated trajectories for particles under navigation

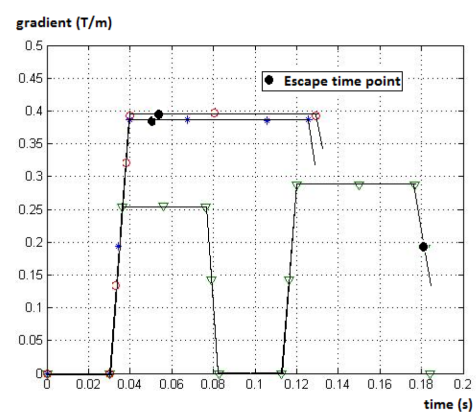


Fig. 7. Magnetic gradient applied to a corresponding particle in fig.6

Also, as stated in the modeling section, we have an assumption that the ellipse-shaped aggregation of microparticles is close to a needle which is always parallel to the main magnetic field. Table IV gives the navigation rates in our simulation as the angle α between the ellipse major axis and y-axis changes.

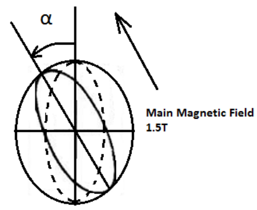


Fig. 8. Rotation of aggregations according to main magnetic field

TABLE IV
NAVIGATION RATE BY ALPHA FROM SIMULATION

α (radian)	Total test samples	Navigated samples	Navigation rate (%)
0	47078	29775	63.2
π/6	47078	30746	65.3
π/4	47078	32299	68.6
π/3	47078	35541	75.5
π/2	47078	47078	100.0

V. DISCUSSION

The control process of a classical fuzzy controller is as simple as a process of table look-up and output. This may bring instability to the system and insurmountable control dead zones due to its limited number of divided classes [8]. However, in our application, since we are not pursuing a 'point to point' servo control, a single fuzzy controller is introduced for its simplicity and independence from environment parameters.

The SIMO fuzzy controller has an output for maximum magnetic gradient maintaining time. This is because that we have confirmed a poisseuille flow model by calculating the Reynolds number. While a particle is travelling at the edge of the flow, its velocity tends to be much lower than that of a particle travelling in the middle, thus the MRI system should have more times for tracking rather than to propel. Hence, the output maintaining time needs to be reduced.

As shown in Fig.2, in our simulation, the magnetic gradient produced by MRI coil for propelling microparticles is designed to have an angle of 45 degrees to the x-axis all the time (if propulsion with steering is desirable). That is to say, the magnetic force induced by the gradient is always trying to pull the particle back in x-axis while propelling in y-axis at the same time, although not significant, so as to obtain more time for navigation. Nonetheless, we still suffer a great loss of particles in one bifurcation. For most of microparticles, the MRI system is only able to track once or twice for its location and velocity, due to the time-consuming tracking process.

There is another important assumption that microparticles will always gather as an ellipse-shaped aggregation with a uniform distribution. Besides, in our model, the interaction force between particles is also assumed to be negligible. In reality, the structure inside the aggregation of microparticles is unpredictable. References [9, 10] show the importance for modeling for aggregations and interaction forces.

VI. CONCLUSION

To steer microparticles inside blood vessels, a SIMO fuzzy controller could potentially be considered as an appropriate controller, according to preliminary simulation results. Despite of certain limitations, this controller has shown its advantages in a fast responding time for a 'real-time' feedback control and its adaptability to cardiovascular navigation with multiple complex, nonlinear, time-varying environment parameters.

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