Suggested Shape for a First Generation Endovascular Untethered Microdevice Prototype

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Abstract – **Shape design optimization for an untethered microdevice meant to navigate in the human bloodstream is studied based on previous variational Finite Element (FE) work done. The suggested shape for a first generation prototype is a prolate spheroid design, thus minimizing the drag and maximising the use of the MRI's magnetic gradients.**

Keywords – **magnetic resonance, magnetic gradient, microdevice, blood circulatory system, magnetic propulsion, fluid dynamics, shape optimization.**

I. INTRODUCTION

As stated in [1], the MR-Sub (Magnetic Resonance Submarine) project plans to use the magnetic gradients generated by clinical Magnetic Resonance Imaging (MRI) systems to propel, track and control, a device containing a ferromagnetic core, in the blood circulatory system. Many future medical applications can be envisioned for such a minimally invasive tool, some of which are explored in the same proceedings by [2].

The goal of the present paper is to set some guidelines for a microdevice shape which could be used as a first generation prototype. The dimensions of the device are dependent of the targeted area as explained in [1]. Nonetheless, we propose to optimally minimize the drag force thus reducing required magnetic gradient produced by the coils of an MRI system. This is a crucial point since we aim at using existing MRI systems without major modifications.

Since the microdevice is propelled by the magnetic gradients from a MRI system, it must be made of some ferromagnetic material and this material will have a tendency to be magnetized in the direction following the MRI's main magnetic field. This may prove to be a problem when the device is moving in a transverse direction. A solution could be to create a free-rotating drag optimized shell around a ferromagnetic core, but given the dimensions of the device, this could imply some microfabrication problems considering it must be made from some bio-compatible material.

We could also factor in the scaling problem: in small blood vessels, blood displays non-Newtonian behaviour. That is, it can no longer be treated as a homogenous fluid and the blood cells have comparable dimensions to the arterial diameter. In turn, this means the device would have to travel next to objects of similar size.

Finally, the device should also be able to carry whatever necessary "equipment" for the surgical task it may be called to perform.

II. STATEMENT

A. Governing equations

Reference [3] showed that the ferromagnetic micro-device is propelled according to:

$$\vec{F}_{magnetic} = \left(\vec{m} \cdot \vec{\nabla}\right)\vec{B} = \left(V_{ferro}\vec{M} \cdot \vec{\nabla}\right)\vec{B}\ , \qquad (1)$$

where $\vec{F}_{magnetic}$ is the magnetic force (N), $\vec{m}$ is the magnetic moment ($A \cdot m^2$), $\vec{B}$ is the magnetic-flux density (T), V_{ferro} is the volume of the ferromagnetic body (m^3) and $\vec{M}$ is the magnetization of the material per unit volume. Since the magnetic force is proportional to the volume of ferromagnetic material, our starting point will be a unit volume. This volume must be optimized as to minimize drag while moving within fluidic cavities such as blood vessels.

B. First approach

In a somewhat intuitive approach, let us consider for simplicity, a blood vessel located far from the heart with a steady flow. This allows us to consider previous studies aimed at drag reduction.

The shape design optimization problem has been studied by numerous authors in a variety of flows. An optimal profile while working with a Stokes flow has been obtained [4]. This study's objective is to determine the shape of a stiff body, *S*, for a given unit volume, which produces minimum drag as the body would move at constant low speed (low Reynolds number, Re) through a viscous fluid. Far from the optimized shape, the flow is considered to have almost uniform speed U_0.

A unique solution is found for *S*, with the addition of the following considerations:

(i) *S* is axisymmetric and has a centre of symmetry

(ii) The front and rear of S must each be tangential to a cone of angle 120^{o}

The solution is a rugby-ball shape, shown in Fig. 1.

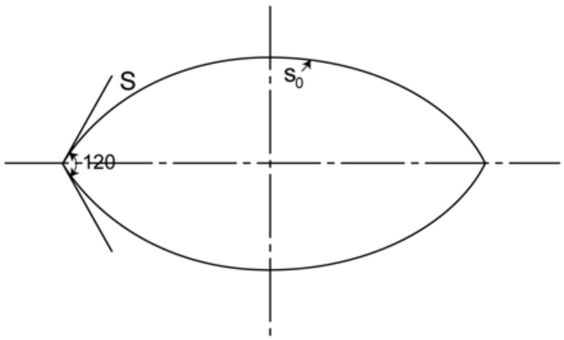

Fig. 1 Prolate spheroid with smallest drag in Stokes flow

A similar solution is also expected when the unit-volume body lies in the center of an infinitely long tube. The optimality condition is, once again, to have a constant normal fluid velocity derivative on S which leads to a slimmer design having the same (axisymmetric) profile as Fig. 1.

As suggested in [5], by linearizing the steady Navier-Stokes equations and applying the same variational method used in the Stokes flow, it is shown that the previous solution is also valid for higher Reynolds number, as long as there is a unique solution to the steady Navier-Stokes equations.

In more recent studies by [6], the minimal drag design is found by introducing a flexible body in a laminar flow. The initial body has a circular geometry: it is essentially a small stiff material located at the center line, surrounded by soft deformable material. The soft material is in fact represented by a B-spline with eight control points. The fluid-structure interaction (FSI) is reflected in the structural deformations which bring the body to its lowest drag shape. The result for $\mathrm{Re}=15$ is an elliptic-like design.

The latter approach could be adapted in future FE simulations at higher flow speed. Let us consider a steady blood flow in the human aorta and let us use:

$$\mathrm{Re}=\frac{\rho\bar{u}d}{\mu}. \tag{2}$$

In (2), Re is the Reynolds number, ρ the blood density ($\rho\approx 1g/cm^{3}$), μ is the viscosity ($0.04\,g/cm\,s$), d is the diameter of the aorta ($d\approx 2.5cm$) and $\bar{u}$ is the average blood velocity estimated to $17cm/s$ (assuming a mean cardiac output of $5L/\min$). With the above data [6], finds $\mathrm{Re}\approx 1063$ but a more realistic pulsatile flow can exceed considerably this value. The transition from laminar to turbulent flow can be estimated at $\mathrm{Re}\approx 2000$. Here the aortic region was considered as fairly turbulent and the aorta is sufficiently large to locally consider a Newtonian viscosity.

This worst-case scenario also shows that the above-considered studies are an excellent starting point to our initial problem since the blood flow isn't always turbulent. A consistent FE model should take into account many more parameters, such as the varying viscosity and some degree of turbulence.

III. SIMULATIONS

A simple prolate spheroid is described by [8]:

$$\frac{x^{2}+y^{2}}{a}+\frac{z^{2}}{c}=1, \tag{3}$$

with $c>a$. The optimal $120°$ angle imposed by [4] could be increased in our particular application to decrease the risk of damaging the inner walls of the blood vessels.

A slim prolate spheroid design needs to be experimentally tested in order to find a and c, the optimum parameters in (3), in accordance with the simulated blood pathways. A blood analog which mimics the mechanical properties of real blood, such as non-Newtonian behaviour, viscosity, etc. should be used in a pulsatile flow contained, preferably, in an elastic tube with flexible walls. A more complex installation could encompass some arterial bifurcations, thus aiding in the qualitative determination of the polar radius c. This must be repeated as to account for the various dimensional scales found in the circulatory system.

A numerical simulation was done using Simulink, by adapting a PID controller simulation model developed for a spherical bead [9]. Three forces are considered to act on this bead: the magnetic gradient (1), a drag force (unbounded region) and the buoyancy force. The prolate spheroid (Fig. 1), which produces only 91% of the drag associated with a sphere of equal volume [4], was compared to the spherical design presently used in the MR-Sub project.

The first simulation (Fig.2), represents an open-loop PID control session and, as expected, the spherical prolate design (dotted line) reaches a higher speed than a sphere.

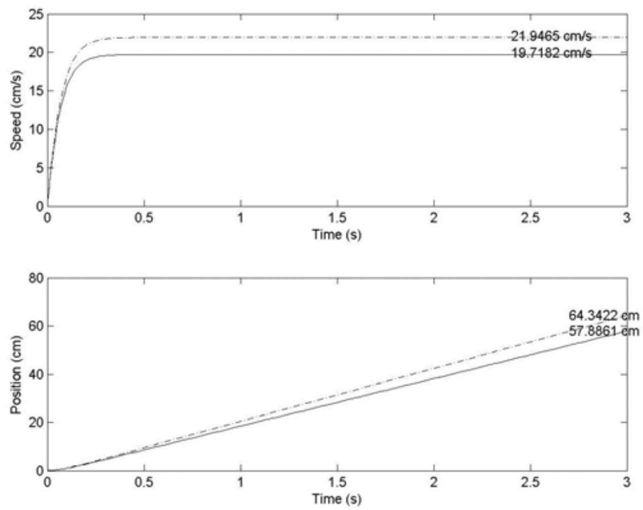

Fig. 2 Position and speed vs time for an open-loop simulation

Obviously, similar results are expected in a closed-loop circuit (Fig. 3), where a target of $0.1\,m$ is reached, starting from the origin.

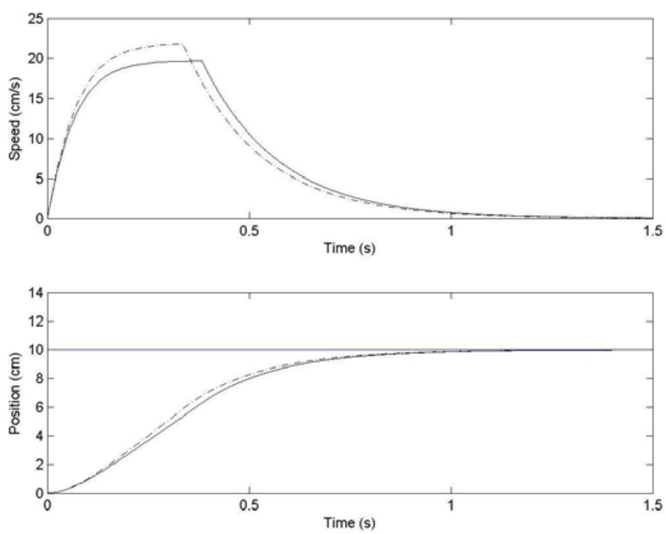

Fig. 3 Position and speed vs time for a closed-loop simulation

Once again the spherical prolate (dotted line) achieved higher speed and it also reached to the target, slightly faster compared to a sphere.

IV. Future Directions

The MR-Sub context proves to be more complex than the situations depicted earlier but, they are nonetheless an interesting starting point. The shape of the micro-device is also submitted to other constraints [1] such as, but not limited to, the possibility to navigate through arterial bifurcations and mainly, to the orientation of the magnetic field [2] and to the task submitted to the device.

As stated earlier, the shape design is an important feature since the driving force is comparable to the drag force, such that the use of clinical MRI systems without major modifications is difficult, unless the shape is optimized.

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