Nature-inspired microfluidic propulsion using magnetic artificial cilia

Khaderi, Syed Nizamuddin

Chapter 10

Summary

This work has addressed the computational design of magnetically-driven artificial cilia for microfluidic propulsion. The objective of the thesis, as mentioned in section 1.5, was two-fold:

1. To identify under what conditions a magnetic film can mimic the asymmetric beat motion of natural cilia.

2. To explore how the flow created by the artificial cilia can be controlled and optimised.

The first question is answered in chapter 3 and the second in the subsequent chapters.

In this thesis a coupled solid-fluid magneto-mechanical model has been developed that captures the deformation of the cilia, the distribution of the magnetic field in and outside the cilia, and the resulting motion of the fluid (chapter 2). The results are shown as a function of the physical parameters that govern the flow created by the cilia in addition to geometric parameters such as the cilia spacing (a), channel height (H) and phase difference between the cilia ($\Delta\phi$). The four physical parameters involved are (i) the magnetic number M_n - the ratio of magnetic to elastic forces, (ii) the fluid number F_n - the ratio of fluid viscous forces to the elastic forces, (iii) the Reynolds number Re - the ratio of fluid inertia forces to the viscous forces and (iv) the inertia number I_n - the ratio of inertia forces of the cilia to their elastic forces.

In Chapter 3, we have shown several configurations that can mimic the motion of natural cilia with pronounced asymmetry. One configuration is based on a curled permanently magnetic cilium. When a magnetic field is applied that is opposite to the magnetization of the permanently magnetic cilium, it undergoes a buckling-kind of instability, mimicking a recovery stroke. When the magnetic field is switched off, the cilium performs the effective stroke and returns to its initial curled position because of elastic forces. Another configuration is based on super-paramagnetic cilia that have a tapered cross section. When these cilia are subjected to a rotating magnetic field, they follow the applied magnetic field and stay straight, performing an effective stroke. On further rotation of the magnetic field, the magnetic couple distribution near the fixed end forces the cilia to come back to the initial position, while the couple near the free end makes the cilia to stay closer to the channel wall. As the cross section is tapered, the magnitude of the couple near the fixed end is larger than at the free end. This causes the cilia to come back with a whip-like recovery stroke. In these configurations the recovery stroke takes place at much higher velocity than the effective stroke. The fluid flow created by the cilia was found to scale linearly with the area swept by the tip of the cilia. We subsequently studied the area swept by the cilia as a function of the dimensionless numbers in the limit of low Reynolds numbers. It was found that a larger magnetic number was needed to sweep a given swept area when the fluid number was increased, and that the inertia number does not significantly influence the area swept by the cilia. Also, when the fluid number was increased, it took a long time for the cilia to come back to their initial position.

The pressure and flow generated by the cilia as a function of cilia spacing and channel height were studied in chapter 4. The analysis was performed using two channel configurations that are of practical importance - an open-loop channel and a closed-loop channel. When the cilia spacing decreases, the resistance offered by the fluid decreases and this causes the fluid flow to increase. When the channel height was increased, the cilia create a Couette flow in the channel that leads to a linear increase in the flow with the channel height. When the cilia spacing is decreased for a given channel length, many cilia are present per unit length of the channel. Consequently, the pressure generated by the cilia increases. On the other hand, when the channel height is increased, the pressure generated drastically decreases. This chapter provides guidelines for the cilia spacing, length and channel height for optimal performance.

The effect of fluid inertia on the flow generated was studied in chapter 5. The fluid inertia has two effects; firstly, the momentum diffusion from the cilia into the channel is delayed and secondly, a part of the energy input to the fluid during the effective stroke is retained after the effective stroke has completed. These effects enrich the fluid-structure interaction of the artificial cilia. The artificial cilia also possess a temporal asymmetry by performing a slow effective stroke and a quick recovery stroke. When the inertial forces are larger than the viscous forces, the flow created is unidirectional and larger than that of the Stokes regime. Interestingly, for certain sets of parameters, due to the temporal asymmetry and large inertia forces, the fluid transported can be in the direction of the recovery stroke. Moreover, we found that the direction of the fluid flow can be changed by simply changing the operation frequency of the cilia.

As the flow created by the artificial cilia depends on the fluid inertia, spatial, temporal and orientational asymmetry, we used a model problem to probe the effect of each of these parameters individually in chapter 6. It was found that, irrespective of the Reynolds number, the fluid flow scales linearly with the swept area. At high Reynolds numbers the fluid transported is unidirectional and larger than that of the Stokes limit, which can be further enhanced by a fast effective stroke. At high Reynolds numbers the orientational asymmetry is sufficient to cause a fluid transport. The effects of the three symmetries were generalised into a definition of configurational symmetry, whose absence might create a flow in microchannels.

In chapter 7, we investigated the effect of the out-of-phase motion of the cilia on the fluid flow using two approaches: by externally prescribing a phase difference between adjacent cilia and by applying a non-uniform magnetic field. In the former case the area swept by the cilia remains nearly the same for all phase differences. In the latter case the cilia sweep a larger area when the applied magnetic field travels in the direction opposite to the effective stroke (antiplectic metachrony) compared to the situation when the magnetic field and the effective stroke are in the same direction (symplectic metachrony). The out-of-phase motion of the cilia results in a pressure distribution in the channel that causes a local vortex formation during the recovery stroke, resulting in a zero negative flow. This leads to an unidirectional flow with a subsequent increase in the net fluid transported. When the area swept remains constant, the cilia create a larger flow which is unidirectional and independent of the phase difference for large cilia spacings. However, for small cilia spacings, they create a larger flow for antiplectic metachrony compared to symplectic metachrony.

The fluid transport due to the out-of-phase motion of symmetrically beating cilia was studied in chapter 8. The out-of-phase beating of cilia leads to a collective non-reciprocal motion, which leads to a net pressure gradient in the direction of the wave and a fluid flow in the opposite direction. The fluid flow is created mainly due to the unidirectional displacement of fluid particles near the cilia, leading to a net flow that is also unidirectional. The flow was found to be maximum when the wavelength of the applied magnetic field is four times the cilia length.

To analyse the effect of the cilia width and out-of-plane motion of the cilia, a three-dimensional numerical method was developed in chapter 9. The cilia are modelled using shell elements and the fluid using boundary elements, while the magnetic field was calculated using the magneto-static boundary conditions. Using this model we verified that an asymmetric motion of the cilia can also be generated using a tapering in the width. The width of the cilia and their spacing in the width direction were investigated for the maximum flow condition. We found that per unit width of the channel and for a given tapering, many narrow cilia spaced close together create a larger flow compared to a few broad cilia. In chapter 7, we analysed the flow caused by cilia that were beating out-of-phase in the beat plane. The three-dimensional model allowed us to study the out-of-phase beating perpendicular to the beat plane. We found that such a motion causes a flow normal to the beat plane which is of comparable magnitude as the flow in the plane of beat. This suggests a way to simultaneously mix and propel fluids in microchannels.

The cilia-based fluid manipulation systems in the recent past are summarised in table 10.1 with the aim of classifying the mechanisms that are responsible for the fluid transport. It can be seen that in the majority of the cases the actuation is performed through a magnetic field. The cilia that were able to successfully demonstrate a flow relied on (i) three-dimensional motion (items 2, 4, 5 and 10 in table 10.1) – mimicking the motion of nodal cilia – (ii) a combination of temporal and orientational asymmetry (item 1 in table 10.1), (iii) spatial asymmetry (items 7, 9 and 11 in table 10.1) and (iv) orientational asymmetry (items 5 and 12 in table 10.1). The orientational asymmetry was used in combination with inertial forces and non-reciprocal motion by Fahrni *et al.* (2009) and Alexeev *et al.* (2008*b*), respectively.

We would like to conclude by providing guidelines on the design of artificial ciliary systems based on the studies performed in this thesis. Let us start by assuming that the material properties of the cilia, the channel height and the fluid viscosity are known. The design parameters are assumed to be the geometry of the cilia (length, thickness and width), magnetic field, cilia spacing and frequency. As we want the cilia to operate under moderate pressure heads, we propose the length of the cilia to be half that of the channel height (chapter 4). From chapter 5, we find that for a given magnetic number, the cilia create large and unidirectional flows for $\mathrm{Re} > 1$ and $F_n < 1$. Using the first condition, we can find the frequency of operation. Using the second condition (on F_n), we can arrive at the ratio of the cilia length to thickness (the aspect ratio). Once we know the aspect ratio, we can find the magnetic field to be applied using the magnetic number.

Table 10.1: *Artificial cilia in microfluidics. E and T refer to experimental and theoretical work.*

No.	Reference	Force field	Length scale	Frequency	Asymmetry	E / T
1	den Toonder *et al.* (2008)	Electrostatic	100 μm	80 Hz	Temporal and orientational	E
2	Evans *et al.* (2007); Shields *et al.* (2010)	Magnetic	10 μm	35 Hz	3D motion	E
3	Oh *et al.* (2009)	Base excitation	400 μm	100 Hz	None	E
4	Vilfan *et al.* (2010)	Magnetic	30 μm	1 Hz	3D motion	E, T
5	Fahrni *et al.* (2009)	Magnetic	300 μm	50 Hz	Orientational and 3D motion	E
6	van Oosten *et al.* (2009)	Photo actuation	10 mm	0.04 Hz	Spatial	E
7	Belardi *et al.* (2011); Hussong *et al.* (2011*b*)	Magnetic	70 μm	20 Hz	Spatial	E
8	Timonen *et al.* (2010)	Magnetic	6 mm	Not reported	3D motion	E
9	Gauger *et al.* (2009)	Magnetic	-	-	Spatial	T
10	Downton & Stark (2009)	Magnetic	-	-	3D motion	T
11	Kim & Netz (2006)	External motors	-	-	Spatial	T
12	Alexeev *et al.* (2008*b*)	Point force	-	-	Orientational	T
13	Ghosh *et al.* (2010)	Point force	-	-	None	T