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An AC Magnetohydrodynamic Micropump: Towards a True Integrated Microfluidic System

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ABSTRACT

An AC Magnetohydrodynamic (MHD) micropump has been demonstrated in which the Lorentz force is used to propel an electrolytic solution along a microchannel etched in silicon. This micropump has no moving parts, produces a continuous (not pulsatile) flow, and is compatible with solutions containing biological specimens.

INTRODUCTION

Micropumps are a critical component for any microfluidic system. Transport of samples and reagents for biological analysis requires the presence of a micropump. To date, work in micropumps can be categorized into two types: mechanical and non-mechanical.

Most mechanical micropumps have the presence of a membrane that can be actuated in different ways: electrostatically [1], pneumatically [2], thermopneumatically [3], piezoelectrically [4, 5], and electromagnetically [6]. In all cases, high voltages are needed to actuate these membranes which produce pulse flow in channels. An advantage of all these different types of membrane pumps is the ability to pump any kind of liquid.

Non-mechanical pumps on the other hand have no moving parts and operate on a variety of mechanisms depending on the intended application. Non-mechanical pumps that have been demonstrated include: electrohydrodynamic pumps [7] to pump dielectric liquids, electrokinetic pumps [8] using electroosmotic and electrophoretic effects for molecular separation, and bubble pumps [9] using any type of liquid.

For biological applications, electrokinetic pumping is by far the most commonly utilized. Although the microfluidic channels for electrokinetic pumping have been reduced in size, high voltage requirements across the channels prevent the microfluidic system from being portable. Macroscopic syringe pumps are also used to pump liquids through microchannels in biological applications.

We have demonstrated a new type of non-mechanical micropump, using the Lorentz force as the pumping mechanism for biological analysis. The AC Magnetohydrodynamic (MHD) micropump investigated produces a continuous flow and allows for complex microchannel design.

THEORY OF OPERATION

Macroscopic MHD pumps have been used for some time to pump liquid metals [10]. In our micropump, an AC electrical current and a perpendicular, synchronous AC magnetic field pass through an electrolytic solution, transverse to a microchannel. This produces a Lorentz force along the channel, causing the solution to flow. This geometry is illustrated in Fig. 1.

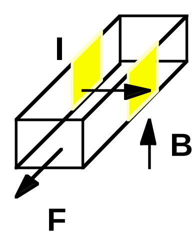

Figure 1. Vector-diagram of MHD pump

The Lorentz force produced is given by

$$\vec{F} = \vec{I} \times \vec{B} w \quad (1)$$

where w is the distance between the electrodes. This results in a pressure given by

$$P = IB / h \quad (2)$$

where h is the height of the electrode. In microchannels, flow is governed by Poiseuille's law,

$$P = QR \quad (3)$$

where Q is the volumetric flow rate and R is the resistance to flow dependent on the geometry of the channel. For rectangular channels, the fluidic resistance is given by

$$R = \frac{16\mu L (w+h)^2}{w^3 h^3} \quad (4)$$

where L is the total length of the channel and μ is the viscosity of the fluid. Substituting (4) into (3) and then equating with (2) gives the flow rate as

$$Q = \frac{IBw^3h^2}{16\mu L(w+h)^2} \quad (5)$$

An MHD micropump using a DC current and a permanent magnet was recently described by Jang and Lee [11]. In such a device, the same electrolytic reaction that enables current conduction also produces gas bubbles that impede fluid flow and causes electrodes to dissolve. This limits the practical application of such a device to non-electrolytic conductors, such as liquid metals. When an AC current of sufficiently high frequency is passed through an electrolytic solution, the chemical reactions are reversed rapidly enough that bubbles never have a chance to form and no electrode degradation can occur. In this case, the time-averaged Lorentz force depends on the phase of the magnetic field, relative to the electrode current, and is given by

$$F = IBw\int_0^{2\pi} \sin\omega t \sin(\omega t + \phi)d\phi \quad (5)$$

The integrand can have a value from -1/2 to 1/2.

FABRICATION

The devices are fabricated by etching a V-groove through a silicon wafer, depositing metal electrodes on an oxide layer that continue down the walls of the groove, and then anodically bonding the wafer between two plates of glass. Holes are drilled in the top glass to allow for contacting the electrodes and for fluidic input and output. A photograph of a fluidic circuit in which liquid is pumped about a closed loop is shown in Fig. 2. An electromagnet, positioned beneath the device, produces a magnetic field normal to the plane of the wafer as shown in Fig. 3. A photograph is shown in Fig. 4 illustrating the MHD system and the mini-electromagnet used.

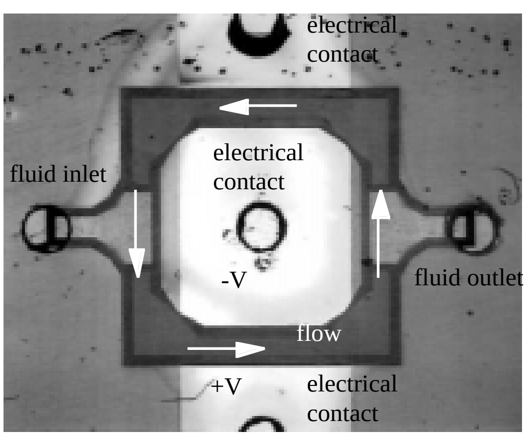

Figure 2. Top view of circular pump. Channel depth of 380 µm and top width of 800 µm. Electrode width of 4mm.

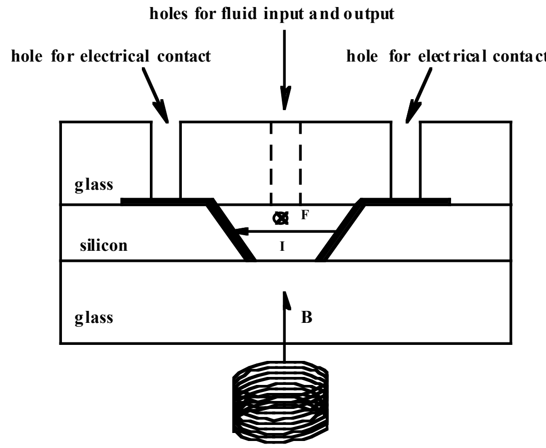

Figure 3. Cross-section of AC MHD micropump set-up. The Lorentz force produced is directed into the paper

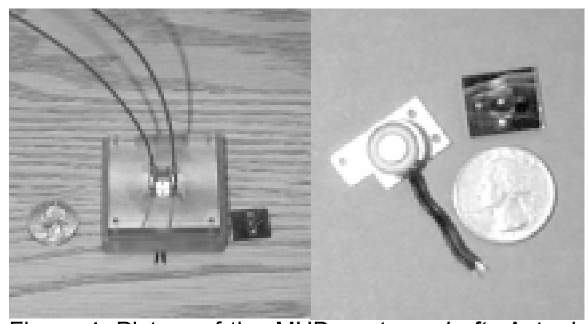

Figure 4. Picture of the MHD system. *Left* Actual MHD package. *Right* The mini-electromagnet and device used.

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EXPERIMENTAL RESULTS

To determine the allowed operating conditions for various solutions, a microchannel was observed under a microscope as current and frequency were varied. At each frequency, the maximum allowed current was determined at which no bubbles were evident. For a typical device, Fig. 5 shows the current, as a function of frequency at which bubbles first appear in a 1M NaCl solution. At 1 kHz, where we have chosen to operate, an AC current amplitude of 90 mA can be passed through the solution with no bubble formation. In addition, 0.1M NaCl, 0.01M NaCl, 0.01M NaOH, and Phosphate Buffered Solution (PBS pH=7.2), were found to have maximum allowed currents at 1kHz of 48, 29, 23, and 13 mA respectively.

The fluidic circuit shown in Fig. 2 was filled with a 1M NaCl solution containing 5µm diameter polystyrene beads. The flow velocity was measured by tracking the motion of the beads using a computer with video capture software. Figure 6 shows the evolution of three beads between two frames 0.87 seconds apart. With a fixed electrode current amplitude of 75 mA, flow velocity was measured as a function of electromagnet current. This is plotted in Fig. 7. With both electrode and electromagnet

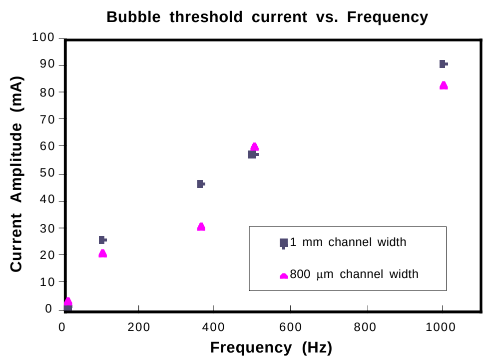
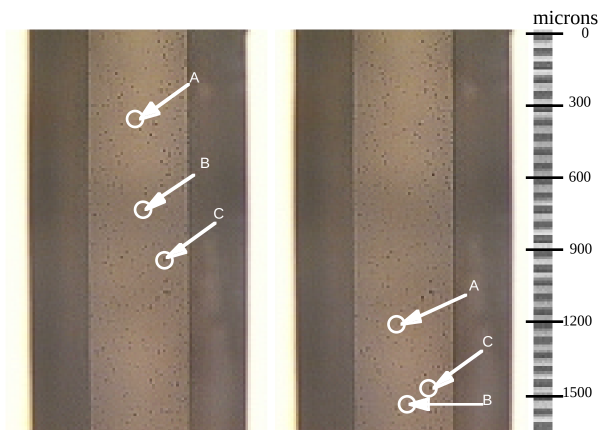


Figure 5. Current at which bubbles first appear as a function of frequency using 1 M NaCl solution for different channel widths.



t = 0 t = 0.87 sec

Figure 6. Video capture of 5 µm polystyrene beads flowing through a microchannel 800 µm width. Flow velocities for the three particles measured in mm/s are : A = 1.06, B = 1.02, C = 0.67

current amplitudes held constant, flow speed and direction could be changed by adjusting the relative phase between the two currents. These measurements are plotted in Fig. 8. Maximum flow could be achieved in opposite directions at 0˚ and 180˚ relative phase, and no flow was observed at 90˚ relative phase.

Again using the same device of Fig. 2, a bead velocity of 1.06 mm/sec was observed with an electrode current of 75 mA and electromagnet current of 375 mA, corresponding to a magnetic field of 251 gauss at the surface of the magnet (quite a bit lower in the channel). If this is assumed to be the average flow velocity, the volumetric flow rate is calculated to be 13 µl/min. Using 0.1M NaCl, 0.01M NaCl, 0.01M NaOH, and PBS pH=7.2, flow velocities of 1.48, 0.52, 0.30, and 0.35 mm/sec, respectively were measured in another device having a 380 µm depth and a 1.1 mm top channel width. This corresponds to volumetric flow rates of 37, 13, 7.5, and 8.7 µl/min.

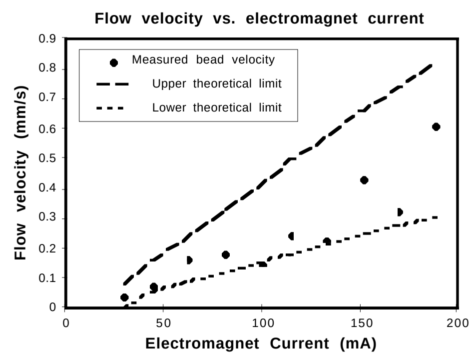


Figure 7. Measured bead velocity as a function of electromagnet current. The upper and lower theoretical limits correspond to the measured magnetic field beneath the device and above the device respectively.

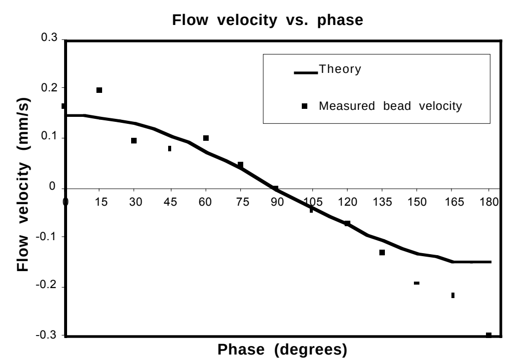


Figure 8. Measured bead velocity as a function of phase. Settings are: frequency at 1 kHz, current amplitude across channel at 75 mA, electromagnet current amplitude at 135 mA.

CONCLUSIONS

We have demonstrated a novel AC Magnetohydrodynamic micropump which produces a continuous flow, has no moving parts, and is compatible with solutions containing biological specimens. Pumping was demonstrated in several different electrolytic solutions.

Using a single electromagnet, multiple pumps on the same chip can be driven independently by varying their electrode current amplitude and phase relative to the magnet, thus enabling routing in complex integrated microfluidic systems.

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