DESIGN & EXPERIMENTATION OF HIGH CURRENT DENSITY DC MAGNETOHYDRODYNAMIC (MHD) MICROPUMP

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Bao Thanh Nguyen

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The Undersigned Faculty Committee Approves the

Thesis of Bao Thanh Nguyen:

Design, Fabrication, & Experimentation of DC MHD Micropump

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Approval Date
ABSTRACT OF THE THESIS

Design, Fabrication, & Experimentation of DC MHD Micropump
by
Bao Thanh Nguyen
Master of Science in Bioengineering
San Diego State University, 2008

A major challenge for integrated Lab-on-a-Chip (LOC) systems is the precise and consistent control of fluid flow. Magnetohydrodynamics (MHD) micropumps which contain no moving parts and capable of generating a continuous flow in any ionic fluid offer an ideal solution for biological applications. MHD micropumping has been demonstrated using both AC and DC currents by a number of researchers with varying degrees of success. However, current MHD designs based on DC (Direct Current) do not meet the flow rate requirements for fully automated LOC application (> 100 ul/min). Gas bubbles, byproducts of electrolysis, particularly hydrogen gas are the main contributor to low experimentally observed flow rate. These tiny bubbles coalesce in the main flow channel and hinder fluid’s flow. Since hydrolysis is inevitable under direct current (DC) excitation, mitigating its negative effects could significantly improve MHD micropump’s performance. In this research, we introduce a novel MHD micropump which effectively increases flow rate by deploying bubbles isolation and releasing mechanism that limit electrolyzed gas’ disruption of flow. BIRS (Bubble Isolation and Release System) MHD, with a flow channel of 800 µm x 800 µm cross-section and 6.4 mm length, could pump 330 µL/min of 1 M NaCl solution at 5 Volts DC. This voltage corresponds to a current density of approximately 200,000 A/m² on the surface of platinum electrodes with length of 6.4 mm, diameter of 200 µm and resistance of 3 Ohms.
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CHAPTER 1

INTRODUCTION

An efficient microfluidic system is a crucial element in designing a successful Lab-On-a-Chip system. Lab-On-a-Chip (LOC) refers to a single Microelectromechanical System (MEMS) device containing multiple laboratory procedures that is capable of analyzing biological or chemical assays at very small scale. Typically LOC systems handle liquid that range from pico-liters to milli-liters in volume. Besides the precise control of fluid flow, low actuation voltage (less than 10 Volt) and high flow rate (greater than 100 µL/min) are two important criteria in micropump designs. Low actuation voltage allows LOC systems to be portable and cost effective. Mean while high flow rates increase speed at which samples are analyzed and better sample preparation.

Current micropumps operate by propulsion forces generated by either mechanical or non-mechanical means. Within these two major classes of micropumps, there are further subcategories describing the physics involved [1]. Mechanical micropumps require moving parts to exert force on fluid, usually in the form of elaborated microgears, valves or flexible membranes. These high aspect ratio and intricate features are challenging to miniaturize. Clogging and fatigue failures are also prevalent. Due to physical limitations of micromachining processes, mechanical micropumps are difficult and expensive to fabricate. By physically acting the fluid, mechanical micropumps are capable of generating higher forces, hence higher pressure head and flow rate, than most non-mechanical micropumps.

On the other hand, non-mechanical micropumps utilize electrical and/or magnetic forces to pump. Fluid’s electrochemistry and electrical properties are imperative in determining pump’s behavior and performance; while such is not the case in mechanical micropumps. Non-mechanical pumps depend on the microelectrodes, of various configurations, that are in direct contact with fluid to propel fluid. The lack moving parts makes these micropumps significantly cheaper and easier to fabricate than the mechanical counterparts. No mechanical parts often translate into longer life cycle. However, the
governing fluid dynamics and electrochemistry are much more complex and not well characterized.

Non-mechanical micropump’s advantages eclipse its shortcomings making it the preferred choice for LOC applications. Within the realm of non-mechanical micropumps, there are six major classes: electrohydrodynamic (EHD), electrokinetic (EK), electrochemical, phase transfer, electro wetting, and magnetohydrodynamic (MHD) [2]; each with its own pros and cons. EHD, EK, and MHD are most actively pursued for LOC applications due to their less disruptive nature on the transported material.

In EHD, the interaction between an electric field and a dielectric fluid exerts an electrostatic force on ions within the fluid of interest. There are several methods to produce induced charges required to create “dielectric fluid”. Net body force generated is the sum of Coulomb’s force, Kelvin polarization force, dielectric force, and electrostrictive force according to equation (1) in their respective order. Most published EHD micropumps utilize Coulomb’s force \((qE)\) as the main driving force. There are a few reported usages of Kelvin polarization forces \((P \nabla E)\). Actuation voltage for reported EHD micropump ranges from 40V to 250V with flow rate ranging from 0.12μL/min to 14mL/min. High actuation voltage makes EHD micropump not a viable option for LOC application.

\[
F = qE + P \nabla E - \frac{1}{2} E^2 \nabla \varepsilon + \frac{1}{2} \nabla \left( E^2 \rho \frac{\partial \varepsilon}{\partial \rho} \right)_T
\]

\(\varepsilon\) – fluid’s permittivity  
\(\rho\) – fluid’s density  
\(E\) – electric field  
\(P\) – fluid’s polarization vector  
\(q\) – charge density  
\(T\) – fluid’s temperature
Electroosmotic (EO), which belongs to the electrokinetics (EK) category, requires charged surfaces at the solid-liquid interface to be operational. These charged surfaces attract their counter ions to accumulate near the solid-liquid interface; thus creating an electrical double layer. Applied potential, hence electric field, would cause migration of ions toward their counter electrode while dragging the bulk of the fluid along the boundary of the electrical double layer. Maximum theoretical flow rate for EO pump is described in equation (2) 

Reported EO micropumps’ applied voltage ranges from 40V to over 6000V DC with flow rate of 0.006μL/min to 7mL/min. Similar to EHD, high actuation voltage prevents EO micropumps from being a competitive option for LOC application.

Maximun flow rate

\[ Q = \frac{\pi a^4}{8 \mu l} \Delta p - \frac{\pi a^2 \epsilon \zeta E_z}{\mu} f \left( \frac{a}{\lambda_D} \right) \]  \hspace{1cm} (2)

With

\[ f \left( \frac{a}{\lambda_D} \right) = 1 - \left( \frac{2I_1 \left( \frac{a}{\lambda_D} \right)}{a / \lambda_D I_0 \left( \frac{a}{\lambda_D} \right)} \right) \]

Debye’s shielding length is given as

\[ \lambda_D = \left[ \frac{\epsilon k T}{e^2 \sum_i z_i n_{e,i}} \right]^{1/2} \]

Where

\( \varepsilon \) – fluid’s permittivity \hspace{1cm} \( \zeta \) – zeta potential drop

\( \mu \) – fluid’s viscosity \hspace{1cm} \( \lambda_D \) – Debye’s shielding length

\( a \) – pipe’s radius \hspace{1cm} E_Z – electric field in axial direction

\( e \) – electron’s charge \hspace{1cm} k – Boltzmann constant

\( Q \) – flow rate \hspace{1cm} \( \Delta p \) – pipe’s pressure drop

\( T \) – fluid’s temperature

\( I_0, I_1 \) – zero and first order modified Bessel functions of the first kind

- Figure 1. Typical ElectroHydrodynamic (EHD) micropump [2].
Figure 2. Typical Electroosmotic (EO) micropump [3].

In 1942 by Hannes Alfven observed a phenomenon in plasma physics and described it as electromagnetic hydrodynamic wave [4]. MHD phenomenon is governed by a special case of Maxwell’s equation known as Lorentz force. This force is generated by the interaction between orthogonal electrical current and magnetic field. Lorentz force, acting upon the bulk fluid, is perpendicular to the plane formed by the orthogonal magnetic and electric fields (Figure 3). Idealized MHD micropump’s behavior can be predicted using Lorentz force of Maxwell’s equation, Navier-Stokes’s equation for incompressible flow, and conservations of mass equations. Reported flow rate for MHD micropumps range from 0.002μL/min to 6mL/min with excitation voltage of 10V to 60V. Published reported values for power consumption and performance demonstrated that MHD is a viable option for LOC application.

Lorentz Law (Maxwell’s equation)

\[ F_L = (J \times B) \cdot L_c \]

Where

\[ J = \sigma \cdot (E + U \times B) \]

Navier-Stokes Equation for a Newtonian incompressible fluid
\[ \rho \left( \frac{DU}{Dt} \right) = -\nabla p + \mu \nabla^2 U + F_L + F \]

Conservation of mass

\[ \frac{\partial \rho}{\partial t} + \nabla (\rho U) = 0 \]

Where

- \( \mu \) – viscosity of fluid
- \( \rho \) – fluid’s density
- \( \sigma \) – electrical conductivity of medium
- \( B \) – magnetic flux density
- \( E \) – electric field
- \( F \) – Body Force
- \( F_L \) – Lorentz force
- \( J \) – current density
- \( L_e \) – length of electrode
- \( p \) – initial pressure
- \( U \) – velocity field of charges
- \( t \) – time

**Figure 3. Typical MagnetoHydrodynamic (MHD) micropump.**

MHD micropumping is a feasible option. However to be fully functional and useful for LOC application, secondary phenomena such as electrolysis, electrokinetic cross flow, and Joule heating need to be addressed. Out of the secondary effects mentioned above, electrolysis is the major most disruptive force to MHD micropump’s performance. Hence, the main focus of this research is the optimization of MHD pumping by minimizing the adverse effects caused by electrolysis (Figure 5). BIRS (Bubble Isolation and Release
MHD has full-length electrodes embedded within the peripheral channels flanking the main flow chamber. Current continuity is maintained via narrow slits at the bottom of the walls separating the compartments. In this manner, the electric field distribution and current density, hence the Lorentz force, could be manipulated by varying the geometry and location of the slits. The solid surfaces above the slits provide favorable locations for electrolyzed bubbles to aggregate [5] and vertically migrate to venting chamber above the pump. For LOC system integration, venting chamber could be hermetically sealed using a semi-permeable seal such as hydrogels, dialysis membranes, or low porous glass frits. Four other designs that eventually led to the BIRS design are also discussed to establish the intended functionality of the BIRS design. Due to cost and time constrains, MHD micropumps presented here are fabricated from polysulfone using CNC (Computer Numerical Control) techniques rather than conventional silicon or glass micromachining. In integrated LOC application, an array of BIRS MHD could be micromachined to increase flow output.

Figure 4. CAD drawings of BIRS MHD Micropump.
Platinum electrodes are situated most lateral from the main flow channel within the electrode compartments. Venting chambers collects electrolyzed gas bubbles and releases them to atmospheric pressure conditions. Narrow gaps along the walls separating flow cell from electrode compartments ensure electrical conduction. The walls provide energetically favorable location for electrolyzed bubbles to accumulate and to be released to venting chamber above the fluid level.

**Figure 5. Micrograph of BIRS MHD Micropump.**
CHAPTER 2

LITERATURE SURVEY

MHD pumping is a relatively new research area for microfluidics despite its success at the macro level [6,7,8]. MHD macropumps are used in metallurgy, power plants, and propulsion engine. Significant interest in MDH micropump development for Lab-on-a-Chip application stems from its low actuation voltages, continuous flow, and lack of moving parts [3]. Absence of moving parts reduces risk of physical damage to transported biological materials such as cells, DNA, RNA, etc, and increase pump’s operational life cycle as well. Bulk micromachining of silicon using LIGA (Lithographie Galvanoformung Abformung), ICP-RIE (Inductively Coupled Plasma Reactive Ion Etching), and anisotropic wet etching together with soft lithography have been used to micromachine MHD micropumps for microsystems. Functional feasibility has been demonstrated by using AC and/or DC current by Lemoff [9], Jang [10], Bao [11], Jeong [12], Homsy et al [13], Wang [14], and Heng [15] with varying successes. Reported micropumps vary from 2mm x 10mm [14] to 10 μm x 10 μm [1] in cross sectional area. Applied magnetic field and current runs the whole gamut from 13 mT [14] to 2.2 Tesla [15] and 1.8 mA [8] to 100 mA [9,11,14], respectively. The observed flow rate ranges from 0.002 μL/min [11] to 6 mL/min [15]. Due to large variation in sizes and applied magnetic field and electric current among the published works, it is difficult to compare the different designs.

The previous attempt at minimizing effects of electrolysis was reported by Homsy et al [13]. A flow rate of 0.5 μL/min was achieved for a 75 μm x 150 μm x 22 mm MHD pump section using a 0.42 T NeFeB permanent magnet and an applied current density of 4000 A/m². Homsy et al used point source electrodes instead of full length electrodes. Lorentz force and electrolysis depend on electrodes’ surface and voltage. Using point source electrodes would account for the low observed flow rate and minimal interference from electrolyzed gas bubbles. This work also did not address the actual release of electrolyzed gas for the pump is complete open to air. An open micropump also neglects a very fundamental design parameter for Lab-on-a-Chip application, which is hermetical sealing.
<table>
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<tr>
<th>Authors</th>
<th>Micromachining Process</th>
<th>Channel Material</th>
<th>Cross-section Geometry</th>
<th>Channel Dimesions (µ)</th>
<th>V, B, I, &amp; Q (µL/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heng, Wang, &amp; Murphy (2000)</td>
<td>UV-LIGA &amp; soft lithography</td>
<td>glass substrate base with PMMA cover plate</td>
<td>rectangular with diffuser/nozzle</td>
<td>d = 200,400 w1 = 250 w2 = 500 w = 875 L = 1000</td>
<td>V = 15 V (AC @ 1 Hz) B =2.1 T I = 75 mA Q = 1900, 6010</td>
</tr>
<tr>
<td>Jang &amp; Lee (2000)</td>
<td>anisotropic bulk etching of top and bottom wafers – epoxy bonded</td>
<td>Silicon</td>
<td>trapezoidal</td>
<td>d = 400 w = 1000 L = 40000</td>
<td>V = 10-60 V (DC) B = 0.44 T I = 1.8 mA Q = 63 (measured) Q = 4 (predicted)</td>
</tr>
<tr>
<td>Lemoff &amp; Lee (2000)</td>
<td>anisotropic etching</td>
<td>silicon with glass cover plate</td>
<td>trapezoidal</td>
<td>d = 380 w = 800 Le = 4000</td>
<td>V = 25 V (AC @ 1 kHz) B = 18.7, 7.4 mT I = 140, 100 mA Q = 18.3, 6.1 (1 M NaCl, 0.1 M NaCl)</td>
</tr>
<tr>
<td>Bao &amp; Harrison (2003)</td>
<td>bulk micromachining with ICP-IRE</td>
<td>silicon with glass cover plate</td>
<td>rectangular</td>
<td>d = 10 w = 10 L = 100</td>
<td>V = 2 V (AC @ 960 Hz) B = 0.45 T I = 100 mA Q = 0.72</td>
</tr>
<tr>
<td>Wang et al (2004)</td>
<td>no micromachining reported</td>
<td>no micromachining reported</td>
<td>Rectangular</td>
<td>d = 2000 w = 10000 Le = 35000</td>
<td>V = 25 V (DC) B = 13 mT I = 140, 100 mA Q = 20.4, 14.35 (1 M NaCl, 0.1 M NaCl)</td>
</tr>
<tr>
<td>Homsy et al (2005)</td>
<td>two step lithography on Pyrex wafers</td>
<td>Pyrex glass</td>
<td>semi-circular</td>
<td>d = 75 w = 150 L =22000</td>
<td>V = ?? (DC) B = 420 mT J = 4000 A/m² Q = 0.5</td>
</tr>
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Table 1. Summary of previously published MHD micropumps.

Note: d – depth of the channel, w – width of the channel (w1, w2 are widths at top and bottom for trapezoidal channels), L – length of the channel, Le – the length of the electrode, V – electric potential, B – magnetic flux density, I – current, J – current density, Q - flow rate.
Figure 6. MHD micropump by Homsy et. al. [8].
CHAPTER 3

MHD MICROPUMP DESIGN CONCEPT

SECONDARY PHYSICS IN MHD MICROPUMP

Electrokinetics Cross Flow

While secondary electrokinetics (EK) flow, Joule heating, and electrolysis’s contributions to the overall performance of MHD macropumps are negligible; non-linear scaling of electromagnetic forces allows secondary them to become significant competitors to Lorentz force at the micro level. Depending on the design and fabrication materials, they could severely counteract the Lorentz force being generated. They could even potentially supplant Lorentz force as the main pumping mechanism.

Secondary EK flow, normal to MHD flow, creates ionic locomotion between parallel electrodes. It consists of two distinct components: electrophoretic (EP) and electroosmotic (EO) forces. EO’s contribution to electrokinetic flow is much higher than EP’s. As discussed in earlier chapter, electroosmotic phenomenon requires charged surfaces to generate the electrical double layer required for pumping. Proper material selection for micropump fabrication could abate EO cross flow by eliminating charges accumulation along the surfaces of pump’s interior.

Joule Heating

Transformation of electrical energy into work inevitable produces heat due to resistive loss. Joule heating, a conductive heat transfer process produces variation in fluid’s temperature and its effective electrochemical properties. Joule heating, itself, is not detrimental to MHD pumping, but its effect on the fluid is. MHD phenomenon, a function of fluid’s conductivity, would also experience local variation caused by this conduction heating. If flow velocity is great enough to disperse heat being generated, it could adequately eliminate local temperature variation. Increasing convection transfer term coupled with reducing resistive heating would be ideal solution to reduce the adverse effects of Joule heating on MHD flow.
**Electrolysis**

Electrolysis is intrinsic to MHD and most other non-mechanical micropumps that operate in DC mode. The transferring of electrons via ions in solution allows electrical current to flow from one electrode to the other. Current flow is the driving force behind the reduction-oxidation process known as electrolysis. Different solutions have different current conduction limit before undergoing electrolysis. Electrolysis itself is a problem. But the effects of its byproducts are the main factors impeding flow.

Initially, electrolyze gas would form micro bubbles on electrodes’ surface. As more micro bubbles are generated, they would coalesce into larger bubbles according to laws of thermodynamics. Assuming constant pressure and temperature, the same number of molecules in gas state occupies approximately 1000 fold of its volume at liquid state. As more gases are generated; they will displace the liquid inside the micropump to form a gas occlusion. The absence of conductive media will disable local MHD pumping for Lorentz force cannot be generated without electrical current. As more electrolyzed gases are generated, they would coalesce with the large gas occlusions until there are no reactants left to produce electrolysis. Meanwhile, pressure within the gas pockets continues to rise as it expands in a confined volume. Reduced pump’s power coupled with increase pressure requirement creates a synergistically compounding problem that will render the whole pump inoperable until the gas pocket is removed.

**MHD Micropump Design Concept**

Since electrolysis is inevitable; the size, rate and location of aggregation, and releasing electrolyzed gases are critical component that determine MHD micropump’s performance. Size and rate of electrolysis highly depends on the chemical interaction of electrolytes and electrodes used. Using electro-chemically optimal electrolytes and electrodes pair, that has highest reduction oxidation potential, would mitigate some of the problems associated with electrolysis. However electrolytes/electrodes selection is a very case specific solution.

The main focus of this research is to develop a general MHD micropump for all LOC applications in which low excitation voltage is major criterion. Hence size and rate of electrolysis bubbles formation is already minimized. Designing mechanisms to control the
location and releasing of electrolyzed products will be the focal point of MHD micropumps presented here.

Platinum electrodes and aqueous sodium chloride are selected in this work because they are most common used in biological applications. For 6400 μm long platinum electrodes with diameter of 200 μm used in these experiments, the current density is calculated to be approximately $2 \times 10^6 \text{ A/m}^2$ at 5V excitation. Electrolysis occurs according to standard reduction oxidation equations given in equations 2a-2d.

Cathode: \[ 2 \text{H}_2\text{O} + 2e^- \rightarrow 2 \text{OH}^- + \text{H}_2 (g) \quad E_{\text{ox}}^{\circ} = -0.83 \text{ V} \quad (2.a) \]
\[ \text{Na}^+ + e^- \rightarrow \text{Na} \quad E_{\text{ox}}^{\circ} = -2.71 \text{ V} \quad (2.b) \]

Anode: \[ 2 \text{Cl}^- \rightarrow \text{Cl}_2 (g) + 2 e^- \quad E_{\text{ox}}^{\circ} = -1.36 \text{ V} \quad (2.c) \]
\[ 2 \text{H}_2\text{O} \rightarrow 2 \text{O}_2 (g) + 4 \text{H}^+ + 4 e^- \quad E_{\text{ox}}^{\circ} = -1.23 \text{ V} \quad (2.d) \]

The products on this oxidation-reduction process depends not only the theoretical reaction but also empirical observations. In reality, sodium is not reduced to solid sodium at the cathode due to its high potential and temperature requirement. At the anode, high chlorine’s concentration causes a down shift of oxidation voltage required; thus favoring its reaction. Oxygen and chlorine have very similar oxidation potential. Oxygen production is unfavorable due to high overvoltage required to initiate the reaction. This overvoltage depends on the chemistry between electrodes used and aqueous sodium. Pure platinum electrodes used in these experiments favors hydrogen and chlorine gas generation. The overall electrolysis that occurs in aqueous sodium chloride would give off primarily hydrogen and chlorine gas as described in equation 3 [17].

\[ 2\text{NaCl (aq)} + 2\text{H}_2\text{O (l)} \rightarrow 2 \text{NaOH (aq)} + \text{H}_2 (g) + \text{Cl}_2 (g) \quad (3) \]

As discussed earlier, electrolysis could be detrimental to flow. The initial gas bubbles’ diameter is much smaller than the channel’s opening. If these bubbles do not coalesce; flow should be relatively unhindered. Therefore the main concept in designing
MHD micropump designs presented here are to limit aggregation of gas bubbles and dispersing them.

To limit bubbles coalescence, electrodes are placed in separate compartments. Conduits, along the walls separating electrodes compartments from main flow cell, function as bubbles trapping mechanism while allowing current continuity within the pump. Geometry and location of conduits are varied among different designs to manipulate electric field distribution, current density, and location of electrolyzed gases’ coalescence. These conduits would also provide a favorable location for gas coalescence. Therefore they would prevent large gas bubbles from entering the main flow cell while allowing smaller gas pockets to enter the flow. Dispersing smaller gas pockets would reduce total amount of gas available for aggregation; thus minimizes their contribution to the overall pump’s performance.

In an open micropump system such as Homsy [8], gas bubbles could escape to atmosphere rather than aggregating along the pump section and obstructing flow. For integrated LOC application, hermetic sealing is an important requirement. This makes dispersing aggregated bubbles a very challenging task. Gas bubbles bursting within the sealed section will create undesirable multidirectional flow similar to phase transfer micropumps. Venting chambers equilibrated with atmospheric pressure are introduced in final design so aggregated bubbles could be released without creating inadvertent back flow.
Table 2. Summary of MHD micropump designs.

Note: Electrodes are noted as ■. All Version A – Conventional single channel/compartment micropump, Version B - Compartmentalized micropump, Version C - Compartmentalized micropump with elevated main channel, Version D - Compartmentalized micropump with buried electrodes, Version E – BIRS micropump. (d = 800 μm, w = 800 μm, l = 6.4 mm).

<table>
<thead>
<tr>
<th>MHD Version</th>
<th>Front View</th>
<th>Top View</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td><img src="image1" alt="Front View" /></td>
<td><img src="image2" alt="Top View" /></td>
<td>Traditional design with electrodes along sidewalls</td>
</tr>
<tr>
<td>B</td>
<td><img src="image3" alt="Front View" /></td>
<td><img src="image4" alt="Top View" /></td>
<td>Modified with electrode compartments</td>
</tr>
<tr>
<td>C</td>
<td><img src="image5" alt="Front View" /></td>
<td><img src="image6" alt="Top View" /></td>
<td>Modified with electrode compartments and raised main channel</td>
</tr>
</tbody>
</table>
D

Modified with electrode compartments and raised main channel and perforations for electrical continuity.

E

Modified with electrode compartments and spacers for gas escape.
CHAPTER 4

FABRICATION & EXPERIMENTAL SETUP

MHD micropumps are designed using Solidworks and fabricated from polysulfone on a Haas CNC machine (Haas Automation Inc., Oxnard, California). The rapid prototyping offered by CNC is the main motivation for adopting this milling process. In total, five designs with increasing sophistication in bubble isolating and discharging capabilities are designed and fabricated. A summary of configuration and dimensions is given in Table 2. Versions A, B, and C are machined as one unit while version D and E require milling two parts each that are subsequently bonded using Methylene Chloride. During experiments, venting chambers are not sealed with semi-permeable membrane, but exposed to atmospheric pressure. The experimental set-up is shown in Figure 7.

For experiments involving version A, B, C, and D, pump sections are hermetically sealed by bonding No. 1 glass coverslips to polysulfone substrate. Custom Teflon insulated platinum electrodes (WPI, Saratoga, Florida), 200 microns in diameter and 6.4 mm length (running for the whole length of the main channel) are powered using HP 6236B power supply (Hewlett Packard, Palo Alto, California). Electrodes are made by embedding Teflon coated platinum wire inside an epoxy filled 22 gage hypodermic needle. Gold connector is then soldered to the platinum wire for stable connection to power supply. Entire outer electrode is coated with paraffin and epoxy to ensure proper electrical insulation; leaving only the desired length exposed. To ensure proper placement inside the micropumps, electrodes are fixed in position by using micromanipulator and epoxy glue.

DC current was generated using GoldStar FG-2002C (LG Precision, Seoul, South Korea) function generator coupled with in-house amplifier. 18 mT permanent magnets, purchased from Industrial Liquidator (San Diego, California), were used to generate the required magnetic field for micropumps. Video recordings are done on Wesco stereoscope (Western Scientific Company Inc., Valencia, California) at 5x magnification using a Zarbeco ZC105 camera (Zarbeco LLC, Randolph, New Jersey). Data analysis is done using Matlab (Mathworks, Natick, Massachusetts).
All experiments are performed using 1 M NaCl solution at 20° C under identical configuration and settings. NaCl solution is equilibrated in micropumps for 10 minutes prior to start of experiments to ensure proper priming. Pumps are activated for 2 minutes and output volume measured using Pipetman® (Gilson Inc., Middleton, Wisconsin). Electrolyzed gas bubbles are monitored and velocities are measured using particle tracking algorithm. Triplicate readings of flow measurements are taken.
Figure 7. Experimental Setup for to measure flow rate.
CHAPTER 5

RESULTS

EXPERIMENTAL RESULTS

Single Compartment Micropump – Version A

To validate CNC’s fabrication against traditional MEMS micromachining, a simple conventional MHD pump with electrodes located within the same chamber along the side walls was machined and tested. This simple design serves as a reference to estimate the effect of unmitigated bubble formation on MHD micropump’s behavior, specifically flow rate. In this configuration, flow rates of 100 and 200 μL/min were obtained for 5 and 20 Volt respectively. These results show that flow rate is not linearly proportional to applied voltage as expected.

Chaotic and rapid bubbles generation in junction with unpredictable bubbles bursting are the main reasons for nonlinear performance. Other phenomena such as Joule heating and electroosmotic flow, observed by other researchers, are additional sources contributing to this nonlinear behavior. Even at lower voltages, random bubbles aggregation within the pump section was observed. This random behavior abruptly devoided fluid from large sections of electrodes and disabled current conduction, hence pumping. Due to a high pressure gradient between the reservoir and its location, gases produced near the inlet prefer to accumulate at the pump’s inlet. In general, this design suffers from excessive bubbling; limiting its usefulness in high current application. The main point taken from this design is that CNC milled MHD micropump can produce similar results as published works that used tradition MEMS fabrication technology.
A – Micrograph of pump section, B – Measured flow rate at 5, 10, 15, and 20 V DC.

Figure 8. Conventional MHD Micropump.
**Compartmentalized Micropump – Version B**

As a first attempt to control electrolyzed gas’s aggregation, electrodes compartments and evenly spaced microfringes (200 μm wide, 400 μm long, and 800 μm deep) were introduced in this design. These microfringes are 800 μm center to center apart. Flow rates of 150 and 300 μL/min were observed at 5 and 20 Volt respectively.

Compared to reference design, there is a slight improvement in flow rate. However flow rate with respect to increase in voltage is still not proportional as expected. Bubbles’ coalescence in the main channel continued to exist. Initially, small bubbles migrated into the main channel and continue toward the exit. However some of these bubbles began to accumulate around the microfringes. These occlusions began to grow as more bubbles aggregated together. Once they can span the space between the fringes, current conduction would be interrupted and deactivate the whole pump in similar manner observed in prior design. Hence, microfringes alone cannot effectively mitigate the adverse effects of bubbles on flow. They can only delay the process.
A – Micrograph of pump section, B – Measured flow rate at 5, 10, 15, and 20 V DC.

**Figure 9.** Compartmentalized MHD Micropump.
Compartmentalized Micropump with Elevated Main Channel – Version C

Location of gas formation and aggregation are quintessential factors in determining how electrolyzed gases would affect flow. Since gases have limited solubility in liquid, increasing time and distance travel in liquid would allow more gas molecule to dissolve in liquid. Therefore the increasing the depth of electrodes compartment would serve this purpose. In this design, electrode compartments are twice as deep as the main channel. Dimensions of other features were retained so quantitative comparison between different designs could be evaluated. Flow rates of 150 and 400 μL/min were observed at 5 and 20 Volt respectively.

This design, observed flow rate is better than previous designs at higher voltages (e.g. at 15 and 20 Volts). Having room for electrolyzed gas to travel and burst seem to improve pump’s performance at higher voltages. At longer operation time, bubbles still behaved as in design B. They aggregate along the edges of microfringes and electrode compartment’s walls. Only a very limited amount of small gas bubbles were able to join the flow smoothly. Electrolysis is a product of the applied current. Thus, the “electrolysis rate” required to generate flow rate of more than 100 μL/min is simply too high. Simple passive bubbles’ retardation mechanism is inadequate to alleviate bubble’s negative impact on flow.
A – Micrograph of pump section, B – Measured flow rate at 5, 10, 15, and 20 V DC.

**Figure 10. Compartmentalized MHD Micropump with Elevated Main Channel.**
Compartmentalized Micropump with Buried Electrodes – Version D

Simple trapping of gas bubbles slightly improved pump’s performance. The fourth design incorporates a more aggressive trapping mechanism. As in design C, there is an elevated main flow section. However the inferior chamber is completely open. A 200 μm thick horizontal layer between the upper flow sections is introduced to preclude electrolyzed bubbles from entering the main flow cell. To maintain current continuity, this separation layer contains perforation holes on the electrodes’ compartments’ bottom that are 250 μm in diameter and paced at 800 μm center to center apart in between microfringes. There are three possible electrodes placement configurations: electrodes are on the same plane either superior or inferior to the separation layer and electrodes are on different planes.

The best result is obtained with anode placed in bottom chamber and cathode the diagonally opposing upper layer. Flow rate of 75 and 325 μL/min are observed for 5 and 20 Volt, respectively. Reversing anode and cathode polarity yields flow rate of 50 and 200 μL/min for 5 and 20 V respectively. In the optimal electrode’s configuration, chlorine gas, produced at the anode and more soluble than hydrogen, has an opportunity to equilibrate with the surrounding and travel with the liquid toward the exit without impeding flow. The more volatile and rapid forming hydrogen can burst easier while they are in the upper layer. Because they do not have liquid exerting pressure trying to compress them. This rapid formation and bursting of hydrogen gas bubbles which lead to discontinuous current conduction in the bottom layer, is probably the main cause for this shift in flow rate when current is reversed.
Version D - Experimental Flow Rate

A – Micrograph of pump section, B – Measured flow rate at 5, 10, 15, and 20 V DC.

Figure 11. Compartmentalized MHD Micropump with Buried Electrodes.
B.I.R.S. (Bubbles Isolation and Release System) – Version E

Observations from previous designs form the backbone of this design. The main problem is not electrolysis but rather the negative consequences of coalescence of gas bubbles. If electrolyzed gas bubbles can be prevented from aggregating, MHD micropump should operate close to its theoretical limit. As observed in previous designs, microfringes promote bubble coalescing around them. However bubbles’ accumulation between microfringes is not desirable; because it would disturb current conduction in the main flow cell. In this design, full length walls are employed to separate the three compartments instead of microfringes. This would prevent gas bubbles from entering the main channel while providing a thermodynamically favorable surface for attachment. Hydrostatic pressure gradient would cause gas bubbles, coalesced on the walls, to gradually migrate upward. Slits at bottom of the walls mentioned above would behave as current continuity conduits. In this fashion, current density can be manipulated by adjusting the height of these horizontal slits. To ensure proper gas discharging while maintaining hermetrical seal, venting chambers above the electrode compartments are introduced. This free space allows electrolyzed gas to equilibrate with atmospheric pressure. In summary, crossing channels are completely removed and replaced by solid barriers with horizontal slits in BIRS MHD micropump.

This design gives the best result at low voltage setting. Maximum flow rate of 330 \( \mu \text{L/min} \) is observed at 5 Volt. In version D, 20 volt is required to achieve comparable flow rate. At high voltages, more bubbles are generated than can be handled by the bubble trapping and isolation region causing major drop in BIRS MHD micropump’s performance. At high voltage, electrolyzed gases coalesce and burst too rapidly; driving liquid out of the electrode’s compartments. This would cause current discontinuity in the main channel. A possible remedy is to increase the size of the bubble isolation chamber for higher voltage application.
A – Micrograph of pump section, B – Measured flow rate at 5, 10, 15, and 20 V DC.

Figure 12. BIRS MHD Micropump.
SIMULATION RESULTS

Numerical models were built using COMSOL 3.3 to predict electric and magnetic field distribution, Lorentz force distribution, and flow rate. These models, however, do not include the effect of bubbles or other secondary phenomena and assumes simple one-phase flow. Two-phase flow models – which are beyond the scope of the current work – are required to full extrapolate the effects of electrolysis. Detailed discussion of the numerical models for MHD micropumps applicable to this research is provided in Patel and Kassegne [16]. Summary of the physics and boundary conditions used in simulations is provided in Table 3. The main scope of this research is not in simulation of MHD micropump. However simulation provides insights on how to improve pump’s design and understand experimental observed phenomena.

Simulation results provide total Lorentz force generated and subsequently flow rates. A conductive fluid of 1.5 S/cm and uniform magnetic flux density of 18 mT were used for simulation. These values represented the conductivity of 1 M NaCl and magnetic strength of permanent magnet used to obtain experimental results. Figure 18 demonstrates predicted velocity profiles of all MHD micropumps presented here along pump’s height at entrance and along pump’s length at center. Single compartment design (version A) exhibits highest performance. However, the model does not include the effect of bubbles. Close spacing between electrodes also contribute to the high theoretical flow rate. Compartmentalized design (version B), BIRS (version E), compartmentalized with elevated channel (version C), and compartmentalized with buried electrodes (version D) are predicted to have flow rate in descending order.

As expected, the FEA simulations show that flow velocity is a function of geometry if effects of electrolysis are not considered. However, the distribution of flow profile, summarized in figures 13 - 17, offer interesting insight into the spatial distribution of the velocity field. Compartmentalized design (version B) and BIRS seem to have ideal flow velocity distribution with highest concentration in the main pump channels. BIRS design exhibits some side flow bleeding which increase with increase in height separation slits. Compartmentalized micropump with elevated main channel (version C) and
compartmentalized micropump with buried electrode (version D) exhibit significant undesired flow in electrode’s compartments.

Table 3. Boundary conditions for generalized 3D rectangular MHD equations for MHD microfluidics pumps [16].

Note: x is the length of the pump, y is the width, and z is the height.

<table>
<thead>
<tr>
<th>Equation Type</th>
<th>Boundary Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Electromagnetics</td>
<td>$B(x, y, o) = \text{known } (+) \quad (x &lt; Le)$</td>
</tr>
<tr>
<td></td>
<td>$B(x, y, d) = \text{known } (-) \quad (x &lt; Le)$</td>
</tr>
<tr>
<td></td>
<td>\textit{Electrodes}</td>
</tr>
<tr>
<td></td>
<td>$\Phi(x, 0, z) = V \quad (x &lt; Le)$</td>
</tr>
<tr>
<td></td>
<td>$\Phi(x, b, z) = -V \quad (x &lt; Le)$</td>
</tr>
<tr>
<td></td>
<td>\textit{Insulations}</td>
</tr>
<tr>
<td></td>
<td>Magnetically insulated elsewhere</td>
</tr>
<tr>
<td></td>
<td>Electrically insulated elsewhere</td>
</tr>
<tr>
<td>Fluid Dynamics</td>
<td>$U(x, 0, z) = 0 \quad (\text{no-slip}), \quad U(x, y, 0) = 0 \quad (\text{no-slip})$</td>
</tr>
<tr>
<td></td>
<td>$U(x, b, z) = 0 \quad (\text{no-slip}), \quad U(x, y, d) = 0 \quad (\text{no-slip})$</td>
</tr>
<tr>
<td></td>
<td>\textit{Inlet}</td>
</tr>
<tr>
<td></td>
<td>$U(0, y, z) = 0$</td>
</tr>
<tr>
<td></td>
<td>\textit{Outlet}</td>
</tr>
<tr>
<td></td>
<td>$P(L, y, z) = 0$</td>
</tr>
<tr>
<td></td>
<td>body force = $F_L$ (Lorentz Force)</td>
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</table>
Qualitative flow velocity distribution and FEA mesh for Version A (Conventional Single Compartment MHD Micropump)

Figure 13. Version A MHD Micropump.
Figure 14. Version B MHD Micropump.
Qualitative flow velocity distribution and FEA mesh for Version C (Compartmentalized with Elevated Main Channel MHD Micropump)

**Figure 15. Version C MHD Micropump.**
Qualitative flow velocity distribution and FEA mesh for Version D (Compartmentalized with Buried Electrodes MHD Micropump)

**Figure 16. Version D MHD Micropump.**
Qualitative flow velocity distribution and FEA mesh for Version E (BIRS MHD Micropump)

**Figure 17. Version E MHD Micropump.**
A – Predicted flow profile along height of channel at midline, B – Predicted flow profile along length of channel at center.

Figure 18. Predicted flow profiles.
Theoretical flow rates determined by FEA are compared against experimental measurements as shown in Figure 18. Flow rate at 5 Volts shows a relatively close agreement between theoretical and observed values. Since, the amount of electrolyzed gas bubbles is relatively few at 5 Volts; flow rate is not expected to be drastically affected. However, as the applied voltage increases to 10, 15, and 20 volts, experimentally measured flow rate is as low as 10% of the theoretically predicted value. In general, the trend indicates that single compartment MHD micropump (version A) has a consistently high discrepancy between theoretical and experimental values. The introduction of bubble coalescence mechanisms in designs B-E, in general, tends to reduce this discrepancy.

Figure 19. Comparison between experimental and theoretical flow rate.

BIRS MHD micropump’s performance diminishes as higher voltages are applied. As explained earlier, this is due to the electrolyzed gas that exceeded BIRS’s capacity. These bubbles would deactivate MHD pumping as in all other designs by preventing current conduction. Flow rate of 330 μL/min for an 800 μm x 800 μm x 6.4 mm MHD micropump is already very high for most micropumps. This operational range satisfies the object of this research; thus there is no pressing need to consider operation at higher voltage.
CHAPTER 6

CONCLUSION & FUTURE WORK

In this research, we introduced a novel high flow rate DC-based MHD micropump with bubble isolation and release system. The negative effect of electrolyzed gases, generated at electrode surfaces, is the main cause for low experimental flow rate when compared with theoretical values. These tiny bubbles coalesce in the main flow channel and obstruct the flow of fluid. Because electrolysis is inevitable with DC excitation; compartmentalized electrode channels with bubbles coalescence retarding and release mechanisms are implemented in this research to prevent undesirable aggregation. BIRS (Bubbles Isolation and Releasing System) MHD is demonstrated to be capable of generating a flow rate of up to 330 µL/min using 1 M NaCl solution in DC mode with potentials of 5 V and a current density of approximately 2x10^6 A/m^2 for a main channel with a 800 µm x 800 µm cross-section and 6.4 mm in length. This pump performs well at low voltages (5 Volts); but degrades at higher voltages due to excessive electrolysis which requires a larger venting system.

The results from experiments were compared with FEA simulations. Numerical models offer interesting and useful insights into spatial distribution of velocity field as determined by the Lorentz’s force. However, quantitative agreement between experiments and FEA show variations between 25% and 300%. These variations are due to the simple one-phase flow assumptions in the FEA models. A numerical model that considers generation and propagation of electrolysis products is currently being pursued as an extension of this work to elucidate better understanding of MHD micropump’s behaviors and to improve pump design.

The main objective of this research was to design micropumps for LOC application with flow rate of 100 µL/min or more. The BIRS MHD micropump achieved this goal and is therefore a suitable choice for this application. Furthermore, the ability to CNC machine a series of these micropumps in an array format where the BIRS pumps share a common
bubble release chamber could offer a cost effective and efficient pumping system applicable for a variety of low voltage LOC and μTAS (micro-Total Analysis System).
REFERENCES


ABSTRACT OF THE THESIS

Design, Fabrication, & Experimentation of DC MHD Micropump
by
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Master of Science in Bioengineering
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A major challenge for integrated Lab-on-a-Chip (LOC) systems is the precise and consistent control of fluid flow. Magnetohydrodynamics (MHD) micropumps which contain no moving parts and capable of generating a continuous flow in any ionic fluid offer an ideal solution for biological applications. MHD micropumping has been demonstrated using both AC and DC currents by a number of researchers with varying degrees of success. However, current MHD designs based on DC (Direct Current) do not meet the flow rate requirements for fully automated LOC application (> 100 ul/min). Gas bubbles, byproducts of electrolysis, particularly hydrogen gas are the main contributor to low experimentally observed flow rate. These tiny bubbles coalesce in the main flow channel and hinder fluid’s flow. Since hydrolysis is inevitable under direct current (DC) excitation, mitigating its negative effects could significantly improve MHD micropump’s performance. In this research, we introduce a novel MHD micropump which effectively increases flow rate by deploying bubbles isolation and releasing mechanism that limit electrolyzed gas’ disruption of flow. BIRS (Bubble Isolation and Release System) MHD, with a flow channel of 800 µm x 800 µm cross-section and 6.4 mm length, could pump 330 µL/min of 1 M NaCl solution at 5 Volts DC. This voltage corresponds to a current density of approximately 200,000 A/m² on the surface of platinum electrodes with length of 6.4 mm, diameter of 200 µm and resistance of 3 Ohms.