Active control of microdroplets and particles in microfluidics

Ali Beyzavi

2009

Ali, B. (2009). Active control of microdroplets and particles in microfluidics. Master's thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/47207>

<https://doi.org/10.32657/10356/47207>

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ACTIVE CONTROL OF MICRODROPLETS AND PARTICLES IN MICROFLUIDICS

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School of Mechanical and Aerospace Engineering
A thesis presented to the Nanyang Technological University
in partial fulfillment of the requirements for the degree of
Master of Engineering
2009

Abstract:

The manipulation of small amounts of fluids in the order of nano and pico liters has emerged as a new scientific field of microfluidics. This field has potentially a huge impact on chemical synthesis, biological analysis, optics and information technology. At the beginning, all microfluidic devices were designed to manipulate continuous flow, while recently there has been a shift towards working with discrete droplets. Droplet manipulation allows the use of non mechanical methods for actuation and in some cases elimination of micro and nanochannels in the chips. The field is new and therefore still needs improvements in the actuation schemes. The purpose of this research is using magnetic force for active control and manipulation of droplets.

The magnetic field of microplanar coils has been modeled and used to study the one and two dimensional actuation of ferrofluid droplet on a free surface. In the last part, design and fabrication of a real low cost, multi functional device for operating a variety of functions on a droplet has been discussed. These devices can be used as a platform for many of lab-on-a chip applications.

To my parents
Fereshteh and Iraj

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Acknowledgments

The author wishes to express his sincere gratitude and appreciation to the following people, who had offered invaluable contribution to this thesis:

- Associate Professor Nguyen Nam-Trung, the author's supervisor, for his sincere encouragements, invaluable guidance and friendship during the past years;
- My parents for their deep loves and continuous encouragements.
- My fellow research students, labmates, and technician staff in thermal and fluids lab and many other people who helped me with useful discussions, suggestions and experimental support;

Nomenclature

Θ_R	Diffraction angel for surface acoustic waves
θ	Contact angel
γ_g	Surface tension for liquid-gas interface
γ_{sg}	Surface tension for solid-gas interface
γ_{sl}	Surface tension for solid-liquid interface
k_{el}	Conductivity of dielectric
v_f	Sound velocities in fluid
v_s	Sound velocities in the substrate
χ	Magnetic susceptibility
η	Viscosity
$\Re$	Real component of the electric field
B	Magnetic field strength
E	Intensity of electric field
F	Force
G	Gibbs free energy
q	Electric charge
T	Temperature

U	Droplet velocity
V	Velocity
$\bar{E}_{el}^*$	Complex conjugate of the electric field
$\bar{m}(w)$	Dipole moment
μ_{eo}	Electroosmotic mobility of charged particle
μ_{ep}	Electrophoretic mobility of charged particle
ρ_{el}	Charge density
ε	Dielectric permittivity of the media
ε_0	Dielectric permittivity of vacuum
ζ	Zeta potential
$K(\omega)$	Clausiou-Mossoti factor
$\tilde{\varepsilon}_m$	Complex permittivity of the solution
$\tilde{\varepsilon}_p$	Complex permittivity of the particle

Chapter 1

Introduction

1.1 Microfluidics and Its Origin

Microfluidics is the science of handling and analyzing of fluids in structures of micrometer scale [1]. Fluidic systems of small dimension have the advantage of reduced reagent consumption and waste generation, fast reaction times, and a large surface-to-volume ratio, offering an intrinsic compatibility between the use of a microfluidic system and surface-based assays [2].

Miniaturization of functional components was the idea originating from electronic and computer sciences. The size of the initial electronic circuits were very large and an electronic device used to accumulate a very large space. Figure 1.1 (a) shows a 5 mega byte electronic storage memory, made in 1956. Figure 1.1 (b) illustrates a 1 giga byte memory, made in 2003. As it can be compared, the effort for miniaturization of the components during these years, has been very efficient: the size, weight and subsequently the cost of the device has decreased while its functional efficiency has increases thousands of times. After such a revolution in electronic and computer sciences, scientists decided to follow the same trend for other devices, to increase their efficiency and reduce their production cost. This is actually one of the final aims of microfluidics and MEMS technologies. Figure 1.1 (c) and (d) compares a chemical lab and a number of devices that can perform chemical/biological reactions in a space as small as a chip.

The first applications of microfluidic technologies have been in chemical analysis.

Figure 1.1: The miniaturization of functional components has immensely increased the efficiency of the devices. (a) A storage memory made by IBM in 1956. (b) The storage memory made in 2003. The size of the device has decreased and its storage capacity has increased 5000 times. (c) A chemical lab in 1990. (d) A number of devices that can perform the chemical/ biological reactions in a much smaller space [3, 4].

Working with small amounts of liquids has several benefits such as shorter time of chemical reactions and using smaller amount of sample and reagents which finally results in reducing the costs.

Microfluidics has emerged from four traditional fields of science: chemical analysis, biodefence, molecular biology and microelectronics. The first motivations of microfluidics can be found in microanalytical techniques such as gas-phase chromatography (GPC), high-pressure liquid chromatography (HPLC) and capillary electrophoresis (CE) which previously revolutionized chemical analysis. These methods combined with the power of laser in optical detection make high sensitivity and high resolution with very small amounts of sample possible [4].

The second motivation for the development of microfluidic systems appeared after the end of the cold war, because of terrorist threats of chemical and biological weapons. The Defense Advanced Research Projects Agency (DARPA) of the US Department of Defense, supported a series of programmes against the threats. These programs aimed to develop field-deployable microfluidic systems to serve as detectors for chemical and biological threats. This support was a motivation for growth of microfluidics in academia [4].

The field of molecular biology was third motivational force. The amazing growth of genomics in the 1980s, followed by the advent of microanalysis related to molecular biology, such as DNA sequencing, required analytical methods with much greater throughput, higher sensitivity and resolution. Microfluidics offered approaches to realize the above challenges [4].

The fourth motivation came from microelectronics. The idea of photolithography to generate fine patterns with high resolution in micro and nano scale was developed in microelectronics and in micro electromechanical systems (MEMS). This technology was very applicable in microfluidics. The early works in microfluidics used Silicon and glass for fabrication of the devices, but later plastic was replaced instead of them. This is because silicon and glass don't have the required properties for working with biological samples. For example silicon is opaque to visible and ultraviolet light and therefore can not allow optical detection. Neither glass nor silicon has permeability to gases that is required for working with mammalian cells [4].

Because of the above motivations the field of microfluidics progressed rapidly. In the

Figure 1.2: Micro fluidic components for handling continuous flow: (a) micro pump [7] (b) micro valve [8] and (c) micro channels in a device for growing stem cells.

Figure 1.3: A microfluidic chemostat used to study the growth of microbial populations [4]. The device contains a high number of pneumatic valves and micro channels.

early development, most of the microfluidic devices were based on continuous flows. In this way all the components of fluid manipulation (such as pumps, valves, channels, etc) were designed and fabricated in micro scale [5, 6] (Figure 1.2). These basic components were then used in several sophisticated microfluidic devices (Figure 5.1).

1.2 Motivations

Design and fabrication of continuous flow-based microfluidic systems with rigid channels, pumps and valves offer very little flexibility in terms of scalability and reconfigurability. These systems are usually application specific. They also require large volumes of liquid for priming the channels prior to use. The manipulation of liquids in small discrete volumes (microdroplets) provides an alternative, especially when the droplets are not confined by microchannels, which provides additional degrees of freedom.

Droplet-based microfluidics has several advantages over continuous-flow systems. The most important advantageous is the ease of fabrication leading to the reduction in costs.

Figure 1.4: A microfluidic device for protein crystallization. (a) A device in which droplets containing proteins are trapped in wells in the microchannels. The droplets are then subjected to many different conditions for crystallization. (b) An optical micrograph of dyed droplets in the wells of the device. (c) Droplets containing crystallized proteins. The droplets are produced in the device, then collected in a glass capillary [4].

Further advantages are reconfigurability and the scalability of architecture. This is why recently there has been a great interest towards investigating on droplet-based system [9]. Figure 1.4 shows device examples for protein crystallization and works by droplet samples. Of course, in more advanced droplet-based devices there is no need for mechanical channels and pumps.

Small droplets can be manipulated based on electrostatics [10], electrowetting [11], thermo capillary effects [12] or magnetism [13] without or within microfluidic channels. Besides electrostatic and thermocapillary forces, magnetic forces have also been used for the manipulation of microdroplets. Magnetic force is a body force and depends on the magnetic moment and gradient of magnetic field. The magnetic moment is directly proportional to the volume which is a disadvantage in micro scale but a high magnetic gradient can be obtained by reducing the gap between the coil wires. Using micromachining technologies

small planar coils can be fabricated for actuating magnetic particles or droplets.

1.3 Objectives

The main objective of the project is design and fabrication of a system for one and two-dimensional manipulation of droplets by magnetic force with the following features:

- New actuation scheme.
- No mechanical pump, no valve.
- Computer interface for controlling the position of droplets.
- Modular design.

1.4 Scope of the Project

To reach the mentioned objective the research can be divided into following tasks:

- Reviewing the actuation schemes for droplets and particles in microfluidics.
- Designing and preparing circuits and electronic interface, writing computer programs for one dimensional control of the droplets.
- Developing the mentioned task for two dimensional control of the droplets.

1.5 Organization

The organization of this thesis is as below:

- Chapter one is the introduction of the current research work.
- Chapter two is a review on manipulation methods for droplets and particles in microfluidic devices. This review explains different methods of active control and manipulation of droplets and particles by electric field, thermocapillarity, acoustic wave, centrifugal force and finally discusses the magnetic manipulation more extensively.

In order to have a good view on advantages and disadvantages of magnetic manipulation of particles and droplets, a wide scope of manipulation methods has been included in this review.

- Chapter three is about modeling and optimization of planar microcoils.
- Chapter four investigates the one dimensional actuation of ferrofluid droplet in a device and models the kinematics of the droplet motion.
- Chapter five develops the one dimensional actuation to two dimensional actuation of the droplet.
- Chapter six is conclusion and future directions.

Chapter 2

Literature Review

2.1 Droplet Actuation by Electric Forces

Recently, the area of electrokinetics has seen a rapid development because of the increased interest in processes and devices that manipulate and assemble micro/nano sized particles and liquid volumes by electric field. The studies in this field are concerned about mainly biological applications such as manipulation of live cells, characterization of minute liquid samples, fabrication of massively parallel bioassays or assembly of nano structures [14].

This section mainly discusses the on chip manipulation of colloidal particles and droplets suspended in liquid, using alternating (AC) and constant (DC) electric fields. Electric field is a good option since the force exerted on droplets or particles can be precisely tuned by adjusting the electric field.

The controlling parameters of an AC signal (magnitude, wave symmetry, wave shape, frequency, and phase) can be adjusted electronically and can affect the behavior of particles between the electrodes. This allows precise adjustment of the driving forces that is hardly possible with other techniques like hydrodynamic, evaporation, or mechanical manipulation.

Electrodes and devices based on electric actuation can be fabricated by micro technologies based on micro lithography, used in conventional microelectronics. One great benefit is that there are no mechanical or optical components involved. Manipulation of particles

Figure 2.1: Electric field around charges.

by optical means for example, requires laser traps that needs complex mechanical, optical and electronic components [14]. To benefit from working with electric field, one has to have the knowledge about the behavior of the particles in an applied electric field.

2.1.1 Basic Concepts

Physics of Electric Field

When one defines the behavior of a quantity in a given region in terms of a set of values, one for each point in the region, this behavior of the quantity is referred to as a field [15]. The value at each point of a field can be either measured experimentally or predicted by solving the governing equations describing the field.

An electric field is a property of the space surrounding an electric charge. This electric field exerts a force on other electrically charged objects. The electric field is a vector with the SI units of Newtons per coulomb or, equivalently, volts per meter (Figure 2.1). The direction of the field at a point is defined by the direction of the electric force exerted on a positive test charge placed at that point. The strength of the field is defined by the ratio of the electric force on a charge to the magnitude of the charge.

$$\vec{E} = \frac{\vec{F}}{q} \quad (2.1)$$

The intensity of the field caused by a charge q in a distance of R from the charge is :

$$\vec{E} = \frac{q}{4\pi\epsilon_0 R^2} \vec{a}_R \quad (2.2)$$

where $\vec{a}_R$ is the unit vector from the charge to the point in which the intensity is to be obtained.

Forces in Electric Field

Two types of particles will be affected in an electric field: charged particles and polarized particles (dipoles). The general case is considered here, where a particle occupies the region Ω in space centered around the point r_0 . The general position inside the particle is denoted as $r + r_0$ [16].

It is also assumed that the particle has a charge of q and a dipole moment of p equal to :

$$q = \int_{\Omega} \rho_{el}(r + r_0) dr \quad (2.3)$$

$$p = \int_{\Omega} \rho_{el}(r + r_0) r dr \quad (2.4)$$

where ρ_{el} is the charge density. If an external electric field is imposed on the system, the force acting on the particle with mentioned specification is [16]:

$$F = \int_{\Omega} \rho_{el} dr (r + r_0) E(r + r_0) \approx \int_{\Omega} \rho_{el} dr (r + r_0) [E(r_0) + r \partial E(r_0)] = qE(r_0) + p \partial E(r_0) \quad (2.5)$$

The first term is the force resulting from the charge of the particle and the second term is the force of the dipole moment. It can be simply concluded that there is always a force acting on a charged particle in an electric field (uniform or non uniform field). For a dipole particle, the force only exists if it is located in a non uniform electric field.

Electrokinetics

Electrokinetics is the general term referring to the bulk relative motion between two charged phases. This phenomena appears when attempts are made to shear off the mobile part of the electric double layer from a charged surface. There are three main phenomenon in electrokinetics: electrophoresis, electro osmosis and dielectrophoresis.

Electrophoresis. If charged particles inside a liquid are placed under an electrical field, positive charged particle move toward cathode and negative charged particles move toward

Figure 2.2: (a) The simple point dipole consisting of a charge $+q$ separated from a charge $-q$ by the distance d . (b) The dipole moments (light gray ovals) and external charges (white circles) inside a body (dark gray). Polarization charge is left behind in the body when the dipole moments sticks out at the surface of the body [16].

the anode. Neutral particles are not affected. This effect is called electrophoresis.

The velocity of a charged particle is proportional to the field strength:

$$v = \mu_{ep} E \quad (2.6)$$

where μ_{ep} is the electrophoretic mobility of charged particle and E is the the field strength.

The electrophoretic mobility increases with the particle zeta potential and decreases with the viscosity of the surrounding medium. The coefficient correlating between μ and ζ depends on the size of the particles. Huckel equation is used for describing the mobility of a "small" particle defined as having a radius r much smaller than the Debye length of the counter-ionic atmosphere, $1/k$:

$$\mu = \frac{2\epsilon\epsilon_0\zeta}{3\eta} \quad (2.7)$$

where ϵ and ϵ_0 , are the dielectric permittivity of the media and vacuum, respectively. If the size of the particles is much larger than the Debye length of the counterionic layer ($r \gg 1/k$), then Helmholtz-Smoluchowski equation is used for the electrophoretic mobility of particles [14]:

$$\mu = \frac{\epsilon\epsilon_0\zeta}{\eta} \quad (2.8)$$

The electrophoretic mobility determined by measuring the particle velocity as a function of the applied external field allows measuring the particle potential. White and O'Brien divided the problem of calculating zeta potential from the electrophoretic particle mobility into two simpler problems [17]: calculation of force needed to move a particle with velocity u with no applied field and the calculation of force required to keep the particle fixed in an electric field E . Using these analytical tools, a numerical scheme for the conversion of particle mobility measurements into zeta potential values for spherical particles was developed. Extensive research has been done to separate DNA or proteins based on electrophoresis [14].

Electro-osmosis. Electro-osmosis occurs when a polar liquid is in contact with a solid under an electric field. When a polar liquid is in contact with a solid, the surface of the solid acquires an electric charge. The surface charge then influences the migration of charges in the liquid near the wall. Ions on the surface are firmly attracted to the wall and form a layer called Stern layer. In this layer ions of liquid are paired with the charges on the surface. The Stern layer influences the charge distribution deeper in the fluid, creating a thicker layer of excess charges of the same sign called diffuse layer. The Stern layer and diffuse layer together are called the electric double layer (EDL)[18]. The Stern layer is immobile because it is near the wall and more affected by the charges. Diffuse layer can move by the electric force in the electric field and because of fluid viscosity it pulls layers of fluid and moves the whole fluid [18]. The flow velocity is:

$$v = \mu_{eo}E \quad (2.9)$$

where μ_{eo} is the electro osmotic mobility of the fluid and E is the electric field strength.

The interesting characteristic in a electro osmotic flow in micro channels is that the velocity profile reaches the constant value at a distance of a few times the Debye length from the walls, as shown in Figure 2.3.

Dielectrophoresis

Equation (2.5) shows the existence of a force acting on a dipole particle, although it has no net charge, located in a non uniform electric field. Dielectrophoresis is based on this force and refers to the motion of polarized particles suspended in electrolyte and subject to

Figure 2.3: The velocity profile (dashed line and arrows) and the negative Debye layer charge density profile (dark gray and full line) in an ideal electroosmotic (EO) flow inside a cylindrical channel of radius a and positively charged walls (thick horizontal lines). The EO flow is induced by the external potential difference resulting in the homogeneous electric field. The velocity profile reaches the constant value at a distance of a few times the Debye length from the walls. No pressure drop is present along the channel in this ideal case [16].

a non uniform electric field. The force on a homogeneous dielectric particle in an aqueous solution is [18]:

$$F = \frac{1}{2} \Re[(\bar{m}(w) \cdot \nabla \bar{E}_{el}^*)] \quad (2.10)$$

where $\Re$ is the real component, $\bar{E}_{el}^*$ is the complex conjugate of the electrical field $\bar{E}_{el}$ and $\bar{m}(w)$ (dipole moment) for a spherical particle of radius r_0 and a solution of permittivity ϵ_m and angular field frequency of ω (rad/sec) is :

$$\bar{m}(w) = 4\pi\epsilon_m r_0^3 K(w) \bar{E} \quad (2.11)$$

where $K(w)$ is the Clausiou-Mossoti factor, defined as:

$$K(w) = \frac{\tilde{\epsilon}_p - \tilde{\epsilon}_m}{\tilde{\epsilon}_p + 2\tilde{\epsilon}_m} \quad (2.12)$$

where $\tilde{\epsilon}_p$ and $\tilde{\epsilon}_m$ are the compelex permittivities of the particle and solution. The complex permittivity for an isotropic homogenous dielectric is :

$$\tilde{\epsilon} = \epsilon - \frac{k_{el}}{\omega} i \quad (2.13)$$

here k_{el} is the conductivity of dielectric [18].

Figure 2.4: Droplets with diameters much smaller than the channel orifice can be formed by "flow-focusing" in microchannels, as shown in the (a) to (f) sequence [19].

2.1.2 Droplet Manipulation by Dielectrophoresis

A number of studies have focused on droplet transport in microchannels. Anna *et al.* (2003) [19] made the system in which droplets with diameters much smaller than the channel width can be formed by "flow-focusing" (Figure 2.4).

Ahn *et al.* designed special channels that are able to break up the droplets into smaller sizes [20]. Link *et al.* fabricated a device in which droplets flowing within channels can be diverted along different tracks by application of dielectrophoresis [21]. In the absence of an electric field, water drops flow into the waste channel (Figure 2.5). By applying an electric field, the droplets are attracted toward the activated electrode and flow into the collection channel. Multiple liquid streams can be filled into a single droplet for rapid mixing. The electro hydrodynamic forces acting on droplets formed using these techniques have been characterized by Zheng *et al.* [22]. Such systems, still operate within channels and don't have some of the flexibility required for micro total analysis (mTAS) applications [14].

Velev *et al.* [10] developed a technique for manipulating liquid microdroplets freely floating on the surface of a heavier liquid using dielectrophoresis (Figure 2.6).

Droplets of water or hydrocarbons are suspended over perfluorinated hydrocarbon oil (F-Oil). This oil has high density and low dielectric permittivity. There are arrays of electrodes below the oil chamber. The electrodes are energized by AC or they are grounded. This makes dielectrophoretic traps that can move the droplets in programmed path, when the electrodes are consecutively switched on and off [14].

Figure 2.5: Switching the droplets in channels by dielectrophoresis. Parts (a) and (b) are the schematic of the device. Parts (c) and (d) show the switching process [21].

Depending on the pattern of energized electrodes, different equilibrium positions of the floating droplets are possible (Figure 2.6). The floating droplet positions itself above the center of the gap between the energized and grounded electrodes. This is because the droplet places itself above the area of highest field intensity where the DEP force is strongest. This situation is illustrated by the left droplet in the simulation and on the real chip shown in Figures 2.6 (a) and (b). However, if only a single electrode is energized in an electrode array where two neighboring electrodes are grounded, the droplet positions itself directly above the energized electrode. The reasons for this positioning is illustrated by the electrostatic simulation for the droplet shown in Figure 2.6 (a). Two areas of high field intensity are created in this electrode configuration in the gaps to the left and to the right of the energized electrode. The droplet minimizes the DEP energy by symmetrically balancing the attractive forces towards the two gaps by hovering above the electrode between them. When the electrodes in the array are turned on and off sequentially, the droplet moves along the path of the energized electrodes. These droplets are suitable for the use as micro reactors for chemical syntheses and can also be used for transport of the resultant precipitated solids [10]. The droplets can be further encapsulated by dodecane oil caps to prevent evaporation [14].

The temperature gradient caused by evaporation from the surface of the droplet causes

Figure 2.6: On-chip manipulation of floating microdroplets using dielectrophoresis [10]; (a) Simulation of the electric field intensity distribution in the vertical plane. For electrodes energized at the end of the array, the droplet is attracted towards the middle of the gap between energized and grounded electrodes, otherwise the droplet positions itself on top of the electrode. The color scale to the right shows the electric field intensity; (b) Optical image of 750 nl droplets containing polymer microspheres (top) and gold nanoparticles (bottom) suspended above two tracks of electrodes; (c) the schematic of the chip.

fluid circulation inside the droplet (Figure 2.7) [23, 14].

2.2 Droplet Actuation by Thermocapillary Effect

In thermal actuation, a temperature gradient is needed along the desired movement direction of the droplet. The temperature difference of the two ends of the droplet can cause the surface tension difference, which will finally actuate the liquid droplet. After introducing some basic concepts in the next section, examples of devices will be reviewed.

2.2.1 Basic Concepts

Surface Tension

The surface tension of an interface is defined as the Gibbs free energy per area for a given pressure and temperature [16]

$$\gamma \equiv \left(\frac{\partial G}{\partial A} \right)_{p,T} \quad (2.14)$$

Figure 2.7: Circulation inside droplet because of thermal effects [23].

In metric system (SI), the unit of γ is J/m^2 .

When a molecule is in bulk, it has chemical bond with the neighboring molecules and therefore has certain amount of binding energy. The molecules on the surface, are in contact with gas from one side, thus they can't form as many bonds as the molecules in bulk. This is why they have higher amount of energy. This is exactly the surface tension; it costs energy to form a surface. It is easy to estimate the order of magnitude of surface tension for a liquid gas interface, based on this assumption. A molecule in the bulk has roughly six nearest neighbors. A molecule in surface has only five, missing the one above it in the gas. The area covered by a single molecule is roughly 0.9 nm^2 , while a typical inter-molecular bond ΔE in a liquid is of the order a couple of thermal energies, roughly $E = 2k_B T = 50 \text{ meV}$. This leads to [16]:

$$\gamma = \frac{2k_B T}{A} = \frac{50 \text{ meV}}{(0.3 \text{ nm})^2} = 90 \text{ mJ/m}^2$$

In water, the origin of surface tension is related to the polar nature of the water molecule. Hot water is more effective in cleaning because the lower surface tension makes it a better "wetting agent" to penetrate into pores instead of bridging them with surface tension. Detergents and soaps decrease the surface tension [16].

Figure 2.8: Contact angel of a droplet [16].

Contact Angle

The contact angle θ is defined as the angle between the solid-liquid and the liquid-gas interface at the contact line where three immiscible phases meet, as illustrated in Figure 2.8. In equilibrium, θ is determined by the three surface tensions γ_{sl} , γ_{lg} and γ_{sg} for the solid-liquid, liquid-gas and solid-gas interfaces by Young's equation:

$$\cos \theta = \frac{\gamma_{sg} - \gamma_{sl}}{\gamma_{lg}} \quad (2.15)$$

Hydrophilic and Hydrophobic surfaces

If the contact angel is greater than 90° , the system is called hydrophobic while those with the angel less than 90° are hydrophilic. Surface roughness affects contact angel of a droplet on a surface. By increasing the surface roughness on a hydrophobic surface, the contact angel of droplet increases while if you do the same on a hydrophilic surface, the contact angel will decrease.

On a hydrophilic surface, the liquid always flows inside the small cavities of the surface. When there is more roughness, the droplet will be in touch with a greater surface [Figure 2.9(a)]. As a result, being in touch with a greater hydrophilic surface attracts the droplet stronger towards the surface and this stronger force decreases the contact angel. When the contact angel decrease it means that the surface is more hydrophilic, so increasing the roughness in a hydrophilic surface makes it more hydrophilic.

When the droplet is in touch with a rougher hydrophobic surface, the story is different.

Figure 2.9: Effect of roughness on the contact angel: By increasing the surface roughness the contact angel; (a) decreases on a hydrophilic surface; (b) increases on a hydrophobic surface [24].

The droplet can not enter the small cavities (as it is a hydrophobic surface). This way the droplet is in contact with smaller area. This smaller area pulls the droplet with weaker force downward [Figure 2.9 (b)], therefore the contact angel will be greater than before. An increase of the contact angel of a droplet means that the surface is more hydrophobic. This is why increasing the roughness of a hydrophobic surface makes it more hydrophobic [24].

Marangoni Effect

In many cases the surface tension is varying in space. Gradient in surface tension can be implied by gradients in the concentration of surfactants such as soap at the interface or temperature gradients γ . Just similar to gradients in the pressure field that implies a gradient force per area, $-\nabla p$, a gradient in the surface tension implies a gradient force per length, $+\nabla\gamma$. The difference in sign between the two gradient forces is because of the fact that pressure forces tend to maximize volume, whereas surface-tension forces tend to minimize area. The surface tension gradient force is known as the Marangoni force:

$$f_{\text{marag}} \propto \nabla\gamma \quad (2.16)$$

In a more general form, the relationship between surface tension γ , surface curvature c and Laplace pressure p can be realized by equation (2.17):

$$p = -\gamma c \quad (2.17)$$

One can have an idea of the order magnitude of temperature induced Marangoni forces

Figure 2.10: Tears of wine [25].

by noting that heating up a water-air interface by 5°C from 20°C to 25°C will decrease the surface tension by 0.8 mJm^{-2} from 72.9 mJm^{-2} to 72.1 mJm^{-2} [16].

The shorter a distance over which one can maintain this temperature gradient the stronger the Marangoni force; but in all cases only in microsystems one can hope for a sufficiently large effect compared to other forces. The first Marangoni flow considered was the tears of wine phenomenon. The phenomenon is observed in a glass of almost any type of wine [16]. An illustration of the tears of wine phenomenon is presented in Figure 2.10

A thin layer of alcohol covers the glass surface, when the wine is poured in a glass. This thin layer evaporates more rapidly than the alcohol in the bulk. Subsequently, a gradient in the concentration of alcohol will be created. The surface tension depends on this concentration, therefore a gradient in surface tension is created. This gradient is associated with a Marangoni force causing the wine climb up the glass surface and accumulate in the top in a band of fluid that thickens. When this band eventually becomes gravitationally unstable, it releases the tears of wine. When the tears roll back into the bulk wine, they decrease the concentration of alcohol, as they have already lost their alcohol. This cycle continues until the alcohol is completely evaporated. The chemical potential energy is transferred to kinetic energy during this process [25].

The Marangoni force can also be used as a micro propulsion system, as some bacteria actually do in nature. The principle is simple, If one emits some surfactant that lowers the surface tension behind a little body then the body will be pushed forward, as the interface

Figure 2.11: Propelling a small boat by maragoni force [25].

tries to minimize the region of high surface tension (without surfactant) while maximizing the region of low surface tension (with surfactant)[16]. Figure 2.11 shows a floating 2.5-cm-long body which contains a small volume of soap, that serves as its fuel in propelling it across the free surface. The soap exits the backside of the boat, decreasing the local surface tension. The resulting surface tension gradient moves the boat forward.

2.2.2 Manipulation and Active Control by Thermocapillary Effect

Manipulation of droplets and plugs in micro scale is mainly based on the change in surface tension. There are several ways to create surface tension difference, as depicted in Figure 2.12 [26].

In the simplest case, liquid can rise in a capillary tube, as long as there is a balance between gravitational forces and capillary force [Figure 2.12 (a)]. A plug placed in an open conical tube moves up to the tip of the cone [27], because of the difference in Laplace pressure (Equation 2.17) between both menisci [Figure 2.12 (b)].

Weislogel [28] studied the movement of a plug in a tube with hydrophobic and hydrophilic inner surface. The plug runs away from the hydrophobic part and moves towards hydrophilic part. The motion stops when the whole plug is in the hydrophilic side. Figure 2.12(d) shows a liquid plug containing surfactants such as chlorosilanes, which modify the wettability of the solid [29]. If the reaction starts on one side of the drop, a surface tension difference will be created that pushes the plug [30]. Surface tension can be also changed by light. Ichimura *et al.* coated the inner part of a tube with some compound, whose sur-

Figure 2.12: Different methods to manipulate droplets by surface tension difference [26].

face energy changes when exposed to the light [31, 26]. Later a gradual lightning causes a surface tension gradient and moves the plug [Figure 2.12(e)].

Thermocapillary effect is the next option to drive a liquid within a surface [32]. Since the surface tension of a liquid usually decreases with a rise of its temperature, a liquid plug can run away from the heated end of a capillary tube [Figure 2.12(f)]. A linear relation between surface tension and temperature can be used for a small temperature range:

$$\gamma_g(T) = \gamma_{go} - K(T - T_0) \quad (2.18)$$

K is the surface tension temperature constant (its unit is $\frac{\text{N}}{\text{m}^\circ\text{K}}$), T is the actual temperature, and γ_{go} is the surface tension at the reference temperature T_0 . So the key of thermocapillary effect is dependency of the surface tension on temperature.

The velocity of the droplet under a temperature gradient can be predicted by the following formula [33]:

Figure 2.13: A device for active control of droplet by thermocapillary effect. [34].

$$V = C \frac{h_0}{\eta} \left[\frac{\partial}{\partial x} (\gamma_{sg} - \gamma_{sl}) + \frac{\partial}{\partial T} (\gamma_{sg} - \gamma_{sl}) \frac{dT}{dx} + \frac{1}{2} \frac{\partial \gamma}{\partial T} \frac{dT}{dx} \right] \quad (2.19)$$

where h_0 is the thickness of the droplet at the center and C is a constant.

Among the mentioned methods to manipulate droplet in this section, only the last two cases are related to the active control and manipulation of the droplet [Figure 2.12 (e) and (f)]. The rest of these methods are passive, and are not discussed in this part.

2.2.3 Device Examples

Ting *et al.* [34] designed a system for active switching and break up of droplets in micro channels (Figure 2.13). A heater is used on one side of a T junction. By turning on the heater the temperature of one branch of the channel will change leading to the change in the surface tension of that branch. Once the droplet reaches the junction, it breaks up into two smaller daughter droplets and the size ratio of the daughter droplets depends on the surface tension of each branch of the channel.

Using the setup in the Figure 2.13, its possible to switch the droplet [Figure 2.14 (b)] or change the size ratio between two daughter droplets [Figure 2.14(a)]. With a little change in the mentioned configuration, making a platform for chaotic mixer is possible [35] (Figure 2.15). As shown in Figure 2.15, there are 2 heaters on the sides of a capillary tube. By turning them on and off, the temperature on each side of the capillary tube changes. This

Figure 2.14: Switching and controlling droplet ratio in the device: (a) Active break up ($T = 32^{\circ}\text{C}$) (b) Active switching ($T = 40^{\circ}\text{C}$). [34].

change leads to the change in the surface tension again and causes the plug to move. The movement of the plug inside the tube induces a fluid flow inside the plug. It is possible to change the frequency of switching and thus a chaotic motion is observed. The criteria to distinguish the chaotic motion is phase portrait diagrams (Figure 2.15). The heaters are turned on and off periodically, so if the motion of the plug is periodic, then the graph must be a close loop. If the graph is not a close loop, there is a chaotic motion. The induced streams inside plug can be useful for mixing in microfluidics. Glockner *et al.* [36] fabricated a cyclic flow control system in closed channels by external heat sources (Figure 2.16). The liquid plug is constrained in a channel with air on both of its sides. When the heater is energized, the surface tension of the channel changes and the liquid moves towards the colder part. When the plug moves, the air pressure on one side decreases and it increases on the other side, so after turning off the heater, the plug returns to its previous position. If the cyclic heat source is connected to a parallel network of micro channels, then microfluidic transport can be controlled in the network [36].

Droplets on a free surface can be manipulated without being constrained on a tube or channel, by treating a surface with hydrophobic and hydrophilic routes and then applying the thermal gradient. Once the droplet is on the surface it will move along hydrophilic lines and is in fact constrained without any physical boundary. Darhuber *et al.* [37] designed a device based on thermocapillary effect to manipulate droplets (Figure 2.17). Chemically modified surface (hydrophobic-hydrophilic) was used with electronically controlled heater array. The hydrophobic or hydrophilic surface helps a droplet to move on a certain path

Figure 2.15: a) Plug movement because of thermal gradient. (b) Phase portrait constructed from the velocity of the plug measured at a time lag of $J = 4$: and different frequencies: (1) $f = 0.1\text{ Hz}$, (2) $f = 0.4\text{ Hz}$, (3) $f = 0.8\text{ Hz}$ and (4) $f = 1\text{ Hz}$. As it is shown, by increasing the frequency, the motion of the plug becomes chaotic [35].

and the heaters make thermal gradient which makes a surface tension gradient and moves the droplet.

Figure 2.16: Cyclic flow control device [36]

Figure 2.17: A device with modified surface to handle droplet by thermal gradient. The hydrophobic/hydrophilic surface confines the droplet to move in certain paths and there is no need of a channel [37].

2.3 Droplet Actuation by Electrowetting

The principle of droplet manipulation by electrowetting is similar to manipulation by thermocapillary effect. A surface tension gradient should be created along the desired direction of movement to move the droplet. In this scheme, this surface tension gradient is the result of electric forces between the droplet and the surface.

2.3.1 Basic Concepts

Electrowetting Mechanism: Electrowetting is the phenomenon related to change of surface tension due to applying voltage on a surface with a droplet. Figure 2.18 shows a droplet on a surface. The surface and droplet can be connected to a voltage supply. When the voltage is connected to the droplet and electrode, an electric field is created. The field at the corners of the electrolyte droplet pulls the droplet down onto the electrode, resulting in lowering the contact angle and increasing the droplet contact area. Lippmann's equation describes the relation between the surface tension and applied voltage [38]:

$$\gamma_{sl}(V) = \gamma_{sl}|_{v=0} - \frac{cv^2}{2} \quad (2.20)$$

where c is the specific capacitance of the dielectric layer, v is the voltage and $\gamma_{sl}|_{v=0}$ is the surface tension of the solid/liquid interface when $v = 0$. It is obvious that by increasing the voltage, the interfacial surface tension in liquid/gas interface decreases. Considering the Young's equation for contact angel (Equation2.15), the contact angel also decreases after applying the voltage.

Figure 2.18: Principle of electrowetting on Dielectric; (a) The droplet before applying voltage; (b) After applying voltage the contact angel has decreased [38].

Based on electrowetting actuation, a series of basic functions on a droplet like transporting, mixing, storing, etc can be realized. By combining these basic functions on a chip a variety of devices for several applications have been designed and fabricated.

2.3.2 Device Examples

Cho *et al.* demonstrated four main actions on droplets on a single chip (Figure 2.19) [38]:

- **Moving droplets:** The chip is composed of an array of addressable electrodes on the bottom, and a covering layer at the top. There is a thin gap between these two layers. The sample droplets are placed in this gap [Figure 2.20 (a)]. The array of electrodes can be controlled electronically. The asymmetric change of the interfacial tension by switching the voltage in electrodes, induces an asymmetric deformation in the liquid meniscus. This deformation causes a pressure difference inside liquid and finally moves the droplet [Figure 2.20(b)].
- **Droplet formation:** If a long liquid column is hydrodynamically unstable, it spontaneously breaks into droplets [39]. To make the column unstable, a certain pressure difference should be created in it. To make this pressure difference a disturbance with the wave length of $\frac{2\pi b_0}{\sqrt{1-\cos\theta}}$ can be used [38]. The chip has a reservoir for liq-

Figure 2.19: Four main actions on droplets: (a) creating; (b) cutting; (c) merging and (d) transporting [38].

Figure 2.20: Structure of the chip: (a) The top and bottom layer and the addressable electrodes; (b) a droplet placed in the thin gap between top and bottom layer [38].

Figure 2.21: Cutting and merging of droplets: (a) to (c) show the cutting process.(d) and (e) show the merging process. [38].

uid. Liquid is pulled out in the form of a long column and then using the mentioned concept, the droplets will be formed.

- **Cutting droplet:** Figure 2.21(a) shows a droplet extended over three electrodes. If the right and left electrodes are energized and the middle electrode is turned off, the contact angel of the both ends of the droplet changes, resulting in cutting the droplet in two parts. Figures 2.21 (a) and (c) show the whole process.
- **Merging droplets:** If two droplets are moved towards each other, they will merge, once they meet each other. Figure 2.21(d) and (e) show two droplets that are merged.

The four mentioned operations have a great potential for making several platforms for lab-on-a chip applications. Mixing and diluting biological samples can be achieved on a chip by electrowetting. Polymerase chain reaction (PCR) chips have been fabricated based on this actuation method [40].

Another interesting application of electrowetting is in optical science. Figure 2.22 (a) shows a droplet on a surface. In equilibrium state, the liquid gas interface of droplet has a spherical shape. This spherical shape and the refractive index of liquid, make the droplet act as an optical lenes. Light passing through the droplet can be focused. The focal length of the lenes depends on its curvature. Using electrowetting effect, the curvature can be tuned electronically. Figure 2.22 (b) is a graph that shows the dependency of focal length and applied voltage in the droplet. This idea has been commercialized to design variable-focus liquid lens for miniature cameras. Figure 2.23 shows the structure of a camera which

Figure 2.22: A droplet acting as a lenes; (a)the curvature of the lenes can be controlled by applying different valtages; (b) the focal length of the droplet lenes versus applied voltage [41].

has a droplet as a lenes as well as the images obtained by this type of lenes [41].

Figure 2.23: Droplet used as a lenes: (a) Optical design of the camera with a liquid as a lenes; (b) the assembled camera and the liquid lenes; (c) image taken by a fixed lenes; (d) image taken by a liquid lenes focused at 50cm; (e)image taken by a liquid lenes focused at 2cm [41].

Figure 2.24: Rayleigh wave spreading on a surface [42].

2.4 Acoustic Manipulation

Surface acoustic waves can be used to actuate droplets on the planar surface of a piezoelectric chip. The liquid can be confined by chemical treatment of the chip surface. A fluidic network can be defined by lithographically created wetting properties of the surface, similar to the wiring of an electronic circuit. Acoustic pressure exerted by the surface wave causes internal streaming in the fluid and eventually actuation of small droplets along certain trajectories. Based on this concept, programmable biochips for a variety of testings on a chip have been achieved.

2.4.1 Basic concepts

Surface Acoustic Waves

As a surface acoustic wave or a Rayleigh wave moves across a surface (Figure 2.24) it moves a surface particle in a circle or ellipse in the direction of propagation. If particles are measured in deeper parts, the particles move in smaller ellipses and reach a "no movement" depth. The maximum distance that Rayleigh waves move particles decreases rapidly with the depth in the material [42]. To make this wave, usually an electric signal is given to a piezoelectric and piezoelectric changes the signal to the mechanical wave.

Acoustic Streaming

The actuation concept for acoustic manipulation of droplets in microfluidics is based on a phenomena called acoustic streaming [43]. Figure 2.25 shows the streams induced in a droplet by acoustic waves. A surface acoustic wave (SAW) is spreading from left to

Figure 2.25: Streams induced inside droplet because of acoustic streaming [44].

right along the x-axis. It reaches the boundary of a liquid at the surface of the substrate in $x = 0$. The propagation of the wave in the droplet, causes a finite difference in the pressure and subsequently density of the fluid. Both pressure and density then oscillate around their respective equilibrium value p_0 , and ρ_0 . The pressure difference directly above the surface of the substrate leads to the excitation of a longitudinal sound wave into the liquid. The wave is launched under a diffraction angle, given by:

$$\Theta_R = \arcsin\left(\frac{v_s}{v_f}\right) \quad (2.21)$$

where, v_s and v_f are the sound velocities of the substrate, and the fluid, respectively. This is because the sound velocities for the solid and liquid substrates are not equal.

The SAW is also responsible for the acoustic radiation pressure

$$p_s = \rho_0 v_s^2 \left(\frac{\Delta p}{\rho_0}\right)^2 \quad (2.22)$$

in the direction of the sound propagation in the fluid. This leads to internal streaming in a droplet, as the boundary of the droplet reflects the actuated fluid to the source. Using a fluorescent dye, the internal streaming inside the droplet can be seen.

Figure 2.26(a), illustrates two snapshots taken 1 s apart from each other approximately. When the amplitude of SAW increases, the acoustic radiation pressure deforms the droplet surface at the other side of the sound entrance. This can be seen in Figure 2.26(b) [44].

2.4.2 Manipulation of Droplets by Acoustic Waves

Based on surface acoustic waves and acoustic streaming, different device examples are described in following:

Figure 2.26: The effect of surface acoustic wave in droplets: (a) for small amplitudes a stream is induced inside droplet, as its shown by fluorescent dye inside; (b) for bigger amplitudes of wave the drop is deformed [44].

Figure 2.27: The device that induces streams inside droplet by acoustic wave: a) The beginning of the experiment, b) 240 ms after starting mixing c) after 1 s d) after 5 s. e)The device itself [43]

Mixing

Strobl *et al.* (2002) [43] made a device in which the stream was induced in the droplet by a SAW [Figure 2.27 (e)]. As discussed above, when the SAW propagates over a surface with a droplet, it can induce fluid streams inside the droplet. These streams can be used for stirring. This technique is useful for droplets with a volume between 5 micro liter and 500 pl. The duration of a mixing experiment depends on the RF power. Increasing the RF power reduces the time for stirring a droplet. Figure 2.27 (a) to (d) shows a sample of mixing by this device where the volume of drop is 5 micro liter and the bright spots are fluorescent dye. The wave is propagated from the left to the right with a frequency of 114 MHz.

Figure 2.28: (a) A droplet is moving from the right to the left, the cross shows the starting position. The droplet consists of a aqueous 1 percent Bromophenol Blue solution; (b) Two droplets are transported on one fluidic track [43].

Figure 2.29: Two droplets moving in different pathes on a surface [43].

Actuating Small Droplets

If the power of acoustic wave increases to a certain value, it can move a droplet on its path. Figure 2.28 shows a simple setup for this purpose. Using surface treatment techniques, it is possible to make hydrophobic and hydrophilic lines that can act as virtual channels. There are several methods to change and adjust the wetting properties of the surface of a chip. The most frequent technique is to silanize certain areas of the surface while keeping the other areas protected by photoresists.

It is also possible to design a chip with a more complex layout, as shown in Figure 2.29. On a chip with several defined lines one can manipulate two or more droplets at the same time and independent from each other. Also, it is possible to guide a droplet to a crossing line and let it switch. Such a layout makes the movement of a droplet possible to every position on a chip[43].

2.5 Droplet Actuation by Centrifugal Force

It is possible to obtain flow rates ranging from less than 10 nl /s to more than 100 $\mu\text{l/s}$ by centrifugal force using a rotating disc. The value depends on disc geometry, rotational rate (RPM), and fluid properties [45]. This pumping method is insensitive to physicochemical properties such as pH, ionic strength, or chemical composition, while AC and DC electrokinetic means of pumping for example, are dependent on such parameters. Several fluids like blood, milk, and urine, solvents (e.g., DMSO) and surfactants have been pumped successfully. A greater range of types and amounts of fluids are possible to be pumped by centrifugal pumping rather than electrokinetic and acoustic pumps. This approach seems to be more suitable than acoustic and electrokinetic approaches.

Moreover, miniaturization and integrating are easily possible for both continuous and droplet based microfluidic systems. A variety of functions like decanting, valving, calibration, metering, mixing, sample splitting, and separation, can be done on this platform [46].

2.5.1 Working Principle

When a disk rotates with the angular velocity of ω , there will be a centrifugal force exerted on the components of the disk. This force is proportional to the rotation rate, geometry and location of channels and reservoirs, as well as fluid properties. The flow rate of aqueous solutions in fluidic CD structures were characterized by Madou *et al.* [47] and Duffy *et al.* [45] and the result were compared with simple centrifuge theory. The average velocity U of the liquid is given as:

$$U = \frac{D_b^2 \rho \omega^2 \bar{r} \Delta r}{32 \mu L} \quad (2.23)$$

and the volumetric flow rate is $Q = UA$ where D_b is the hydraulic diameter of the channel (defined as $4A/P$, where A is the area of cross-section and P is the perimeter of the channel), ω is the angular velocity of the CD, ρ is the density of the liquid, $\bar{r}$ is the average distance of the liquid confined in the channels from the disc center, Δr is the radial extent of the fluid, L is the length of the liquid in the capillary channel and μ is the viscosity of the solution (see Figure 2.30).

Figure 2.30: Schematic illustrations for the description of CD microfluidics. Two reservoirs connected by a microfluidic chamber [46].

Flow rates in the range of 5 nl/s to greater than 0.1 ml/s, have been achieved by different combination of rotational speeds (from 400 to 1600 rpm), channel widths (20 to 500 μm) and channel depths (16 to 340 μm) [46].

2.5.2 Components

Based on the mentioned principle a number of fluidic components have been fabricated to demonstrate some basic operations. By combining these basic functions, sophisticated devices for decanting, sample splitting, separation have been made. By integrating these components it is possible to miniaturize many of lab functions on a CD chip. This chip is called lab CD. Here only two most important of these basic functions are discussed: Mixing and valving as rest of the functions are based on them.

Mixing of Fluid

Diffusional mixing can be implemented by joining two microfluidic channels together into a single long serpentine fluidic channel. Stoichiometric mixing in the channel by keeping equal flow rates of the two streams joining into the mixing channel can be obtained by Proper design of channel length and reagent reservoirs [48].

In addition to diffusion, mixing can be also created by turbulence on the CD by emptying two narrow streams into a common chamber. The splash of fluids to the chamber wall

Figure 2.31: A lab on a CD. A variety of mixing operations can be done on such a platform. (a) The valving diagram; (b) The rotary disc with arrays of valves and chambers shown in (a); (c) The fabricated device [45].

causes their effective mixing. Figure 2.31 shows a mixer made by Duffy *et al.* [45].

Figure 2.32: Two types of valves used in CD type chips; (a) Hydrophobic valve made by a constriction in a chamber made of hydrophobic material; (b) Hydrophobic valve made by the application of hydrophobic material to a zone in the channel; (c) Hydrophobic channel made by the application of hydrophobic material to a zone in a channel made with structured vertical walls; (d) Capillary valve made by a sudden expansion in channel diameter, such as when a channel meets a reservoir [46]

Valving

Two main types of valves are used in CD labs: Hydrophobic valves and capillary valves. Both of the types have been integrated into CD chips [49, 50, 51, 52].

Hydrophobic Valves

The schematic of three types of hydrophobic valves is illustrated in Figure 2.32. In all the cases the hydrophobic surface prevents the flow of the liquid. For liquid to pass these barriers, the CD must rotate above a critical speed, where the centrifugal forces exerted on the liquid column overcome the pressure needed to pass the valve.

Capillary Valves

Capillary valves have been used on CD chips in several research works [51, 53]. The physics behind is based on the surface tension, which increases when the cross section of a hydrophilic capillary expands suddenly, as shown in the Figure 2.32(d). The maximum pressure at the capillary barrier expressed based on the interfacial free energy, when the

capillary cross section is axisymmetric, is given by:

$$P_{cb} = \frac{4\gamma_{al} \sin \theta_c}{D_b} \quad (2.24)$$

where θ_c is the contact angel, γ_{al} is the surface tension between air and liquid and D_b is the hydraulic diameter.

On the other hand, the liquid pressure at the meniscus, from the centrifugal force acting on the liquid, can be described as follows:

$$P_m = \rho \omega^2 \bar{r} \Delta r \quad (2.25)$$

So the criteria for the valve to be opened is that $p_m > p_{cb}$. Thus the minimum rotational velocity for opening the valve can be obtained using the above equations.

Based on these basic components and some other techniques, a variety of devices with this technology have been made. The examples are : molecular diagnoses chips on CD, metering (volume definition), chromatography, etc [46].

Figure 2.33: Magnetic field around a magnet bar and a wire with current.

2.6 Manipulation by Magnetic Force

As reviewed in previous section, manipulation of free droplets on solid surfaces or channels using different methods can provide versatile platforms with significant applications in biotechnology, and the medical science. The drawback with a number of previous methods is sample evaporation, the requirement for large voltages (electrowetting) or great thermal gradient (thermocapillary), and complicated surface modifications.

magnetic force is an alternative option for manipulation of droplets or particles. In spite of electric field, the effect of magnetic field on particles does not depend on ionic concentration, PH or temperature. The interesting point about magnetic field is that, in spite of general belief, the objects that are neither magnetic, nor have any charge also can be manipulated by magnetic force, so magnetic force can actuate a wide range of particles and materials [54].

2.6.1 Basic concepts

Magnetic Field

A magnetic field is the region in the neighborhood of a magnet, electric current, or changing electric field, where magnetic forces are observable. Figure 2.35 shows the magnetic field around a magnet and a wire with current.

In a magnet, the lines of the field start from north pole and end to the south pole. In a

wire with electric current, direction of the field can be determined by right hand rule.

Magnetic Material

There are three main categories of materials according to their magnetic properties [15]:

- Ferromagnetic materials,
- Paramagnetic materials,
- Diamagnetic materials,

Materials that respond strongly to a magnetic field are known as ferromagnetic. Magnetic permeability shows the ability of a material to be magnetized or to amplify the magnetic field in its vicinity. The permeability of Ferromagnetic materials is as big as 1,000 (or more) times that of the free space (vacuum). Paramagnetic materials are weakly attracted by a magnetic field and have permeabilities slightly greater than that of free space. A few materials, such as bismuth or antimony have a negative permeability and are repelled by a magnetic field. These materials are called diamagnetic. As the force experienced by paramagnetic and diamagnetic materials is very small, they are usually called nonmagnetic materials and therefore ferromagnetic materials are called magnetic materials.

The Origin of Magnetism, Definition of Magnetic Hysteresis and Saturation

The origin of magnetism should be investigated in the atomic level. Electrons have electric charge and turn around both themselves (spin) and nucleus in the atom (Figure 2.34). This motion of electric charge creates a weak magnetic field. The direction of this field depends on the spin of electron. According to the exclusion principle in quantum physics, two electrons that are in the same energy level can not have the same spin. For many materials, all the electrons are paired in an energy level, so that their magnetic fields cancel each other. These material are diamagnetic. In some atoms, however, the number of electrons with spin in one direction is not equal and this results in a net magnetic field for the atom (paramagnetic and ferromagnetic materials). The magnetic fields of the atoms in a certain region tend to align in the same direction, so that they strengthen each other. Such a region is called a domain. The domains have different size and orientantion in an un

Figure 2.34: The spin and the orbital rotation of the electron as the origin of magnetization in material.

Figure 2.35: Schematic illustrations for magnetic domains inside a ferromagnetic material: (a) Before alignment; (b) Alignment by an external magnetic field.

magnetized sample. When an external magnetic field is applied, domains will be aligned with the direction of the field, as shown in Figure 2.35.

When the domains disappear in all other directions, the remaining domains rotate to align with external field. After the rotation is complete, no further magnetization takes place. A saturation point is reached in this condition. In the case of ferromagnetic materials, if the external field is then reduced to zero, the sample still retains most of its magnetism; this is known as hysteresis. In paramagnetic materials after removing the external field, the alignment will be disturbed quickly and the magnetization is no longer present in the material [15].

2.6.2 Forces in a Magnetic Field

Three types of objects can be affected by a magnetic field: magnetic materials, non magnetic but electrically charged objects and diamagnetic materials.

- Magnetic materials (ferromagnetic materials) : In order to manipulate particles or droplets by magnetic field, a magnetic field gradient is required to exert a translation force. A uniform field can only create a torque, but no translational motion. The magnetic force exerted on a point-like magnetic dipole or magnetic moment m can be expressed as the derivative of the magnetic energy [55, 56]

$$F_m = \frac{1}{\mu_0} \nabla(m \cdot B) \approx \frac{1}{\mu_0} (m \cdot \nabla) B \quad (2.26)$$

The second part of the equation is applicable when the magnetic moment of the particle is not varying in space, i.e ($\nabla \cdot m = 0$). This is only a correct assumption when the moment is permanent, or the magnetization of the magnetic particle is completely saturated. For example, for a constant moment m in the x direction, $m \cdot \nabla = m_x \frac{\partial}{\partial x}$ and a force will be exerted on the moment, provided that there is a magnetic field gradient in the x direction [57]. In the case of superparamagnetic nanoparticles in a nonmagnetic medium, one can write for the moment $m = V\mu_0 M = V\mu_0 \chi H$, with m being the magnetization of the particle and χ is the magnetic susceptibility. Using the relation $B = \mu_0 H$, Eq. (2.26) becomes:

$$F_m = \frac{V\chi}{\mu_0} (B \cdot \nabla) B \quad (2.27)$$

where V is the volume of the object. For example, when the magnetic particles are constrained to move in a microchannel in the x direction, the x component of the force becomes:

$$F = \frac{V\chi}{\mu_0} (B_x \frac{\partial}{\partial x} + B_y \frac{\partial}{\partial y} + B_z \frac{\partial}{\partial z}) B_x \quad (2.28)$$

This equation shows that both the magnitude of the magnetic field (for inducing a large moment) and the magnetic field gradient need to be large, to have a strong magnetic actuation force. However, one should be reminded, that if both the field

and the moment have a spatial dependence, the first part of Eq. (2.26) needs to be considered and the force expression becomes more complicated [56].

- Non magnetic but electrically charged objects:

If there is a particle with the velocity of V and charge of q in a magnetic field, there is a force acting on the particle equal to [58]:

$$F = qV \times B \quad (2.29)$$

- Diamagnetic materials:

Contrary to our intuition, the objects that are apparently nonmagnetic can be manipulated by a magnetic field and can stabilize free levitation under the field of a permanent magnet [59, 54]. This phenomena is mainly related to eddy current in diamagnetic materials. When a moving (or changing) magnetic field passes a conductor, or vice-versa, eddy current will be induced inside the conductor. The relative motion of magnetic field and the conductor causes a circulating flow of electrons, or current, within the conductor (Figure 2.36). According to Lenz's law these eddies of current create electromagnets with magnetic fields that oppose the effect of the applied magnetic field. The strength of the opposing magnetic field is proportional to the conductivity of the material, relative velocity of motion, and intensity of magnetic field. Figure 2.37 shows a living frog levitated by a magnetic field [54].

2.6.3 Magnetic Beads and Magnetic Fluids

Magnetic beads are very small, super paramagnetic particles with a size range of 0.8 to 5.0 μm in diameter. They are able to maintain a magnetic domain under excitation. These beads have high magnetic susceptibility and show magnetic dipoles when placed in a magnetic field, but do not exhibit residual magnetization when the external field is removed. Hence, after removal of the external magnetic field, the separated beads are dispersed in the suspension.

Magnetic bead solutions or magnetic fluids are stable dispersions of encapsulated magnetic particles in an aqueous or organic carrier medium. The difference between the two

Figure 2.36: As the metal plate moves in the magnetic field, eddy currents are induced in the plate.

types of fluids is defined by the size of the particles. Ferrofluids and magneto rheological fluids are the two main types. Magneto rheological fluids are stable suspensions of micron sized magnetically polarisable particles suspended in a carrier fluid. These fluids vary their viscosity with the applied magnetic field and can solidify in the presence of a sufficiently big magnetic field (Figure 2.38). In comparison, colloidal ferrofluids that are typically a suspension of magnetite particles of approximately 10 nm in diameter, can flow even in the most intense applied magnetic fields. Ferrofluids are usually defined as magnetically soft material, that means the magnetism vector follows the applied field without hysteresis, and a saturation occurs by a small applied field.

Synthesis and Characterization of the Ferrofluid

Previously, magnetic bead solutions were produced by grinding magnetite (Fe_3O_4) with long-chain hydrocarbons and a grinding agent [62]. Later, magnetic fluids were produced by precipitating an aqueous $\text{Fe}^{+3}/\text{Fe}^{+2}$ solution with a base, coating these particles with a layer of oleic acid and then pouring them in a non-aqueous fluid [63]. Both types of

Figure 2.37: The frog levitated by magnetic force [54].

Figure 2.38: Magnetorheological fluids :a)when there is no magnetic field b)when the fluid is solidified under magnetic field [60]

process can produce very small magnetic particles with a coating of surfactant in a non-aqueous liquid. The coating of the particles makes them hydrophobic. However, in many applications of magnetic beads water has been used as the continuous phase. A water-based magnetic fluid was realized by the conversion of iron products to magnetic iron oxide in an aqueous medium under controlled pH conditions [64]. Polymer-coated magnetic beads can be produced by in situ precipitation of magnetic materials in the presence of a polymer [65].

Using this method, magnetic beads surrounded by a hydrophilic polymer shell have been made. It is possible to choose for the polymer the water-soluble dextran [66], poly-(ethylene imine) [67], poly-(vinyl alcohol) [68], poly-(ethylene glycol) ([69], etc. Figure 2.40 (a) shows a transmission electron microscopy (TEM) image of dextran nanoparticles; note that the dextran is not visible in the TEM image and that the particles have an irregular

Figure 2.39: Top and side view of a droplet of ferrofluid [61]

Figure 2.40: (a) A transmission electron microscopy (TEM) photograph of dextran nanoparticles; note that the dextran is not visible in the TEM picture and that the particles have an irregular (non-spherical) shape. (b) A TEM photograph of more regular polystyrene particles with encapsulated magnetite nanoparticles [70, 65].

(non-spherical) shape. Figure 2.40 (b) shows a TEM image of more regular polystyrene particles with encapsulated magnetite nanoparticles [65]. Single basic types of magnetic beads have their individual advantages and disadvantages. Polystyrene coated magnetic particles are known for their very good size distribution and spherical shape. However, due to their hydrophobic surface, a high amount of unspecific protein and antibody binds on the particle surface, so it should to be modified chemically. For adsorbing proteins and DNA on the surface, Magnetic silica particles are a very efficient choice, but the types with small size distribution and an ideal spherical shape is hardly available [71]. Magnetic polysaccharide particles are important for many in vivo applications. They are biocompatible and available in a size range below 300 nm [72, 70]. Their drawback is that the particles are irregular in shape and they are sensitive to mechanical stress. Also, magnetic poly-(lactic acid) particles play an important role in in vivo applications [73]. They are biodegradable and their degradation time in the blood is adjustable by their molecular weight and exact

chemical composition. However, because of their hydrophobic surfaces, these particles stick to plastic surfaces like pipette tips and plastic recipients, resulting in problems with particle handling and analytical errors [65].

2.6.4 Manipulation and Active Control of Droplets and Particles by Magnetic Force

Nguyen *et al.* introduced a method for manipulating magnetic droplets on a free surface [74]. The device consists of a PCB that has coils on both sides. The coils on the top are connected to the same current direction. The magnetic field created by them forms a potential well which traps the droplet, so the coils create a virtual channel in which the droplet will be bounded. The droplet itself is magnetized by two permanent magnets and is affected by the magnetic field of the coils.

The coils at the bottom of the board are connected to the current with opposite direction, so the magnetic field caused by them has a gradient along the x direction and drives the Ferrofluid droplet. The direction of the droplet motion is controlled by switching the current direction in these driving coils.

Song *et al.* designed and fabricated a device for manipulation and mixing of aqueous micro drops, using ferrofluid as a carrier liquid [75]. When a nonmagnetic fluid that is immiscible with the ferrofluid, is placed on top of ferrofluid and a magnetic field is applied, a magnetic pressure jump at will be created in the interface of the nonmagnetic fluid and ferrofluid due to the magnetization gradient.

By changing the configuration of the magnetic field, it is possible to control the spatial change in magnetic pressure jump dynamically from point to point, thus inducing a dynamic deformation of the interface, i.e., shape change and movement of nonmagnetic fluid (droplet in this case). Figure 2.41 shows the elements of the device. Ferrofluid is spread on the surface of silicon in zigzag shape and an aqueous droplet is placed on top of that. A strip magnet was used to keep the shape of the ferrofluid lines. By rotating the magnetic stirrer, the gradient of magnetic field is changed and thus a pressure is created to move the droplet. The above concept can be used for mixing the droplets with a little change. Figure 2.42 shows the elements of the mixer. The second layer of strip magnet is placed to

Figure 2.41: Schematics of ferrofluid-based platform for the manipulation of free water droplets: (a) periodic line generation by a single strip magnet and (b) dynamic pattern generation of ferrofluid under the rotation of a magnetic stirrer [75].

Figure 2.42: (a) Schematic diagram of the experimental setup and b) Optical images of droplet mixing using discontinuous ferrofluid pattern. Droplet size: 8 microliter [75].

reduce the continuity of magnetic fluid, to ease the mixing of droplets. The important benefit with this method is that the samples don't need to be magnetic or tagged by magnetic particles. A wide range of samples can be manipulated by this system. Lee *et al.* fabricated an integrated cell manipulation system on CMOS chips [76]. Their device consists of an array of micro coils and electronic circuit to control the magnetic field generated by each coil. Each coil is designed to have the maximum intensity of the magnetic field in the middle as seen in Figure 2.43. Cells are tagged with magnetic particles and magnetic field exerts force to them. By controlling the intensity of the magnetic field in each coil and the location of potential well, the cell is moved to any desired location in the domain that is favorable. Figure 2.44 shows that by controlling the point where the interaction of two neighbor magnetic field has maximum magnitude (trap point), the position of the cell can be controlled. Lyuksyutov *et al.* designed and fabricated a chip for manipulation of

Figure 2.43: The device composed of microcoil integrated with CMOS IC to manipulate cells [76].

non magnetic droplets [59]. In this chip first eddy current is induced in the droplet. These eddy currents magnetize the non magnetic droplets (oil and water) and then the droplet is manipulated by magnetic force. Based on this concept, droplets can be moved along the channel [Figures 2.45 (a) to (c)] or mixed together [Figure 2.45(d) to (f)]

Shikida *et al.* made a device for magnetic handling of droplets utilizing surface tension and wettability [77]. The device can handle three main actions: reaction, separation and dilution for samples suspended in the droplets. The samples are tagged with magnetic beads immersed in oil. Moving the permanent magnet under the channel displaces the droplet. Reaction is shown in Figure 2.46 (a). By moving the magnet, the sample suspended in the buffer crosses the glass gate and joins the reagent. For separation, the droplet is moved near the gate and once it goes under the glass gate the sample is separated from the main droplet and remains there, mainly because of poor wettability under the glass gate [Figure 2.46(b)]. For dilution the mechanism is similar to reaction [Figure 2.46 (c)].

Lehman *et al.* designed a chip for two dimensional magnetic manipulation of micro droplets [13]. The chip is placed over a permanent magnetic sheet to magnetize the beads inside droplet. There are series of planar coils on the PCB chip. These coils can produce a magnetic field to manipulate the magnetized droplets [Figure 2.47 (a)]. The droplets are on

Figure 2.44: By controlling the position of potential well made by adjoining coils, it is possible to move cells (tagged by magnetic beads) to any location inside the plane [76].

a Teflon sheet. A series of hydrophobic/hydrophilic patterning are performed on the Teflon to make a barrier for droplets. This barrier restricts the droplet to go on favorite directions. As the friction between the droplet and Teflon reservoir is negligible, the magnetic force can be equal to viscous drag force: $F_m = F_d$ and $F_d = 3\pi D_h \eta V$ Where D_h is the hydraulic particle diameter, η is the viscosity of the oil and V is the droplet velocity. Figure 2.47 (b) shows that with this system its possible to mix droplets (for dilution, for example) or separate them. In Figure 2.47 (b), the big droplet is a washing site and the small droplet is containing the sample. With one passage of sample through the washing site, it will be diluted by 6 times. So after N rounds of washing the sample will be diluted by 6^N times.

Dorvee *et al.* performed a special process on silicon to gain magnetic beads which can be used as carriers of several kinds of fluids in droplet format [78]. The process to gain the beads is shown in Figure 2.48 .

These beads can be used to be loaded with liquid droplets and unloaded to deliver liquid on bio chips like what is shown in Figure 2.49 (a) to (e): In Figure 2.49(a) the beads are filled with water and then moved by magnetic force next to a capillary tube. The beads have amphiphilic characteristics, this is why they can carry liquids. In Figure 2.49(b) and (c) the water inside the droplet is extracted out by capillary effect and in figure 2.49(d) the remained droplet is moved back by magnetic force in its first location and in Figure 2.49(e) refilled with blue liquid. So these beads can be a carrier for droplets on microfluidic

Figure 2.45: The droplet levitated in a magnetic field [59].

applications. The other characteristic of these beads is that they reflect light by different intensities based on they are filled with liquid or are empty and also based on what type of liquid is inside them [see Figure 2.49 (f) and (g)].

Greivell *et al.* designed a ferrofluid magnetic pipet [79] [Figure 2.50 (a) to (c)]. The ferrofluid is used as a plug so if it moves, it will push the liquid in the front. A magnet was used to move the ferrofluid plug. As the magnet moves the ferrofluid follows, thus the fluid in the tube is moved up. Hatch *et al.* (2001) developed the above concept and designed a magnetic pump [80] [Figure 2.50 (b)]. Two ferrofluid plugs are inserted in a circular path and a permanent magnet is used to rotate one plug and another magnet to keep the other plug in a fixed position. When the magnet moves, the translating plug follows it and pushes the fluid in front of itself from one side and sucks the fluid in the back side. As the second plug is fixed, the pushed fluid only can go out of the circular path and this is how the liquid is pumped from point A to point B in one cycle. When the translating plug reaches the stationary plug, the magnet still moves and causes formation of another rotary plug and the cycle goes on. The whole sequence is shown in Figure 2.50 (d) to (f).

Yamahata *et al.* used the ferrofluid for pumping with the following mechanism [81]: The plug of ferrofluid is surrounded by oil, to avoid the contact by pumping liquid. When the permanent magnet moves up and down causes the plug to move up and down. The down

Figure 2.46: Some basic operations on a chip : (a) Reaction; (b) Dilution; and (c) separation mechanism [77].

Figure 2.47: The array of micro coils on a chip that can move the droplet for mixing or dilution. [13].

Figure 2.48: The process to make the microbeads [78].

Figure 2.49: The beads as a carrier of fluid in liquid scale. In (a) The beads are filled with water and then moved by magnetic force next to a capillary tube. In (b) and (c) the water inside the droplet is extracted out by capillary effect and in (d) the remained droplet is moved back by magnetic force in its first location and in (e) refilled with blue liquid. (f) and (g) Using light reflection its possible to distinguish whether the carrier is filled or empty [78].

Figure 2.50: Magnetic pipet and the magnetic pump based on that. (a) By moving the ferrofluid plug its possible to move the liquid; (b) The sequence of pumping by ferrofluid plug ((i) to(iv)) [80].

movement of plug sucks water from the inlet. The upward movement of plug pushes the water to the outlet (Figure 2.51). As there are 2 check valves on inlet and outlet on reverse directions, the pumped liquid never goes to unwanted directions during the process.

Rong *et al.* made an on chip magnetic bead separator for bio cell sorting [82]. Their device is composed of micro channels integrated with 3 pairs of magnetic tips. The operational principle of the magnetic bead separator is illustrated in Figure 2.52. Before excitation, the magnetic bead solution flows from the inlet through the microchannel. The flow is evenly distributed between outlet A and B [Figure 2.52(a)]. When a driving signal is

Figure 2.51: Ferrofluid plug in a capillary used as a magnetic pump [81].

Figure 2.52: A magnetic separator. (a) The state with no excitation (b) Two tips on the right are excited and the beads are converted to outlet A and C. (c) The two tips on the top are excited and the beads are converted to the outlet B and C. [82]

applied to one of the inductor pairs, the magnetic beads will be switched to a specific channel under the magnetic forces exerted on them. As shown in Figure 2.52(b), if a driving signal is applied to the left pair of the inductors, the magnetic beads will be sorted to the left channel and flow out through outlet A. The magnetic beads can also be switched to outlet B if the driving signal is applied to the respective pair of the inductors [Figure 2.52(c)]. Doyel *et al.* (2002) reported a self assembling magnetic matrices for DNA separation [83]. If a suspension of magnetic particles limited in a thin gap [Figure 2.53 (a)] is subjected to a constant, homogenous field, the magnetic particles will be self organized in the gap, in the shape of regular array of columns [Figure 2.53 (b)]. When the field is off, the array returns to the liquid suspension state. This array acts as a filter for separating DNA. The shape and the space between columns of this array depends on the intensity of applied field as well as the size of the magnetic beads. So by changing these two parameters it is possible to make filters with different sizes (different separations are possible). The other benefit of this method is that by removing the magnetic field, the suspension will be like a liquid and can be washed away easily.

Fluorescent intensity is a technique used for detecting DNA. The graphs in Figure 2.53 (c) and (d) show the fluorescent intensity at the 10 mm from the injection zone versus time which proves successful separation of DNA by this method.

Nguyen *et al.* fabricated a chip for rapid polymerase chain reaction [3]. The device

Figure 2.53: DNA separation by suspension of magnetic beads : (a) The gap filled with magnetic suspension (b) formation of matrices under magnetic field (c) Fluorescence intensity at 10 mm from the injection zone versus time. Separation of a mixture of λ -phage DNA and λ -DNA digested with Xho I (Pharmacia) (the numbers refer to the size of the DNA fragments in kbp), at fields strengths of 4.8 (right), 7 (middle), and 10 left V/cm. (D) Separation of λ -phage DNA concatemers at a field of 3.2 V/cm. Other conditions are identical to those in (C) [83].

consists of microchannels made in PMMA by laser machine (Figure 2.54). Heaters and thermal sensors have been used to create required thermal cycles for DNA amplification. A plug of ferrofluid is placed inside the channel and carries the initial samples of DNA. The plug is then rotated in a circular path by a rotary permanent magnet. During the rotation thermal cycling affects the sample and cause the required reactions for amplifying the DNA. Previous samples of PCR devices were based on continuous flow pumping that required high bonding strength. Using a magnet as the actuator eliminates the need for expensive pumps and provides the advantages of low cost, small power consumption and lower strength of bonding. Furthermore, the microchip has a much simpler design and smaller footprint compared to the rectangular serpentine design. Experimental evidences show that successful PCR has been achieved in less than 4 minutes.

Figure 2.54: Polymerase chain reaction device. The ferrofluid plug carries the DNA samples and is actuated by a rotary permanent magnet. The device is capable of duplicating DNA in a very short time (4 minutes for a specific sample [3])

Chapter 3

Modeling and Optimization of Planar Microcoils for Magnetic Actuation

3.1 Introduction

As mentioned in the literature review, microfluidics started with manipulation of continuous flow in micro and nano channels but recently a paradigm shift from continuous flow systems to droplet-based systems has been observed. In the next two chapters, design and fabrication of a device for manipulating ferrofluid droplet by magnetic force will be discussed. This chapter specifically is dedicated to modeling and optimization of planar microcoils. The result will be later used in design and modeling of a device for magnetic actuation of droplets.

Recently, magnetism has found many applications in microfluidics. Typical examples are magnetohydrodynamic micropumps [84, 85], ferrofluid droplets for sample transport in small scales [74, 86, 87], magnetically doped Poly dimethyl siloxane (PDMS) to be used as actuator [88] and magnetic micro particles for improved mixing [88]. In life sciences, magnetism also has been used for trapping, sorting and separating cells and transporting biological objects [76, 89, 90]. Magnetic particles have been used for DNA separation as an alternative for gel matrix [83, 91].

Using magnetic force for manipulation of particles or droplets offers a number of advantages. Objects can be transported or actuated by external magnets or coils that are not in

contact with the fluid. In spite of the electric field, the effect of magnetic field on particles is generally not influenced by surface charges, pH, ionic concentrations or temperature [88].

Two basic components of the device that is going to be designed and fabricated are ferrofluid droplet and planar micro coils. magnetic beads and ferrofluid are possible options to work with magnetic force. Magnetic beads with diameters ranging from 250 nm to 6 μ m can not be dispersed homogeneously inside a droplet under the influence of magnetic force. The reason is that Brownian forces cannot overcome the magnetic forces acting on these large beads. In the worst case the droplet will be divided into two parts: one with magnetic beads and one without magnetic beads. As a result the droplet without magnetic beads can not be manipulated in the magnetic field. Fortunately the particles in a ferrofluid have diameters on the order of few nanometers. At this scale, the random Brownian force is larger than the magnetic force and there will be no problem of dispersion. Therefore, ferrofluid droplets can potentially be used as carrier for handling samples in a droplet-based lab-on-a-chip device.

Microcoils have a great potential to be used in miniaturized devices as functional components such as valves [92], magnetic particle carriers [13, 93], relays [94] and even sensors for near-surface material properties [95]. Microcoils have been used for digital microfluidics, where magnetic droplets can be positioned and transported by external magnetic fields [74, 86, 76]. The magnitude of the magnetic flux can be well controlled by the current passing through the coil. The current itself is to be controlled by external electronic circuits, micro controllers or PC. Thus, microcoils allow further flexibilities in designing the devices for digital microfluidics. The other benefit of planar microcoils is the existence of an analytical model to calculate their magnetic field and flux. The model enables the prediction of the device behavior and the optimization of its design. The first step towards using planar coils in microfluidic devices is the development of an efficient model.

Busch-Vishniac discussed the use of magnetically driven micro actuators and the possibility of generating magnetic forces strong enough for actuation [96]. Recently several works have been reported on modeling of magnetic actuators. Ko et al. reported a numerical analysis for a micro machined magnetic actuator using finite element method to identify critical parameters that affect the performances of a micro actuator. Their numerical results indicate that the dimensions of the core and the magnetic material have an influence on the

performance of planar actuators [97]. Lin et al. used a numerical model with FARADAY software to study the effect of coil parameters in stimulating live tissues [98]. They found the governing parameters in the coil design for optimum field penetration such as coil configuration, diameter, and number of turns in one loop of the coil. Feustel et al. reported a numerical simulation for optimization of planar electromagnetic actuators [99]. Jansen et al. used an analytical model for calculating the torque exerted by a stator coil on the array of permanent magnets [100]. Garcia et al. modeled the magnetic field of only one straight segment of current carrying wire with finite length [101] and proposed a numerical procedure for calculating the magnetic field of the coils. A number of studies were also conducted to calculate the inductance and impedance of magnetic structures. Hurley et al. calculated the impedance for planar structures with spiral windings [102]. Mohan et al. obtained some simple expressions for inductance of spiral coils [103].

Many of the previous works are based on numerical solutions. Our contribution in this part of the research is extending the analytical model of the magnetic field of a single segment of wire to calculate the magnetic field strength, flux and resultant force of a rectangular, planar electromagnetic microcoil. The optimization of the coil parameters is discussed to maximize the field strength and force for application in magnetic actuation of droplet-based microfluidic devices.

3.2 Model of planar microcoils

3.2.1 Magnetic field of a straight wire segment with finite length

Figure 3.1(a) shows the wire arrangement of a planar rectangular coil. Each straight segment of the coil is termed here as a wire segment. In modeling, the width and the height of the wires are not considered in the modeling, as the length of the wire is much much bigger than the two other dimensions. The gap size g is defined as the center to center distance between two neighboring parallel segments. Four consecutive wire segments form a turn. Thus, if the number of segments is n , the number of turns is approximately $n/4$. As shown in Fig. 3.1, planar coils are composed of a series of current-carrying wires with finite length, Figure 3.1(a). To calculate the magnetic field of a planar coil, the magnetic field

created by each wire of a finite length should be obtained in the calculation domain. The superposition of the magnetic fields of all these segments results in the total magnetic field of the coil.

Applying Biot-Savart's law to a segment of wire carrying current I , parallel to $x-z$ plane results in [101]:

$$\begin{aligned}
 H_x &= 0 \\
 H_y &= -\frac{zI}{4\pi} \int_{x_1}^{x_2} \frac{dl}{[\sqrt{(x-l)^2 + (y-a)^2 + z^2}]^3} \\
 &= -\frac{zI}{4\pi[(a-y)^2 + z^2]} \times \left[\frac{x_2 - x}{\sqrt{(x_2 - x)^2 + (a-y)^2 + z^2}} \right. \\
 &\quad \left. - \frac{x_1 - x}{\sqrt{(x_1 - x)^2 + (a-y)^2 + z^2}} \right] \\
 H_z &= \frac{(y-a)I}{4\pi} \int_{x_1}^{x_2} \frac{dl}{[\sqrt{(x-l)^2 + (y-a)^2 + z^2}]^3} \\
 &= -\frac{(a-y)I}{4\pi[(a-y)^2 + z^2]} \times \left[\frac{x_2 - x}{\sqrt{(x_2 - x)^2 + (a-y)^2 + z^2}} \right. \\
 &\quad \left. - \frac{x_1 - x}{\sqrt{(x_1 - x)^2 + (a-y)^2 + z^2}} \right]
 \end{aligned} \tag{3.1}$$

where x, y, z are the coordinates of the point where the magnetic field is to be calculated, x_1 and x_2 are the x coordinate of the two ends of the segment of wire, I is the electric current passing through the wire, a is the distance of the wire from the $x-z$ plane and dl is the differential spatial element on the wire segment, Figure 3.1(b).

Equation (3.1) describes the magnetic field for the horizontal segments of a planar coil. The magnetic field of the vertical segments can be calculated in the similar manner. The calculation can be simplified by a coordinate transformation. According to Figure 3.1(c), (3.1) can be formulated for a vertical segment using the following coordinate transformation:

$$(x, y, z) \rightarrow (y, x, -z) \tag{3.2}$$

After the coordinate transformation, the magnetic field components of a vertical seg-

Figure 3.1: Wire arrangement for modeling a planar coil: (a) Wire arrangement of a planar rectangular coil. (b) Arrangement of a horizontal wire segment with finite length. (c) Coordinate transformation to formulate the magnetic field of a vertical segment of wire.

ment can be determined as:

$$\begin{aligned}
 H_x &= \frac{zI}{4\pi[(a-x)^2 + z^2]} \times \left[\frac{y_2 - y}{\sqrt{(y_2 - y)^2 + (a-x)^2 + z^2}} - \frac{y_1 - y}{\sqrt{(y_1 - y)^2 + (a-x)^2 + z^2}} \right] \\
 H_y &= 0 \\
 H_z &= -\frac{(a-x)I}{4\pi[(a-x)^2 + z^2]} \times \left[\frac{y_2 - y}{\sqrt{(y_2 - y)^2 + (a-x)^2 + z^2}} - \frac{y_1 - y}{\sqrt{(y_1 - y)^2 + (a-x)^2 + z^2}} \right].
 \end{aligned} \tag{3.3}$$

3.2.2 Magnetic field of a rectangular planar coil

The magnetic field vectors of all segments of a rectangular planar coil can be determined by superposition of the field vectors of all the horizontal and vertical segments:

$$H_{x,\text{coil}} = \sum_{i=1}^n H_{x,i}, H_{y,\text{coil}} = \sum_{i=1}^n H_{y,i}, H_{z,\text{coil}} = \sum_{i=1}^n H_{z,i}, \tag{3.4}$$

where n is the total segment number of the coil. In (3.1) and (3.3), I is the current passing through all wire segments with consideration of its relative direction in each segment.

Figure 3.2: Modeling results for magnetic field distribution of a single planar coil on the central line of the coil ($n=47$, $g = 200 \mu\text{m}$, $I = 0.3\text{A}$, $y = 0 \text{ mm}$, $z = 1 \text{ mm}$).

The magnetic flux $\mathbf{B}$ can subsequently be calculated as:

$$\mathbf{B} = \mu_0 \mathbf{H} (1 + \chi_m) \quad (3.5)$$

where χ_m is the magnetic susceptibility of the medium in which the magnetic field is to be calculated and $\mu_0 = 4\pi \times 10^{-7}$ is the permeability of free space.

The above model was implemented in Matlab (MathWorks Inc, USA). Figure 3.2 shows the typical results of the model for a rectangular coil. The field strength was calculated for a line running along the x -axis and through the center of the coil.

The modeling results show that the total magnetic field of a planar coil has almost the same trend and magnitude as the z -component of its magnetic field strength. To validate the model, the z -component of the magnetic field in of a microcoil with the same parameters as in the model was measured by a gaussmeter (GM05, Hirst magnetic instrument, UK). Details on the fabrication of the coil and the experimental setup are reported later in Section 3.3. Both theoretical and experimental results of a micro coil with 94 segments, $200\text{-}\mu\text{m}$ gap and a current of 0.55 A are shown in Figure 3.3 .

Figure 3.3: Comparison between the modeling result and the experimental result of the z -component of the magnetic field of a planar coil. The magnetic flux was measured with a Gaussmeter and converted into magnetic field strength using (3.5) ($n = 94$, $g = 200 \mu\text{m}$, $I = 0.55 \text{ A}$, $\chi_m = 0$, $\mu_0 = 4\pi \times 10^{-7}$, $y = 0 \text{ mm}$, $z = 1 \text{ mm}$).

3.2.3 Influence of coil parameters on its magnetic field and magnetic force

In this section, the developed analytical model is used to study the influence of coil parameters on its magnetic field. The gap between the wires, the number of wire segments and the current magnitude are varied, and their influences on the resultant magnetic field and magnetic force is discussed. The force acting on a magnetic object located in a magnetic field can be expressed as [56]:

$$\mathbf{F} = \frac{V\Delta\chi}{\mu_0}(\mathbf{B} \cdot \nabla)\mathbf{B} \quad (3.6)$$

where V is the volume of the object, μ_0 is the magnetic permeability of the free space and $\Delta\chi$ is the difference between magnetic susceptibility of the object and the surrounding medium. Extending (3.6) for each force component leads to:

$$\begin{aligned} F_x &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_x}{\partial x} + B_y \frac{\partial B_x}{\partial y} + B_z \frac{\partial B_x}{\partial z} \right) \\ F_y &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_y}{\partial x} + B_y \frac{\partial B_y}{\partial y} + B_z \frac{\partial B_y}{\partial z} \right) \\ F_z &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_z}{\partial x} + B_y \frac{\partial B_z}{\partial y} + B_z \frac{\partial B_z}{\partial z} \right) \end{aligned} \quad (3.7)$$

The Maxwell's equation

$$\frac{1}{\mu} \nabla \times \mathbf{B} = 0 \quad (3.8)$$

leads to

$$\frac{\partial B_y}{\partial z} = \frac{\partial B_z}{\partial y}, \frac{\partial B_x}{\partial z} = \frac{\partial B_z}{\partial x}, \frac{\partial B_x}{\partial y} = \frac{\partial B_y}{\partial x}. \quad (3.9)$$

Thus, equation (3.7) can be rewritten for a planar coil with n segments as:

$$\begin{aligned} F_{x,\text{total}} &= \frac{V\Delta\chi}{\mu_0} \sum_{i=1}^n \left(B_{x,i} \frac{\partial B_{x,i}}{\partial x} + B_{y,i} \frac{\partial B_{y,i}}{\partial x} + B_{z,i} \frac{\partial B_{z,i}}{\partial x} \right) \\ F_{y,\text{total}} &= \frac{V\Delta\chi}{\mu_0} \sum_{i=1}^n \left(B_{x,i} \frac{\partial B_{x,i}}{\partial y} + B_{y,i} \frac{\partial B_{y,i}}{\partial y} + B_{z,i} \frac{\partial B_{z,i}}{\partial y} \right) \\ F_{z,\text{total}} &= \frac{V\Delta\chi}{\mu_0} \sum_{i=1}^n \left(B_{x,i} \frac{\partial B_{x,i}}{\partial z} + B_{y,i} \frac{\partial B_{y,i}}{\partial z} + B_{z,i} \frac{\partial B_{z,i}}{\partial z} \right) \end{aligned} \quad (3.10)$$

3.3 Experimental setup and results

3.3.1 Experimental setup

Planar coils were fabricated by photolithography and etching on a double-sided PCB with precoated photoresist (Farnell, USA). The patterns of the coils were designed with AutoCAD (Autodesk Inc., USA) and printed on a transparent mask. Using a UV light exposure unit, the patterns were transferred to the photoresist of the PCB. After developing the photoresist, the PCB was etched by a ferric chloride solution. Coils with different numbers of wire segments and different gap sizes were fabricated. For a gap of 200 μm , coils with 20, 48 and 94 segments were fabricated. For a segment number of 48, coils with different gap sizes of 200 μm , 650 μm and 1150 μm were also fabricated. Figure 3.4(a) shows the PCB with coils characterized in our experiments.

Figure 3.4(b) shows the setup for the characterization measurement of the coils. Three sets of verification experiments were carried out for the coils. In the first set of experiments, the z -component of the magnetic field of a microcoil with a gap of 200 μm , and 94 segments were measured at different currents. In the second set of experiments, the gap size was kept constant at 200 μm while the segment number varies as 20, 48 and 94. The current in use was 0.4 A. In the third set of experiments, the segment number is fixed at 94 and the gap size varies as 200 μm , 650 μm and 1150 μm . In this set of experiments, the current was fixed at 0.5 A. The same gaussmeter described in the previous section was used to measure

Figure 3.4: Validation experiments for the magnetic field of a micro coil: (a) microcoils with different parameters fabricated on a PCB, and (b) the experimental setup.

the magnetic flux. The position of the sensor probe was adjusted by a precision positioner along x axis. The measured flux was converted to the field strength using (3.5) with $\chi_m = 0$, $\mu_0 = 4\pi \times 10^{-7}$ for air as the surrounding medium. Several sets of measurement were conducted for a same condition to measure the magnetic field of the coils. The graphs represent the result of the mean value for 3 sets of measurements. The maximum difference between each of the sets with the mean value was less than 5%. The fluctuations in sensor readings had the maximum value of 8%.

3.3.2 Results and discussions

The modeling and experimental results are presented and compared in Figures 3.5, 3.6 and 3.7. Figure 3.5 shows the theoretical and measured distribution of the z -component of the field strength along the x -axis. Figure 3.6 compares the data of the maximum field strength at the coil center. Figure 3.7 depicts the distribution of the term $(\mathbf{B} \cdot \nabla)\mathbf{B}_x$, which according to (3.10) represents the x -component of the force on a magnetic particle at the given position. The results were taken along the x -axis through the coil center. The theoretical model shows that increasing the current through the planar coil increases both the magnitude and the gradient of the magnetic field, Figure 3.5(a). According to (3.1) and (3.3), the magnetic field strength is linearly proportional to the magnitude of the electric

current. Both analytical model and experimental data show that when the current increases from 0.1 A to 0.55 A, the maximum field strength at the center of the coil increases from 50 A/m to 800 A/m, Figure 3.6(a). Thus, a higher current will lead to larger magnetic force, Figure 3.7(a). From the relation of (3.10), the maximum magnetic force can be expected to be a square function of the actuating current. Besides its magnitude, the direction of the current also affects the magnetic field. If the direction of the current changes, the direction of magnetic field will be reversed.

It is not always possible to use a higher current to obtain a larger magnetic force, because a higher current also generates more heat in the coils. Exceeding a certain temperature limit, the heat can burn and destroy the coil. Another option to increase the magnetic field is increasing the number of wire segments in the coil. As shown in Figure 3.5(b), increasing the number of wire segments increases the strength of magnetic field but not significantly its gradient. The reason for the small change in field gradient is the larger coil size associated with the higher segment number. The more segments the coil has, the stronger is the field. Figure 3.6(a) shows that the relation between the field strength and the number of segments is not strictly linear. Since the field gradient does not increase significantly with the number of wire segments, the maximum magnetic force caused by the coil is expected to be linearly proportional to the coil number as depicted in Figure 3.7(b).

The gap size between the coil wires also affects the magnetic field. Figure 3.5(c) shows that the smaller the gap is, the bigger is the magnitude and gradient of the magnetic field. The theoretical curve of the field strength is wavy when the gap size increases to 1000 μm . At this large gap size, the fields generated by the wire segments do not overlap well and appear individually. The measurement can not capture thus wavy characteristics because of the relatively large size of the sensor chip on the order of 1 mm. The maximum field strength at the center of the coil is inversely proportional to the gap size, Figure 3.6(c). Thus, according to (3.10), the maximum magnetic force is expected to be inversely proportional to the square of the gap size, Figure 3.7(c). Thus, for actuation applications, the smaller the gap between coil segments, the larger is the magnetic force. The limitation for decreasing the gap is only the resolution of the micro fabrication process. Another advantage of decreasing the gap size is that more wires can be placed in each unit of length. Thus microcoils have the advantage of both large segment number and small gap size.

3.4 Conclusion

Magnetic actuation has shown itself as a promising tool for a variety of miniaturized devices such as microvalves, micropumps and droplet actuators. Planar microcoils have a huge potential for being used in magnetic digital microfluidic devices. We reported an analytical model for calculating the magnetic field of a planar rectangular coil. Based on this model, the force acting on a magnetic particle located in the field can be calculated. Microcoils with varying parameters were fabricated in a commercially available PCB using photolithography and wet etching. A series of experiments was carried out to validate the theoretical model. Analytical and experimental results show that smaller gap size between wire segments, large number of wire segments and higher electric current can increase the magnitude of the induced magnetic field. Considering practical applications of microcoils, decreasing the gap between the wire segments is the best solution for a maximum magnetic force. Although magnetic force is a volume-based force and is not favorable in microscale, coils fabricated by microtechnology may have very small gap size and a large number of wire segments, while the coil footprint is kept small. This fact promises the effective use of microcoils in actuation applications, especially in magnetic droplet-based microfluidics.

Figure 3.5: Influence of coil parameters on the magnitude and the gradient of the magnetic field: (a) Field strength at different currents ($n = 94$, $g = 200 \mu\text{m}$, $y = 0 \text{ mm}$, $z = 1 \text{ mm}$). (b) Field strength with different segment numbers ($I = 0.4 \text{ A}$, $g = 200 \mu\text{m}$, $y = 0 \text{ mm}$, $z = 1 \text{ mm}$). (c) Field strength with different gap sizes ($I = 0.5 \text{ A}$, $n = 47$, $y = 0 \text{ mm}$, $z = 1 \text{ mm}$).

Figure 3.6: Dependence of the maximum field strength at the coil center on (a) current ($n = 94$, $g = 200 \mu m$, $y = 0$ mm, $z = 1$ mm), (b) segment number ($I = 0.4$ A, $g = 200 \mu m$, $y = 1$ mm, $z = 1$ mm), and (c) gap size ($I = 0.5$ A, $n = 47$, $y = 0$ mm, $z = 1$ mm).

Figure 3.7: Magnetic force distribution in x direction on the line running through the center of the coil (a) as a function of current (b) as a function of wire number and (c) as a function of gap size between coil wires.

Chapter 4

One-Dimensional Actuation of Ferrofluid Droplet by Planar Microils

4.1 Introduction

In this section, actuation of ferrofluid droplet, using a device composed of planar microcoils, is described. The model of planar coils from previous chapter is used to describe the actuation mechanism and formulate the motion of the droplet under magnetic force.

Following assumptions have been made in modeling the problem:

1. It is presumed that the droplet deformation under magnetic field is negligible and thus, the ferrofluid droplet is considered as a hard sphere. The drag force has been calculated based on this assumption. The suggested model does not consider the droplet deformation effect.
2. The thermal distribution on the surface of Teflon sheet by microcoils is assumed to be negligible and it does not create thermocapillary effect to affect the motion of droplet.
3. It is assumed that the change of viscosity with temperature change is negligible.
4. The assumptions and simplifications for modeling the magnetic field of the pair of planar microcoils is similar to that mentioned in chapter three.

5. The magnetic flux created by the microcoils are distorted around the droplet. This distortion of the field lines are considered to be negligible.

4.2 The device and the model of magnetic field

In a magnetic field, magnetic particles are always attracted to the field maxima [54]. The principle in manipulation of the droplet in the device is based on changing the location of magnetic field maxima. Since the ferrofluid droplet always moves towards the field peak point, if the location of the peak point is changed from right to left, then the droplet can be moved back and forth. To do this, a device composed of a series of planar microcoils is made (Figure 4.1). There are coils on both sides of a printed circuit board (PCB). A reservoir is made around the coils on the top part to contain the silicon oil.

A ferrofluid droplet is placed over a piece of teflon sheet on the top side of PCB. The droplet is immersed in silicon oil and is moved by magnetic force. The coils on top make a virtual channel to limit the droplet along x axis and the coils on the bottom actuate the droplet. In order to understand the actuation mechanism of the ferrofluid, a model to describe the magnetic field of two coils is first developed. The magnetic field is modeled based on the previous chapter.

4.2.1 The model for the magnetic field of a pair of planar coils

In the previous chapter, a model describing the magnetic field of a single planar coil was developed. Here, by superpositioning the magnetic field of each coil, the field in the actuation domain can be calculated.

$$\begin{aligned} H_{x,\text{total}} &= H_{x,\text{coil 1}} + H_{x,\text{coil 2}} \\ H_{y,\text{total}} &= H_{y,\text{coil 1}} + H_{y,\text{coil 2}} \\ H_{z,\text{total}} &= H_{z,\text{coil 1}} + H_{z,\text{coil 2}} \end{aligned} \tag{4.1}$$

Validation of the Model

In order to validate the model for the two coils, the magnetic field on the line connecting the center of two coils has been measure, using a Gauss meter (GM05, Hirst magnetic instrument, UK). The experimental and analytical results can be compared in Figure 4.2.

Figure 4.1: The schematic of the device made on a PCB to manipulate a ferrofluid droplet. Coils on top make a virtual channel and the the coils on the bottom actuate the droplet. (a) The 3D view of the device. (b) The details of the device components. (c) A close view from the microcoils after etching on PCB. [74].

Figure 4.2: Comparing the analytical model of magnetic field of two coils with experimental data from sensor measurement. $I=0.3\text{A}$, $n=47$, distance between two coil centers=6 mm, distance of the measurement points from the coils plane=0.8 mm.

4.2.2 Actuation Mechanism

Using the developed model, it is possible to explain in details, how each component of the device works and makes the actuation possible.

Driving coils

Two coils on the back side of the PCB are used to move the droplet. According to the simulation, if the coils are run with the opposite direction of current, a magnetic field with two peak points is created (the curve of H_{total} in Figure 4.3).

The ferrofluid droplet is always attracted to the peak points of magnetic field. Here the peak points are located in the center of the coils, therefore if the droplet is nearer to each of the coil centers, it will be attracted to the center of that coil. In order to make one absolute peak point in the domain, two bars of permanent magnets are used in addition to the coils [Figure 4.1 (b)]. These permanent magnets make a homogenous magnetic field

Figure 4.3: Simulation of the magnetic field of driving coils. The plots are for the field of the coils without any permanent magnets. The current directions are opposite in two coils. (a) arrangement of the coils and the coordinate. (b) Magnetic field in three directions. $n=47$, $I=0.8$ A, $g=50\text{ }\mu\text{m}$.

in the points that are far enough from their poles. This homogenous field will be added to the field of the planar coils. After this superpositions, there will be only one absolute field maxima and ferrofluid droplet is attracted to that point only [H_{total} curve in Figure 4.4 (b)]. The other benefit of using permanent magnets is that they amplify the magnetic force on the object. As discussed in previous chapter, magnetic force depends on both magnetic gradient and magnitude of the field. The homogenous field of permanent magnets increases the magnitude of the field and therefore increases the magnetic force on the droplet.

If the direction of the current in two driving coils is changed, then the location of the absolute peak point also changes. Using the electronic circuits and experimental setup described later, the location of the magnetic peak has been changed and the droplet was moved back and forth along the x axis.

The virtual channel

Two coils on the top of the PCB are used to make a virtual channel. To confine the droplet along x axis, the magnetic peak should be located in the middle of the channel. according to the simulations by the developed model, if the two coils are run by the same current direction and same magnitude, there will be two peak points further from the center of the channel, that is not useful for the aimed function (Figure 4.5). When the homogenous field of permanent magnets are added, a peak point in the middle of the channel appears that can confine the droplet. This peak point acts like a trap and keeps the droplet in the middle of the channel.

4.3 Forces acting on the droplet moving in silicon oil

Using the described device, the ferrofluid droplet is immersed in oil and is moved by magnetic force from microcoils and permanent magnets. In order to model the kinematic behavior of the droplet, forces acting on the droplet should be known. It is assumed that the ferrofluid droplet keeps the shape of sphere, when it is in oil. The contact angle of the droplet immersed in oil on the teflon surface was measured using a tensiometer (FTA200, First Ten Angstrom). The angel was about 127° (Figure 4.7). This relative large contact angle justifies the assumption of the spherical droplet shape.

There are two forces acting on the droplet when it is moving in silicon oil: drag force

Figure 4.4: Simulation of the magnetic field of two driving coils in the device as well as permanent magnets. (a) arrangement of the coils and the coordinate. (b) Magnetic field in three directions. $n=47$, $I=0.8$ A, $g=50$ μm .

Figure 4.5: Simulation of the magnetic field of two coils with the same current direction, without permanent magnets. (a) arrangement of the coils and the coordinate. (b) Magnetic field in three directions. $n=47$, $I=0.8$ A, $g=50 \mu\text{m}$.

Figure 4.6: Simulation of the magnetic field of two coils with the same current direction added with the homogenous magnetic field of permanent magnets. (a) arrangement of the coils and the coordinate. (b) Magnetic field in three directions. $n=47$, $I=0.8$ A, $g=50\text{ }\mu\text{m}$.

Figure 4.7: The ferrofluid droplet immersed in oil keeps the shape of sphere.

and magnetic force. Considering the assumption of sphere shape for the droplet, the drag force over it will be [104]:

$$F_{\text{drag}} = 3\pi D\eta U \frac{1 - 2\eta/3\eta_{ff}}{1 + \eta/\eta_{ff}} \quad (4.2)$$

where D is the diameter of the droplet, η is the dynamic viscosity of the carrier fluid (silicon oil), η_{ff} is the dynamic viscosity of the ferrofluid and U is the velocity of the droplet in oil. if $\eta_{ff} \gg \eta$, then the formula can be simplified as:

$$F_{\text{drag}} = 3\pi D\eta U \quad (4.3)$$

Magnetic force of a single coil acting on magnetic particles was discussed in previous section. In a system composed of two coils, the magnetic force can be calculated by superposition of magnetic force of each single coil. Extending the main equation of magnetic force for the system of two coils leads to the following equations for the magnetic force of two coils:

$$\begin{aligned} F_x &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_x}{\partial x} + B_y \frac{\partial B_x}{\partial y} + B_z \frac{\partial B_x}{\partial z} \right) \\ F_y &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_y}{\partial x} + B_y \frac{\partial B_y}{\partial y} + B_z \frac{\partial B_y}{\partial z} \right) \\ F_z &= \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial B_z}{\partial x} + B_y \frac{\partial B_z}{\partial y} + B_z \frac{\partial B_z}{\partial z} \right) \end{aligned} \quad (4.4)$$

and the magnetic force of two coils is:

$$\begin{aligned}
F_{x,\text{total}} &= F_{x,\text{coil1}} + F_{x,\text{coil2}} \\
F_{y,\text{total}} &= F_{y,\text{coil1}} + F_{y,\text{coil2}} \\
F_{z,\text{total}} &= F_{z,\text{coil1}} + F_{z,\text{coil2}}
\end{aligned} \tag{4.5}$$

Since the droplet is actuated one dimensionally along x axis and its motion in other directions is confined by the virtual channel, in the modeling, only F_x will be used. The magnetic field of the permanent magnets should be considered. Sensor measurements show that the magnetic field of the permanent magnets is around 5000 to 8000 A/m in the actuation domain, depending on their distance from the device.

Figure 4.9 is the result of Matlab simulation for the derivative of magnetic field and force acting on a ferrofluid droplet with the diameter of 1mm.

As shown, the force acting on the droplet varies with its location. In the Matlab simulation, none of the force components were neglected and as the graphs show, F_x and $\frac{\partial B_z}{\partial x}$ have the same trend. This is because H_x and H_y have an order around 500 to 1000 A/m, depending on the used current, while the permanent magnets generate a homogenous field around 10 times greater than them in z direction. Therefore total H_z will be about 10 times greater than H_x and H_y . Thus F_x can be approximated as below:

$$F_{x,\text{total}} = \frac{V\Delta\chi}{\mu_0} \sum_{i=1}^n \left\{ B_z \frac{\partial}{\partial x} (B_z) \right\} \tag{4.6}$$

The distance between two coils, affects the shape of force profile. Different force profiles by different distances between coil centers are illustrated in Figure 4.8. In order to actuate the droplet between two coil centers, the force between two coil centers should not change its sign, otherwise the actuation in the full range is not possible.

Figure 4.10 shows derivative of the magnetic field and force acting on the droplet when the direction of the current is changed in the coils. As the model proves, by inverting the current direction in coils, the location of the magnetic peak changes and the direction of the force changes as well. Both figures 4.9 and 4.10 show that the force in the center of the coils is zero. This is expected because the magnetic field has its maximum or minimum in the coil centers and since the force is proportional to the derivative of the magnetic field, when the field is maximum or minimum, its derivative is zero and subsequently the force is zero.

Figure 4.8: Simulation of the magnetic force acting on a ferrofluid droplet. The diameter of droplet is 1 mm. (a) coil arrangement. (b) the profile of the force with respect to different distances between coil centers. In the last case that $L = 8$ mm, actuation in the full range is not possible, because the force has changed its sign between coil centers.

One way to understand and evaluate the stability or instability of magnetic particles in a magnetic field is evaluating the direction of the magnetic force in vicinity of each point. For the coil that has maximum field strength in its center [Figure 4.9 (a), the coil in the left], the point in the coil center is a stable point for the magnetic particles. In the right side of the coil center, the force is negative. In the left side of the same point, the force is positive. Therefore if the droplet wants to move in right or left, the force returns it to the point on the center of the coil and in the center, the force is zero. This is why the droplet has stability in this point. On the other hand, for the coil that has minimum field strength in its center [Figure 4.9(a), the coil in right], the center of the coil repels the droplet and is a very unstable point for magnetic particles. In the right side of the coil center, the force is positive and in the left side of the center, force is negative. Therefore, although in the center itself the force is zero, if the droplet moves for a very small distance from center, the force immediately pulls it to right or left. This is why the center of the coil with minimum B is an unstable point.

4.3.1 Modeling the motion of a ferrofluid droplet immersed in silicon oil

The purpose of this section is making a mathematical model that can predict the displacement versus time and velocity versus time curves of ferrofluid droplet in the oil.

The forces acting on the droplet were described in the previous part. Using the Newton's second law for the droplet results in the following equations:

$$\begin{aligned}
 \sum F_x &= m\mathbf{a}_x \\
 F_{x\text{ magnetic}} - F_{x\text{ drag}} &= m\mathbf{a}_x \\
 \frac{V\Delta\chi}{\mu_0} \left(B_x \frac{\partial}{\partial x} B_x + B_y \frac{\partial}{\partial x} B_y + B_z \frac{\partial}{\partial x} B_z \right) \\
 - 3\pi D\eta U \frac{1 - 2\eta/3\eta_{ff}}{1 + \eta/\eta_{ff}} - m\mathbf{a}_x &= 0
 \end{aligned} \tag{4.7}$$

where B_x , B_y , B_z are the magnetic fields in three directions resulted by the coils, m is the mass, U is velocity and a is the acceleration of the ferrofluid droplet. For the magnetic force, it is possible to use the analytical formula of the force from Matlab codes directly,

Figure 4.9: The derivative of magnetic field with respect to x and magnetic force of a pair of planar coil. (a) coil arrangement. (b) the curves of the force and derivatives of the field. $I=0.7$ A, $n=47$ and field of the permanent magnets= 1000 A/m.

Figure 4.10: Changing the direction of the current, changes the direction of the force on the objects between two coils. (a) coil arrangement. (b) the curves of the derivative and force.

Figure 4.11: The solution of the force balance equation, presenting the velocity and displacement curves of ferrofluid droplet under magnetic actuation. $\eta=100\text{cst}$, $f=0.01\text{ Hz}$, $I=0.88\text{ A}$, $D=2.4\text{ mm}$.

but since this analytical formula is sum of the analytical formulas of all the wire segments for three directions, it has a very long form that dramatically affects calculation time. For convenience in calculation, a polynomial function was fit into the force curve, using Matlab's curve fitting tool box.

The final force balance equation will be:

$$\frac{6}{\rho\pi D^3} \left(p_1 x^6 + p_2 x^5 + p_3 x^4 + p_4 x^3 + p_5 x^2 + p_6 x + p_7 \right) - \frac{18\eta}{\rho D^2} \frac{dx}{dt} - \frac{d^2x}{dt^2} = 0 \quad (4.8)$$

where p_1 to p_7 are the constants determined by curve fitting algorithm for the polynomial used instead of the magnetic force curve.

This equation was numerically solved by Runge-Kutta method. A typical solution with specific parameters is presented in Figure 4.11.

4.3.2 Experimental setup, fabrication process and Control scheme

Fabrication process and materials

Microcoils were made on the PCB by the same process and materials described in the previous chapter. A reservoir was made around the coils and filled with silicon oil (SigmaAldrich, USA). A piece of teflon sheet (professional plastics, USA) is placed over the coils to reduce the friction between ferrofluid and the surface. Because of the low surface energy of Teflon and the slightly lower density of silicone oil the ferrofluid droplet stays on the bottom surface and retains a spherical shape (Figure 4.7). Electrical terminals are mounted on two sides of the PCB to transfer control signals to the coils.

The ferrofluid (APG S10n, Ferrotec, USA) used in the experiments is a suspension of magnetic beads in synthetic ester oil. At 25°C, the dynamic viscosity, density, surface tension and susceptibility of ferrofluid are 0.406 kg/m.s, 1330 kg/m³, 0.032 N/m and 1.6 respectively. The silicon oil has the density of 960 kg/m³, kinematic viscosity of 50 and 100 cst (two types of oils have been used), dynamic viscosity of 0.048 kg/m.s and 0.096 kg/m.s and surface tension of 0.0203 N/m. The assumption of a solid sphere for the ferrofluid droplet is acceptable as carrier fluids are less viscous than the ferrofluid. The large contact angle of ferrofluid on teflon, immersed in oil that is around 127 degrees, justifies the assumption of the spherical droplet shape.

Experimental setup and Control scheme

The experimental setup, switching circuit for the driving field and the pulse diagrams are illustrated in Figure 4.12.

Coils on the top are connected to a constant current [Figure 4.12(b)]. Coils on the bottom are run with a periodic signal [4.12 (c)].

A miniature electromagnetic relay (Fujitsu, Takamisawa RY5WK) was used in switching device. The trigger signal comes from a signal generator for coils and is also used for triggering the CCD camera. Thus, the CCD camera is synchronized with the driving magnetic field. Accurate time measurement is achieved by fixing the frame rate of the CCD camera. The position of the ferrofluid droplet is extracted from the recorded images using the codes developed in Matlab program. The codes are presented in the appendix. First, the gray scale image is converted into a binary image to identify droplet region. The centroid of the droplet is calculated and the position of the droplet is determined by tracking its centroid.

Using this setup, the following curves for velocity and displacement have been ob-

Figure 4.12: Experimental setup to study the behaviour of ferrofluid. (a) and (d) show the connections and hardware. (b) and (c) show the pulse diagrams [86].

Figure 4.13: Comparing the numerical solution of the model equation and the experimental data. (a) Comparing velocity curves. (b) Comparing displacement curves. Experimental data is taken from [86].

tained. Several sets of data were available from experiments but the graphs are plotted by using only one set of data as the sets were very similar.

4.4 Results and discussion

Figure 4.13 compares the numerical solution and experimental results. In the beginning of the motion, the droplet accelerated because the magnetic force is increasing with displacement. This acceleration increases the velocity of droplet and subsequently the drag force increase. After some time the drag force is dominant over the magnetic force and the droplet decelerates and finally stops.

The motion of the ferrofluid is mainly affected by two forces: magnetic force and viscous drag force.

Magnetic force depends on the parameters on the design of coil (gap size between coil wires, number of wires), the current running the coils and the size of the ferrofluid droplet. In the previous chapter, the effect of coil parameters for a single coil was evaluated. Figure 4.14 shows the effect of changing diameter and running current on the magnetic force of a pair of planar coils. As it is shown, a higher driving current and a bigger droplet diameter increase the magnetic force.

Figure 4.14: Evaluating the effect of changing current of the coils and the diameter of the droplet on the magnetic force over the droplet. (a) arrangement of two coils. (b) Increasing the current increases the magnetic force. (c) Increasing the diameter of the droplet increases the magnetic force.

Drag force depends on the size of the droplet and the viscosity of carrier fluid. Figure 4.15 shows the effect of different parameters on the kinematics of the droplet motion immersed in silicon oil.

In Figure 4.15 (a) diameter of the droplet is changed and viscosity and current are fixed. Both the experimental data and numerical model show that the bigger the droplet diameter is, the higher velocity it can achieve in the domain. This is because the magnetic force on the droplet depends on the diameter with the power of three, while the drag force depends on the droplet diameter with the power of two. Therefore, the increase in magnetic force due to the increase in droplet size is dominant over the increase in drag force. Thus, the droplets with the bigger size, obtain higher velocities.

Figure 4.15 (b) shows that a higher running current for coils, increases the speed of the droplet and shortens the time to reach the maximum velocity. This is because the magnetic force of a coil depends on the current passing through the coil. The higher the current is, the greater the magnetic force over the droplet will be.

Figure 4.15 (c) shows the effect of changing the viscosity of the carrier fluid, while the current and size of the droplet are fixed. The higher the viscosity is, the lower the speed of the droplet will be. This is because, the drag force over the droplet is directly proportional to the viscosity of the carrier fluid. A higher viscosity increases the drag force and as a result, the velocity of the droplet decreases.

In all three sets of data, it can be compared that the experimental data and numerical model almost agree. The difference can be due to several reasons.

The droplet deforms when it is moved back and forth. However, the model assumes that the droplet keeps the shape of a sphere. Coils on the PCB generate heat when they are on and if the distribution of their heat is not homogenous on the surface of the teflon sheet, the thermocapillary effect can affect the motion of the droplet. Moreover, the heat changes the viscosity of the carrier fluid and the drag force is directly proportional to the viscosity.

Another reason is the difference between the analytical values of the magnetic field that are used in the model and the experimental values. As Figure 4.2 shows, the developed analytical model is slightly different from the experimental values. Furthermore, the magnetic force is approximated by a polynomial, using curve fitting procedure. The curve fitting always has some errors.

Since the recorded images only give us the spacial position in xy -plane, the exact z position of the droplet can not be measured. Although the force component $F - z$ is small, it can elevate the ferrofluid droplet from the Teflon surface. The estimated position of $z = 5.5$ mm were therefore taken for all numerical results presented in the Figures.

4.5 Conclusion

One dimensional actuation of ferrofluid droplet, using a device composed of planar microcoils was described. Based on the model of a single planar microcoil in previous chapter, the model for a pair of planar coils was developed and the actuation mechanism was described. The forces acting on the droplet immersed in oil were calculated and a mathematical model to explain the kinematics of the droplet motion was made. The result from the model were compared with the experimental data. There is an agreement between the model and the experimental data, validating the model. There are some differences between the model and experimental data due to the droplet deformation, heat generated in the coils and errors in curve fitting.

Based on one dimensional actuation, two dimensional actuation of ferrofluid droplet has been achieved and will be explained in the next chapter.

Figure 4.15: The effect of different parameters on the kinematics of ferrofluid motion immersed in silicon oil. (a) Diameter of the droplet is changed and viscosity and current are fixed. (b) Current is changed and Diameter of the droplet and viscosity are fixed. (d) The viscosity is changed and the current and diameter of the droplet are fixed. The values of fitted function on the magnetic force are corrected by the factor of 0.05. This value is fixed for all the numerical result. Experimental data taken from [86].

Chapter 5

Two Dimensional Actuation of Ferrofluid Droplet by Planar Microcoils

5.1 Introduction

In this chapter, the actuation concept used in chapter 4 will be expanded to be used for two dimensional actuation. A device composed of planar microcoils on PCB is made to manipulate a ferrofluid droplet immersed in oil on a platform of 5×5 mm.

5.2 The device and actuation mechanism

Figure 5.1 shows the device and its components in details. There is a reservoir around the coils for silicon oil. Since all the coils are in touch with the oil, the heat generated can be absorbed. A ferrofluid droplet is placed over a teflon sheet. The reservoir is filled with silicon oil. The droplet was moved by magnetic force on the area between coil centers.

The same principle as in previous section is applied to manipulate the ferrofluid droplet. Magnetic particles are attracted to the magnetic field peak points. The device for two dimensional actuation has four coils on the top part of a PCB. These coils generate magnetic field. The configuration of the magnetic field is controlled by electronic circuits and PC programming in a way that the location of the magnetic peak changes in a two dimensional platform. Since the ferrofluid droplet follows this peak point, it will be moved in 2D

Figure 5.1: The device used for two dimensional actuation of ferrofluid droplet. (a)The device after full assembly. (b) Illustration of the coils on top with the details. (c) the close view of the micro coils etched on the PCB.

Figure 5.2: The magnetic field of a single coil with new design.

patterns in the platform.

The shape of the planar coils used for this device is different from the previous device for 1D actuation. Magnetic field has its maximum in the center of each single coil. Figure 5.2 shows the measurement data for two lines passing through the center of a single coil with new design. In order to move the droplet in a loop, coils are run in a sequence. In each sequence, the location of magnetic peak is set in the center of one coil. Since this location is shifted from one coil to another one in a loop, the droplet follows it and moves in the desired path, that is a full loop in the domain. When the sequence is repeated for several cycles, the droplet makes several loops. Figure 5.3 shows the sensor measurement data for each sequence between coil centers.

As the Figure shows, the magnetic field created by single coils has two points with maximum intensity, that is the center of the two adjacent coils. This field alone can not actuate the droplet in the full range, because there are two peak points and the droplet goes to the coil center that is nearer to that. In order to make a single peak point in the

domain, two permanent magnets are used. These permanent magnets create a homogenous magnetic field in z direction in the points that are far enough from their poles. Figure 5.4 shows the field between coil centers when the magnetic field of permanent magnets is also added. As it can be seen, there is only one absolute peak point between each pair of coils.

If the direction of the current is inverted in each coil, then the location of the peak point and the direction of the droplet motion changes.

5.3 Control scheme

The hardware and control circuit

In order to actuate and control the droplet, electric pulses are sent to the coils sequentially. Figure 5.5 shows the setup to control the droplet motion.

To generate control pulses, Codes are developed in Matlab program. Since the pulses from PC are in the form of voltage signals, They should be converted to current signals, using an amplifier circuit. Figure 5.6 shows the circuit in detail.

There are five inputs to the amplifier circuit. The input from digital to analogue (D/A) card, is for adjusting the amplitude of the current signal. This input is in the form of voltage signal. When this input changes from 0 to 5 volts, the output current changes from 0 to 2 Amperes. This task is mainly done by the power operational amplifier (OPAMP) LM675 in the circuit illustrated in Figure 5.6. The input from the digital card is for changing the direction of the current. This input is connected through a resistor to one of the relay pins. When this pin is activated, the direction of the current in the output of the relay will be inverted. The heat generated in the power OPAMP and resistors, makes fluctuations in the output of the circuit. To remove this problem, the OPAMPs are connected to heat sink and two fans are mounted over the circuit box for cooling.

The pulses to control the amplitude and sign of the output signal are generated by Matlab program. the codes are available in the appendix. Using the data acquisition tool box in Matlab, the signals are sent to the D/A card and digital card.

Controlling the droplet motion

Figure 5.7 shows the pulse diagrams and coils configuration to move the droplet in a loop. The circuit is designed with four output channels but only two channels are used.

For more convenience in control and accuracy in having same current in each two adjacent coils, the coils are connected in pairs by connecting the center of each two coils together. In each period of time, droplet moves from center of one coil to the center of another coil [Figure 5.7 (c)].

5.4 Experimental Setup and results

5.5 Experimental Setup

The planar coils are made on PCB with the same materials and lithography process explained in chapter 4. Figure 5.8 shows the experimental setup used for actuating the droplet and recording the droplet motion. The motion of the droplet was recorded by a CCD camera. The images were processed by the codes developed in matlab, using the image processing toolbox. Each movie of the droplet motion was stored in the form of a matrix with arrays representing the frames of the movie. Then each image was converted to a binary image to distinguish the boundary of the droplet from background easily. The centroid of the droplet was calculated in every frame and trajectories of droplet motion were obtained by tracking the centroid of the droplet.

5.6 Results

Figure 5.9 shows the trajectory of the droplet motion between two adjacent coil centers as well as the actuation sequence for two coils only. The direction of the droplet motion was controlled by the computer programme. As depicted, the droplet was moved back and forth. Figure 5.10 shows the curves of position, velocity, acceleration and net force acting over the droplet versus time. These data are all obtained by image processing codes. Figure 5.10 (b) is the change of the droplet position during the time with respect to the coordinate system illustrated in Figure 5.10 (a). The periodic curve, represents the reciprocative motion of the droplet between two coil centers. Figure 5.10 (c) is the curve of velocity versus time. In each period, the curve has one maximum and one minimum. These points represent the peak velocity when the droplet is moving forward or backward. Figure 5.10 (d) depicts

the acceleration of the droplet in the whole actuation period. The peak acceleration occurs when the droplet changes the direction of its motion. Figure 5.10 (e) shows the distance of the droplet from the initial position. When the droplet reaches the center of the other coil, the curve has its maximum value. Figure 5.10 (f) represents the net force acting on per kilogram of the droplet. As the net force is directly proportional to the acceleration, it has the maximum value on the same points where the acceleration is maximum. Thus the droplet experiences the maximum net force when it is changing its direction of motion. Since the actuation between each two pairs of coils is possible, the droplet can make a full loop by moving between four coil centers. Figure 5.11 shows the droplet motion in a full loop between coil centers. Figure 5.12 shows the curves of distance from the initial position, velocity and displacement versus time for the droplet moving in a full loop. In Figure 5.12 (a), certain periods of change of the position with respect to time are seen. These curves show that the droplet has moved from the center of one coil to the center of another one. Figure 5.12 (b) shows the velocity of the droplet in two directions in different cycles. The droplet reached the maximum value of approximately 1 mm/s in positive and negative directions of x and y axes. Figure 5.12 (c) demonstrates the change of the net distance from the origin with respect to time. In all the three graphs, the periodic curves demonstrate a two dimensional, cyclic, controllable motion of the droplet in a close loop.

5.7 Conclusion

Two dimensional, programmable actuation of the ferrofluid droplet in a close loop by PC interface was demonstrated. The actuation concept is based on changing the location of magnetic field peak points in the domain. Magnetic objects are attracted to the peak points of the field. If the location of the field maxima is changed, the droplet follows this change and therefore it can be moved in a desired path.

A device composed of four coils was fabricated on a PCB by photo lithography techniques. In every time step, each pair of the coils is run with opposite current directions. This, creates a magnetic field with two peak points. When the field of microcoils is added with the uniform field of permanent magnets, one single local peak point appears that attracts the droplet. By switching the location of this peak point between four coil centers in

a close loop, the droplet was moved in a close loop.

The device can be used for variety of applications in lab on a chip platforms. The droplet can be loaded with chemical/biological samples and be used as a carrier. When the droplet moves in the chamber, an internal flow is induced both in the droplet and the carrier fluid (silicon oil). Internal flow in the ferrofluid droplet can be used for mixing of samples encapsulated inside ferrofluid droplet. External flow induced in the carrier fluid also can be used for mixing of different liquids in the chamber.

Figure 5.3: The magnetic field of microcoils in the device. The field has its maximum intensity in coil centers. This distribution of magnetic field is not still proper for actuation and needs to be modified to have a single peak point between each two adjacent coils. The line connecting the coil centers shows the path in which the magnetic flux was measured by gauss meter.

Figure 5.4: The magnetic field between coil centers in presence of permanent magnets. The uniform magnetic field of permanent magnets in z direction is added to the magnetic field of microcoils. The resultant field has only one peak point between two adjacent coils and the droplet is attracted to this point. By changing the location of this peak point in a loop, the droplet will move in a loop. The line connecting the coil centers shows the path in which the magnetic flux was measured by the gauss meter.

Figure 5.5: The setup to control the droplet motion in the device. The control signals are generated in PC, sent to the circuit to be amplified and finally are sent to the coils.

(a)

(b)

Figure 5.6: (a) Overall diagram of the amplifier circuit box, showing terminals, inputs and outputs. (b) The details of the amplifier circuit only.

Figure 5.7: (a) Pulse diagrams for four coils. positive current means current going in the coil. Negative current means current going out of the coil. (b) Arrangement of four coil. (c) Direction of droplet movement in a loop.

Figure 5.8: Experimental setup to actuate droplet in a 2D platform.

Figure 5.9: (a) Motion of the droplet between two coil centers. (b) The trajectory of motion. (c) The current signals sent to the coils (d) The arrangement of coils.

Figure 5.10: Curves related to characterizing the droplet motion

Figure 5.11: (a) Actuation of the ferrofluid droplet in a full loop on a 2D platform. (b) The trajectory of the droplet on the domain. (c) Coil arrangement. (d) Pulse diagram to run the coils.

Figure 5.12: The curves characterizing the 2D droplet periodic motion. (a) The position versus time. (b) The velocity versus time and (c) The distance from initial position versus time.

Chapter 6

Conclusion and Future Directions

6.1 Conclusion

This thesis demonstrated the active control and manipulation of droplets in one and two dimensional platforms by using magnetic field.

As a tool in magnetic manipulation, first the magnetic field of a planar microcoil was modeled and the parameters in coil design were optimized. It was both experimentally and analytically shown that higher number of loops, smaller gap size and higher driving current increase the magnitude of the magnetic field and subsequently the force of a planar micro coil.

One dimensional actuation of ferrofluid droplet in a device composed of microcoils was discussed and modeled. In order to actuate the droplet, a non homogenous magnetic field was created by two planar coils. The magnetic field was amplified and modified to have one single peak point by adding two permanent magnets. Since the magnetic objects are always attracted to the magnetic field maxima, the ferrofluid droplet was moved along an axis by moving the location of the peak point in the domain. Two other coils were used to make a virtual channel to keep the droplet along the axis.

In the next step, a device was made for two dimensional actuation of ferrofluid in a close loop. Four coils were designed on a PCB to make non homogenous magnetic field. After adding the magnetic field of a set of permanent magnets, there is only one peak point in the magnetic field between each two coil centers. The location of the peak point is changed

from the center of one coil to the center of another coil and since the droplet follows the peak, it was moved in a loop between coil centers.

The last part of the thesis is about designing a multi functional low cost device for handling and operating a variety of functions over a droplet. Switching, splitting, merging and sorting droplets is expected to occur in the device.

The devices can be used in lab on chip platforms for a variety of chemical, biomedical and biological applications.

6.2 Future directions

This research demonstrated the prove of concept for actuation of droplets by magnetic force. More work in the future can be done to develop and commercialize the device and make sophisticated lab on chip platforms. In continuation with this research, flow fields induced in the domain can be studied.

When the droplet moves in the chamber filled with the oil, an internal flow is induced both in the droplet and the carrier fluid (silicon oil). Internal flow in the ferrofluid droplet can be used for mixing of samples encapsulated inside ferrofluid droplet. External flow induced in the carrier fluid also can be used for mixing of different liquids in the order of micro liters.

A real low cost, multi functional device for handling droplets in micro channels was fabricated. This device is designed for four main functions of switching, splitting, merging and sorting ferrofluid droplets by magnetic force created by microcoils. As the next step for future work, this device can be tested. A mathematical model that can predict the whole phenomena (switching, merging, splitting and sorting) can be developed in the future. The relationship between current, size of the sorted droplets, switching time and many other parameters can be investigated by the model.

This digital microfluidic chip can have variety of applications in lab on chip devices for biomedical, chemical and biochemical applications.

Appendix A

publication

A.1 Journal Papers

- N.T. Nguyen, A. Beyzavi, K.M.Ng and X.Huang. *Microfluid nano Fluid*. Kinematics and deformation of ferrofluid droplets under magnetic actuation,2007.
- Ali Beyzavi, Nam-Trung Nguyen, *Journal of Micromechanics and Microengineering* . Modeling and optimization of the magnetic field of planar Microcoils.,2008.
- Ali Beyzavi, Nam-Trung Nguyen, *Journal of Physics D: Applied Physics*. One-dimensional actuation of a ferrofluid droplet by planar microcoils.,2008.
- Two-dimensional, programmable actuation of a ferrofluid droplet by planar microcoils. under preparation.

Appendix B

Magnetic field of a current carrying wire considering its thickness and width

In a more general form, there is an analytical formula to calculate the magnetic field from a current carrying wire considering its thickness and width. If the wire has the thickness of t and the width of w , the magnetic field will be [76]:

$$B_x = -\frac{\mu_0 I}{4\pi wt} \left\{ \begin{aligned} & -\alpha \log \left(\frac{\alpha^2 + \delta^2}{\alpha^2 + \gamma^2} \right) - 2\delta \tan^{-1} \left(\frac{\alpha}{\delta} \right) + 2\gamma \tan^{-1} \left(\frac{\alpha}{\gamma} \right) + \\ & \beta \log \left(\frac{\beta^2 + \delta^2}{\beta^2 + \gamma^2} \right) + 2\delta \tan^{-1} \left(\frac{\beta}{\delta} \right) - 2\gamma \tan^{-1} \left(\frac{\beta}{\gamma} \right) \end{aligned} \right\} \quad (\text{B.1})$$

$$B_y = -\frac{\mu_0 I}{4\pi wt} \left\{ \begin{aligned} & -\gamma \log \left(\frac{\gamma^2 + \beta^2}{\gamma^2 + \alpha^2} \right) - 2\beta \tan^{-1} \left(\frac{\gamma}{\beta} \right) + 2\alpha \tan^{-1} \left(\frac{\gamma}{\alpha} \right) + \\ & \delta \log \left(\frac{\beta^2 + \delta^2}{\delta^2 + \alpha^2} \right) + 2\beta \tan^{-1} \left(\frac{\delta}{\beta} \right) - 2\alpha \tan^{-1} \left(\frac{\delta}{\alpha} \right) \end{aligned} \right\} \quad (\text{B.2})$$

$$B_z = 0 \quad (\text{B.3})$$

where $\alpha = x - w/2$, $\beta = x + w/2$, $\gamma = y - t/2$, $\delta = y + t/2$.

Figure B.1: A segment of wire with the thickness of t and width of w .

Appendix C

Magnetic field measurement and Hall effect

In the experiments, the magnetic field was measured frequently. Here, the base of the measurement and the theory based on which the sensor works is briefly described.

Hall Effect Theory

If there is a thin conductor element, carrying current as shown in Figure (C.1), to see the hall effect, every time the voltage between points a and b is measured. These are two points on the sides of the element that can make a line perpendicular to the direction of the current flow. when there is no magnetic field around the element, the voltage between points a and b is zero (Figure C.1 (a)). In presence of a magnetic field perpendicular to the conductor plane, a Lorenz force is exerted to the current. This force disturbs the current distribution and causes a voltage across points a and b (Figure C.1 (a)). This voltage is called Hall voltage and is directly proportional to the magnetic field applied. In the gauss meters, this voltage is first amplified and then measured to calculate the magnetic field perpendicular to the surface of the element. The effect of magnetic field on the current carrying element, leading to appearance of voltage across the element is called "Hall effect".

Figure C.1: (a) When there is no magnetic field, the voltage across the conductive element carrying current is zero. (b) In presence of a magnetic field, a voltage is induced across the filament that is directly proportional to the magnetic field according to Hall effect theory.

Appendix D

Matlab codes

In this part, the matlab codes developed for different parts of the thesis are presented.

D.1 Matlab codes for Image processing

The following codes are developed to track the path of a droplet, moving in the domain.

```
clear all clc
timeinterval=1;%time between two frames;
aviname='D:\ALI\research\2d magnetic droplet control\ccd
camera\coils in pairs\1ls2-9v4cm.avi'; thresh=0.35; mov1 =
aviread(aviname); mov2 = aviread(aviname,20); imshow(mov2.cdata);
[x1 y1]=getpts; [x2 y2]=getpts; fprintf('choose a point on the edge
of the coil')
[x3 y3]=getpts;%choosing point on coils. as we know the side dimension=6mm,
%it is used for calculating ratio of pixel to mm
fprintf('choose a point on the edge of the coil') [x4 y4]=getpts;
ratiopixel=6/abs(y3-y4);%pixel to mm ratio
framelist=length(mov1); framelist=framelist-1
u=[];time=[];ax=[];x=[]; v=[];ay=[];y=[]; start=3;
for index=1:(framelist-start) % load up the thresholded-image
%stack with 'nFrames' images
    fprintf(1,'*'); % so I know it hasn't crashed while it's running...
    mov = aviread(aviname,index+start);
    %filename
    time(index)=index*timeinterval;
    bw=threshold(mov.cdata,index,thresh,x1,x2,y1,y2);
    %pause
    bw=imcomplement(bw);
    imshow(bw);
```

```

%pause
L=bwlabel(bw); % 'particle' assignment
stats=regionprops(L,'centroid'); % retrieve some of
%the 'particle' properties
Lstats=length(stats);
if index==1
    u(index)=0;
    ax(index)=stats.Centroid(1);
    ay(index)=stats.Centroid(2);
    ay(index)=-ay(index)
    x(index)=0;
    y(index)=0;
    s(index)=0
    %ratio(index)=(stats.BoundingBox(3))/(stats.BoundingBox(2));
else
    ax(index)=stats.Centroid(1);
    ay(index)=stats.Centroid(2);
    ay(index)=-ay(index)
    u(index)=ratiopixel*(ax(index)-ax(index-1))/timeinterval;
    v(index)=ratiopixel*(ay(index)-ay(index-1))/timeinterval;
    x(index)=ratiopixel*(ax(index)-ax(1));
    y(index)=ratiopixel*(ay(index)-ay(1));
    accx(index)=(u(index)-u(index-1))/timeinterval;%acceleration in x drction
    accy(index)=(v(index)-v(index-1))/timeinterval;%acceleration in y drction
    acct(index)=((accx(index))^2+(accy(index))^2)^0.5%total acceleration
    s(index)=((x(index)-x(1))^2+(y(index)-y(1))^2)^0.5%distance from the
    %initial location
    Vtotal(index)=(u(index)^2+v(index)^2)^0.5 %total velocity
end
end % loop over all frames.
%meanx=mean(x(5:framelist));
%end
%plot(time(1:(framelist-start)),u(1:(framelist-start)));

figure subplot(2,1,1);
plot(time(2:(framelist-start)),x(2:(framelist-start)),'b'); hold on
plot(time(2:(framelist-start)),y(2:(framelist-start)),'r');

xlabel('time t (s)'); ylabel('Position (mm)'); legend('x','y');
hold off
% %
subplot(2,1,2);
plot(time(1:(framelist-start)),u(1:(framelist-start)),'b'); hold on
plot(time(1:(framelist-start)),v(1:(framelist-start)),'r');
xlabel('time t (s)'); ylabel('Velocity (mm/s)'); legend('velocity

```



```

in x axis','velocity in y axis'); hold off %%

%-----
figure subplot(3,1,1);
plot(x(1:(framelist-start)),y(1:(framelist-start))) xlabel('X
(mm)'); ylabel('y (mm)');

subplot(3,1,2);
plot(time(1:(framelist-start)),s(1:(framelist-start)),'b');
xlabel('time t (s)'); ylabel('distance from initial position(mm)');

subplot(3,1,3);
plot(time(1:(framelist-start)),Vtotal(1:(framelist-start)),'b');
xlabel('time t (s)'); ylabel('total Velocity(mm/s)');

%-----
figure subplot(2,1,1);
plot(time(1:(framelist-start)),accx(1:(framelist-start)),'b'); hold
on plot(time(1:(framelist-start)),accy(1:(framelist-start)),'r');
xlabel('time t (s)'); ylabel('acceleration (mm/s2)');
legend('x','y'); hold off
% %
acct=0.001*acct

subplot(2,1,2);
plot(time(1:(framelist-start)),acct(1:(framelist-start)),'b');
xlabel('time t (s)'); ylabel('force (N)'); legend('Net force on
drople(multiply numbers in mass of droplet)'); hold off

```

D.2 Matlab codes for generating command signals and transferring the signals to A/D and digital card

```

%-----digital card registration and adding ports-----
clear all; daqregister('advantech')

rehash toolboxcache

clear all;

out=daqhwinfo;

```

```

out.InstalledAdaptors;

out=daqhwinfo('advantech');

dio=digitalio('advantech',0);

hwinfo=daqhwinfo(dio)

hwinfo.Port(1)%DIO channel0

hwinfo.Port(2)%DIO channel1

hwoutlines=addline(dio,0:3,2,'out');
%-----analogue card registration and adding channels
daqregister('advantech') rehash toolboxcache
% clear all;
AO = analogoutput('advantech',1); chan0 = addchannel(AO,0); chan1 =
addchannel(AO,1); chan2 = addchannel(AO,2); chan3 =
addchannel(AO,3); set(AO,'TriggerType','Manual')
%---duration of each part of signal in second.maximum 4 parts for each signal--

du(1)=5 %duration of first part of signal in seconds
du(2)=5 du(3)=5 du(4)=5 rate=31.4 du=fix(du*rate)
duto=(du(1)+du(2)+du(3)+du(4))%total duration of whole signal
%-----defining the sign of signal in each part-----
%----determining sign of signal for channel one

s1(1)=1 %1 means positive signal- 0 means negative signal.s(2)=0 means
% the sign of second part of signal is negative and so on.
s1(2)=0 s1(3)=1 s1(4)=1
%----determining sign of signal for channel two
s2(1)=1 s2(2)=0 s2(3)=1 s2(4)=1
%as we work with two channels, we use default value of 1 for channel 3 and 4.
%-----making the digital signal-----
ds1=linspace(1,1,duto)%the main matrix of digital signal for
% channel1 which has all elements
ds2=linspace(1,1,duto)%the main matrix of digital signal for
%channel2 which has all elements
ds3=linspace(1,1,duto)
ds4=linspace(1,1,duto)

%-----setting sign for channel 1-----

if s1(1)==0
    for i=1:du(1);%this loop makes the elements of matrix
        % %related to first part of digital signal zero , and

```

```

        %this, makes this portion negative in final out put.
        ds1(i)=0;
    end;
end;
if s1(2)==0
    for i=du(1)+1:du(2)+du(1);
        ds1(i)=0;
    end;
end;
if s1(3)==0
    for i=du(2)+du(1)+1:du(3)+du(2)+du(1);
        ds1(i)=0;
    end;
end;
if s1(4)==0
    for i=du(3)+du(2)+du(1)+1:du(3)+du(2)+du(1)+du(4);
        ds1(i)=0;
    end;
end;
%-----setting sign for channel 2-----
if s2(1)==0
    for i=1:du(1);%this loop makes the elements of matrix
    %related to first part of digital signal zero , and
    %this, makes this portion negative in final out put.
        ds2(i)=0;
    end;
end;
if s2(2)==0
    for i=du(1)+1:du(2)+du(1);
        ds2(i)=0;
    end;
end;
if s2(3)==0
    for i=du(2)+du(1)+1:du(3)+du(2)+du(1);
        ds2(i)=0;
    end;
end;
if s2(4)==0
    for i=du(3)+du(2)+du(1)+1:du(3)+du(2)+du(1)+du(4);
        ds2(i)=0;
    end;
end;
%----making analogue signal-----

as1=linspace(3,3,duto)';%analogue signal for channel 1
as2=linspace(3,3,duto)';

```

```

as3=linspace(3,3,duto)';
as4=linspace(3,3,duto)';%analogue signal for channel 4

%-----writting data on output ports-----
for i=1:duto
    putdata(AO,[as1(i) as2(i) as3(i) as4(i)])
start(AO)
trigger(AO)
waittilstop(AO,200)
    putvalue(dio,logical([ds1(i),ds2(i),ds3(i),ds4(i)]));
end
%----making the out put zero after finishing signal-----
putvalue(dio,logical([0,0,0,0]));
putdata(AO,[0,0,0,0])
start(AO)
trigger(AO) waittilstop(AO,200) delete(dio) clear dio
%ActualRate = get(AO,'SampleRate')
delete(AO)
clear AO

```

D.3 Matlab codes for solving the motion equation of the droplet in silicon oil

```

clc clear
global Beta %global variable to be used in function file-its an coefficient
global alpha

k1=1    %0.05      %factor to modify polynomial coefficients
k2=1    %input('give k2 around 0.14 ')
D=0.0024; %diameter of ferrofluid droplet

ro=960 %kg/m3 density of silicon oil (sorrounding fluid)

Noo=100*10^(-6)% converted Cst to m2/s , as 100cst=10^(-4)

eta=ro*Noo % [eta]=kg/m.s dynamic viscosity , [ro]=kg/m3 , [noo]=m2/s

Ro_ff=1330 %kg/m3 density of ferrofluid

alpha=k1*6/(Ro_ff*pi*D^3)

Beta=k2*18*eta/(Ro_ff*D^2)

```



```

%-----
%options = odeset('MaxStep',0.01)
%options = odeset('InitialStep',0.1,'MaxStep',0.1)
oldOpts = odeset;
newOpts = odeset(oldOpts,'maxstep',1)%0.00015
[time1,Y] = ode15s(@ali2,[0 50],[0.00016 .0],newOpts);% ode45 solves
%equation numerically(runge-kutta) ,0=initial time, 50 maximum time ,
%l=initial displacement ,0=initial velocity
%[time2,Y2] = ode15s(@ali3,[50 100],[0.00599 .0],newOpts)
%-----plotting experimental and numerical data-----
%-----plotting velocity
figure
plot(time1,Y(:,2),'.')% plotting the velocity curve
hold on
%plot(time2,Y2(:,2),'.')% plotting the velocity curve
hold on
load V100I880D2427F001
u=u*10^-3
plot(time,u,'-');
title('velocity');
xlabel('t (sec)');
ylabel('u (m/sec)');
%-----plotting displacement-----
figure
plot(time1,Y(:,1),'.')% plotting the displacement curve

hold on plot(time2,Y2(:,1),'.') hold on
load V100I880D2427F001
x=x*10^-3+0.00015
plot(time,x,'-');
figure plot(Y(:,1),Y(:,2))
%---codes of the function program, ali2:
function dy = ali2(t,y) global Betta; global alpha;

% p1 = -9.299e+009; %(-1.227e+010, -6.329e+009)Data for Hp=500A/m,
%D=1.2 mm, I=1amp,dist=6.3mm,width=0.00015,gap=0.00015,
%
% p2 = 1.337e+008; %(7.736e+007, 1.9e+008)
%
% p3 = -6.009e+005; %(-1.009e+006, -1.923e+005)
%
% p4 = 735.4; %(-678, 2149)
%
% p5 = 0.1435; %(-2.222, 2.509)
%
% p6 = 0.001286; %(-0.0004159, 0.002988)
%
% p7 = 1.207e-006; %(8.227e-007, 1.592e-006)
% p1 = -7.53e+009 %(-9.282e+009, -5.779e+009)Data for Hp=500A/m, D=1.2 mm,
%I=0.75amp,dist=6.3mm,width=0.00015,gap=0.00015,
%
% p2 = 1.182e+008 %(8.502e+007, 1.514e+008)

```

```

%      p3 = -6.423e+005 %(-8.831e+005, -4.014e+005)
%      p4 =      1422 %(588.6, 2255)
%      p5 =      -1.538 %(-2.933, -0.1439)
%      p6 =      0.002025 %(0.001022, 0.003028)
%      p7 =      5.57e-007 %(3.304e-007, 7.836e-007)
% p1 = -2.541e+010 %(-3.133e+010, -1.95e+010)Data for Hp=500A/m, D=1.8 mm,
% I=0.75amp,dist=6.3mm,width=0.00015,gap=0.00015,
%      p2 =      3.99e+008 %(2.87e+008, 5.111e+008)
%      p3 = -2.168e+006 %(-2.981e+006, -1.355e+006)
%      p4 =      4799 %(1987, 7611)
%      p5 =      -5.192 %(-9.898, -0.4858)
%      p6 =      0.006834 %(0.003448, 0.01022)
%      p7 =      1.88e-006 %(1.115e-006, 2.645e-006)
% p1 = -6.024e+010 %(-7.425e+010, -4.623e+010)Data for Hp=500A/m, D=2.4 mm,
% I=0.75amp,dist=6.3mm,width=0.00015,gap=0.00015,
%      p2 =      9.459e+008 %(6.802e+008, 1.212e+009)
%      p3 = -5.138e+006 %(-7.065e+006, -3.211e+006)
%      p4 =      1.138e+004 %(4709, 1.804e+004)
%      p5 =      -12.31 %(-23.46, -1.152)
%      p6 =      0.0162 %(0.008172, 0.02423)
%      p7 =      4.456e-006 %(2.643e-006, 6.269e-006)
%
%      p1 = -6.797e+010 %(-8.671e+010, -4.923e+010)
%      p2 =      1.022e+009 %(6.667e+008, 1.378e+009)
%      p3 = -5.093e+006 %(-7.671e+006, -2.516e+006)
%      p4 =      9099 %(181.1, 1.802e+004)
%      p5 =      -6.406 %(-21.33, 8.516)
%      p6 =      0.01383 %(0.003093, 0.02457)
%      p7 =      6.929e-006 %(4.503e-006, 9.354e-006)

%Fx_fit = p1*x.^6 + p2*x.^5 + p3*x.^4 + p4*x.^3 + p5*x.^2 + p6*x.^1 + p7

dy = zeros(2,1); % a column vector
dy(1) = y(2); dy(2) = alpha*(p1*y(1).^6 + p2*y(1).^5 + p3*y(1).^4 +
p4*y(1).^3 + p5*y(1).^2 + p6*y(1).^1 + p7)-Betta*y(2);

```

Bibliography

- [1] J.B David, A.M Glennys, and G.M Walker. Physics and Applications of Microfluidic in biology. *Annu. Rev. Biomed. Eng.*, 4:261–286, 2002.
- [2] A. Mans, D.J Harrison, E.M.J Verpoorte, J.C Fettingier, A. Paulus, H. Lüdi, and H.M Widmer. Planar chips technology for miniaturization and integration of separation techniques into monitoring systems: capillary electrophoresis on a chip. *Journal of chromatography*, 593(1-2):253–258, 1992.
- [3] Y. Sun, YC Kwok, and NT Nguyen. A circular ferrofluid driven microchip for rapid polymerase chain reaction. *Lab on a Chip*, 7(8):1012–1017, 2007.
- [4] G.M. Whitesides. The origins and the future of microfluidics. *Nature*, 442(7101):368–373, 2006.
- [5] N.T. Nguyen and T.Q. Truong. A fully polymeric micropump with piezoelectric actuator. *Sensors and Actuators B: Chemical*, 97(1):137–143, 2004.
- [6] J.L. Perry and S.G. Kandlikar. Review of fabrication of nanochannels for single phase liquid flow. *Microfluidics and Nanofluidics*, 2(3):185–193, 2006.
- [7] T.Q Truong, X.Y Huang, and N.T Nguyen. Polymeric stack assembled micropump with su-8 check vales. *International Journal of Computational Engineering Science*, 4(2):249–252, 2003.
- [8] N.T. Nguyen, T.Q. Truong, K.K. Wong, S.S. Ho, and C.L.N. Low. Micro check valves for integration into polymeric microfluidic devices. *Journal of Micromechanics and Microengineering*, 14(1):69–75, 2004.

- [9] R. Mukhopadhyay. Diving into droplets. *Anal Chem*, 78(5):1401–4, 2006.
- [10] O.D Velev, B.G Prevo, and K.H Bhatt. On-chip manipulation of free droplets. *Nature*, 426(6966):515–6, 2003.
- [11] L.Y. Yeo and H.C. Chang. Static and spontaneous electrowetting. *Modern Physics Letters B*, 19(12):549–569, 2005.
- [12] A.A. Darhuber and S.M. Troian. Principles of microfluidic actuation by modulation of surface stresses. *Annual Review of Fluid Mechanics*, 37(1):425–455, 2005.
- [13] U. Lehmann, S. Hadjidj, VK Parashar, C. Vandevyver, A. Rida, and M. Gijs. Two-dimensional magnetic manipulation of microdroplets on a chip as a platform for bioanalytical applications. *Sensors and Actuators B*, 117(2), 2006.
- [14] O. D. velev and H.Bhatt. On chip micromanipulation and assembly of collidal particles by electric fields. *soft matter*, 124:384–392, 2006.
- [15] B.Guru and H.Hiziroglu. *Electromagnetic field theory fundamentals*. Cambridge, England, 2004.
- [16] H. Bruus. *Theoretical Microfluidics, Lecture notes*. DTU press, Denmark, 2006.
- [17] R.W. O’Brien and L.R. White. Electrophoretic mobility of a spherical colloidal particle. *J. Chem. Soc., Faraday Trans*, 2(74):1607–1626, 1978.
- [18] N.T. Nguyen and S.T. Wereley. *Fundamentals and Applications of Microfluidics*. Artech House, Boston, 2002.
- [19] S. L. Anna, N. Bontoux, and H. A. Stone. Formation of dispersions using ”flow focusing” in microchannels. *Applied Physics Letters*, 82(3):364–366, 2003.
- [20] K. Ahn, C. Kerbage, T.P. Hunt, R.M. Westervelt, D.R. Link, and D.A. Weitz. Dielectrophoretic manipulation of drops for high-speed microfluidic sorting devices. *Applied Physics Letters*, 88:024104, 2006.

- [21] D.R. Link, S.L. Anna, D.A. Weitz, and H.A. Stone. Geometrically Mediated Breakup of Drops in Microfluidic Devices. *Physical Review Letters*, 92(5):54503, 2004.
- [22] J. Zeng and T. Korsmeyer. Principles of droplet electrohydrodynamics for lab-on-a-chip. *Lab Chip*, 4(4):265–77, 2004.
- [23] S.T. Chang and O.D. Velev. Evaporation-Induced Particle Microseparations inside Droplets Floating on a Chip. *Langmuir*, 22(4):1459–1468, 2006.
- [24] J. Bico, U. Thiele, and D. Quere. Wetting of textured surfaces. *Colloids and Surfaces A*, 206:41–46, 2002.
- [25] MIT lecture notes available MIT opencourseware (OCW). section related to Surface tension, Lecture4.
- [26] J. BICO and D. QUÉRÉ. Self-propelling slugs. *Journal of Fluid Mechanics*, 467:101–127, 2002.
- [27] H. Bouasse. Capillarite, phenomenes superficiels. *Delagrave*, 1924.
- [28] M.M. Weislogel. Steady spontaneous capillary flow in partially coated tubes. *AIChE Journal*, 43(3):645–654, 1997.
- [29] C.D. Bain, G.D. Burnett-Hall, and R.R. Montgomerie. Rapid motion of liquid drops. *Nature*, 372(6505):414–415, 1994.
- [30] P.G de Gennes. Wetting: statics and dynamics. *Reviews of Modern Physics*, 57(3):827–863, 1985.
- [31] K. Ichimura et al. Light-Driven Motion of Liquids on a Photoresponsive Surface. *Science*, 288(5471):1624–1626, 2000.
- [32] A. Mazouchi. Thermocapillary migration of long bubbles in cylindrical capillary tubes. *Physics of Fluids*, 12(3):542, 2000.
- [33] F. Brochard. Motions of droplets on solid surfaces induced by chemical or thermal gradients. *Langmuir*, 5(2):432–438, 1989.

- [34] T.H. Ting, Y.F. Yap, N.-T. Nguyen, T.N. Wong, J.C.K. Chai, and Yobas. Active control for droplet based microfluidics. In *Proceedings of SPIE - The International Society for Optical Engineering 6416*, art. no. 64160E, 2007.
- [35] Z. Jiao, N.T. Nguyen, and X. Huang. Chaotic motion of microplugs under high-frequency thermocapillary actuation. *Journal of Micromechanics and Microengineering*, 17(1):180–185, 2007.
- [36] P.S. Glockner and G.F. Naterer. Thermocapillary control of microfluidic transport with a stationary cyclic heat source. *Journal of Micromechanics and Microengineering*, 15(12):2216–2229, 2005.
- [37] A.A. Darhuber, J.P. Valentino, S.M. Troian, and S. Wagner. Thermocapillary actuation of droplets on chemically patterned surfaces by programmable microheater arrays. *Microelectromechanical Systems, Journal of*, 12(6):873–879, 2003.
- [38] S.K. Cho, H. Moon, and C.J. Kim. Creating, transporting, cutting, and merging liquid droplets by electrowetting-based actuation for digital microfluidic circuits. *Microelectromechanical Systems, Journal of*, 12(1):70–80, 2003.
- [39] T.B. Jones, M. Gunji, M. Washizu, and M.J. Feldman. Dielectrophoretic liquid actuation and nanodroplet formation. *Journal of Applied Physics*, 89:1441, 2001.
- [40] M.G. Pollack, P.Y. Paik, A.D. Shenderov, V.K. Pamula, F.S. Dietrich, and R.B. Fair. Investigation of electrowetting-based microfluidics for real-time PCR applications. *Proc. 7th Int. Conf. Miniaturized Chemical and Biochemical Analysis Systems (MicroTAS)*, pages 619–622.
- [41] S. Kuiper et al. Variable-focus liquid lens for miniature cameras. *Applied Physics Letters*, 85(7):1128, 2004.
- [42] A. Wixforth, C. Strobl, C. Gauer, A. Toegl, J. Scriba, and Z. v Guttenberg. Acoustic manipulation of small droplets. *Analytical and Bioanalytical Chemistry*, 379(7):982–991, 2004.

- [43] C.J. Strobl, A. Rathgeber, A. Wixforth, C. Gauer, and J. Scriba. Planar microfluidic processors. *Ultrasonics Symposium, 2002. Proceedings. 2002 IEEE*, 1, 2002.
- [44] A. Wixforth. Acoustically Driven Programmable Microfluidics for Biological and Chemical Applications. *Journal of the Association for Laboratory Automation*, 11(6):399–405, 2006.
- [45] D.C. Duffy, H.L. Gillis, J. Lin, NF Sheppard, and G.J. Kellogg. Microfabricated centrifugal microfluidic systems: characterization and multiple enzymatic assays. *Anal. Chem*, 71(20):4669–4678, 1999.
- [46] M. Madou, J. Zoval, G. Jia, H. Kido, J. Kim, and N. Kim. Lab on a CD. *Annu Rev Biomed Eng*, 8:601–28, 2006.
- [47] M.J. Madou and G.J. Kellogg. LabCD: A centrifuge-based microfluidic platform for diagnostics. *Proceedings of SPIE*, 3259:80–93, 1998.
- [48] M.J. Madou. *Fundamentals of Microfabrication: The Science of Minuturization*. CRC Press, 2002.
- [49] G. Ekstrand, C. Holmquist, A.E. Orlefors, B. Hellman, A. Larsson, and P. Andersson. Microfluidics in a rotating CD. *Micro Total Analysis Systems*, pages 311–314, 2000.
- [50] A.L. Tiensuu, L.L. OveOhman, and O. Larsson. Hydrophobic valves by ink-jet printing on plastic CDs with integrated microfluidics. *Micro Total Analysis Systems 2000*, pages 575–578.
- [51] M. Madou, Y. Lu, S. Lai, J. Lee, and S. Daunert. A centrifugal microfluidic platform, a comparison. *Proc. TAS 2000 Conf*, pages 565–570.
- [52] J. Zeng, D. Banerjee, M. Deshpande, J. Gilbert, D. Duffy, and G. Kellogg. Design analyses of capillary burst valves in centrifugal microfluidics. *Proceedings of Micro Total Analysis Systems, Kluwer Academic Publisher, Dordrecht, The Netherlands*, pages 579–582, 2000.

- [53] R.D. Johnson, I.H.A. Badr, G. Barrett, S. Lai, Y. Lu, M.J. Madou, and L.G. Bachas. Development of a fully integrated analysis system for ions based on ion-selective optodes and centrifugal microfluidics. *Anal. Chem*, 73(16):3940–3946, 2001.
- [54] M.D. Simon. Diamagnetic levitation: Flying frogs and floating magnets(invited). *Journal of Applied Physics*, 87(9):6200, 2000.
- [55] M. Zborowski, L. Sun, L.R. Moore, P.S. Williams, and J.J. Chalmers. Continuous cell separation using novel magnetic quadrupole flow sorter. *Journal of Magnetism and Magnetic Materials*, 194(1-3):224–230, 1999.
- [56] M.A.M. Gijs. Magnetic bead handling on-chip: new opportunities for analytical applications. *Microfluidics and Nanofluidics*, 1(1):22–40, 2004.
- [57] S. Chikazumi and S.H. Charap. *Physics of Magnetism*. Krieger, 1978.
- [58] N.T. Nguyen and S. Wereley. *Fundamentals and Applications of Microfluidics*. Artech house, 2006.
- [59] I.F. Lyuksyutov. On-chip manipulation of levitated femtodroplets. *Applied Physics Letters*, 85(10):1817, 2004.
- [60] G. Madhavan. ” Smart” fluids solidify their presence. *Potentials, IEEE*, 25(1):19–21, 2006.
- [61] S. Kodama. Dynamic ferrofluid sculpture: organic shape-changing art forms. *communication of the ACM*, 51(6):79–81, 2008.
- [62] US Patent. (No. 3215572), 1965.
- [63] US Patent. (No. 3843540), 1974.
- [64] US Patent. (No. 4019994), 1974.
- [65] K. Landfester and L.P. Ramirez. Encapsulated magnetite particles for biomedical application. *Journal of Physics: Condensed Matter*, 15(15):S1345–S1361, 2003.
- [66] US Patent. (No. 4452773), 1984.

- [67] US Patent. (No. 4267234), 1981.
- [68] J. LEE, T. ISOBE, and M. SENNA. Magnetic properties of ultrafine magnetite particles and their slurries prepared via in-situ precipitation. *Colloids and surfaces. A, Physicochemical and engineering aspects*, 109:121–127, 1996.
- [69] M. Suzuki, M. Shinkai, M. Kamihira, and T. Kobayashi. Preparation and characteristics of magnetite-labelled antibody with the use of poly (ethylene glycol) derivatives. *Biotechnol. Appl. Biochem*, 21:335–345, 1995.
- [70] C. Gruttner and J. Teller. New types of silica-fortified magnetic nanoparticles as tools for molecular biology applications. *Journal of Magnetism and Magnetic Materials*, 194(1):8–15, 1999.
- [71] US Patent. (No. 6255477), 2001.
- [72] S. Miltenyi, W. Muller, W. Weichel, and A. Radbruch. High gradient magnetic cell separation with MACS. *Cytometry*, 11(2):231–8, 1990.
- [73] U.O. Hafeli and G.J. Pauer. In vitro and in vivo toxicity of magnetic microspheres. *J. Magn. Magn. Mater*, 194:76–82, 1999.
- [74] N.T. Nguyen, K.M. Ng, and X. Huang. Manipulation of ferrofluid droplets using planar coils. *Applied Physics Letters*, 89:052509, 2006.
- [75] W.B. Song, Z. Ding, C. Son, and B. Ziaie. Aqueous microdrop manipulation and mixing using ferrofluid dynamics. *Applied Physics Letters*, 90:092501, 2007.
- [76] H. Lee, Y. Liu, D. Ham, and R.M. Westervelt. Integrated cell manipulation system-CMOS/microfluidic hybrid. *Lab Chip*, 7(3):331–7, 2007.
- [77] M. Shikida, K. Inouchi, H. Honda, and K. Sato. Magnetic handling of droplet in micro chemical analysis system utilizing surface tension and wettability. *Micro Electro Mechanical Systems, 2004. 17th IEEE International Conference on MEMS*, pages 359–362, 2004.

- [78] J.R. Dorvee, A.M. Derfus, S.N. Bhatia, and M.J. Sailor. Manipulation of liquid droplets using amphiphilic, magnetic one-dimensional photonic crystal chaperones. *Nature Materials*, 3(12):896–899, 2004.
- [79] N.E Greivell and B. Hannaford. The design of a ferrofluid magnetic pipette. *Biomedical Engineering, IEEE Transactions on*, 44(3):129–135, 1997.
- [80] A. Hatch, AE Kamholz, G. Holman, P. Yager, and KF Bohringer. A ferrofluidic magnetic micropump. *Microelectromechanical Systems, Journal of*, 10(2):215–221, 2001.
- [81] C. Yamahata, M. Chastellain, V.K. Parashar, A. Petri, H. Hofmann, and MAM Gijs. Plastic micropump with ferrofluidic actuation. *Microelectromechanical Systems, Journal of*, 14(1):96–102, 2005.
- [82] R. Rong, J.W. Choi, and C.H. Ahn. An on-chip magnetic bead separator for bio-cell sorting. *Journal of Micromechanics and Microengineering*, 16(12):2783–2790, 2006.
- [83] P.S. Doyle, J. Bibette, A. Bancaud, and J.L. Viovy. Self-Assembled Magnetic Matrices for DNA Separation Chips. *Science*, 295(5563):2237–2237, 2002.
- [84] J. Jang and S.S. Lee. Theoretical and experimental study of MHD (magnetohydrodynamic) micropump. *Sensors & Actuators: A. Physical*, 80(1):84–89, 2000.
- [85] J. Zhong, M. Yi, and H.H. Bau. Magneto hydrodynamic (MHD) pump fabricated with ceramic tapes. *Sensors & Actuators: A. Physical*, 96(1):59–66, 2002.
- [86] N.T. Nguyen, A. Beyzavi, K.M. Ng, and X. Huang. Kinematics and deformation of ferrofluid droplets under magnetic actuation. *Microfluidics and Nanofluidics*, 3(5):571–579, 2007.
- [87] Z.G. Guo, F. Zhou, J.C. Hao, Y.M. Liang, W.M. Liu, and W.T.S. Huck. Stick and slide ferrofluidic droplets on superhydrophobic surfaces. *Applied Physics Letters*, 89:081911, 2006.
- [88] N. Pamme. Magnetism and microfluidics. *Lab on a Chip*, 6(1):24–38, 2006.

- [89] M. Zborowski, P.S. Malchesky, T.F. Jan, and G.S. Hall. Quantitative separation of bacteria in saline solution using lanthanide Er (III) and a magnetic field. *J Gen Microbiol*, 138(Pt 1):63–8, 1992.
- [90] I. Safarik and M. Safarik. Use of magnetic techniques for the isolation of cells. *Journal of Chromatography B*, 722:33–53, 1999.
- [91] C. Goubault and J.L. Viovy. Quantitative Microfluidic Separation of DNA in Self-Assembled Magnetic Matrixes. *Analytical Chemistry*, 76(13):3770–3776, 2004.
- [92] A. Meckes, J. Behrens, and W. Benecke. Electromagnetically driven microvalve fabricated in silicon. *Solid State Sensors and Actuators. Transducers'97 Chicago., 1997 International Conference on*, 2.
- [93] A. Rida, V. Fernandez, and MAM Gijs. Long-range transport of magnetic microbeads using simple planar coils placed in a uniform magnetostatic field. *Applied Physics Letters*, 83:2396, 2003.
- [94] W.P. Taylor, O. Brand, M.G. Allen, T. Relays, and C.A. Hawthorne. Fully integrated magnetically actuated micromachined relays. *Microelectromechanical Systems, Journal of*, 7(2):181–191, 1998.
- [95] S.C Mukhopadhyay, S. Yamada, and M. Iwahara. Investigation of near-surface material properties using planar type meander coil. *JSAEM Studies on Applied Electromagnetics and Mechanics*, 11:61–69, 2001.
- [96] I.J. Busch-Vishniac. The case for magnetically driven microactuators. *The Journal of the Acoustical Society of America*, 92:2353, 1992.
- [97] C.H. Ko, J.J. Yang, J.C. Chiou, S.C. Chen, and T.H. Kao. Magnetic Analysis of A Micromachined Magnetic Actuator Using the Finite Element Method. *Proceedings of SPIE*, 3893:127–136, 1999.
- [98] V.W.H. Lin, L.N. Hsiao, and V. Dhaka. Magnetic coil design considerations for functional magneticstimulation. *Biomedical Engineering, IEEE Transactions on*, 47(5):600–610, 2000.

- [99] A. Feustel, O. Krusemark, and J. Müller. Numerical simulation and optimization of planar electromagnetic actuators. *Sensors & Actuators: A. Physical*, 70(3):276–282, 1998.
- [100] J.W. Jansen, E.A. Lomonova, A.J.A. Vandenput, and C.M.M. Van Lierop. Analytical model of a magnetically levitated linear actuator. *2005 Conference Record - IAS Annual Meeting (IEEE Industry Applications Society)*.
- [101] A. Garcia, JA Carrasco, JF Soto, F. Maganto, and C. Morón. A method for calculating the magnetic field produced by a coil of any shape. *Sensors & Actuators: A. Physical*, 91(1-2):230–232, 2001.
- [102] W.G. Hurley, M.C. Duffy, S. O'Reilly, and S. CianOMathuna. Impedance Formulas for Planar Magnetic Structures with Spiral Windings. *IEEE Transactions on industrial electronics*, 46(2):271, 1999.
- [103] S.S. Mohan, M. del Mar Hershenson, S.P. Boyd, and T.H. Lee. Simple accurate expressions for planar spiral inductances. *Solid-State Circuits, IEEE Journal of*, 34(10):1419–1424, 1999.
- [104] F.M. White. *Viscous fluid flow*. McGraw-Hill New York, 1974.