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Electropermanent magnet actuation for droplet ferromicrofluidics

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Abstract

Droplet actuation is an essential mechanism for droplet-based microfluidic systems. On-demand electromagnetic actuation is used in a ferrofluid-based microfluidic system for water droplet displacement. Electropermanent magnets (EPMs) are used to induce 50 mT magnetic fields in a ferrofluid filled microchannel with gradients up to 6.4×10^4 kA/m². Short 50 μ s current pulses activate the electropermanent magnets and generate negative magnetophoretic forces that range from 10 to 70 nN on 40 to 80 μ m water-in-ferrofluid droplets. Maximum droplet displacement velocities of up to 300 μ m/s are obtained under flow and no-flow conditions. Electropermanent magnet-activated droplet sorting under continuous flow is demonstrated using a split-junction microfluidic design.

Keywords

Droplet Microfluidics; Droplet Sorting; Electropermanent Magnets; Ferrofluids; Magnetic Actuation; Ferromicrofluidics

INNOVATION

Magnetic actuation in microfluidic devices is an attractive method of droplet manipulation, given the minimal impact magnetic fields have on biological materials or fluids. Despite its use in passive microfluidic systems, the development of active magnetic actuation has been hampered by its weak forces. This work demonstrates the use of switchable electropermanent magnets (EPMs) to generate higher magnetic field gradients, which generate negative magnetophoretic forces of about 10 nN on 40 μ m-diameter droplets in a ferrofluid emulsion.

INTRODUCTION

Droplet microfluidics has emerged as its own distinct field, with several advantages over single-phase microfluidics and a growing number of applications^{1,2}. Several approaches have been explored for droplet actuation, including mechanical³, hydrodynamical^{4,5},

thermal⁶, electrical^{7,8}, magnetic^{9,10}, optical¹¹ and acoustic¹². Of these, magnetic actuation has seen limited development. A major reason is that the available magnetic field sources are not well suited for localized and on-demand actuation. The most ubiquitous source, permanent magnets, provide strong magnetic fields but are generally too large for localized actuation using field gradients and lack on-demand switching capability. These are heavily used in passive applications such as magnetically labeled sorting¹³. Electromagnets, on the other hand, provide on-demand switching capability and can be formed by fabricating planar coils for localized actuation¹⁴. A major drawback of electromagnets is their poor scaling. As the coil diameter decreases, the resistance of the wires increases and significant Joule heating can occur, due to the large currents required. Electromagnets also require continuous current to maintain the magnetic field.

In this work, we propose an alternative magnetic source that provides on-demand switching, stronger magnetic fields and localized actuation for microfluidics: EPMs. EPMs are magnetic devices that combine the on/off switching capability (magnetization and demagnetization) of an electromagnet with the magnetic strength of a permanent magnet¹⁵. These devices, most commonly used in industrial applications¹⁶, have recently been developed in smaller form factors for several applications including programmable matter¹⁷, micromotors¹⁸ and reconfigurable smartphones¹⁹.

This paper establishes the feasibility of using EPMs to actuate droplets in an emulsion ferromicrofluidic system. Ferrofluid-based droplet microfluidics, like any other droplet-based system, use two immiscible liquids (phases) to form emulsions¹. Two-phase systems consist of a continuous and a dispersed phase, and can be configured for oil-in-water or water-in-oil emulsions, depending on the material properties and application. For this study, a water-in-oil configuration will be used as aqueous droplets have a wider range of applications such as polymerase chain reaction (PCR) systems, cell sorting and drug response studies^{20,21}.

To demonstrate EPMs as actuators for droplet ferromicrofluidics, we have fabricated and tested a droplet-sorting configuration that can serve as the basic building block for a single droplet-sorting platform. In contrast to other droplet actuation mechanisms, such as dielectrophoresis⁷ or acoustophoresis¹², negative magnetophoretic forces are attractive for cell sorting since magnetic fields have no biological effect over cells, nor are they affected by ion distribution or pH of solutions²². In distinction to existing ferromicrofluidic magnetic actuation methods, such as passive sorting of aqueous droplets in oil-based ferrofluid using permanent magnets^{9,23} and active cell manipulation in water-based ferrofluid using microfabricated electromagnets²⁴, we demonstrate active sorting of aqueous droplets in a water-in-oil-based ferrofluid.

METHODS AND MATERIALS

Negative magnetophoretic force

Negative magnetophoresis ($\mathbf{F}_m$) is a repulsive force non-magnetic bodies (such as a water droplet) experience inside a magnetic medium in an inhomogeneous magnetic field²⁵;

$$\mathbf{F}_m = \int_V \mu_0 (\mathbf{M} \cdot \nabla) \mathbf{H} dV. \quad (1)$$

$\mathbf{M}$ and $\mathbf{H}$ denote the magnetization of the medium and the external magnetic field, respectively and μ_0 the permeability of free space. For dilute ferrofluids, with volume concentrations below 20%, both $\mathbf{M}$ and $\mathbf{H}$ can be assumed to be collinear since particle-to-particle interactions are minimal and do not interfere with particle rotation and alignment. Using vector identities, Equation (1) can be simplified (see Supplementary Information) to

$$\mathbf{F}_m = \int_V \mu_0 M \nabla H dV, \quad (2)$$

where M and H are the magnitude of $\mathbf{M}$ and $\mathbf{H}$ respectively. From Equation (2), the three major properties of the force are: (1) the magnetization, which depends on nanoparticle type, size and concentration within the ferrofluid (see Supplementary Information); (2) the gradient of the magnetic field, which depends on the geometry of the magnetic source; and (3) the volume of the non-magnetic body. These properties define the design space for negative magnetophoresis actuation in microfluidic systems.

Negative magnetophoresis driven bodies in solution experience two forces known as Stokes drag force ($\mathbf{F}_d$) and hydrodynamic lift force ($\mathbf{F}_l$). For spherical droplets, the drag force is given by²⁶

$$\mathbf{F}_d = -3\pi\eta_{oil}dv. \quad (3)$$

This force is proportional to the diameter of the droplet (d), the flow velocity (v) and the dynamic viscosity of the continuous oil phase (η_{oil}). Droplets moving inside a microfluidic channel experience an enhancement in drag force due to the proximity to the channel walls. This enhancement can be separated into parallel ($\mathbf{F}_{d\parallel}$)²⁷ and perpendicular ($\mathbf{F}_{d\perp}$)²⁸ drag force components as follows:

$$\mathbf{F}_d = \mathbf{F}_{d\parallel} + \mathbf{F}_{d\perp} = -3\pi\eta_{oil}dv(\lambda_{d\parallel} + \lambda_{d\perp}). \quad (4)$$

See Supplementary Information for the description of the coefficients ($\lambda_{d\parallel}$ and $\lambda_{d\perp}$). The negative sign in Equations (3) and (4) indicates that the force is opposite to the body's velocity. For spheres moving under Poiseuille flow in a rectangular microchannel, the lift force can be approximated by²⁹

$$\mathbf{F}_l \left(\frac{y}{w} \right) = \frac{3\rho_{oil}d^4v_m^2}{8\pi w^2} f \left(\frac{y}{w} \right), \quad (5)$$

where ρ_{oil} is the density of the continuous oil phase, v_m the maximum velocity of the Poiseuille flow, w the channel width and $f(y/w)$ is a function of the lateral position. The lift

force, defined as a lateral force, is symmetric about the channel centerline and has three equilibrium positions. The equilibrium at the channel centerline ($y/w = 0.5$) is not stable, but the other two positions, approximately at $y/w = 0.2$ and 0.8 , are stable. Depending on the starting lateral position of the droplet, the lift force will be positive, negative or zero with respect to the negative magnetophoretic force. Figure 1 shows the force balance diagram for an aqueous droplet in oil-based ferrofluid in the presence of an inhomogeneous magnetic field.

Electropermanent magnets

EPMs are made of a combination of two permanent magnets, with the same remnant magnetization but different coercivity (one hard — very difficult to demagnetize, one soft — easy to demagnetize), two ferromagnetic poles, and a coil (Fig. 2a). The EPM switching sequence is shown in Fig. 2c–f. When the magnets have opposite magnetization, the magnetic flux is contained inside the poles and hence the device is “off” (Fig. 2c). To generate a net magnetic flux outside the device, the magnetizations must be collinear. A short positive current pulse is sent through the coil to generate a magnetic field that reverses the magnetization of the soft magnet, thus turning the device “on” (Fig. 2d). In contrast to an electromagnet, the EPM stays “on” without a current (Fig. 2e). To turn the device “off”, a short negative current pulse is sent through the coil to reverse the magnetization of the soft magnet (Fig. 2f).

Magnetic actuation simulations

Finite element simulations of the droplet ferromicrofluidic system were performed using COMSOL Multiphysics® (Burlington, MA). The magnetic field was simulated for the OFF and ON states (Fig. 3a,b). For the OFF state (Fig. 3a), the soft magnet is simulated with a remnant magnetic field equal in magnitude but with opposite direction to the hard magnet. At this stage, the magnetic flux at the pole tips is weak in the y -direction and strong in the z -direction, hence there is minimal magnetization in the ferrofluid channel. For the ON state (Fig. 3b), the soft magnet is simulated with small remnant magnetic field, as experimental data shown below demonstrates that the current pulse almost completely demagnetizes this magnet but does not invert the magnetization. At this stage, the magnetic flux at the pole tips is strong in the y -direction and therefore, there is maximal magnetic field in the ferrofluid channel.

To understand the effect of EPMs on droplets, we studied the individual parameters that influence the negative magnetophoretic force: magnetic field, magnetic field gradient and magnetization. Figure 3c–e shows the simulation of these parameters for the ON state in the ferrofluid channel in the vicinity of a single EPM pole. The force density, Fig. 3f, represents the distribution of the lateral force (y) inside the ferrofluid channel and is given by

$$f_y = -\mu_0 M \frac{dH}{dy}. \quad (6)$$

From Fig. 3f, we can see that the region of strong force has roughly the same width as the EPM pole and is at its maximum at the inner edge of the pole. The magnetic force on a water droplet can be found by integrating Equation (6) across the volume of a ferrofluid droplet, since the magnetophoretic force for a non-magnetic object immersed in a magnetic medium has the same magnitude as that of a magnetic object of the same volume. As seen in Fig. 3g, the force scales with the droplet volume and pushes aqueous droplets away (y -direction) from the EPM and toward the opposite wall of the channel. These EPMs can deliver forces in the 10 nN range for droplets about 40 μm in diameter.

Electropermanent magnet assembly

EPMs were assembled by hand following Knaian's design¹⁵. The ferromagnetic poles were cut by wire electrodischarge machining from a 350 μm -thick Hiperco 50 sheet (EFI Specialty Metals, Los Alamitos, CA). This Iron–Cobalt–Vanadium alloy has ideal magnetic switching properties: low coercivity and low A.C. core loss³⁰. These properties enable fast magnetic switching with minimal losses. Furthermore, this material has a high magnetic saturation that allows the poles to magnetize to the full extent of the magnetization of the permanent magnets without saturating. N40 grade Neodymium–Iron–Boron (NIB) rods were used for the hard magnets (Amazing Magnets, Anaheim, CA). These magnets measure 3.2 mm long and 1.6 mm in diameter, have a high coercivity (H_c) of 907 kA/m and a remnant magnetic field (B_r) of 1.25 T. Grade 5 Aluminum–Nickel–Cobalt (Alnico) rods were used for the soft magnets (Magnet Kingdom, Medford, NY). These magnets measure 12.7 mm long and 1.6 mm in diameter, have a low H_c of 50 kA/m and a B_r of 1.25 T. To match the length of the NIB magnets, the Alnico rods were cut and ground lengthwise with a rotary tool. To make the EPM sandwich assembly, the pole pieces and the two magnets were glued in place with Loctite Hysol E60-HP epoxy adhesive (Henkel, Westlake, OH) and hardened overnight at room temperature. Excess glue was then sanded to render a smooth surface for the coil. The coil was made by winding 100 turns of 40 AWG bondable magnet wire (MWS Wire Industries, Westlake Village, CA). Finally, a drop of ethanol was used to activate the bond layer of the wire, after which the EPMs were dried for 15 minutes at room temperature. The activated bond layer fuses the windings together, adding mechanical stability. The completed EPM assembly is shown in Fig. 2b.

Electropermanent magnet characterization

The electrical properties of the EPMs were characterized. The coil is represented with a simple equivalent circuit of an inductor with series resistance. The D.C. resistance, R_{coil} , was measured with a digital multimeter (34401A, Keysight Technologies, Santa Rosa, CA) with a value of 4.2 Ω . The inductance, L_{coil} , was measured using the frequency sweep method (see Supplementary Information) with a function generator (33220A, Keysight Technologies, Santa Rosa, CA) and an oscilloscope (DSO5014A, Keysight Technologies, Santa Rosa, CA) with a value of 22.4 μH . The temporal response of the current flow through the EPM's coil can be modeled from these parameters. The initial time response to a step input for an inductor with series resistance is exponential with a time constant $\tau_{coil} = L_{coil}/R_{coil}$ or 5.3 μs . This value corresponds to the time it takes to reach 63.2% of the maximum current. At 5τ , or 26.5 μs , the current reaches 99.3% of its maximum value, thus setting the maximum switching speed for these EPMs.

EPM switching requires bidirectional high current pulses. A MOSFET H-Bridge configuration, commonly used to drive D.C. motors, was used for the driver circuit (see Supplementary Information). The current was supplied from a single D.C. power supply (E3633A, Keysight Technologies, Santa Rosa, CA). The switching pulses, positive and negative, were generated from dedicated function generators controlled by LabVIEW (National Instruments, Austin, TX) and amplified using operational amplifier gain stages.

Ferrofluid preparation

Droplet ferromicrofluidics requires a two-phase system consisting of a continuous and a dispersed phase. Deionized water was used for the dispersed phase and an oil-based ferrofluid for the continuous phase. The oil-based ferrofluid was prepared by mixing EMG 1200 dry iron oxide nanoparticles (Ferrotec, Santa Clara, CA) with mineral oil (Molecular biology grade, Sigma Aldrich, St. Louis, MO). With an average diameter of 10 nm, these nanoparticles exhibit superparamagnetic properties that along with the surfactant coating prevent particle agglomeration, even in the presence of a magnetic field. EMG 1200 nanoparticles have a fatty acid surfactant coating that allows their solubility into the mineral oil, thus forming a stable colloidal suspension. EMG 1200 pellets were ground into a fine powder using a mortar. The powder was mixed with the mineral oil in an ultrasonic bath for 15 minutes at 80°C. The final ferrofluid mixture had a concentration (ϕ) of $4.5 \pm 0.6\%$ (v/v), a density (ρ_{oil}) of 1.027 ± 0.025 g/mL, a viscosity (η_{oil}) of 21.4 ± 2 mPa·s, and a saturation magnetization (M_s) of 20 ± 2.8 kA/m (for calculations, see Supplementary Information). To complete the two-phase system, Span 80 nonionic surfactant (Sigma-Aldrich, St. Louis, MO) was added to the ferrofluid at a concentration of 5% (w/w) to prevent droplet coalescence.

Fluidic device setup

Microfluidic chips were made of two layers of PDMS (Sylgard 184, Dow Corning, Midland, MI) using standard soft-lithography techniques. The 3 mm top channel layer was cast in an SU-8 mold (MicroChem, Westborough, MA), patterned on a silicon wafer, by pouring a 5:1 (v/v) ratio of base to curing agent and cured for 2 hours at 60°C. The 1 mm bottom layer was made by pouring a 20:1 (v/v) ratio on to a blank silicon wafer and curing for 2 hours at 60°C. Next, the chips were diced and the ports punched with a 22G blunt needle. Finally, both layers were aligned and thermally bonded overnight at 60°C.

Droplet generation

Droplets were generated by flow-focusing in a cross-junction (Fig. 4a) driven by syringe pumps (Pump 11 Elite, Harvard Apparatus, Plymouth Meeting, PA). This design uses two oil-based ferrofluid channels to pinch off the water stream, causing droplet breakup at the nozzle region. The droplet generation rate, size and spacing can be tuned by modulating the ratio between the oil flow rate Q_o and the water flow rate Q_w (see Supplementary Information). Typical flow rates for the oil are between 1 and 10 $\mu\text{L}/\text{min}$ and between 0.1 and 1 $\mu\text{L}/\text{min}$ for the water. Generated droplets are spherical, so long as their diameter is smaller than the height of the channel, $h = 55 \mu\text{m}$ (Fig. 4b). Larger droplets take the shape of a spherical frustum (Fig. 4c). These droplets have a large contact area with the top and

bottom of the channel, which allows optical access to the droplet, in contrast to smaller droplets surrounded by the opaque ferrofluid.

RESULTS

Electropermanent magnet actuation

The EPM's temporal response was tested with a 100 μ s, square current pulse. The exponential current response was measured through a series 0.1 Ω resistor using an oscilloscope (Fig. 5a). A time constant of 6.6 μ s and a maximum current of 5.15 A characterize the response. This measured time constant is longer than the calculated 5.3 μ s due to the lower effective resistance of the coil when connected to the H-Bridge configuration. The minimum actuation time of the EPM, defined as $t_{act} = 5\tau_{coil}$ is 33 μ s.

The magnetic field of the EPM was also characterized. First, 100 μ s current pulses were used to magnetize and demagnetize the EPM. The magnetic field at the edge of the EPM pole was measured for each current level using a gaussmeter (GM 1-ST, AlphaLab, Salt Lake City, UT). Results, shown in Fig. 5b, present hysteretic behavior typical of a soft magnet. Positive currents pulses, with an actuation voltage (V_s) sweep from 0 to 20 V, were used to magnetize the EPM from 0 to 23.4 mT. Negative current pulses, with a V_s sweep from 0 to -20 V, were used to demagnetize the EPM from 23.4 to -2.4 mT. The extreme values of the magnetic field indicate that the soft magnet is being modulated from negative saturation (max $-B_r$), where it becomes slightly stronger than the hard magnet, to almost the point of demagnetization ($B_r \sim 0$). Measuring the field of an EPM assembled without a soft magnet confirmed that this was the case. The measured field strength matched the maximum field strength obtained with modulation, proving the soft magnet is demagnetizing but not inverting the internal magnetization. Using the 0.6 mm thickness of the Hall-Effect probe of the gaussmeter, the magnetic fields at the edge of the poles and at the surface of the magnets were extrapolated and used to estimate the B_r for each magnet. The B_r for the hard magnet was estimated at 1.1 T and for the soft magnet 1.17 T and 0.3 T for the OFF and ON stages, respectively. These values were used in the finite element model.

The effect of pulse length on the maximum magnetic field was investigated. Starting with a fully demagnetized EPM, current pulses ranging from 10 to 100 μ s were applied and the resulting magnetic field was measured at the edge of the poles. As seen in Fig. 5c, intermediate levels of magnetic field can be achieved by varying the length of the pulse. The magnetic field reaches 90% of the maximum value with a 50 μ s current pulse.

The EPM has an effective resistance of 3 Ω when connected to the H-Bridge and the power supply. During an EPM switching (50 μ s, 20V), the EPM consumes 124 W of power and requires 5.8 mJ of energy (see Supplementary Information). The EPM requires no static power after activation, since the magnetization is maintained until a reverse pulse is applied.

Droplet actuation

To test the actuation force on droplets, we mounted an EPM adjacent to a microfluidic channel with the poles perpendicular to the channel (Fig. 6a). An EPM with asymmetric poles was used to test the effect of a single pole. The distance from the edge of the pole to

channel, p , was 200 μm . A flow-focusing configuration was used to generate droplets. As droplets flowed past the pole of the EPM, their trajectories were recorded with a high-speed camera (Zyla 5.5 sCMOS, Andor Technology, Concord, MA) at a frame rate of 500 fps and an inverted microscope (Nikon Instruments, Melville, NY). Figure 6b shows a sequence of images of a 70 μm droplet moving past an activated EPM pole. The axial (v_x) and lateral (v_y) components of the droplet velocity were calculated by tracking their frame-by-frame position with an image-processing algorithm in ImageJ (NIH, Bethesda, MA). The resulting velocity components are plotted in Fig. 6c. The axial component starts at a constant velocity of 4.2 mm/s far away from the EPM and decreases as the droplet passes by the activated EPM pole and is pushed towards the opposite wall, eventually reaching a new steady velocity of 2.8 mm/s. This behavior is expected for a Poiseuille flow where the maximum flow is at the center of the channel and zero at the walls. Since the droplets are large, compared to the channel size, they are driven by non-zero velocity streamlines even when part of the droplet is in contact with the channel wall. The lateral velocity component starts at zero, as the droplets are self-centered by the hydrodynamic lift force. The lateral force increases as the droplet approaches the pole. During this period, the droplet acceleration indicates the magnetic force is stronger than the sum of drag and lift forces. The droplet reaches a maximum lateral velocity of 0.31 mm/s near the inner edge of the EPM pole. At this point, the magnetic force is equal to the sum of drag and lift. After this point, the lateral velocity decreases as the perpendicular component of the drag force becomes dominant. The lateral velocity eventually reaches zero as the droplet contacts the channel wall.

This experiment was repeated with multiple droplet sizes, all presenting similar behavior. For smaller droplet sizes, higher flow rates were required since droplet size is inversely proportional to flow rate in a flow-focusing droplet generator. At high flow rates, droplet tracking led to lower droplet velocity resolution. To increase the resolution of small droplet tracking, we developed a stopped-flow setup (Fig. 6d). Using a syringe attached to the chip outlets, back pressure was applied to stop the flow and allow droplet placement in front of the EPM pole. Once the droplet was in place, the EPM was activated to study droplet displacement dynamics during EPM switching. Figure 6e shows a sequence of images of a 40 μm droplet's motion after EPM activation (100 μs , 20 V). Figure 6f shows the measured droplet velocity as it moves towards the opposite wall. Once the EPM is activated, the droplet experiences acceleration and reaches a terminal velocity of approximately 0.3 mm/s within 20 ms. During this period the magnetic force is equal to the drag force, since there is no lift force with the flow stopped. The velocity remains constant until the perpendicular component of the drag force becomes dominant and reduces the velocity. The velocity eventually reaches zero as the droplet contacts the channel wall. This experiment was repeated for multiple droplet sizes.

The drag and lift forces were calculated for all droplets in both experiments. For the droplet sizes and flow velocities we use in this paper, the lift force is at least an order of magnitude smaller than the other forces, so it was neglected. The results are summarized in Fig. 6g plotted along the magnetic force. All droplets smaller than $h = 55 \mu\text{m}$ were measured under stopped-flow conditions, while the larger droplets were measured under flow. Also, for droplets with diameters larger than h , their effective diameter is used in Fig. 6g, which is the diameter of a sphere of the same volume as the actual droplet volume. As we can see, the

drag force matches the magnetic force for small droplets but diverges for large droplets. A reason for this discrepancy is likely that larger droplets have a greater surface area in contact with the channel walls; therefore, the additional friction will reduce the overall velocity. Nonetheless, droplet actuation was demonstrated with actuation forces ranging from several nN to up to 70 nN and displacement velocities up to 0.3 mm/s.

EPM-actuated water droplet sorting was demonstrated using a split channel configuration as shown in Fig. 6h,i. In this configuration, asymmetrical outlet channels were used to create two paths with different fluidic resistances. The bottom channel, wider than the top channel, has a lower resistance. Without magnetic actuation, all droplets flow toward the bottom channel (Fig. 6h). With the EPM activated, droplets are diverted toward the opposite wall, eventually flowing out the top channel (Fig. 6i, Supplementary Video 1). This configuration can be used to sort two different types of targets and can be cascaded in stages for multiple target selection.

DISCUSSION

This work demonstrates the feasibility of using EPMs for actuating droplets in a ferromicrofluidic emulsion. The negative magnetophoretic force is an attractive alternative to the more established dielectrophoretic and acoustophoretic force-based droplet manipulation techniques. Since magnetic fields do not affect biochemical processes in the droplet, the negative magnetophoretic force is limited only by the maximum field gradient that can be practically achieved. For dielectrophoretic forces, high electric fields may induce charging at the surface of the droplet³¹ and can also affect the pH of the droplet solution. High acoustic fields could induce cavitation³², potentially damaging cells.

The principal advantage of miniature EPMs over conventional electromagnets is that they require no static power to maintain the ON state, eliminating concerns about Joule heating of the microfluid and excessive system power. In addition, our non-optimized, hand-assembled EPMs demonstrate higher switching speeds and higher field gradients than are feasible using planar microcoil electromagnets¹⁴. External electromagnets are capable of generating larger magnetic field strengths than is possible for planar microcoils, but their large size translates into lower magnetic field gradients and their larger inductances leads to slower switching times.

Droplet ferromicrofluidic systems using switchable EPMs will require increasing the negative magnetophoretic force. In particular, high-throughput droplet sorting will require both stronger and more localized magnetic forces. The EPM-ferrofluid system has several parameters that can be modified to increase F_m . Although larger droplets are desirable if optical access is needed, as seen in Fig. 6g, the displacement force and velocity of large droplets reaches a saturation point, due to the reduction in the volume of ferrofluid surrounding the droplet. Using a ferrofluid with higher particle concentration directly increases magnetization and the magnetic force (see Supplementary Fig. 4) but also increases viscosity and drag force, thus limiting the net effect. Modifying the shape of the EPM poles can enhance the local magnetic gradients and increase F_m . Sharper edges not only increase gradients, but reduce the width of the actuation window along the channel. The

current droplet actuation design has an actuation window similar in size to the width of the poles, which is about 350 μm . For single-droplet sorting, the actuation window must be around 100 μm for droplets in the 50 μm range and inter-droplet spacing of about 100 μm .

EPMs can also benefit from the advances in material technology and microfabrication, since they scale favorably¹⁵. Advances in magnetic material deposition³³, permanent magnet microfabrication³⁴ and coil integration³⁵ can lead to finer magnetic field localization and lower switching power, since the resistance of the coil decreases with scaling. Since EPMS only require power when switching, they are attractive for miniaturizing complex droplet-based digital ferromicrofluidic systems³⁶.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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Figure 1.

Force balance diagram. Negative magnetophoretic force F_m is counterbalanced by the Stokes drag force F_d and the hydrodynamic lift force F_l . A water droplet immersed in magnetic oil, such as a ferrofluid, will experience a repulsive force in the presence of an inhomogeneous magnetic field.

Figure 2.

EPM assembly and operation with simplified magnetic flux depiction (black dotted lines and arrows). (a) Diagram of an electropermanent magnet (EPM) sandwich configuration with dimensions: $a = 4$ mm, $b = 3.6$ mm, $s = 3.2$ mm and $d = 350$ μ m. (b) Image of assembled EPM. (c) Magnets have opposite magnetization: EPM is deactivated (d) Positive current is applied to EPM aligning the magnetizations: EPM is activated. (e) Magnetizations stay aligned after current is turned off; EPM stays activated. (f) Negative current is applied to the EPM reversing the magnetizations: EPM is deactivated.

Figure 3.

Finite element modeling of droplet ferromicrofluidic system. The EPM is simulated using the remnant magnetic field of the magnets. The microchannel is simulated at separation of 200 μm . (a) Magnitude of lateral magnetic flux density $|B_y|$ and normalized magnetic flux field (red arrows) for EPM in the OFF state. There are no magnetic flux lines inside microchannel as the field is stronger in the z -direction; thus, the magnetic field is contained within the EPM. (b) $|B_y|$ and magnetic flux for EPM in the ON state. The strong lateral component (y -direction) of the magnetic flux at the pole tips induces a field inside the

microchannel. **(c)** Magnitude of lateral magnetic field component (B_y) inside ferrofluid channel with detail of total magnetic flux (red arrows). **(d)** Gradient of the magnitude of the magnetic field H in the lateral direction. **(e)** Magnitude of magnetization M . **(f)** Lateral magnetic force density (f_y) inside the ferrofluid microchannel. **(g)** Lateral magnetic force ($F_{m,y}$) for spherical droplets with a range of diameters. Droplets are located at the center of the channel and at a distance of $y = 300 \mu\text{m}$ from the edge of the poles and then swept in the x -direction. The location of maximum force is at the inner edge of the EPM pole.

Figure 4.

(a) Droplet generation using flow-focusing of oil-based ferrofluid and water in a cross-junction geometry. The image was taken with the high-speed camera coupled to the inverted microscope. (b) Cross-sectional view of a spherical droplet with $d < h$. (c) Cross-sectional view of a spherical frustum droplet with $d > h$.

Figure 5.

Switching performance of EPMs. (a) Pulse response (blue) of the EPM coil current to a 100 μs voltage pulse (red). (b) Magnetic field measurements at the edge of the EPM pole as actuation voltage (V_s) is swept. Positive V_s activates the EPM while negative V_s deactivates them. (c) Measured magnetic flux density magnitude at the edge of the EPM pole for different current pulse lengths. The magnetic field saturates with pulses longer than 50 μs .

Figure 6.

Experimental results of EPM actuation on water droplets in oil-based ferrofluid. The EPM pole was mounted at a distance of $p = 200\ \mu\text{m}$. **(a)** Diagram of the EPM configuration for droplet displacement experiments under flow. **(b)** Displacement sequence for a $70\ \mu\text{m}$ water droplet with EPM edge location highlighted by the red dotted line. The experiment used the following flow rates: $Q_o = 1.6\ \mu\text{L}/\text{min}$ and $Q_w = 0.05\ \mu\text{L}/\text{min}$. **(c)** Axial and lateral velocity profile of a $70\ \mu\text{m}$ droplet under flow along the channel. **(d)** Diagram of the EPM configuration for droplet displacement experiments under no flow. **(e)** Displacement

sequence for a 40 μm droplet after EPM actuation. **(f)** Lateral velocity profile of a 40 μm droplet across the width of the channel. **(g)** Lateral force comparison for multiple sized droplets. The magnetic force $\mathbf{F}_m$ simulated with COMSOL Multiphysics[®]. The drag force $\mathbf{F}_d$ was calculated from Equation (4) using maximum droplet velocity. Droplets smaller than 55 μm represent no-flow experiments, while the larger droplets represent in-flow experiments (see text). **(h)** Split channel configuration with EPM OFF. Droplets of 70 μm flow toward the bottom channel due to lower fluidic resistance. **(i)** Split-channel configuration with EPM ON. Droplets of 70 μm are displaced toward the upper channel due to magnetic force.