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The influence of magnetic carrier size on the performance of microfluidic integrated micro-electromagnetic traps

Boris Le Droff · Liviu Clime · Teodor Veres

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Abstract Efficient manipulation and capture of magnetic carriers in fluid stream require appropriate magnetic confinement devices whose performances are strongly dependent on the nature of the magnetic carriers. In this sense, we have performed a systematic investigation of the magnetic capture efficiencies for five commercially available superparamagnetic particles pumped along rectangular microfluidic channels using microelectromagnetic traps composed of planar circular current-carrying microwires and cylindrical ferromagnetic posts. In addition, in order to obtain a quantitative description of particle movement, we have implemented a numerical model for the dynamics of magnetic objects subjected to magnetic field gradients in conventional continuous-flow microfluidic devices. Fully 3D trajectories of the particles, effective cross-sectional areas of the microchannel as well as micro-electromagnet trapping efficiencies are compared to experimental measurements and a very good agreement is obtained. Finally, a simple and effective analytical model to determine the critical velocity, i.e. when the magnetic trapping device is no longer able to capture and hold 100% of the magnetic superparamagnetic particles, is also presented.

Keywords Lab on a chip · Microfluidics · Magnetic capture · Superparamagnetic particles

1 Introduction

Magnetophoresis is a well-known technique in biotechnology since it provides a rapid and convenient method for separation, sorting, purification or filtering of bio-analytes (DNA, cells, bacteria,...) grafted on magnetic carriers (Whitesides et al. 1983; Verpoorte 2003; Gijs 2004; Hsing et al. 2007). Recently, the integration of these nowadays standard laboratory technologies into micro-total analysis systems (μ -TAS) has been receiving significant attention since such miniaturized devices allow to minimize the amounts of reagents and the sample volume needed while increasing the analysis throughput (Pamme 2006). Moreover, the possibility to magnetically confine and concentrate target analytes in microscopic volumes for in situ detection offers a great potential for sensitive detection of minute amounts of biological species. For example, we have recently demonstrated that this principle can be used to detect target DNA in the zeptomolar range directly on magnetic particles in non-flowing conditions, using very sensitive optical methods, while the micro-beads are being magnetically confined in a small volume by a micro-electromagnetic trap (μ -EMT) (Dubus et al. 2006). However, efficient manipulation and capture of magnetic carriers in flow conditions require a careful design of the magnetic confinement devices as well as appropriate flow conditions.

Previous reports in the literature have shown the possibility to effectively manipulate and control the motion of superparamagnetic beads in μ -TAS by using either permanent magnets or electromagnets, or combinations of both. Permanent magnets provide strong magnetic fields which are very important for these types of carriers as they lead to increased magnetic moments of the particles and consequently increased magnetic driving forces (Kim and Park 2005; Zaytseva et al. 2005). Microfluidic integrated

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magnetic manipulation has also been demonstrated with different designs of magnetic devices relying upon ferromagnetic microstructures with sharp edges coupled with the use of macroscopic permanent magnets or electromagnets (Deng et al. 2002; Mirowski et al. 2004; Rida and Gijss 2004). Although the use of macroscopic magnets offers the advantage of a much stronger magnetic field, they are not suitable when microscaled magnetic confinement devices or further miniaturization and multiplexing in μ -TAS are required. On the contrary, the use of microfabricated electromagnets allows to create strong local magnetic field gradients that can easily be either switched on and off or modulated by controlling the intensity of the current powering the micro-coils (Choi et al. 2001; de Vries et al. 2005; Smistrup et al. 2005; Dubus et al. 2006; Ramadan et al. 2006a, b; Smistrup et al. 2006). Moreover, by using several integrated micro-electromagnets (Deng et al. 2001; Brzeska et al. 2004; Lee et al. 2004; Liu et al. 2007), modulable magnetic field profiles for effective manipulation and positioning of magnetic carriers can easily be achieved. Although these efforts have to some extent addressed several challenges related to the magnetic manipulation in microfluidic channels, there are still many remaining issues related to magnetic bead confinement with micron or even sub-micron resolution, as required for the development of ultrasensitive biosensors.

Previous works on magnetic bead manipulation were essentially concentrated on the magnetic capture of micron-size carriers due to their high magnetic load and consequently large magnetic moment. However, in certain applications, smaller particles may offer some additional advantages. For example, due to the relatively high surface to volume ratio, magnetic nanoparticles offer a larger binding capacity while forming more stable suspensions. In the context of optical biosensors, the use of nanoparticles should also result in less light scattering leading to better signal-to-noise ratios and therefore improving the detection limits. Nevertheless, the magnetic force being proportional to the particle volume, any reduction in size of the magnetic particles leads to an important decrease of the magnetic load and consequently to reduced separation and trapping efficiencies. For applications requiring the detection of low concentration of biological or chemical targets, capture with an efficiency as close as possible to 100% becomes a very important issue. It is therefore essential to be able to predict the magnetic trappability of magnetic carriers in continuous flow microfluidic devices. For practical microfluidic lab-on-a-chip devices, this can be achieved by appropriate magnetic confinement devices and by taking into account the nature and the intrinsic magnetic properties of the magnetic carriers.

In this paper, we investigate both theoretically and experimentally the effect of the size of the magnetic

particles (from 4.5 μm down to 200 nm) on their trappability in axially symmetrical micro-electromagnets. The magnetic field and magnetic gradient generated by the μ -EMT as well as the differential equations for the motion of particles in the microfluidic channels are numerically solved for different flow velocities and compared to the experimental results. A simple analytical model for the critical velocities of the particles in magnetic trapping devices is described and tested for our experimental setup. The paper is organized as follows. The process developed for the fabrication of the confinement device as well as the experimental setup used to measure the bead velocities in a microfluidic channel are described in Sect. 2. In Sect. 3 we briefly present the numerical model for the dynamics of non-interacting superparamagnetic particles subjected to magnetic field gradients, drag and gravitational forces. Comparisons between experimental and theoretical results as well as some concluding remarks are presented in Sects. 4 and 5, respectively.

2 Experimental

2.1 Magnetic particles

Most of the commercially available magnetic particles used for biochemical applications consist of nanometer-sized magnetic particles dispersed into polymer or silica matrices to form beads of various sizes. Five types of commercial magnetic monodisperse particles functionalized with carboxyl groups with sizes ranging from 200 nm up to 4.5 μm and magnetic loads from 20 to 70% (as purchased from Invitrogen/Dynal, Oslo, Norway and Ademtech, Bordeaux, France) have been used for this study. Table 1 presents some relevant physical and geometrical properties of these particles. With the exception of the 4.5- μm diameter particles, all others were CY3-labeled in order to facilitate the visualization of their trajectories. All particles ($\sim 10^5$ beads/mL) were suspended in phosphate buffered

Table 1 Physical properties of the commercial magnetic beads (information supplied by the manufacturers)

Brand name	Diameter (μm)	Susceptibility (no units)
M-450 ^a	4.5	1.6
M-280 ^a	2.8	0.976
MyOne ^a	1	1.458
MasterBeads ^b	0.5	1.5
Adembeads ^b	0.2	1.425

^a From Invitrogen/Dynabeads

^b From Ademtech

saline solution (PBS) containing 0.1% of Tween[®] 20 (Sigma Chemical Co.) to prevent their aggregation.

2.2 Design and fabrication of μ -EMT devices

The micro-electromagnets used in the magnetic confinement devices regularly consist of planar current-carrying micro-wires. A first generation of such micro-electromagnets consists of simple ring-shaped conductive loops, as those designed by Lee and collaborators (2001). Since the functionality of this simple design in a μ -EMT for the detection of minute amounts of DNA in static flow conditions (Dubus et al. 2006) has already been demonstrated, in the present paper, we test an improved design in which the micro-electromagnet contains a ferromagnetic Ni micropost in the middle of the loop. A similar design (but more difficult to fabricate, with a NiFe post inside the microfluidic channel) has also been recently proposed (Ramadan et al. 2006a, b). In our approach, the ferromagnetic structure is not in contact with the fluid carrying the magnetic particles, allowing the formation of a monolayer of trapped magnetic particles thus minimizing the optical interferences (i.e. light scattering) during the confocal detection (Dubus et al. 2006). In such design, the lines of magnetic field are concentrated inside the ferromagnetic structure creating strong local gradients and consequently increased magnetic forces. Another advantage of the addition of ferromagnetic posts in the design of these electromagnets is that they allow the reduction of the applied current intensity and consequently the heat dissipated by Joule effect.

Figure 1 summarizes the microfabrication process of both microfluidic integrated μ -EMTs fabricated on $2 \times 3''$ microscope glass slides. Transparent substrates were used to facilitate the optical observation and the fluorescent detection on inverted microscopes. Prior to the microfabrication steps, the microscope glass slides were cleaned by immersion in piranha solution (3:1 $\text{H}_2\text{SO}_4\text{:H}_2\text{O}$), blow dried with N_2 , rinsed with deionised water and then dehydrated for 1 h at 200°C. A conducting seed layer of Cr (10 nm) is used as adhesion layer followed by a layer of Au (70 nm), both deposited by using an electron-beam evaporator (step i). In the next step, AZ-9260 photoresist (Clariant Corporation, Somerville, NJ, USA) was spun on the wafer to define, using UV photolithography, the pattern of the electroplating photoresist mold of the microcoil and the contact pads (step ii). Before electroplating, a reactive ion etch descum process was applied in order to remove all the photoresist residues from the development process. Next, the mold was filled with 10 μm of electroplated gold (Technic Gold 25 ES, Technic Inc., Providence, RI, USA) using a DC current density of 2 mA/cm^2 and at a

temperature of 50°C (step iii) followed by photoresist stripping with acetone (step iv). To complete the fabrication of μ -EMT (with a central Ni micropost), the process was repeated from step (ii) to step (vii) using a double coat of AZ-9260. The resist mold was then filled with 35 μm of electroplated nickel (Sulfamate Nickel Plating SN-10, Transene Company Inc., MA, USA) at a DC current density of 10 mA/cm^2 and at a temperature of 40°C. In the final step of the fabrication process, the non-electroplated area of the gold seed layer was sputter-etched and the chromium adhesion layer was selectively etched using a commercial solution (Chromium Etchant CR-4S, Cyantek Corporation, Inc., Fremont, CA, USA). Inset in Fig. 1 shows the so obtained μ -EMT.

Finally, a thick insulating layer (50 μm) of premixed poly(dimethylsiloxane) (PDMS, Sylgard 184, Dow Corning) was spun on top of the electroplated microstructures. The planarity of the PDMS surface was then checked using a Dektak profilometer. The optimal thickness of the insulating layer needed to optimize the profile of the magnetic field generated by the μ -EMT was established by numerical simulations (described in detail in the next section). With this microfabrication process, embedded ferromagnetic structures as well as conductors with large cross-sectional areas can easily be achieved. Although the magnetic fields created by electromagnets increase with the intensity of the applied electrical currents, all the experiments in this paper were performed at relatively low-intensity DC electrical currents (~ 250 mA) in order to limit the Joule heat dissipated in micro-coils.

Microfluidic channels having 20 μm in depth and 100 μm in width were fabricated in PDMS using a replica molding technique (Xia and Whitesides 1998). Inlet and outlet holes were punched in the replica and finally aligned and placed in conformal contact with the PDMS surface of the μ -EMT substrate, forming a tight and reversible seal.

2.3 Measurement setup

The microfluidic device was placed on an inverted microscope (TE2000 Eclipse, Nikon, Canada). A back-thinned electron multiplier CCD camera (Model C9100-12, Hamamatsu, Japan) coupled to the microscope was used to capture motion sequences of the fluorescent microbeads. Fluidic connections were made by silicon tubing inserted into the holes of the device in order to connect a syringe mounted on a syringe pump (model 200, KD Scientific Inc., MA). The magnetic beads were introduced in the microfluidic channel by aspirating the sample solution from the outlet reservoir at flow rates ranging from 0.01 up to 10 $\mu\text{L h}^{-1}$. The images of the magnetic bead trajectories

Fig. 1 Fabrication scheme of micro-electromagnetic traps; *i* 10/70 nm of Cr/Au seed layer is deposited onto a $2 \times 3''$ glass slide substrate by e-beam evaporation; *ii* 15 μm thick layer of AZ-9260 photoresist is spun onto the substrate and UV patterned; *iii* 10 μm of gold is electroplated in the photoresist mould; *iv* the photoresist is stripped; *v* a second UV lithography step is performed on a double coat of AZ-9260; *vi* 35 μm thick film of Ni is electroplated; *vii* the photoresist is stripped, and the uncovered seed layer is etched; *viii* a thick insulating layer of PDMS is spun onto the microfabricated electromagnetic device; *ix* the microfluidic channel is aligned with the $\mu\text{-EMT}$. *Inset* is a SEM image of the $\mu\text{-EMT}$ before the deposition of the insulating layer

under the influence of the magnetic field were captured with an integration time of 10 ms. Motion tracking analysis software (SimplePCI, Compix) was then used to track magnetic microbeads across time-separated image sequences in order to obtain their actual line trajectory, their number as well as their velocities. The magnetic trapping (capture) efficiency T at given flow velocity V of the liquid is defined as

$$T(V) = \frac{N}{N_T} \times 100\% \quad (1)$$

where N corresponds to the number of particles trapped by the $\mu\text{-EMT}$ whereas N_T is the total number of counted particles entering the microscope field of view during the capture sequences. Each experiment was performed by passing through the device an average amount of about 100 particles at constant average flow rates and applied magnetic field. The magnetic field was then removed (by switching off the electrical current), the beads flushed out by high flow rate pumping and the experiments repeated for various flow rates. The parameter chosen in this study to characterize the efficiency of capture for superparamagnetic beads is the velocity at which the $\mu\text{-EMT}$ is no longer able to capture and hold 100% of the magnetic particles, which is called critical (or cut-off) velocity and denoted by V_c .

In order to find the correlation between the physical properties of superparamagnetic particles, the capture efficiency T and the critical bead velocity V_c , numerical simulations of the dynamics of these particles are compared to experimental measurements, the numerical model used in this comparison is being described in the following section.

3 Numerical simulation

The liquid flow in the microfluidic channel of the above described $\mu\text{-EMT}$ is modeled as a Poiseuille flow in a rectangular cross-section pipe (Happel 1965) of width $w_c = 100 \mu\text{m}$ and height $h_c = 20 \mu\text{m}$ (see Fig. 2). The magnetic field is computed by considering a Ni micropost (P) with a diameter of $D = 40 \mu\text{m}$ and height $h_s = 35 \mu\text{m}$ whose top surface is placed at $\Delta h = 15 \mu\text{m}$ below the bottom wall of the microfluidic channel (C) (Note that with this geometry, the thickness of the insulating layer spun on top of the electroplated microstructure is $h_s + \Delta h = 50 \mu\text{m}$).

To compute the magnetic field inside the microchannel, one can take advantage of the axial symmetry of the electromagnet and find the magnetic field by solving the 2D problem

$$\nabla \times (\mu^{-1} \nabla \times A_\phi) = j_\phi \quad (2)$$

in the plane yOz , where A_ϕ is the circumferential component of the magnetic vector potential $\vec{A}$ (so that $\vec{H} = \mu^{-1} \nabla \times \vec{A}$), j_ϕ the current density carrying the electric loop and μ the absolute magnetic permeability of the post. The computation domain $D_1 D_2 D_3 D_4$ (Fig. 2) is chosen to be a rectangle of sides $D_1 D_4 = 1 \text{ mm}$ and $D_1 D_2 = 0.8 \text{ mm}$, large enough in order to impose $A_\phi = 0$ as Dirichlet boundary conditions on $D_1 D_2 \cup D_2 D_3 \cup D_3 D_4$. Since the solution A_ϕ has to be symmetrical with respect to Oz axis, a Neumann-type boundary condition $\partial_y A_\phi = 0$ is imposed on $D_1 D_4$. To

Fig. 2 Longitudinal section of the second generation μ -EMT with a plane defined by electromagnet and microchannel symmetry axes. The micropost (P) symmetry axis coincides with Oz and it is magnetized by the magnetic field generated by a metallic (and non-magnetic) solenoid circular wire S. A scalar cut plane of the magnetic force acting upon Dynabeads with 2.8 μm in diameter in the vertical

symmetry plane is drawn as a contour fill plot in the region corresponding to the microfluidic channel. Numerical values of the magnetization vector are about 9 kA/m at the center of the post (O) and uniformly vary from 20 up to 10 kA/m on the post generatrices. D is the diameter of the magnetic post, s is the width of the square section of the electrical coil, and d is the gap between both

reflect the experimental conditions, a current density of $24 \times 10^8 \text{ A/m}^2$, that is a current intensity of 240 mA is set for the source term j_ϕ in the subdomain $C_1C_2C_3C_4$ corresponding to the electric coil. Since electroplated Ni is a relatively soft ferromagnetic material and the magnetic field generated by the electric coil is relatively weak, we consider the material in the magnetic post P as linear with a relative magnetic permeability $\mu_r = 1,000$ (Ramadan et al. 2006b). The inherent problems associated with the evaluation of the spatial derivatives of the magnetic field on linear triangles can be avoided by using higher order elements in the finite element scheme for Eq. (2) and advanced least-square-based interpolation algorithms.

In the absence of any inertial contribution to the motion of micro- or nano-objects in viscous liquids (Mikkelsen and Bruus 2005), the law of motion $\vec{r}(t)$ for these particles can be obtained from the static equilibrium condition between the magnetic $\vec{F}_{\text{mag}}$, buoyancy $\vec{F}_g$ and drag $\vec{F}_{\text{drag}}$ forces (Gijs 2004; Mikkelsen and Bruus 2005; Smistrup et al. 2005),

$$0 = \vec{F}_{\text{drag}} + \vec{F}_{\text{mag}} + \vec{F}_g \quad (3)$$

Considering the size of these particles, the effect of the Brownian motion can safely be ignored (Lee et al. 2001).

For relatively large-radius beads (at least tens of nanometers) the drag force is well approximated by the Stokes law (Happel 1965):

$$\vec{F}_{\text{drag}} = 6\pi\eta R(\dot{\vec{r}} - \vec{V}) \quad (4)$$

where η is the liquid viscosity, R the particle radius and $\vec{V}$ the liquid velocity. The liquid velocity field $\vec{V}$ in Eq. (4) is obtained from the Fourier sum corresponding to a Poiseuille flow in a rectangular cross-section channel (Happel 1965).

$\vec{F}_{\text{mag}}$ in Eq. (3) can be expressed as

$$\vec{F}_{\text{mag}} = \mu_0 \frac{4\pi R^3}{3} \chi_a (\vec{H} \cdot \nabla) \vec{H} \quad (5)$$

where χ_a is the magnetic susceptibility of the beads and μ_0 the absolute magnetic permeability of free space. It is worthwhile to mention that χ_a represents an “apparent” susceptibility, that is the slope dM/dH of the magnetization curves near the origin and not the intrinsic susceptibility of the magnetic material of the beads. Consequently, geometrical (demagnetization) effects as well as the heterogeneity in the composition of the magnetic beads are already included in the numerical value of χ_a (Fonnum et al. 2005).

The effect of the gravitational field is included through the buoyancy term

$$\vec{F}_g = -\frac{4\pi R^3}{3} (\rho_b - \rho) g \hat{z} \quad (6)$$

with ρ_b and ρ standing for the densities of beads and liquid, respectively, g the gravitational acceleration and $\hat{z}$ the unit vector of Oz axis.

As we can see in Eq. (5), $\vec{F}_{\text{mag}}$ depends on both magnetic field intensity $\vec{H}$ and its gradients ∇H_α with $\alpha \in \{x, y, z\}$, so in order to obtain large values of the magnetic actuation forces, both strong magnetic fields and large gradients of its components are necessary.

With the above defined interaction terms, Eq. (3) becomes a first-order ordinary differential equation with $\vec{r}(t)$ as unknown functions for each launched particles. The microbeads are launched from different upstream points ($\gamma < 0$) of the microchannel and the equation of motion (3) solved iteratively by using the Runge–Kutta–Fehlberg algorithm with adaptative step-size control (Galassi et al. 2006). The simulations run until $z \geq z_{\text{min}}$ (microchannel floor) and $t \leq T_{\text{MAX}}$ for each launched particle, where T_{MAX} stands for the maximum time interval allowed for the simulation. If $z \leq z_{\text{min}}$ while $t \leq T_{\text{MAX}}$ then the particles are considered as trapped. If $y \geq y_{\text{max}}$ while $t \leq T_{\text{MAX}}$ then the particles are considered as escaped. Then the magnetic trapping efficiency T is evaluated at different flow speeds by simply counting the number of particles trapped and escaped from the microelectromagnet. The profiles $T(V)$ and the corresponding critical (cut-off) parameters V_c are thus computed for all the magnetic beads listed in Table 1. In order to evaluate the influence of the gravitational field on these quantities, Eq. (3) was solved firstly with the buoyancy term $\vec{F}_g$ included (full model) and then by neglecting it ($\vec{F}_g = 0$).

4 Results and discussion

Magnetic beads were introduced in the microfluidic channel as described in the experimental section. As expected, with no electrical current applied to the μ -EMT (i.e. in the absence of any magnetic field), no noticeable deflections in the particle trajectories are observed. When a maximum current of 240 mA is applied to the μ -EMT, the beads experience a magnetic force, which affects their trajectories and lead them toward regions of local maxima of the magnetic field. In the absence of the magnetic post, the magnetic field is generated only by the electric microcoil and this maximum of the magnetic field is localized on its symmetry axis Oz (dot-dashed line in Fig. 3a). In this case, the trapped particles will be confined preponderantly in a region near Oz , where the horizontal components of the magnetic field gradient vanish. The influence of the ferromagnetic post on the trapping process is twofold: first, the magnetic field becomes stronger (on the channel floor it is roughly doubled—Fig. 3b) and consequently the

Fig. 3 The effect of adding the Ni cylindrical post on the magnetic field generated by the μ -EMT along Oy' (a) and Oz' (b) directions

magnetization in individual beads will be enhanced. Second, the magnetic field presents steeper spatial variations in certain regions around the microelectromagnet—especially near the micropost edges. Both of the above mentioned effects lead to larger values of the magnetic actuation forces in certain regions of the microfluidic channel (see the contour fill plot in Fig. 2) and consequently to larger capture efficiencies. Moreover, since the distribution of the local field maxima may change, the spatial distribution of the trapped particles will also change in the presence of the electromagnetic post such that they will agglomerate near the micropost edges, where the magnetic field presents maximal values, rather than above the center of the loop, as it is the case in the absence of the magnetic post. This is confirmed by both numerical simulations (images a and b in Fig. 4) and experimental measurements (images c and d in the same figure). As we can see on these figures, depending on their position with respect to the μ -EMT, magnetic particles are either trapped by the

Fig. 4 Theoretical (**a** lateral view, **b** top view) and **c** experimental (top view) visualizations of particle trajectories for $2.8\ \mu\text{m}$ beads under the influence of the magnetic field created by the μ -EMT and the drag force in microfluidic flow (The arrow indicates the flow direction. Scale bar = $100\ \mu\text{m}$). Image **d** is an optical image showing the position of the magnetic beads (dark spots) right above the magnetic post (bright circle)

microelectromagnet (see particles flowing in the middle of the microfluidic channel which are stopped when passing over the Ni post) or slightly deflected from their original trajectory and then escaped from μ -EMT. The optical image in Fig. 4d, shows the magnetic beads (dark spots) trapped right above the Ni post (bright circle).

In an effort to understand how the flow rates affect the capture efficiency we have computed the capture cross-sections of the microchannel at four different flow rates for the $500\ \text{nm}$ diameter magnetic beads (Fig. 5). We can clearly see the existence of two regions, where particles are either trapped (dark regions) or escaped (light regions) from the microelectromagnet. As the flow velocity increases, more particles escape from magnetic confinement and the area corresponding to escaped particles increases accordingly. The upper square corners of the microfluidic channel are the first regions where magnetic capture is difficult to achieve, because they are far from the μ -EMT and therefore subjected to lower magnetic forces. On the contrary, beads near the bottom of the microfluidic channel, i.e. close to the μ -EMT, are captured easier, even at relatively high velocity. Further improvements of microfluidic magnetic trapping devices would have to consider rounded upper corners for the microfluidic channels (Unger et al. 2000) or thinner microfluidic channel in the case of smaller particles, in order to bring the particles much closer to the electromagnet where the magnetic force is stronger and the capture efficiency much better at equivalent flow rate.

Since the main focus of the present paper is to find the critical (cut-off) velocities of the particles and not the accurate profiles of the $T(V)$, we limited our study to V_c for the particles listed in Table 1 (Fig. 6). Both experimental and numerical values are relatively close, inherent discrepancies being attributed to the different number of

Fig. 5 Theoretical magnetic capture cross section for $500\ \text{nm}$ magnetic beads (MasterBeads) flowing at various velocities V in the microfluidic channel. The cross indicates the flow direction, w_c and h_c are the width and the height of the microfluidic channel, respectively. Particles are either trapped (dark area) or have escaped (light area)

launched beads (much smaller in experiment). It is also worth mentioning that the magnetic susceptibility of the various magnetic materials used can vary depending on the supplier and it has already been demonstrated that some of these commercial magnetic beads show some variations in their magnetic properties (Baselt et al. 1998). Smaller beads undergo smaller magnetic actuation forces and their trappability is decreased accordingly: the critical velocity decreases from $\sim 200\ \mu\text{m s}^{-1}$ for the larger beads down to $\sim 0.33\ \mu\text{m s}^{-1}$ for the 200-nm diameter beads (note that for the later beads, their critical velocity was too small to accurately measure it). Intermediate values ranging from a few $\mu\text{m s}^{-1}$ up to tens of $\mu\text{m s}^{-1}$ are obtained for the other beads. (This corresponds to flow rates $< 1\ \mu\text{L h}^{-1}$ in the present device configuration.) These results show that a decrease of more than two orders of magnitude in the critical velocity may seriously limit the possibility of speeding up the flow rates in microfluidic channels without compromising on the trapping efficiency. This

Fig. 6 Experimental and theoretical critical velocity V_c to reach 100% capture efficiency for several magnetic bead sizes and best linear fit of experimental data with E_m as fit parameter: $E_m = (8.1 \pm 0.6) \times 10^{10} \text{ A}^2/\text{m}^3$. Error bars indicates standard deviation of repeated experiments

considerably increases the time needed for processing sample volumes, this time being a very important factor for efficient analytical detections.

It is interesting to note that, when the buoyancy term is neglected in the equation of motion (3), a linear dependence between the velocities of the beads and the square of their radius is expected. In the following, we verify if the critical parameter V_c follows the same law, namely if

$$V_c = E_m \cdot \frac{2\mu_0\chi_a R^2}{9\eta} = E_m \cdot \xi \quad (7)$$

where $E_m \propto \langle (\vec{H} \cdot \nabla) \vec{H} \rangle$ is a phenomenological parameter measured in A^2/m^3 and associated with the strength (or efficiency) of the electromagnetic trap. It depends not only on the strength of the electromagnet but on the spatial position and configuration of the microfluidic channel as well. This is the reason why, in order to verify this hypothesis, we plotted in Fig. 6 the experimental as well as the theoretical critical velocities V_c as function of the parameter $\xi = 2\mu_0\chi_a R^2/9\eta$, directly related to the magneto-phoretic mobility of particles (Gijs 2004). The theoretical values on this plot were computed in two situations corresponding to whether gravitational effects are taken into consideration (open circles) or they are completely neglected (open squares). As we can see the cut-off velocities obtained by numerical simulations in the absence of the buoyancy term ($F_g = 0$ in Fig. 6) are very close to a linear dependence as suggested by Eq. (7). However, some deviations from this law are observed when the effects of the gravitational field are considered (full model), deviations due probably to the additional R^3

dependence induced by the buoyancy term (6). Although the experimental data are a little bit scattered (full disks), the points in Fig. 6 suggest indeed a power law dependence ξ^n for critical velocities. The slope of the least square regression line of experimental data gives $E_m = (8.1 \pm 0.6) \times 10^{10} \text{ A}^2/\text{m}^3$ for the efficiency of our trapping device. In order to find the degree n of the dependence $V_c(\xi)$ in Fig. 6, an evaluation of the parameter E_m is mandatory. A first try in our analysis consists of the average value $\langle H_z d_z H_z \rangle$ along the electromagnet symmetry axis inside the microchannel by using

$$\begin{aligned} \left\langle H_z \frac{dH_z}{dz} \right\rangle &= \frac{1}{h_c} \int_0^{h_c} H_z(z') \frac{dH_z}{dz'} dz' = \frac{H_z^2(h_c) - H_z^2(0)}{2h_c} \\ &= 9.01 \times 10^{10} \text{ A}^2/\text{m}^3 \end{aligned} \quad (8)$$

and the finite element solution of Eq. (2). As we can observe in the equation above, the numerical value of $\langle H_z d_z H_z \rangle = 9.01 \times 10^{10} \text{ A}^2/\text{m}^3$ is very close to that of E_m found from the slope of experimental data in Fig. 6. The fact that $E_m \cong \langle H_z d_z H_z \rangle$ suggests a numerical value of n very close to 1, i. e. a linear dependence of V_c with respect to ξ .

In fact, Eq. (8) suggests a very simple way to evaluate the efficiency of an axially symmetric microelectromagnet and consequently the critical velocities for any superparamagnetic bead by using Eq. (7) if the phenomenological parameter E_m of the electromagnet is known. As shown in Eq. (8), E_m is given by only the extreme values of the magnetic field on the floor and the ceiling of the microfluidic channel. Furthermore, the evaluation of the critical velocities can be simplified when the field values can be computed analytically, for example when simple current distributions (without ferromagnetic posts) are employed.

As the susceptibility of the beads employed in this study does not vary too much from one sample to another, the observed dependence points out the strong influence of the particle size and therefore the limitations in the design of devices for magnetic manipulation of this kind of magnetic carriers. Therefore, the design and synthesis of new magnetic nanocarriers with enhanced magnetic properties becomes of great interest for such applications.

5 Conclusions

A systematic investigation of particle trappabilities under the influence of the magnetic fields created by μ -EMTs has been performed for various commercial magnetic beads whose diameters range from 0.2 to 4.5 μm and compared to a predictive model with excellent agreement. The magnetic trapping efficiency and effective cross-sectional area of the channel obtained by numerical simulations and

compared to experimental measurements provide an important insight into the dynamics of superparamagnetic particles in continuous flow microfluidic channels. The critical velocity of superparamagnetic beads is directly proportional to their magneto-phoretic mobility, the proportionality factor being related to the efficiency of the trapping device. For axially symmetrical electromagnets, this efficiency can simply be computed by the difference of the squared magnetic field intensities on the ceiling and floor of the microfluidic channel, along the symmetry axis. Although reducing the size of magnetic particles might offer some advantages, it significantly reduces the magnetic trapping efficiency and therefore the dynamic range for the microfluidic flow. Therefore, factors that determine the optimum bead size will be a compromise between the settling time for the suspended beads (for larger beads), the magnitude of the magnetic force needed for the capture of the magnetic beads against the fluid speed as well as the signal-to-background ratio when optical detection is used.

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