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A Two-Magnet System to Push Therapeutic Nanoparticles

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Abstract

Magnetic fields can be used to direct magnetically susceptible nanoparticles to disease locations: to infections, blood clots, or tumors. Any single magnet always attracts (pulls) ferro- or paramagnetic particles towards it. External magnets have been used to pull therapeutics into tumors near the skin in animals and human clinical trials. Implanting magnetic materials into patients (a feasible approach in some cases) has been envisioned as a means of reaching deeper targets. Yet there are a number of clinical needs, ranging from treatments of the inner ear, to antibiotic-resistant skin infections and cardiac arrhythmias, which would benefit from an ability to magnetically “inject”, or push in, nanomedicines. We develop, analyze, and experimentally demonstrate a novel, simple, and effective arrangement of just two permanent magnets that can magnetically push particles. Such a system might treat diseases of the inner ear; diseases which intravenously injected or orally administered treatments cannot reach due to the blood-brain barrier.

Keywords

Targeted drug delivery; magnetic force; injection; magnetic particles; drug delivery system; nanoparticles; ferrofluid; permanent magnets

Introduction

During magnetic drug targeting (1-4), magnetically-responsive particles coated with or containing therapeutic molecules are usually injected systemically and are then pulled into tissue targets by applied magnetic fields. This drug delivery approach is potentially useful for treatment of cancer, stroke, infection and other diseases because it allows therapeutic substances to be concentrated to disease sites (e.g. to solid tumors, blood clots, or infections) with reduced systemic concentrations and related undesirable side effects (5-10). The magnetically-susceptible objects are often micro- or nano-scale iron oxides or other magnetizable particles coated with biocompatible materials and conveying therapeutic payloads. Other encapsulating objects, such as polymer, microsphere, micelle, and nanocapsule delivery systems, can contain magnetically active materials coupled with therapeutics thus forming composite magnetic delivery carriers (7,11,12). Any single magnet, external or surgically implanted, will always attract such particles towards it. This is due to the physics of magnetic fields (the magnetic part of Maxwell's equations) and the magnetization response of the particles (1-4,13,14).

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In many cases, depending on the location of the target tissue and any anatomical obstacles such as bones, limbs, or the head; pulling the particles in with an external magnet is either not optimal or not possible. For example, the inner ear is behind the blood-brain barrier. Microcirculation vessels that supply the inner ear have walls that are impermeable even to small drug molecules (15). There are a variety of inner ear pathologies that could benefit from targeted delivery. Acute noise-induced injury, uncontrolled labyrinthitis (dizziness), and profound tinnitus (loud ringing in the ears) affect millions of people (16). Here injectable or ingestible drugs are ineffective or inadequate because the blood-brain barrier prevents delivery. To bypass this barrier, a magnetic force has been used to pull therapeutic payloads on nanoparticles into the inner ear of guinea pigs through the round window membrane (RWM) of these animals (17). But, to pull nanoparticles into the inner ear of humans, a pulling magnet would have to be placed on the opposite side of the head at a long working distance. That magnet would need to be extremely strong (magnetic fields and forces drop off quickly with distance (18-21)) hence large, expensive, and unwieldy. Based on the guinea pig studies above, the needed magnetic strength to create the same forces at the required working distance of 30 - 50 cm for humans would exceed current FDA safety limits (8 Tesla for adults, 4 T for children) (22-24). Instead, the ability to *push* particles magnetically from the same side as the target ear, over a much shorter 5 – 10 cm working distance, would facilitate effective treatment with safe magnetic fields (Figure 1).

Past work on improving magnet design for magnetic therapy has focused on increasing magnetic gradients by choice of materials and geometries (25-27), and on design of Halbach arrays for near-skin treatments (4,28) and for Stereotaxis's Niobe system (29,30) used to precisely manipulate catheter and guidewire surgical tools in magnetically assisted cardiology treatments. Embedding and anchorage of magnetic material into blood vessel walls, as a way of creating enhanced localized magnetic field gradients under the action of an externally applied uniform magnetic field, has been considered in (31-33). However, to our best knowledge, there are as yet no published results on a magnetic system to that can push in or 'magnetically inject' nanoparticles.

The physical reason any single magnet, whether it is permanent or electrical, will always attract particles is because the particles experience a force from low to high applied magnetic field strength: the magnetic force on ferro-, para-, or super-magnetic particles is

always in the direction of the gradient of the magnetic field squared $\vec{F}_{mag} = k \nabla \left(|\vec{H}|^2 \right)$ with $k > 0$ (1,14). Since for a single magnet the magnetic field strength $|\vec{H}|$ is greatest nearest to the magnet, the particles are drawn in towards it. In order to push particles away, a local magnetic field minimum must be created at a distance – around this minimum particles will experience forces that go outwards to the surrounding region of higher magnetic field. It is straight-forward to choose the orientation of just two magnets so that the magnetic field cancels at a displaced node. The end result is a simple and practical device that, when placed correctly relative to the nanoparticles (magnets then node then particles), can effectively push them away.

Theory and Modeling

Our system uses a static magnetic field to push away particles. Static magnetic fields are governed by the magneto-static portion of Maxwell's equations (13)

$$\nabla \times \vec{H} = \vec{j} \quad (1)$$

$$\nabla \cdot \vec{B} = 0 \quad (2)$$

where $\vec{B}$ is the magnetic field [in Tesla], $\vec{H}$ is the magnetic intensity [Amperes/meter], and $\vec{j}$ is the current density [A/m²] (it is zero in our case). In a material, $B = \mu_0 (H + M) = \mu_0 (H + \chi H)$ where M is the material magnetization [A/m], χ is the magnetic susceptibility, and $\mu_0 = 4\pi \times 10^{-7}$ N/A² is the permeability of vacuum. Magnetic fields propagate almost unchanged through animal or human tissue (the magnetic susceptibility of tissue is close to zero $\chi \sim 10^{-6} - 10^{-4}$ (34,35)). In contrast, the magnetic susceptibility of magnetite (Fe₃O₄), which is commonly used to form the cores of magnetic nanoparticles (36), is $\chi \sim 20$. Compared to tissue, this susceptibility is 5 to 7 orders of magnitude greater and hence these particles are strongly influenced by magnetic fields (1,37). Due to their nanoscale size they behave in a super-paramagnetic fashion: when the magnetic field is removed they quickly demagnetize (38).

A single particle will experience a force that depends upon the magnetic field and field gradient applied at its location (39-42)

$$\vec{F}_M = \frac{4\pi a^3}{3} \frac{\mu_0 \chi}{(1+\chi/3)} \vec{H} \frac{d\vec{H}}{d\vec{x}} = \frac{2\pi a^3}{3} \frac{\mu_0 \chi}{(1+\chi/3)} \nabla \left(|\vec{H}|^2 \right) \quad (3)$$

where a is the radius of a nanoparticle [in meters] and ∇ is the gradient operator [with units 1/m]. The first relation is more familiar in the magnetic drug delivery literature and clearly shows that a spatially varying magnetic field ($d\vec{H}/d\vec{x} \neq 0$) is required to create a magnetic force. The second relation is equivalent by the chain rule and states that the magnetic force on a para-, ferro-, or super-magnetic particle is always from low to high magnetic field and is proportional to the gradient of the magnetic field strength squared. The magnetic force is also proportional to the volume of the particle. If the applied magnetic field is sufficiently strong to saturate the particle, then $\vec{H} d\vec{H}/d\vec{x}$ in equation (3) is modified to $\vec{M}_{sat} d\vec{H}/d\vec{x}$ where $\vec{M}_{sat}$ is the saturated magnetization of the particle. Since $\vec{M}_{sat}$ lines up with $\vec{H}$, this does not change the direction of the force, only its size.

Based on equation (3), to create an outward push force, $|\vec{H}|^2$ must be made to increase going away from the system of magnets. A simple way to achieve this is to create a local magnetic field minimum at a distance – the magnetic field strength will increase outwards from this minimum and create outward forces. Figure 2 shows how such a minimum can be created at a distance using just two permanent magnets. (The figure is a schematic: numerical simulations of Maxwell's equations are shown in Figure 3 and Figure 4, experimental results are shown in Figure 9.) A single magnet will have the field lines shown. When the magnet is tilted clockwise, along a chosen field line, there will be a location where the magnetic field is purely vertically up (point A). A second identical magnet, flipped and tilted counter-clockwise, will have a like location (point B) where the magnetic field is vertically down and has the same magnitude. If these two magnets are positioned as shown, so that points A and B overlap at point C, the magnetic fields add together (Maxwell's equations are linear) and exactly cancel at C to provide a zero magnetic field ($|\vec{H}|=0$). The assumption here is that the magnet material coercivity is sufficiently high that the magnetic field from the first magnet does not substantially alter the magnetization of the second

magnet (and vice versa). Since the magnetic fields do not cancel at other points surrounding C, this point is a location of a locally minimum (zero) magnetic field strength $|\vec{H}|$. The forces go from low to high magnetic field strength, so in the region beyond C they will push particles away from the two magnets.

It is also possible to create a cancellation in other ways, for example by choosing a point A where the magnetic field is purely horizontal and then both tilting and reversing the polarity of the second magnet to flip the sign of the magnetic field at point B. When these two magnets are correctly overlaid there is again a cancellation of the magnetic field only at point C. The presence of this cancellation node, and the resulting push out forces, has been verified experimentally for both cases – but the experimental results shown below are focused on the first (magnets magnetized through their width) case.

To understand and analyze the magnetic field, and hence forces, around our magnetic push system, we have used Engel-Herbert and Hesjedal's analytical 3-dimensional solution for the magnetic field around a uniformly magnetized rectangular permanent magnet (18). By rotating and adding two such solutions, we attain an analytic expression for the magnetic field around our system. Comsol was used to visualize the magnetic field lines, magnetic field strength, and the resulting magnetic forces, and these are shown in Figure 3 in a planar slice through the midplane of the magnets. Figure 4a shows the magnetic field strength in a cut-away 3-dimensional view. The node is visible as a black (local low magnetic field) region to the right of the shown magnet.

The magnetic push force created by a node cancellation is weaker than a regular pull force, but, for the ear application (Figure 1), it only has to act over a much shorter working distance (5-10 cm instead of 30-50 cm). Since magnetic fields, and thus forces, fall off quickly with distance from magnets, this is still an advantage. In the simulations above, the maximum push occurs at approximately 3.6 cm and creates a magnetic field squared gradient of about $1.25 \times 10^{-4} \text{ T}^2/\text{m}$. The same magnet, but pulling at a longer distance of 20 cm, creates just $1.6 \times 10^{-6} \text{ T}^2/\text{m}$ – this will create a force almost a hundred times weaker by equation (3). Pushing across a short distance is much more effective, and requires far less magnet strength, than pulling over a long distance.

In order to position the node at a desired location, the two magnets must be orientated and spaced correctly. This can be done by understanding how the angle of the magnetic field varies away from each magnet, and then tilting and spacing the magnets appropriately to create the magnetic field cancellation at a desired displaced node location. In Figure 5B, the colored contour lines mark all points of constant magnetic field angle around a single magnet. All points along the contour that correspond to the pink curve (bolded) are locations where the magnetic field vector points up and back at an angle of +120 degrees from the horizontal. When the top magnet in the magnetic injector systems is rotated -30 degrees (clockwise), the magnetic field at all these points will be purely vertical. Thus these are all potential node locations (there is a choice of many). The bottom magnet can then be rotated +30 degrees (counter-clockwise) and placed so that its corresponding -120 degree curve intersects the top magnets +120 degree curve at a desired displacement distance. This type of analysis predicts the location of the node point in Figure 3 and can be used to choose magnet placements for desired node displacements (magnets placed closer together and angled further in towards each other create a stronger node closer in).

The magnetic push is achieved by creating a local region of diminished magnetic field strength – there is no need to have the magnetic field cancel exactly to zero. This means that the push system is robust to geometry errors, machining tolerances, and magnetization imperfections. As an example, Figure 6 shows the magnetic field in and around the push

node for the nominal magnet configuration versus two situations where one of the magnets is misaligned out of plane. (Tilting the magnets in plane simply moves the node according to the analysis of Figure 5.) For the 10° degree misalignment, which far exceeds the $< 2^\circ$ degree orientation errors we have in our experimental setup, the push node is still present. It takes an out-of-plane misalignment error of 25° degrees (panel C) to destroy the node. Likewise, the node is not sensitive to small variations in magnet dimensions. We have also verified that the push node is stable for the alternate magnet design shown in panels D, E, and F of Figure 2.

Experimental Results

In order to show the feasibility of pushing magnetizable particles we have performed two experiments. The first verification test used a well dispersed solution of magnetite (Fe_2O_3) nanoparticles (12-15 nm diameter, Product PM001, Liquids Research, Bangor, Gwynedd, UK) in a nano-pure water solution (18 M Ω , Millipore, Direct Q3). This ferro-fluid was poured into a 10 cm diameter Petri dish (Kimax, Fischer Scientific) and filled with nano-pure water. A first simple proof-of-concept magnetic push device, illustrated in Figure 7, was created by hot gluing and taping neodymium rare earth magnets ($2.54 \times 1.27 \times 0.64$ cm, SuperMagnetMan, www.supermagnetman.com), magnetized along their short axis direction (through the 0.64 cm dimension), to wood shims ($\sim 2.54 \times 12.7 \times (0.08-0.64)$ cm) thick, HomeDepot). The measured pole face magnetic field values were ~ 0.29 Tesla. Two of these constructs were then held together by hand (pushing the magnets together and holding them to complete the test prototype device). The device was then placed into close proximity with the Petri dish and images were acquired in a time series (at $t = 0, 5, 10$, and ~ 120 seconds).

The results for this first push system are shown in Figure 9A and B. A distinct void of nanoparticles was observed between the magnet push system and the displaced node that it created – in this region particles were attracted to the two magnets as predicted by the simulations shown in Figure 3. Beyond the node, the particles were visibly pushed out in a ‘front’, i.e. in an increased density or darker area of particles (indicated for improved clarity by the white lines in Figure 9B). With increasing time this front moved away from the magnet push system and spread out, indicating that a push force was created away from the magnets on the solution of nanoparticles in the region beyond the displaced node.

This initial experiment left some ambiguity with respect to the cause of the plume. In review, we postulated that the front could also have been caused by a hydrodynamic flow effect: the fast inward movement of particles in the inner region (between the node and the magnets) may have created fluid eddies in other regions of the Petri dish, and these eddies may have created the outward plume that was observed. To eliminate this possibility, and also to be able to better test different push system designs (different magnet sizes, angles, and separation distances) we created the second experimental test system and measured the pushing of a single drop of ferrofluid.

The second magnetic push system was designed to create a variable (movable) wedge configuration to orient the magnets for testing. As shown in Figure 8, the unit is constructed of three polycarbonate polymer sheets; the first is the base measuring $2.54 \times 26.67 \times 26.67$ cm, and two arms with dimensions $3.18 \times 17.02 \times 18.29$ cm. The arms have a slot to insert a magnet no larger than $2.54 \times 10.16 \times 10.16$ cm, and for magnets with smaller dimensions, a form is utilized to create an epoxy block matching the slot size to securely hold an array of magnet sizes for planned testing. The receiving slots have covers over them that are screwed on with custom brass nuts over a 0.64 cm thick polycarbonate cover. The arms are joined by custom brass hardware and hinges to minimize the magnetic interactions and maintain sufficient strength to keep the magnets in place for testing. Finally, to vary the relative angle

between the magnets (or magnets in epoxy inserts) a custom hinged mechanism was designed and fabricated from brass. The angle is varied by turning a control dial at the end of a threaded rod to draw the tips of the magnets together, or push them apart. Again, the threaded rod and hinges are made from brass; the dial is a stock plastic wheel, made from non-ferrous materials to minimize magnetic interactions and reduce measurement interference. The only ferro-magnetic piece is a pin to hold on the plastic wheel, which is far enough back from the working area of the device that interactions should be minimal. To facilitate a smooth introduction of the two magnets at the tips (which may push apart or attract depending on the configuration under test) a brass slide is also adjoined to one of the arms. It is a custom clamp mechanism with threaded screws to hold the insert in the clamp and maneuver the magnet/epoxy insert towards (or away) from the vertex of the two arms. This allowed for controlled introduction and removal of the second magnet/epoxy insert, once the initial magnet/epoxy insert was in place.

To unambiguously demonstrate pushing of nanoparticles without any fluid flow effects, we tested the ability of the magnets to push away a single droplet of ferrofluid. We used the same neodymium rare earth magnets as before ($2.54 \times 1.27 \times 0.64$ cm, magnetized by ~ 0.29 Tesla across their width) and a commercially available ferrofluid (TargetMag, Chemicell). The ferrofluid contains 8% by volume of 100 nm diameter multi-core particles. Each particle contains a 70-75 nm in diameter starch encapsulated magnetite core that consists of a fused cluster of single-domain crystals. These magnetic particles were chosen for their size and high magnetic susceptibility ($\chi \approx 72$). A 3.8 cm diameter Petri-dish (Kimax, Fisher Scientific) was used to contain the ferrofluid. The Petri dish was filled with a high viscosity mineral oil (Heavy Mineral Oil, Super Tru – USP/Lot 105492), which served as a suspending medium for the droplet (as done in (43)). A single droplet of the ferrofluid (2-3 μ l volume, ~ 1 mm in diameter) was placed into the oil by a microsyringe (Socorex, Wheaton). A ruler (3.2 mm red markings, C-thru Ruler Company) was placed under the dish to enable visual observation of the displacement of the droplet. As expected, the push occurred just beyond the zero magnetic field node.

With this second experimental paradigm, when the two magnets were brought into the correct location near the Petri dish, i.e. with the cancellation node placed behind the droplet, we observed a movement of the droplet to the right away from the two magnets (Figure 9C). To keep pushing as the droplet moved away, the position of the displaced node was adjusted by decreasing the relative distance between the magnets and the edge of the Petri dish to translate the node further out into the oil. The drop moved ~ 4 mm over the time of the experiment (~ 90 s), away from the magnetic device. Without magnets, the droplet did not move. This showed an unambiguous ability to push a magnetically active (superparamagnetic) material a macroscopic distance away from the magnets at a measurable distance from them ~ 17.5 mm from the furthest edge of the magnet system.

There is a good match between the above experiments and our simulations. The location of the node in both experiments matches our predicted node displacement, and the nanoparticle recirculation observed in Figure 9B is consistent with the force directions shown in Figure 3. The forces created by the magnetic push are sufficient to move both 12 – 15 nm particles in solution as well as a ferrofluid droplet composed of 70 – 75 nm particles in oil on a Petri dish. Preliminary animal experiments have recently been performed to verify the magnetic push system *in vivo*. Like the prior pull experiments (17), these push experiments were conducted in rodent models for tissue targeting that emulates that of the human. During these experiments the magnetic push system was held at a distance of 2-3 cm from the eye in rats to mimic the working distance that must be achieved for treatment of the human eye. The results of these *in-vivo* experiments will be reported in future publications.

Conclusion

We have demonstrated, both in simulations and with two experiments, that a simple arrangement of just two magnets can be used to push away or ‘magnetically inject’ nanoparticles. This is useful for a variety of clinical needs where pulling the particles in towards external or internally implanted magnets is either not desirable or is not possible. In particular, it would allow therapeutic magnetic nanoparticles to be pushed in through the round window membrane into the inner ear, thus bypassing the blood-brain barrier. Bypassing this barrier by pulling therapeutic nanoparticles through the RWM has already been demonstrated in guinea pigs. To do the same in humans, to pull in nanoparticles by a magnet placed on the opposite side of the head, will require applying extremely strong magnetic fields across the human head that exceed FDA safety limits. The demonstrated push capability can create magnetic forces of the same magnitude but using safe magnetic fields simply due to the fact that the system can be held much closer to the RWM on the same side of the head as the diseased ear. Based on this first proof-of-concept demonstration, we are now optimizing this system to be hand-held (for clinical ease of use) and to position and maximize the magnetic forces to match the working distances and push strengths necessary to treat human ear and eye diseases.

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Figure 1.

The inner ear is behind the blood brain barrier: micro-circulation vessels that supply blood to the cochlea have tight capillary junctions that prevent therapies from reaching many inner ear diseases. An ability to push magnetizable therapeutic nanoparticles particles through the round window membrane would allow us to bypass the blood-brain barrier and transport therapy directly into the inner ear. The envisioned treatment is shown above, from left to right: the magnetic push system; human ear anatomy; a gel filled with magnetic nanoparticles that has been injected into the middle ear (light blue with black dots, in the tympanic cavity); the round window membrane (black oval); and a magnetic push force (yellow arrow) to deliver therapeutic particles through the RWM into the inner ear (cochlea).

Figure 2.

The key concept: two permanent magnets can push particles away. A) Schematic field lines around a single magnet magnetized along its width. B) Two magnets. We tilt the top magnet down till the magnetic field $\vec{H}$ is along the y axis at the desired node location (green dot). We flip and tilt the other magnet up. C) When these two magnets are correctly overlaid their magnetic fields add to exactly cancel at the node point C (big dot) but they do not cancel around that point (orange annulus) thus forces go outwards from $\vec{H} = 0$ at the node to $\vec{H} \neq 0$ surrounding it (the pink force arrow). An alternate arrangement: D) A single magnet magnetized along its length. E) Here the bottom magnet is both tilted up and its polarity is reversed. This flips the sign of the magnetic field at point B and cancels the horizontal magnetic field at point A for the top magnet (panel F). In both cases, the magnets' orientations and their separation can be varied to position the outward-force region as needed. (Note that the magnetic fields, not magnetic field lines, add together. The gray curves in panel C and F are only meant as a guide for the eye. See also Figure 3.)

Figure 3.

The magnetic field and forces around two magnets magnetized across their width and rotated inward at 30 degrees, matching the experimental configuration shown in Figure 8. A) Magnetic field lines are illustrated by the gray curves, the direction of the magnetic field is shown by the green arrows. B) The strength of the magnetic field is shown by a logarithmic color scale from 10^{-1} Tesla (white) to 10^{-4} Tesla (black, at the node); 20 contour of constant magnetic field strength are also marked. Black arrows show the resulting force directions that will be created on magnetic particles at each location. The maximum push force occurs just beyond the $(x,y) = (0.033, 0)$ cancellation node. This node would have to be placed on the left of the gel shown in Figure 1 to push particles to the right through the round window membrane. Length units are marked in meters.

Figure 4.

The magnetic field strength in a vertical and horizontal slice around our magnet push system. Regions of high magnetic field are shown in white and yellow, regions of smallest magnetic field strength are in black, and the right magnet is shown by a green wire frame (the left magnet is hidden behind the vertical slice). The push node is visible as a small black egg shape that is tight in the horizontal direction and has a larger extent vertically. Forces on the particles go outwards, from black to red. A second node is visible inside the magnet system (the yellow dot with the red center inside the high-field white region). Length units are marked in meters.

Figure 5.

The magnetic field directions around a single permanent magnet magnetized through its width. A) Magnetic field lines (gray) and the direction of the magnetic field (green arrows). B) Each colored contour shows the location of all points that have a magnetic field at that angle. The $+120^\circ$ curve is bolded in pink: the node point for our 30° push system lies along this curve. (The multicolored curve marked 'jump' shows where the angle convention changes from $+180^\circ$ back down to -180° . Green magnetic field arrows are repeated for clarity.)

Figure 6.

The predicted magnetic field strength in front of the two-magnet push system for A) the nominal magnet configuration, B) when the right magnet is rotated 10° degrees out of the horizontal plane, and C) when it is misaligned by 25° degrees. The color and contours corresponds to the log of the magnetic field strength $\log|\vec{H}|$; closed contours around a black region show the push node in the first two panels. The large 25° degrees misalignment of the third panel eliminates the node.

Figure 7.

The first magnetic push device prototype: two magnets attached to wooden strips by glue and black tape. A) Side view of a magnet attached to a single shim. B) Top view of the two angled magnets. C) The system in use against the side of a Petri dish.

Figure 8. The second magnetic push prototype. A) Top and side view schematic. B) View from the back showing the angle adjustment mechanism. C) View from the front. D) Zoomed top view of the front and working region. The configuration and size of the magnets is shown, along with the location of the null point that they create in the Petri dish.

Figure 9.

Experimental validation of the magnetic push system. A) Photograph of the first simple experiment: two magnets held on a wood wedge. B) The series shows a nanopure water solution with initially uniformly dispersed magnetite nanoparticles at the initial time (i) and at times 5 seconds (ii), 10 seconds (iii), and several minutes (iv) after introduction of the magnetic push system on the left. Circles at top of images i) – iv) indicate particles collecting at the dish edge due to the magnets pulling them in behind the node, white lines delineate the plume of nano-particles, and arrows show its pushing motion away from the magnets in front of the node. Series C i)-iii) shows a micro liter drop of ferrofluid being moved away from the magnet system. The Petri dish is resting flat on a table, the view is from the above, and the ends of the magnetic push system can be seen at the left edge of the three images. As the droplet moved to the right, the relative position between the magnets and the drop was kept “constant” to keep the ferrofluid in the maximum push region of the system, i.e. the angle of the system and its distance to the edge of the Petri dish was adjusted to keep the drop moving forward on a relatively straight path. This series took place over a ~90 s interval and the drop was pushed ~ 4 mm from its starting point. The ruler red grid lines are spaced 3.2 mm apart.