DOI: 10.1002/

Article type: Full Paper

Title: Versatile Modular Microfluidic Platforms Enabled by Novel Magnetorheological Elastomer Microactuators

Shi-Yang Tang, Xuchun Zhang, Shuaishuai Sun, Dan Yuan, Qianbin Zhao, Sheng Yan, Lei Deng, Guolin Yun, Jun Zhang, Shiwu Zhang, and Weihua Li**

S.-Y. Tang and X. Zhang made equal contribution to this work

Dr. S.-Y. Tang, Dr. S. Sun, D. Yuan, Q. Zhao, Dr. S. Yan, L. Deng, G. Yun, Prof. W. Li
School of Mechanical, Materials, Mechatronic and Biomedical Engineering, University of
Wollongong, Australia

E-mail: shiyang@uow.edu.au; weihuali@uow.edu.au

X. Zhang, Prof. S. Zhang
Department of Precision Machinery and Instrumentation, University of Science and Technology
of China, China

Prof. J. Zhang
School of Mechanical Engineering, Nanjing University of Science and Technology, China

Keywords: Microfluidics; Microactuators; Magnetorheological elastomer; lab-on-a-chip; modular systems

Abstract: Microfluidic systems enable rapid diagnosis of diseases, biological analysis, drug screening, and high-precision materials synthesis. In spite of these remarkable abilities, conventional microfluidic systems are microfabricated monolithically on a single platform and their operations rely on bulky expensive external equipment. This restricts their applications outside of research laboratories, and prevents development and assembly of truly versatile and complex systems. Here, we present novel magnetorheological elastomer (MRE) microactuators including pumps and mixers using an innovative actuation mechanism without the need of delicate elements such as thin membranes. Modularized elements are realized using such actuators, which can be easily integrated and actuated using a single self-contained driving unit to create a modular, miniaturized, and robust platform. We investigate the performance of the microactuators *via* a series of experiments, and develop a proof-of-concept modular system to demonstrate the viability of the platform for self-contained applications. The presented MRE microactuators are small size, simple, and efficient, offering a great potential to significantly advance the current research on complex microfluidic systems.

1. Introduction

Microfluidic systems allow for precise control over flow variables such as velocity, shear stress, temperature, and concentration of chemicals in microscale, making them very useful for handling small volumes of samples and reagents for various complex analysis and reactions.^[1] However, driving the flows in conventional microfluidic systems normally requires off-chip bulky and expensive equipment such as syringe pumps and centrifuges, which significantly hampers their utility as powerful tools outside of laboratories. To address this limitation, numerous microactuators have been developed and integrated for driving and controlling fluids, enabling the realization of the so-called ‘self-contained’ microfluidic systems for performing a complete on-chip analysis without the need of external equipment.^[1c] Such integrated systems have the potential to enable various assays and diagnostics in underdeveloped regions that lack proper facilities, and may also offer private point-of-care diagnostic opportunities to facilitate health monitoring from home.^[2]

Existing self-contained microfluidic systems often adapt a monolithic design, where several or all parts including actuators, reactors, and sensors are integrated onto a single chip.^[1c,3] This approach requires a rather complicated multi-step fabrication process, and such a common-to-all approach may have to compromise some performance of each component to accomplish the entire device. In addition, once fabricated, the platform can hardly be reconfigured to add additional functionalities, limiting the system to a narrow range of applications. To fully unleash the benefit of self-contained systems, and expand their potential users to such as biological and clinical labs without the access to microfabrication facilities, it is conceivable to incorporate new functionalities to the system by reconfiguring the platform using modularized components with both portability and versatility.

Among self-contained microfluidic systems, one crucial feature is to be able to precisely drive and control sample fluids for later detection and analysis. This need be achieved using micropumps and mixers harnessing various active or passive mechanisms. For pumping liquids, capillary action has been widely used as a passive approach for flow driving within a microchannel.^[4] This method is simple but requires chemical or plasma treatments of microchannels for maintaining the hydrophilicity of the surface.^[4-5] In addition, a phenomenon called surface energy gradient has also been used for passively pumping liquid in microchannels.^[6] However, this method does not allow for continuous pumping of liquid and the flow rate is small. For active systems, mechanical micropumps utilizing piezoelectric,^[7] electrostatic,^[8] pneumatic,^[9] electromagnetic,^[10] and acoustic streaming^[11] effects have been investigated and integrated into microfluidic systems. However, applying these pumps in a self-contained system with a miniaturized size is very challenging as they have rather complicated fabrication processes and still require bulky and complicated external driving systems such as gas tanks, pressure regulators, function generators, power amplifiers, and high voltage converters. Although commercially available motorized pumps have been incorporated in self-contained systems,^[12] they are rather expensive and integration of multiple pumps within a portable size is still not feasible. Alternatively, electrokinetic micropumps driving liquids using electrohydrodynamic,^[13] electroosmotic,^[14] or electrowetting^[15] effects have been demonstrated in microfluidics. However, the performance of these pumps is highly sensitive to the conductivity of liquids, and they produce relatively low flow rates even with high operating voltages. We have previously demonstrated a liquid metal enabled micropump, in which continuous electrowetting (CEW) effect is induced on a droplet of Galinstan liquid metal for pumping the surrounding liquids with a high flow rate without conventional mechanical moving parts.^[16] However, this pump gradually releases gallium ions and can only pump ionic liquids with a pH value larger

than 6.5. Therefore, developing a micropump without bulky peripherals, and is powerful enough to produce high flow rate within microfluidic systems with low power consumption, is essential for many emerging applications in modular and integrated microfluidic systems.

Mixing in microfluidics can be induced by introducing chaotic advections using mechanical, thermal or electrical driving forces.^[17] A passive approach for obtaining mixing is to induce secondary flows by altering the structures or geometries of the flow channels.^[18] On the other hand, active mixing can be achieved using acoustic,^[19] thermal,^[20] CEW,^[21] and electrokinetic^[22] effects. Similar to the problems associated with micropumps, current micromixers often require bulky peripherals to drive them, or have limited liquids that can be mixed. Therefore, there is still a need to develop a micromixer that possesses features of high efficiency with low power consumption, compact in size, easy to operate, and can be modularized to be readily integrated into other components.

Magnetorheological elastomer (MRE) is a composite material consisting of a non-magnetic polymeric matrix with embedded micro-sized ferromagnetic particles, and its mechanical properties can be controlled by the presence of magnetic fields.^[23] MRE offer many advantages over other materials including low cost, readily tunable mechanical properties, inexpensive processing, and can be fabricated in a wider variety of shapes and sizes.^[23-24] Therefore, MRE has been applied in microfluidic devices to enable applications including fluid control, particle manipulations, and forming interconnects.^[24] Currently developed concept of using MRE for fabricating microactuators such as pumps and mixers are based on actuating delicate structures such as thin membranes and cilia,^[25] this requires rather complicated fabrication process, and each actuator requires an independent driving system, making the overall platform complicated and less robust. Most importantly, these pumps can hardly be integrated into a self-contained

system due to the fact that they lack the ability to overcome large pressure drop within microchannels. [25a]

To overcome the listed limitations, here, we investigated novel MRE microactuators including pumps and mixers using unconventional actuation mechanism without the need of thin membranes, for realizing modularized elements that can be easily integrated and powered using a single driving unit to create a modular and robust platform. We investigated the parameters affecting the operation of the proposed microactuators, and a proof-of-concept modular system was established to demonstrate the viability of the platform for self-contained microfluidic applications.

2. Results and Discussion

2.1. Working Mechanisms and Experimental Setup

The MRE was produced by mixing carbonyl iron microparticles (diameter of 2-5 μm) with PDMS (elastomer/curing agent ratio = 20:1) with a weight ratio of 4:1, and the obtained mixture was poured and cured onto a SU-8 patterned silicon wafer mould at 90 $^{\circ}\text{C}$ for 15 min to fabricate the microactuators. This process is simple and straightforward, where the actuation is realized and controlled using magnetic fields. As an example, the cross-sectional schematic shown in **Figure 1A** depicts the working mechanism of a MRE micropump, in which the pump chamber is deformed upon the exposure to an external magnetic field provided by permanent magnets, inducing a temporary collapse of the chamber by bridging the MRE and the glass substrate. The collapsed MRE seals the chamber and the liquid from the inlet can be pumped towards the outlet by moving the magnet along the chamber (Fig. 1A). This process is expected to be repeated for realizing a continuous pump of liquids.

In order to realize such a MRE micropump, we firstly studied the surface morphology of the obtained MRE using a scanning electron microscope (SEM). Interestingly, despite the large weight ratio between the microparticles and PDMS, we observed a relatively smooth surface at the bottom of the chamber, where the microparticles were sparsely distributed and fully encapsulated by PDMS, as shown in Fig. 1B. **Such an encapsulation of iron microparticles at the bottom surface makes the bonding between the MRE actuators and a glass substrate possible after plasma treatment. However, by cutting the sample we observed a much denser distribution of iron microparticles within the MRE above the bottom surface of the chamber (Fig. 1C); this provides a large magnetic field-induced force due to its high iron content. We conducted a control experiment to test the stability of the MRE within a strong magnetic field (~ 400 mT), and**

no migration of iron microparticles was observed upon the exposure to the magnetic field for 48 hours (see Supporting Information S1). In terms of biocompatibility, release of iron microparticles into the surrounding liquid can be minimized due to their encapsulation by PDMS (see Fig. 1B), and the micro-sized iron particles used in this study are much less prone to be uptaken by biological cells compared to nano-sized particles.^[26] However, contamination induced by the release of $\text{Fe}^{2+}/\text{Fe}^{3+}$ ions after a prolonged exposure to media could still occur. This problem can be resolved by coating the surface of MRE with an additional layer of biocompatible material such as polystyrene.^[27] Besides, iron is generally considered as a biocompatible material as human tissues contain a certain amount of iron carried by haemosiderin, ferritin and transferrin.^[28]

For inducing the movement of permanent magnets to actuate the MRE micropump, we designed an innovative self-contained platform to continuously drive the magnets in an orbital manner, as shown in Fig. 1D. Five cylindrical shaped rare-earth permanent magnets (diameter and thickness of 10 and 7 mm, respectively) were placed in the holes on a Polymethylmethacrylate (PMMA) circular holder (diameter and thickness of 50 and 6 mm, respectively). Low-carbon steel sheets (diameter and thickness of 15 and 2 mm, respectively) were attached to the bottom of the holes to hold the magnets. The detailed top-view of the setup is given in Supporting Information S2. The measured magnetic field intensity on the top of the magnet was ~ 350 mT. The magnet holder was rotated by a DC motor (15 V input voltage) equipped with a reduction gear box; and the speed of rotation is controlled using a pulse width modulation (PWM) controller. The entire system was assembled on a PMMA frame to establish a miniaturized and integrated platform (Fig. 1D inset). The actual image for the completed self-contained system is given in Fig. 1E, in which MRE microactuators such as pumps and mixers can be placed and actuated on the top of the system; and most importantly, the microactuators can be readily integrated with other microfluidic

channels, enabling a versatile modular platform for a wide range of applications. The detailed investigations of the MRE microactuators and integrated platform are elaborated in the following sections.

Figure 1. Working mechanism of the MRE microactuators and experimental setup. (A) Schematic of the working mechanism of a MRE micropump. SEM images for the MRE at (B) the bottom surface and (C) within the MRE above the bottom surface. (D) Exploded schematic of the platform for actuating the MRE microactuators, the inset shows the assembled system. (E) Actual image of the integrated and self-contained platform for modular microfluidics.

2.2. MRE Micropump

Utilizing the mechanism and platform elaborated in Section 2.1, we conducted proof-of-concept experiments to show the performance of our developed MRE micropump in open and closed-loop systems. **Figure 2A** shows the continuous pumping of liquid in an open-loop system in response to the rotation of the magnet holder under the glass substrate (thickness of 1 mm) at a speed of 120 RPM (also shown in Movie S1). The MRE pump has a thickness (t_{MRE}) of 2 mm, and has a chamber with the length (l), width (w) and height (h) of 15 mm, 5 mm, and 75 μm , respectively (Fig. 2A inset). To visualize the pumping effect, we connected the inlet and outlet to pipette tubes and added 100 μL of water mixed with blue dye to the inlet. The liquid in the inlet was quickly transferred to the outlet upon the activation of the DC motor, yielding a large flow rate of $\sim 270 \mu\text{L}/\text{min}$. To further demonstrate the capability of our pump, we connected the outlet of the pump to a microfluidic channel (length $\times$ width $\times$ height = $220 \times 0.5 \times 0.05 \text{ mm}$) with a relatively large hydrodynamic resistance of $\sim 4 \times 10^{13} \text{ Pa}\cdot\text{s}/\text{m}^3$, as shown in Fig. 2B. A flow rate of $\sim 10 \mu\text{L}/\text{min}$ was achieved despite the large pressure drop along the microchannel, indicating that the developed MRE micropump is powerful and can be readily adapted by other microfluidic systems. The electrical current measured for driving the motor at a speed of 120 RPM was $\sim 0.3 \text{ A}$, making the overall power consumption of $\sim 4.5 \text{ W}$.

Based on the promising results obtained from the open-loop system, we integrated the MRE micropump to a closed-loop transparent PDMS microchannel (length $\times$ width $\times$ height = $100 \times 1 \times 0.075 \text{ mm}$), as shown in Fig. 2C. The detailed fabrication process is given in Supporting Information S3. To visualize the pumping of liquid, we added 50 μL of blue dye to the reservoir and set the rotating speed of the magnet holder to 120 RPM. Circulation of liquid was observed within the channel and a flow rate of $\sim 110 \mu\text{L}/\text{min}$ was obtained (Fig. 2C, also shown in Movie

S2). The beauty of such a closed-loop system is that different liquids can be continuously added or removed from the reservoirs without disturbing the pumping effect. This is demonstrated by adding additional 25 μL of yellow dye to the reservoir, in which the two dyes were immediately mixed with the assistance of the pump, producing a green-colored liquid and circulating within the channel. Such a closed-loop medium circulating system is promising for the study of organ models in body-on-a-chip applications.^[29]

Figure 2. Pumping liquids using MRE micropumps in open and closed-loop systems. (A) Snapshot images showing the pump of liquid in an open-loop system, the inset shows the bottom

view of the MRE pump. (B) Image showing pump of liquid into a microfluidic channel using the MRE pump. (C) Snapshot images showing pumping liquid using the MRE pump in a closed-loop system. Scale bar is 1 cm.

In our system, the deformation of the pump chamber upon the exposure to a magnetic field can be modeled as the deflection of a clamped rectangular membrane under a uniform load, which the maximum deflection z of the chamber is given as:^[30]

$$z = \alpha \frac{pw^4}{D} \quad (1)$$

where α is the deflection coefficient; p is pressure exerted on the MRE membrane; and D is the constant flexural rigidity, which can be expressed as:

$$D = \frac{Et_{MRE}^3}{12(1-\nu)} \quad (2)$$

where E and ν is the Young's modulus and Poisson's ratio of the MRE, respectively. The pressure p in (1) can be obtained by simplifying the forces exerted on the MRE to a magnetic force experienced between two parallel plates, and eventually z can be calculated as:

$$z = \alpha \frac{6\mu w^4 B^2 (1-\nu)}{\mu_o E t_{MRE}^3} \quad (3)$$

where B is the magnetic flux density, μ and μ_o is the relative permeability and the permeability of space, respectively. We conducted a series of experiments to investigate the parameters that may affect the performance of the MRE micropump. Theoretically, the flow rate Q generated by the pump is determined by parameters including the operating condition of the driving system, geometry of the pump chamber, hydrodynamic load, force generated by the magnetic field, as well as the mechanical properties of MRE, which can all be correlated into the following relation:

$$Q \propto \frac{\omega zh}{R_h} \quad (4)$$

where ω is the angular velocity of the motor, and R_h is the hydrodynamic resistance of the system.

We firstly investigated the pumping performance of water as a function of the operating condition including the rotating speed of the magnet holder ω , the magnetic field strength B , and the number of magnets used. In our study, instead of operating the pump without any load, we connected the MRE pump to a microchannel (length $\times$ width $\times$ height = $30 \times 0.6 \times 0.04$ mm) with a hydrodynamic resistance of $\sim 1.14 \times 10^{13}$ Pa·s/m³ to simulate the real condition when employing such a pump in microfluidic systems. Keeping the size of the pump chamber as a constant ($l \times w \times h = 15 \times 5 \times 0.075$ mm), **Figure 3A** shows that a higher rotating speed leads to a faster travelling of the magnets, and consequently induces a more effective pumping effect, reaching an average flow rate of ~ 65 μ L/min at the speed of 150 RPM. The relatively large errors obtained at higher rotating speeds of 120 and 150 RPM could due to the introduction of air bubbles into the pumping chamber at larger flow rates when conducting media exchange during experiments, causing instability of the pumping performance.

In our system, the ability for inducing deformation of the pump chamber significantly affects the pumping performance. When smaller magnets were used to reduce B from ~ 350 to ~ 150 mT, we observed a significant compromise of the pumping performance due to the decrease of z in equation (3), reducing the flow rate from ~ 65 to ~ 20 μ L/min at 150 RPM. Furthermore, we found that changing the number of magnets on the holder does not affect the pumping performance significantly, as shown in Fig. 3B, indicating that the moving speed of the magnets actually determines the flow rate instead of their number.

Next, we investigated the pumping performance by changing the width w and the height h of the pump chamber. Using a holder with five magnets and operating it at a speed of 120 RPM, at a constant h of 75 μm we found that higher flow rates were obtained after increasing w from 3 to 5 mm, as shown in Fig. 3C. However, further increase the width to 6 mm did not give rise to a better pumping performance (Fig. 3C). Similarly, at a constant w of 5 mm an increase of flow rate was achieved after changing h from 25 to 75 μm , as shown in Fig. 3D. Nonetheless, a significant drop of the pumping performance was observed after further increasing h to 100 μm (Fig. 3D). The observed saturation or decrease of the pumping performance using chambers with larger sizes could be attributed to the fact that gaps always exist close to the side walls of the chamber when inducing the collapse of MRE (Fig. 1A). This generates backflow of liquid and such an effect is more significant when the chamber size is becoming larger, and therefore compromises the pumping performance.

At a constant operating condition (five magnets at 120 RPM, and $l \times w \times h = 15 \times 5 \times 0.075 \text{ mm}$), the pumping performance was further studied by altering the hydrodynamic resistance R_h exerted on the pump. This was conducted by changing the length of the downstream microchannel or the dynamic viscosity of the pumping liquid, as given in Figs. 3E and F, respectively. The obtained pump curve (Fig. 3E) indicates that our MRE pump is capable of overcoming the large pressure drop in microfluidic systems. In addition, by adding glycerol into water to increase the viscosity, Fig. 3F shows that our pump is able to drive a liquid which is up to ~ 46 times more viscous ($\sim 0.0413 \text{ N}\cdot\text{s}/\text{m}^2$ with 75% glycerol) than water ($8.9 \times 10^{-4} \text{ N}\cdot\text{s}/\text{m}^2$) in a microchannel.

Moreover, we also investigated the effects of the mechanical properties of MRE on the pumping performance. We observed that reducing the stiffness of the MRE can result in a larger deformation z under the same magnetic field, and therefore inducing a better pumping

performance. This was achieved by using MRE made from PDMS with an elastomer/curing agent ratio of 20:1 (measured shear stiffness $k_{shear} = \sim 52 \text{ kN}\cdot\text{m/rad}$) instead of the conventionally used ratio of 10:1 ($k_{shear} = \sim 77 \text{ kN}\cdot\text{m/rad}$). Despite the fact that reducing the content of iron microparticles from 80% to 50% could significantly lower the stiffness of MRE ($k_{shear} = \sim 27 \text{ kN}\cdot\text{m/rad}$), the magnetic force exerted on the pump is also decreased due to the less content of iron. This induces a significant drop of flow rate by $\sim 77\%$ compared to the cases observed with 80% iron content. Additionally, the thickness of the MRE pump t_{mre} also affect the pumping performance, we found that the flow rate was decreased by $\sim 17\%$ when a thicker MRE pump of 3 mm was used. It must be noted that although inducing deformation is easier within a thinner MRE pump (say 1 mm), the high density of MRE makes the chamber susceptible to collapse even before introducing magnetic field, and consequently compromise the stability and performance of the pump.

In these sections, we have demonstrated the fundamentals of the operation and the capabilities of our MRE micropump. The pump offers unique advantages including simple fabrication process, small size, low-cost, low power consumption, and is powerful enough to pump various liquids with high flow rates within microfluidic systems. Such a simple pump is ideal to be modularized for integration to enable a complex, yet practical functions. This will be discussed in the following sections.

Figure 3. Characterization of the pumping performance with respect to different parameters. (A) Plot of flow rate vs. magnet holder rotating speed. (B) Plot of flow rate vs. number of magnets on the holder, the insets shows the images of the holders. (C) Plot of flow rate vs. pump chamber width w . (D) Plot of flow rate vs. pump chamber height h . (E) Plot of flow rate vs. hydrodynamic resistance R_h . (F) Plot of flow rate vs. glycerol content.

2.3. MRE Micromixer

After the successful demonstration of using MRE actuators as micropumps, employing the similar principles we further explored their potential to be used as micromixers in a microfluidic system. Using the similar fabrication methods given in Supporting Information S3, a MRE micromixer with two inlets was fabricated, as shown in **Figure 4A**, in which a circular MRE mixing chamber with a diameter and height of 5 mm and 75 μm , respectively, was embedded into a transparent microchannel (length $\times$ width $\times$ height = 20 $\times$ 1 $\times$ 0.075 mm). Fig. 4B depicts the working mechanism of the MRE micromixer, in which the presence of a magnet under the glass substrate is able to generate a force to induce the deformation of the chamber. Oscillation of the MRE can be generated with the periodical presence and absence of magnets underneath, inducing chaotic advections within the chamber and triggering a mixing effect between the incoming flows.

In order to study the performance of the MRE mixer, we connected the inlets of the mixer to a syringe pump to control the flow rate. Water solutions mixed with yellow and blue dyes were injected into the mixer through different inlets. When a flow rate of 20 $\mu\text{L}/\text{min}$ was applied to each inlet before activating the mixer, the flow was purely laminar due to its low Reynolds number ($Re < 2$) and the mixing only happens at the interface between incoming flows, as evidenced by the formation of a clear boundary between the yellow and blue colored solutions along the middle of the microchannel, as shown in Fig. 4C and its inset. When the magnet holder rotated at a speed of 60 RPM to activate the mixer, the two incoming flows started to mix as the boundary between them disappeared. This forms a stable and uniform gradient of color distribution, changing from yellow to green, and eventually to blue across the channel (Fig. 4C, also shown in Movie S3). Such a capability of generating a gradient concentration of chemical

could be very useful for the study of the dynamics of cellular response to a chemical microenvironment.^[31]

A higher mixing efficiency was obtained when we further increased the rotating speed of the holder to 90 and eventually to 150 RPM, as evidenced by the gradually enhanced homogenous distribution of colors across the microchannel (Fig. 4C, also shown in Movie S3). The mixing performance was analyzed using a MATLAB image analyzing code to calculate the intensity of RGB (red, green and blue) across the microchannel at 3 mm downstream of the MRE mixer. Fig. 4D shows the obtained profile of the green color across the channel for the mixer activated with different rotating speeds of the magnet holder. It is clear from the profiles that the two incoming flows formed a relatively linear distribution of concentration across the channel at 60 RPM, and eventually were fully mixed at 150 RPM, demonstrating the versatility of the developed MRE micromixer for realizing multiple functionalities. The developed MRE micromixer is small and simple to fabricate with an excellent versatility, possessing the potential to be modularized and integrated with other systems.

Figure 4. Mixing liquids using MRE micromixer in a microchannel. (A) Top and bottom views of the MRE micromixer integrated into a microchannel. (B) Working mechanism of the MRE micromixer. (C) Mixing of the incoming flows with a holder rotating speed of 0, 60, 90, and 150 RPM. The insets show the magnified images for the downstream of the microchannel. (D) Obtained profiles of the green color intensity across the microchannel.

2.4. A Modular Microfluidic Platform Enabled by the MRE Microactuators

Harnessing the capability and versatility demonstrated for the developed MRE microactuators, we further investigated employing them as individual modules to form a miniaturized and integrated platform with customized functions. As an example, we established an integrated system composed of two micropumps and one micromixer, as shown in **Figure 5A** (see [Supporting Information S4 for the detailed schematic](#)). All the MRE microactuators were assembled on top of the magnet holder and driven by the same sets of magnets at 120 RPM, while the inlets, outlet, pumps and mixers were connected using Teflon tubes. To demonstrate the functionality of the system, we used Pumps 1 and 2 ($l \times w \times h = 15 \times 5 \times 0.075$ mm) to deliver yellow and blue dyes, respectively, from the inlet test tubes to the micromixer, and the mixed solution with a green color was pumped towards the outlet test tube. The inset of Fig. 5A shows the top view of the system without connecting the outlet to the test tube. We clearly observed the two incoming liquids from the pumps were mixed within the mixer, yielding a green colored solution at the outlet (also shown in Movie S4).

A proof-of-concept experiment was performed using such a modularized system for on-chip synthesis of silver nanoparticles with the ability of controlling size distributions. Two solutions were prepared for the on-chip synthesis; Solution 1 was prepared by mixing ammonium

hydroxide (NH_4OH , 10 mM) into silver nitrate (AgNO_3 , 2 mM) solution to form $[\text{Ag}(\text{NH}_3)_2]^+$ complex, and Solution 2 contains sodium hydroxide (NaOH , 4 mM) and D-glucose (10 mM). Solutions 1 and 2 were added to the test tubes and pumped using Pumps 1 and 2, respectively, into the mixer to reduce $[\text{Ag}(\text{NH}_3)_2]^+$ into silver nanoparticles. We demonstrated that the flow rate ratio between Pumps 1 and 2 can be readily adjusted by using different combinations of pumps with different chamber sizes. In this study, by keeping the length ($l = 15$ mm) and width ($w = 5$ mm) of the pump chambers unchanged, we utilized three combinations of pumps with different heights h for silver nanoparticle synthesis, which are listed as below:

Pump combination 1: Pump 1 - $h = 25$ μm ; Pump 2 - $h = 75$ μm ; flow rate ratio = 1:3.

Pump combination 2: Pump 1 - $h = 75$ μm ; Pump 2 - $h = 75$ μm ; flow rate ratio = 1:1.

Pump combination 3: Pump 1 - $h = 75$ μm ; Pump 2 - $h = 25$ μm ; flow rate ratio = 3:1.

The flow rate ratios were calculated from the experimental data given in Fig. 3D. We added 50 μL of 5% polyvinylpyrrolidone (PVP, molecular weight = 10k) into 500 μL of the obtaining nanoparticle suspensions to stabilize the particles. We performed the dynamic light scattering measurements to characterize the size distributions of the nanoparticles. Interestingly, nanoparticles synthesized using the three different pump combinations had very distinctive size distributions, with an average size of ~ 25 nm using *Pump combination 1*, and increased to ~ 75 nm using *Pump combination 3*, as shown in Fig. 5B. The inset of Fig. 5B shows that larger particle size results in a darker yellow color of the suspension, and their sizes were further confirmed using a transmission electron microscope (TEM), as given in Figs. 5C-E. The difference in size distribution could be due to the fact that higher concentration of glucose (pumped by Pump 2) resulted in a higher reaction rate that consequently increased the supersaturation,

higher supersaturation normally results in smaller primary particles and the presence of PVP prevents aggregation.^[32]

In addition to using multiple micropumps for driving liquids in parallel, we also demonstrated pumping liquids in series using cascaded pump configurations, as shown in Supporting Information S5. Furthermore, for applications which require a continuous monitoring of the reactions occur within the microchannels, the developed self-contained system can be readily integrated with a portable microscope, allowing for the real-time analysis and detection on a chip, as shown in Supporting Information S6. In summary, the described modular platform enabled by the MRE microactuators can be readily integrated and reconfigured into a single miniaturized platform to achieve versatile applications.

Figure 5. A modular platform enabled by the developed MRE microactuators. (A) Image for an integrated and reconfigurable platform composed of two MRE micropumps and one MRE micromixer, this inset shows the top view of the platform. (B) Size distributions of the silver nanoparticles synthesized using the modular platform with different pump combinations; the inset shows the image of the obtained silver nanoparticle suspensions with different size distributions. TEM images for the nanoparticles obtained using (C) *Pump combination 1*, (D) *Pump combination 2*, and (E) *Pump combination 3*.

3. Conclusion

We have validated the mechanisms behind of the operation, and demonstrated the capabilities of our MRE microactuators for pumping and mixing liquids in microfluidics. We demonstrated that our MRE microactuators offer unique advantages including simple fabrication process, low cost, and low power consumption, while being capable of achieving high pumping or mixing performance in microchannels with an excellent controllability. Most importantly, the operation of multiple MRE microactuators can be conducted onto a single miniaturized platform driven by a set of permanent magnets, making our system reconfigurable that enables a versatile and multifunctional system simply by choosing different combinations of actuators and microchannels. This facilitates the development of a modular microfluidic platform that has the potential to achieve complex lab-on-a-chip systems for applications such as micro total analysis systems (μ TAS), chemical synthesis, as well as the studies of cellular and molecular biology.

4. Experimental Section

Chemicals, Reagents, and Preparation: Carbonyl iron microparticles (2-5 μm), AgNO_3 powder, NH_4OH solution, NaOH pellets, D-glucose, glycerol, and PVP were purchased from Sigma Aldrich, Australia. PWM speed controller (12 V, 8 A), DC motor (12 V, 50 kg/cm at 55 RPM), Digital Tachometer, and permanent magnets were purchased from Jaycar Electronics, Australia.

Fabrication and Characterization: PDMS based MRE microactuators and PDMS microchannels were fabricated using standard photolithography techniques.^[3] SEM images were obtained using a JEOL JSM-6490LA scanning electron microscope. PMMA frames and magnet holder were fabricated using a CO_2 laser engraving system (Versa Laser System, Model VLS3.50, Universal Laser System, Ltd.). For qualitatively characterizing the stiffness of prepared MREs with different iron particle contents and elastomer/curing agent ratios, MRE disks with a diameter and thickness of 20 and 1 mm, respectively, were prepared and their shear stiffness was measured using a rotational parallel-plates Rheometer (MCR 301, Anton Paar Companies, Germany). A zeta-sizer (Zetasizer Nano ZS, Marvern Instrument, USA) was used to measure the size distribution of the obtained silver nanoparticles. The rotating speed of the magnet holder was measured using a tachometer. Experiments are repeated three times and the data presented as mean $\pm$ standard error. The image processing toolbox of MATLAB software package (MathWorks, 2010a) was used to obtain the profile of flows inside the microchannels.

Supporting Information

Supporting Information is available from the Wiley Online Library.

Acknowledgements

Dr. Shi-Yang Tang is the recipient of the Vice-Chancellor's Postdoctoral Research Fellowship funded by the University of Wollongong. The authors acknowledge use of the facilities and the assistance of Mr. Tony Romeo and Dr. David Mitchell at the UOW Electron Microscopy Centre.

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