Magnetic Composite Polymer Membrane Actuators
with Applications to Microfluidic Devices

by
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Abstract

This thesis demonstrates new materials and microfabrication techniques for integrating membrane-type magnetic composite polymer (M-CP) actuators into microfluidic devices and systems. A membrane actuator with a powerful stroke volume that displaces 7.4 µL of water under a 110mT external magnetic field is developed and demonstrated in a hybrid M-CP/thermoplastic microfluidic device and in an all-PDMS microfluidic device. To achieve injection mouldable M-CP devices, a new M-CP is developed that consists of an injection mouldable off-thiol-ene-epoxy (OSTE+) polymer resin that is embedded with 25 weight-% rare earth magnet particles to be permanently magnetized. To support the rapid prototyping of PDMS and OSTE+ polymer microfluidic devices, a new type of micromould is developed that uses laser ablation of tape to deliver low cost, ultra-rapid moulds. These developments facilitate future commercial mass production through integration with thermoplastic polymers favored by the microfluidics industry in a scalable, “design-to-manufacture” scheme.

Keywords: Magnetic composite polymer; hybrid microfluidic systems; rare-earth magnetic powder; off-thiol-ene-epoxy; rapid soft lithography moulds; magnetic microfluidic actuators
Dedication

This thesis is dedicated to my wonderful family. There were times where you carried me through this degree. Thank you for your kindness, your support, your encouragement, and all the food you sent over.
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I would like to extend my sincerest thanks to my supervisor, Dr. Bonnie Gray, whose direction and encouragement helped me grow into the engineer I am today. I wouldn’t have done my Masters with anyone else.

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This has been a long road, and I’ve had help at every turn from the amazing people in my life. I am beyond grateful.
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<td>LOC</td>
<td>Lab-on-a-chip</td>
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<tr>
<td>M-CP</td>
<td>Magnetic Composite Polymer</td>
</tr>
<tr>
<td>MEMS</td>
<td>Micro-electro-mechanical Systems</td>
</tr>
<tr>
<td>M-OSTE+</td>
<td>Magnetic Off-thiol-ene-epoxy</td>
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<tr>
<td>PCR</td>
<td>Polymerase Chain Reaction</td>
</tr>
<tr>
<td>PDMS</td>
<td>Polydimethylsiloxane</td>
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<tr>
<td>PMMA</td>
<td>Polymethymethacrylate</td>
</tr>
<tr>
<td>OSTE+</td>
<td>Off-thiol-ene-epoxy</td>
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<tr>
<td>µROAMS</td>
<td>Microfluidic, Reconfigurable On-site Analyser for</td>
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Chapter 1. Introduction

Microfluidics is a discipline concerned with the manipulation of small fluid volumes, from microliters to picoliters. The applications of microfluidics systems are cross-disciplinary, from biology [1] to medical diagnostics [2] and wearable devices [3]. Several key principles and philosophies rule: miniaturization, portability, and precise control. These principles offer many advantages for microfluidic systems over traditional macro-scale technology and have resulted in an interdisciplinary field that combines engineering, physics, chemistry, biology, and medicine, delivering devices that support all of these fields. Precise control in microfluidic systems offers, for example, the ability to count cells [4], or to synthesize targeted drug delivery systems on the nanoscale [5]. Miniaturization enables the use of less sample and less reagent for many tests, resulting in less waste and faster results due to lowered processing time. Reduction of scale also enables portability, as smaller systems consume less power and less space than traditional macro-scale systems. Lab-on-a-chip systems (LOCs) leverage these principles to shrink an entire process of a chemical test or biological process onto one integrated chip. For instance, the polymerase chain reaction (PCR) analysis of DNA can be miniaturized for PCR lab-on-chip devices [6].

Microfluidics has its basis in the microelectronic fabrication industry and in materials such as silicon and glass. However, as microfluidic devices have become more widespread, commercial microfluidic microfabrication has moved toward the development of mass produced polymer microfluidic chips. Microfluidic devices made from polymers are low cost options that can be mass fabricated in thermoplastics such as polymethlymethacrylate, polycarbonate, cyclic-olefin co-polymer, and polyethalyne. These thermoplastics are injection moulded, hot embossed [7], or extruded [8].

In academia, the elastomer polydimethysiloxane (PDMS) arguably remains the most common microfluidic material [9]. It is flexible, optically clear, and biologically inert [7]. While PDMS devices are standard in academia, they struggle to transition into commercially viable products due to the relatively high cost of fabrication for volume manufacture [10], [11] and fabrication process control. These limitations prevent many microfluidic devices and systems developed in academic labs from being adopted and implemented commercially. In addition, PDMS can be problematic due to its gas
permeability and adsorption of small molecules such as DNA to its surface. It is not a matter of simply switching from PDMS to thermoplastics; such a material switch often proves to be very difficult and results in a re-design of structures, devices, and systems [11].

These challenges need to be addressed in academic labs, with research dedicated to creating new methods incorporating industry standard thermoplastics and other possible replacement polymers. The gap between industry and academia must be closed from the academia side, to enable the benefits of microfluidics to be fully realized and for translation of research into commercial product.

1.1. Addressing the gap

Because of the microscale properties of fluids, factors such as surface tension, energy dissipation and fluidic resistance play a more important role in the function of a microfluidic system than they would in a macroscale device[12]. Movement of fluid through microfluidic channels is particularly reliant on the specific material properties of the device to work, such as surface roughness and the coefficient of friction. These properties vary from material to material and make the material choice essential. For instance, passive pumping using capillary action is strongly reliant on the surface tension between the fluid and the surface of the channel, which is vastly different depending on the material (glass/PDMS/PMMA) [12]. Active devices such as microactuators (eg. pumps, valves, mixers that use external forces) also rely on specific material properties to move fluid in a device. For instance, a pneumatic micro-valve relies on the flexibility of a membrane to deflect under an applied pneumatic pressure and close a channel. Functional elements such as micro-heaters made from nanocomposite polymers, which are polymers doped with nanoparticles to obtain a functionality such as conductivity or magnetism, provide an additional challenge, as the matrix polymer of the composite polymer cannot be modified without altering the function of the composite polymer. Due to the specific material property requirements, it is essential to consider the manufacturability of the final product during the design of the initial proof of concept device, as a change in the device material may not be possible without a major effect on the device function [11], [13].

In the opinion of many in the industry, transitioning from materials of academia to the materials of commercial production varies from difficult to nearly impossible depending
on the care taken in the design phase to develop devices that consider vital aspects of manufacturability [14], [15]. Once the original designer no longer individually assembles devices by hand, simplicity and repeatability become major factors in a successful design. Some imperatives when designing for volume manufacture, according to Becker [15], are to:

- “Reduce the number of parts by combining functions
- Reduce the number of microstructured parts by reducing fluidic levels as much as possible
- Use a robust design, i.e. avoid vulnerable structure details if possible
- Minimize assembly steps
- Design so it is only possible to assemble the components in an unambiguous way
- Use popular standard/preferred sizes
- Eliminate or simplify alignments/adjustments
- Do not specify tolerances tighter than necessary
- Do not specify surface roughness smoother than necessary”

This “design-to-manufacture” process [16] is essential to consider when aiming for microfluidic system that advances past the proof-of-concept phase and into volume manufacture. Even more consideration must be taken with the integration of nanocomposite polymers into mass produced devices. The functionality of the nanocomposite polymer is strongly reliant on the matrix polymer and any modifications to the composite polymer risk the integrity of the design. Magnetic composite polymers (M-CPs), which are generally PDMS doped with magnetic micro- or nanoparticles, respond to an applied magnetic field with actuation [17]. Because of the Young’s Modulus of PDMS, they are particularly well suited as flexible microactuators. Any replacement/modification of the PDMS matrix polymer could affect the flexible actuation of the polymer and impact the design. A possible route to integration of M-CPs into commercial thermoplastic microfluidic systems is through hybrid bonding [11], in which two unlike materials are bonded together into one microfluidic system. This allows the nanocomposite polymer structures to be integrated into mass produced thermoplastic substrates without modification of the composite polymer.

1.2. Contributions and Objectives

This thesis presents novel designs for magnetic composite polymer (M-CP) actuators and the novel materials and fabrication methods developed to support these
designs. The primary contribution of this thesis is the development of M-CP membrane actuators that can displace large volumes of water, which are designed to be suitable for use as a pump or a valve with minimal change in geometry, in a microfluidic system using water-based reagents. A membrane actuator, realized as a hybrid device with a flexible M-CP actuator layer bonded to a rigid PMMA channel layer, is an important advancement in the integration of PDMS-based composite polymers into thermoplastic microfluidics. A new ultra-rapid fabrication method has been developed for PDMS soft lithography to deliver novel micromoulds, called “sticker moulds”, with favorable surface geometries without the lead-time required by SU-8 micromoulds. This thesis also contributes a new magnetic composite polymer made from rare earth magnetic powder and OSTEMER 324 Flex, an off-thiol-ene-epoxy (OSTE+) polymer resin that is flexible, can be irreversibly bonded to many thermoplastics and has demonstrated the capability to be injection moulded [18].

In summary, this thesis makes the following contributions to the field of microfluidics and microfluidic actuation techniques:

• Design, fabrication, and characterization of an M-CP membrane actuator with a powerful stroke that can displace large volumes of water and has the potential to be used as a pump or a valve without any change in membrane geometry.
• Demonstration of a hybrid M-CP/PDMS/PMMA device that can be batch fabricated, an essential step forward in mass producing M-CP microfluidic devices integrated with commercially-relevant thermoplastic materials.
• Characterization of the physical and functional properties an off-thiol-ene-epoxy (OSTE+) polymer resin doped with rare-earth magnetic powder (Nd$_{0.7}$Ce$_{0.3}$)$_{10.5}$Fe$_{83.9}$B$_{5.6}$, and designs to integrate the magnetic OSTE+ (M-OSTE+) into thermoplastic devices.
• Development of a new “sticker mould” micromould that is ultra-fast and extremely low cost, requiring no UV-photolithography equipment for rapid soft lithography.

1.3. Roadmap

Chapter 1 introduces the research topic, the motivation, and the outline of the thesis. Chapter 2 delineates the background on the materials used, examines the prior art, and gives an overview of fabrication methods used in many of the chapters, including the fabrication methods for magnetic PDMS and details on the adhesive bonding method.
Chapter 3 presents the novel sticker moulds that act as a supporting technology for ultra-rapid PDMS soft lithography. Chapter 4 details the design, fabrication, and characterization of the M-CP membrane actuators. Chapter 5 details the design, fabrication, and initial characterization of test structures made with the M-CP membrane actuators. Chapter 6 presents the newly developed magnetic composite polymer made with rare-earth magnetic powder and off-thiol-ene-epoxy as the matrix polymer. Chapter 7 focuses on future work, presents the design of the M-CP membrane actuator for a normally-closed valve, and includes a summary of some miscellaneous knowledge of polymer fabrication that could be useful to future readers. Chapter 8 wraps up the thesis with a conclusion.
Chapter 2.  Background

This chapter provides the requisite background information to understand the subsequent chapters. This chapter describes the magnetic polymers used to make the actuators in Section 2.1, the adhesive bonding method that bonded the layers of the devices together in Section 2.7, the prior art on M-CP membranes in Section 2.3, and background on the micropumps (in Section 2.4) and microvalves (in Section 2.5) used for microfluidic systems. The motivation behind the fabrication methods developed in this thesis is discussed in Section 2.6, with further context regarding the purpose of the multiuse membrane given in Section 2.7 for a general reconfigurable microfluidic system and in Section 2.8 for the applications. Section 0 describes the adhesive bonding method used in this thesis to bond the unlike PDMS and PMMA substrates and Section 2.10 details the background on OSTE+ polymer resins that are explored as a new matrix polymer for magnetic composite polymers.

2.1. Magnetic Composite Polymers

For most microfluidic systems, fluid must be moved through the device either by passive or active means. Passive microfluidic devices require no external forces to induce fluid flow. Typically, passive devices rely on capillary flow, or channel geometry, to control the flow of fluid in the microfluidic system. Active flow control relies on some sort of external force to be applied to the system, in the form of a pump, valve, or mixer.

Many different actuation techniques are investigated for active microfluidics applications. Pneumatic actuation, electromechanical actuation, magnetic actuation and many others have long histories of use in the field. Magnetic actuation is attractive for many microfluidic systems [17], due to the high-force, high-stroke actuation it offers, but until recent years its applications have been limited by the reliance on difficult-to-integrate bulk permanent magnets and easy-to-delaminate magnetic thin films [19]. Magnetic composite polymers (M-CP) offer the potential for magnetic actuation in a fabrication method that is easy to integrate into an all polymer fabrication flow.

M-CPs are made by embedding magnetic particles into a highly flexible polymer matrix [19]. M-CPs have proven easy to integrate into polymer microfluidic devices [20] and offer the potential for fully flexible and highly responsive magnetic actuators integrated
into electronic-microfluidic systems [21]. Membranes made from M-CP may be used in microfluidics for, e.g., pumps and valves [20]. Many researchers have investigated M-CPs for micro-electro-mechanical systems (MEMS) and microfluidics applications such as M-CP membrane and cilia actuators for devices such as pumps, valves, and mixers[20], [22], [23].

The magnetic particles used in this thesis are the rare earth magnetic powder (Nd$_{0.7}$ Ce$_{0.3}$)$_{10.5}$Fe$_{83.5}$B$_{5.6}$ (MQFP-12-5 magnetic powder from Magnequench International Inc.), which is a hard magnetic material, and result in an M-CP that is permanently-magnetized for bi-directional (attract/repel) strong actuation. In contrast, M-CP membranes fabricated from soft magnetic materials (such as iron powder) only have weak unidirectional actuation (attraction) in differently poled magnetic fields [24]. This means that soft magnetic M-CPs cannot be “pushed” by a magnetic field and will only be “pulled” towards a magnetic field of any polarity, with a weaker response than that of permanently magnetized hard magnetic M-CPs. This advancement from the soft magnetic materials of prior art [25]–[27] to hard magnetic materials enables the high-force, high stroke actuators presented in Chapter 4.

The main matrix polymer used to realize the M-CP is polydimethylsiloxane (PDMS). There are many benefits to PDMS such as its optical clarity, biocompatibility, and low cost. PDMS is ubiquitous in flexible polymer microfluidics [7], [8], [28], [29] and PDMS suspended with magnetic particles integrates very easily with undoped PDMS [21]. A high doping ratio of 80% rare earth iron by weight is used in this thesis and in prior work from the MicrolInstrumentation Lab results in a highly responsive actuator [20], [20], [22], [30].

While PDMS has excellent properties for microfluidic devices, the material is not easily transitioned into mass produced devices, nor are PDMS-based M-CPs (as discussed in Chapter 1). Chapter 6 explores another base polymer that can be used in an injection moulding fabrication scheme. An off-thiol-ene-epoxy polymer (OSTE+), is a polymer resin that Sandstrom et al. [18] have demonstrated can be injection moulded. Their group investigated the OSTEMER 322 [31], which is a stiff polymer resin similar to the OSTEMER 324 Flex [32] that I investigate in this thesis. Chapter 6 of this thesis details the makings of a magnetic composite polymer with OSTEMER 324 as its base, with 25% rare earth iron powder by weight. The possibilities of OSTE+ polymers are vastly underexplored for composite polymer devices and no prior art exists.
2.2. Magnetic actuator dimension considerations

A major relationship to consider when working with M-CP actuators is the inverse relationship between actuator flexibility and magnetic response. As found in previous work by Rahbar and Gray, the magnetic flux density is directly related to the thickness of the M-CP structure [24]. Intuitively, the more magnetic material included in the M-CP device, the stronger the response is to an applied magnetic field. The effect of using hybrid M-CP/PDMS structures is also explored by Rahbar and Gray [24], where they conclude that structures made with all M-CP deflect less than structures of the same geometry made with a PDMS structure and embedded M-CP. This is due to the lower Young’s Modulus of PDMS than the M-CP, which allows the PDMS structures to deflect more.

This has major mechanical considerations for M-CP membrane designs and is used in this thesis to tailor the flexibility of the structures to the needs of the proposed designs. Thick M-CP structures are combined with thin, flexible M-CP membranes, or embedded in PDMS, to obtain favorable magnetic and mechanical behaviors.

In particular, the membrane actuator presented in Chapter 4 combines a thin M-CP membrane with a thick central boss. This design shows a strong deflection under an applied magnetic field due to the large amount of magnetic material contained in the central boss, yet it is flexible due to the thin M-CP membrane. There was some consideration given to making the flexible membrane out of PDMS rather than M-CP as PDMS is more flexible, with an experimentally determined Young’s modulus of approximately 0.522, compared to that of M-CP at 1.623 [30]. Initial experimental observations indicated that the PDMS membranes with an M-CP boss deflected too strongly under an applied magnetic field. Their long stroke could be a disadvantage in pumping or mixing applications where a higher frequency oscillation is required. The higher flexibility also meant that they were too susceptible to fluid pressure and would deform under a moderate fluid pressure, rendering them slower and less responsive. The M-CP is still highly elastic, with an elastic range of up to 1.6MPa [30], and the membranes operate in the kPa pressure range, meaning that the M-CP membranes should still be reliable for many actuator applications.

The dimensions of the membrane chosen for this work are based on the experimental results of Rahbar and Gray [24] on the relationship between the deflection
of an M-CP membrane with varying thickness and constant width, and based on the experimental results of Hilbich et al. [33] on the deflection of an M-CP membrane of varying width. Both of these works concern square membranes; the work of this thesis is done with a circular membrane. Further illumination on the behaviour of a circular membrane can be found using Kirchhoff’s solution for circular plates. Wang et al. [34] present the solution for the deflection \( w_0 \) of a clamped circular plate under an axisymmetric linearly varying load as:

\[
w_0 = \frac{q_0 R^4}{14400D} \left[ 129 - 290 \left( \frac{r}{R} \right)^2 + 225 \left( \frac{r}{R} \right)^4 - 64 \left( \frac{r}{R} \right)^5 \right] \tag{1}
\]

where \( q_0 \) is the load at the centre of the plate, \( R \) is the radius from the centre of the plate to the support, \( r \) is the radius of the plate, and \( D \) is the flexural rigidity. The flexural rigidity is defined as:

\[
D = \frac{Eh^3}{12(1-v^2)} \tag{2}
\]

where \( E \) is the Young’s modulus, \( h \) is the height of the plate, and \( v \) is the Poisson’s ratio.

2.3. M-CP Membranes

When a new magnetic composite polymer is developed, a common way to demonstrate its potential as a microfluidic actuator is to fabricate it as a membrane and then to characterize the deflection of the membrane [25], [26], [33], [35], [36]. The deflection of the membrane under an applied magnetic field shows that the membrane moves and characterizes the magnetic field it requires.

The M-CP developed in the Microinstrumentation Lab (rare earth magnetic powder \((\text{Nd}_{0.7} \text{Ce}_{0.3})_{10.5}\text{Fe}_{83.9}\text{B}_{5.6}\) suspended in PDMS at 75-80% with respect to weight) has been characterized as a membrane by Hilbich et al. [33]. They fabricated several different M-CP membrane geometries, with varying central magnet geometry and found that 3 to 8 mm square membranes with square central magnetic features ranging from 0.8 to 3.6 mm are able to deflect 100 micrometers or more in magnetic fields ranging from 52 to 6.2 mT [33]. A 100 micrometer deflection was used as a benchmark for significant deflection and
an indication that their membrane could be useful as an actuator. In their work, there is no characterization of the membrane with liquid (i.e. deionized water) and an actuator design such as a pump or a valve was not proposed. So while the membrane showed strong deflection under air conditions, there was no verification of how the membrane would function as an actuator that moves water.

The work in this thesis builds on the membranes presented by Hilbich et al. [33] by characterizing the displacement of deionized water by a similar membrane actuator (Chapter 4) and presenting designs and initial characterization for pump and valve designs (Chapter 5). During the design of the membrane actuator presented in this thesis (Chapter 4), significant consideration was given to the relationship established between the dimensions of the membranes and the magnetic fields that deflect them further than 100 microns [33] and is discussed in Section 4.1.

2.4. Micropumps

Micropumps are classified by their pumping principle and can be broadly classified as mechanical pumps, or non-mechanical pumps [12]. Non-mechanical or passive pumps rely on microscale properties such as capillary action to move fluid [37]. Mechanical pumps are further divided based on whether mechanical energy is added continuously or intermittently to the device for its operation [12]. Systems in which mechanical energy is continuously added are called dynamic systems and encompass, for example, ultrasonic and centrifugal pumps [37]. Displacement pumps involve intermittent application of mechanical energy to move different parts of the device. This category contains check-valve pumps, peristaltic pumps [38], valveless rectification pumps [39], and rotary pumps [12]. The actuators in this thesis are primarily designed for displacement pumps.

Check-valve pumps and valveless rectification pumps rely on the motion of a flexible membrane that is deflected by an actuator to create a stroke volume. Check-valve pumps have one way valves that only allow fluid to flow in one direction and prevent fluid from flowing until a critical pressure is reached [12]. Valveless rectification pumps rely on the geometry of a nozzle and diffuser (or valvular conduit), where the pressure loss coefficients favor flow in the forward direction [12]. Peristaltic pumps rely on the timed actuation of multiple membranes in series [12], [38], [40].
As mechanical displacement pumps typically rely on an actuator deflecting a membrane to create a stroke volume, a strong actuator response, with a flexible pump membrane is essential to an effective pump. Magnetic actuation can offer relatively large actuation forces, with high levels of control, and can act on nearby materials without needing direct contact [12][19][20]. Magnetic actuation in current art is achieved mainly through thin magnetic films attached to polymers [19], [17] and bulk permanent magnets embedded into polymers [12], [19], [41]. Magnetic actuators based on thin films may lack magnetic material and therefore have a weak magnetic response. Magnetic actuators featuring bulk permanent magnets suffer from poor adhesion of the magnetic element which can cause delamination or cracking from the polymer substrate during fabrication or use [19].

In contrast, magnetic composite polymers (M-CP) are an excellent alternative for flexible magnetic actuators. M-CPs are made by embedding magnetic particles into a highly flexible polymer matrix [19] (as discussed in Section 2.1) and M-CP elements integrate easily into polymer microfluidic devices [20]. An M-CP membrane would provide a high-force, high-stroke actuation with good membrane flexibility in a micropump.

An extensive number of papers have been published that demonstrate M-CP membranes and their deflections. These papers list micropumps as their proposed application, but generally have no pumping schemes characterized [25], [33], [35], [42]. There are few papers published that feature a fully characterized polymer mechanical displacement micropump made with an M-CP membrane [23]. For example, check valve pumps are frequently mentioned in M-CP membrane papers with no fluidic characterization. For instance, in their conference paper Feldmann et al. [43] presented an M-CP micropump working with check valves, but list no flow rate and the authors published no further papers on the subject. Because the check valves can be presented as external to the proposed design, authors can demonstrate the deflection of their M-CP membrane without going into detail about their pumping scheme. Fluidic testing can show that a simple membrane may not work as a pump once fluidic considerations such as the critical pressure, dead volume and stroke volume are taken into account. For instance, Santra, Holloway and Baitich [41] present a check valve pump with a permanent magnet embedded in the membrane. While this seems like a simple way to turn a membrane into a pump, in practice the pump has a low stroke and is unable to pump water (only air) due to bubbles forming in the dead volumes [41].
Diffusion pumps (a subset of valveless rectification pumps) feature a central moving membrane, like check valve pumps, but have a simpler geometry because they have nozzle and diffuser rather than moving check valves. Their actuation scheme is also simple, as their magnetic field does not need to move, only alternate. However, due to the lack of valves, each stroke of the membrane moves fluid in both directions, with a net motion pulling into the pump chamber through the nozzle and out through the diffuser.

![Diagram of diffusion pump](image)

**Figure 1**: Function of a diffusion pump over the stroke of the membrane adapted from [44].

Amirouche and Zhou [39] present a simple mechanical diffusion pump based on the vibration of a thin bulk NdFeB magnet embedded into a PDMS membrane. Maximum pumping rates of 320 µL/min were observed, with a max back pressure of 950 Pa. The planar design of the pump allows for easy integration with other PDMS-based microfluidic systems.

Said et al. [23] demonstrate a fully characterized M-CP based diffusion pump, but a major detraction of this work is that the “magnetic particles” used by the group are NdFeB magnets that have been crushed up until the majority of the grain sizes are 50-100 µm [36]. With an extremely low max flow rate of 6.5 nL/min, the pump is intended for applications that require small, controlled fluid delivery such as continuous drug delivery. The deflection of their 3x3 mm membrane is only 12.87 µm, compared to 100 µm deflection obtained from NdFeB M-CP membranes of similar dimensions [33].

### 2.5. Valves

There are two categories of microvalve: passive and active [45]. Active microvalves use external actuation to move between the closed and open valve states. Common active microvalves are pneumatic, thermopneumatic, piezoelectric, electrostatic, or electromagnetic microvalves [12], [45]. Each actuation method has unique advantages and disadvantages. Pneumatic valves tend to be able to withstand higher pressures, but
have slow response times; electromagnetic valves have a lower pressure range, but a faster response time [12]. In a magnetic valve, a permanent magnetic can also apply the external magnetic field required to change the valve’s state. Actuation via permanent magnet provides the benefits of large actuation forces, low power consumption, and high levels of control. An M-CP microvalve actuated by an external permanent magnet provides the additional benefit of a flexible, bidirectional microvalve that can be actuated with relatively low magnetic fields.

Passive microvalves control flow due to their structure, with no external actuation. An example of a passive microvalve is the burst valve, which prevents fluid flow until one side of the valve exceeds a certain pressure threshold and bursts to allow fluid flow [46]. A common valve used in microfluidic systems is the check valve. A check valve is a one-way valve that only allows fluid to flow in one direction and in a microfluidic system is frequently used in combination with an oscillating membrane to make a check-valve pump [41], [45] (shown in Figure 2). There are several common geometries of check valves, such as ball check valves or diaphragm check valves, but the general principle of check valves is that there is a mobile part that blocks flow of fluid in the channel under specific fluid conditions. Typically, this fluid condition is the direction of flow, within a certain pressure range. Above this pressure range, the valve may crack, and below this pressure range, the valve may not seal and prevent flow. A diaphragm check valve is normally closed, but under the correct pressure differential, the valve opens and allows flow as shown in Figure 2.
The membrane actuator presented in Chapter 4 is used in Chapter 5 to create a simple, easy to fabricate normally-closed valve. The M-CP membrane at rest is sealed against a valve seat (normally-closed) and under the correct pressure differential the membrane deflects open to open the channel, in a similar method to the check valve shown in Figure 2. Since the valve is made from M-CP, the valve can be deflected open by an applied magnetic field, adding additional control to the system. This valve design combines the geometry of passive valves with active magnetic actuation to make an active, normally-closed valve that can be opened against its passive geometry.

There are few published M-CP valves made from hard magnetic materials. A flap type valve has been presented by Rahbar et al. [20] using the same rare earth magnetic composite polymer described in Section 2.1. The flap valve was highly successful, able to withstand high pressures and route fluid effectively. The flap valve was difficult to fabricate and involved aligning several layers by hand, leaving much room for improvement on simple and scalable fabrication procedures. Bute et al. [47] also present a flap style valve that is permanently magnetized and designed for flow switching operations. As with micropumps (described in Section 2.4), magnetic membrane valves made from soft magnetic composite polymers are rare as the magnetic response of soft magnetic materials is weak and requires a high applied magnetic field [48]. Elastomeric membranes
(such as PDMS) with embedded bulk magnets are much more common [49], but suffer from adhesion issues, with the bulk magnet being prone to delamination and complicated fabrication/integration [19].

2.6. Fabrication ideology

Many M-CP actuators, such as the valve presented by Rahbar et al. [20], display excellent function, but need to be assembled by hand and cannot be made in large batches. Without a scalable fabrication procedure, these devices will not transition from academia to mass production.

Mass produced thermoplastic microfluidic devices are dominated by fabrication methods such as injection moulding and hot embossing. These fabrication methods leverage the glass transition temperature of thermoplastics to rapidly make a large number of devices. PDMS and other thermoset elastomers are not suitable for these methods of fabrication and must be fabricated separately and integrated into the workflow of thermoplastic manufacture after the thermal treatment. A scheme in which all of thermoplastic fluidic channels of a microfluidic system can be integrated with all the layers of PDMS and PDMS-based actuators with one simple bond is desirable. As discussed in Section 1.1, it is imperative to reduce the number of parts in the system and to reduce the complexity of these structures and their assembly process such that the device design is robust and unambiguous.

The work of this thesis endeavours to design with manufacturability in mind in four ways:

- Reduce the number of parts by demonstrating a multiuse membrane actuator that can be used as a pump, valve, and mixer. Further discussion of the advantages of multiuse membranes is in Section 2.7.

- The actuators are simple, planar membranes embedded in thick supporting polymer that are highly robust even when batch fabricated.

- A hybrid layer based system is used, in which the PMMA structures and the PDMS-based structures are fabricated separately and bonded together in a final step. This reduces the amount of PDMS used, which does not
have an excellent commercial reputation (and is only widely used as a commercial product by Fluidigm [10], [50]), while still enabling its use as a flexible membrane actuator in the thermoplastic microfluidic system.

- There is one interface between the PMMA and the PDMS, and an array of thermoplastic structures can be easily aligned and bonded to a corresponding array of PDMS devices with no seams between the subsystems of the device.

PDMS and the thermoplastic used in this thesis, PMMA, do not readily bond to each other due to mismatched surface properties. Several bonding methods are discussed in Section 2.7. Adhesive bonding through thin transfer adhesives is the most scalable bonding method of the methods presented and is the chosen bonding method for the multilayer hybrid devices demonstrated in this thesis. As the adhesive can be applied to the thermoplastic, then ablated into the exact shape needed, there is minimal alignment needed. The two surfaces are bonded instantly and large arrays can be bonded layer by layer without seams (which tend to leak).

2.7. Multiuse membrane and reconfigurable structure

A multiuse M-CP membrane that can be used for pumps and valves is advantageous because it reduces the number of different parts to be fabricated. Since the M-CP is PDMS-based, it also reduces the number of parts made out of PDMS, which will likely the limiting factor of any device fabrication flow since the PDMS fabrication process is more complex than injection molding for volume manufacture.

An array of M-CP membranes fabricated into a single layer using a PDMS support structure (shown in Figure 3) is an excellent way to simplify the assembly of a microfluidic device. An actuator layer that isolates the required M-CP components into one layer of a multichannel system is easier to align and bond to a thermoplastic substrate than a device that requires each actuator to be individually aligned and bonded to the device. Bonding a PMMA substrate and a single sheet of M-CP membranes embedded in a PDMS layer is an excellent way to prevent leakage, as there are no seams between the M-CP/PDMS layer and the thermoplastic substrate.
A multiuse membrane fabricated into an array has another advantage: it is general. This allows an array of M-CP membranes to be fabricated as one layer and adhered as one sheet to a thermoplastic substrate, without significant consideration of the channel geometry of the thermoplastic substrate. Then the design of the system can be modified by altering the thermoplastic channel layer without needing to consider modifying the fabrication of the M-CP actuator layer. For instance, an array of M-CP membrane actuators could all be used as mixers, or all as pumps, etc. without any changes to the M-CP array by changing the channel design of the thermoplastic substrate. In a reconfigurable device, such as the µROAMS project (discussed in Section 2.8), the multiuse design enables the entire system to be reconfigured without changing the M-CP actuator layer. In this way, a more complex system can be obtained without an increase in fabrication complexity.

2.8. Applications

The microfluidic designs presented in this thesis fit into a larger project: the Microfluidic, Reconfigurable On-site Analyser for Multiplexed Samples, or µROAMS project. Arrays of micro-actuators such as valves and mixers are used in the conceptual device (shown in Figure 4) to route fluid through the platform, mix reagents, and perform multiple tests on samples. Micropumps could be integrated into the device to provide on-chip active pumping to aid in developing a high throughput microfluidic platform that is still small and
portable. The μROAMS project is designed for portable water testing, and water-based reagents. The microfluidic components presented in this thesis are also primarily suited route to water-based fluids, and are tested using water. This limitation is due to the limitations of the physical properties of PDMS, which has poor solvent and acid/base resistance [9].

Figure 4: Conceptual drawing of multifunctional microfluidic platform (this image is reprinted with permission from Strategic Grant 396755-10, “An on-site, reconfigurable, multi-sample microfluidic platform for rapid parallel sample-manipulation”, B.L. Gray (PI) and L. Shannon).

Development of arrayable microactutors such as those described in Chapter 5 could be used in a microfluidic systems such as the μROAMS project. The microactuators presented in this thesis could also be used in other Lab-on-a-chip (LOC) systems, such as for polymerase chain reaction (PCR), as both PDMS and PMMA are biocompatible, although surface contamination issues could be problematic.

Ideally, the components of the system could be mass fabricated at a low cost such that the final system could be disposable, or partially disposable. PDMS can be purchased for approximately $100/kg [51] and PMMA can be purchased for approximately $1-5/kg [9]. The rare earth magnetic powder (MQFP-12-5 magnetic powder from Magnequench International Inc.) was supplied to the Microinstrumentation Lab for free, but for a mass fabricated application would have to be purchased, meaning that the M-CP components
of the mass produced system would be the costliest element, and should be reduced in a disposable device scheme.

A disposable device would avoid the issue of cleaning the device for reuse, which requires developing an cleaning procedure for the device and is not trivial. If the contaminated components could be disposed of and the rest of the system (such as the electronics) could be reused, then the risk of contaminating samples will be minimized, while also reducing the waste generated by the system. Safe disposal or recycling of the system should also be considered.

2.9. Adhesive Bonding of PDMS and PMMA

Polydimethylsiloxane (PDMS) and Polymethylmethacrylate (PMMA) are two materials commonly used in microfluidics, but the two are rarely used together. PDMS and PMMA have different surface properties, which inhibits bonding and the creation of hybrid devices. There are many methods to bond the two together, including surface modification with silanes [52]–[54], sol-gel bonding [55], and adhesive bonding [56]. Channel deformation, persistence of hydrophobicity, and channel blockages are important considerations when choosing a bonding method for a microfluidic device.

My bachelor thesis investigated bonding these two materials and discussed the advantages and disadvantages of several prominent bonding methods. I concluded that the surface properties of PMMA and PDMS are incompatible and the best way to bond them together is using adhesive bonding with transfer adhesives [57].

In particular, the transfer adhesive ARclear 8026 is a 25 µm thick unsupported transfer adhesive [58]. The adhesive comes as a thin sheet between two supporting liners. The ARclear 8026 can be cut to the correct size before the film is removed from the liners. For this application, the transfer adhesive was cut exactly to size using a laser cutter. In previous tests, this adhesive was found to bond PDMS to PMMA and withstand pressures up to 2.76kPa [57]. This method is highly effective to bond the layers together at low pressures, as it is quick, specific, does not block any channels, loses minimal efficacy when water is applied, and does not affect the surface of either device.
2.10. OSTE+ polymers

An off-stoichiometry thiol-ene-epoxy polymer (OSTE+) is a dual-cure ternary monomer system containing epoxy. Figure 5 shows a visualization of the prepolymer mix of thiol, allyl (which are the “ene” groups), and epoxy monomers.

Figure 5: OSTE+ prepolymer composition. Modified with permission from Haraldsson et al. [59].

To begin the curing process, the prepolymer is exposed to UV light. The exposure time depends on the thickness of the prepolymer and the resulting polymer is a flexible, soft, solid elastomeric-like material. The OSTE+ polymer is, at this stage, partially polymerized and has a reactive surface that is ideal for surface modification or bonding. The epoxy groups in the polymer allow the polymer to bond to a wide range of surfaces [60]. All of the allyl groups bond with a fraction of the thiol groups, leaving a portion of unreacted thiol groups [60], shown in Figure 6.
Figure 6: OSTE+ polymer after the first cure. Modified with permission from Haraldsson et al. [59].

The remaining thiol groups react with the epoxy groups in the second cure, which may be accelerated with either UV light, heat, or both. The reaction will complete without acceleration by UV light or heat, but will take several days [18]. Figure 7 shows the polymer matrix at the end of the second cure. After the reaction, the OSTE+ polymer is an optically clear polymer, with a Young’s Modulus that is anywhere from 2.5 MPa to 3 GPa, depending on the exact ratios of the thiol, allyl, and epoxy [59].

Figure 7: OSTE+ polymer after the second cure. Modified with permission from Haraldsson et al. [59].

Several different formulations of the OSTE+ polymer are available from the company OSTEMR, which manufactures these polymers. The Young’s Modulus of the OSTE+ polymers cannot be tuned in the purchased polymer, as changing the Young’s Modulus involves changing many of the functional groups in the prepolymer [61].
A simple microfluidic device with OSTE+ polymers of varying stiffness has been demonstrated. “Rubbery” OSTE+ membranes are shown by Hansson et al. [61] to easily integrate with OSTE+ polymers with a high Young’s modulus to form a nearly homogenous device with hybrid stiffness. The “rubbery” and “stiff” OSTE+ polymers are roughly analogous to the commercial products “OSTEMER 324 Flex” [32] and “OSTEMER 322 Crystal Clear” [31] from Mercene Labs AB. This demonstration is essential, because of the difficulties to bonding together layers of a microfluidic device.

The OSTEMER 322 Crystal Clear has a scalable fabrication scheme that allows it to be injection moulded [18]. As the OSTEMER 324 Flex has similar properties as a prepolymer, this fabrication scheme is also applicable to the flexible OSTE+ and could be relevant to OSTE+ doped with magnetic particles (M-OSTE+).
Chapter 3. **Sticker moulds for rapid soft lithography**

Soft lithography, using SU-8 moulds, is a wide-spread and established way to rapidly prototype microfluidic devices. However, SU-8 moulds have some downsides in terms of being somewhat costly, time consuming and with major design limitations on their thickness. Furthermore, SU-8 does not always make a good mould for composite materials due to mould release issues [62]. It is also essential to have moulds capable of large features with a high thickness for magnetic polymers: the more magnetic material used and the thicker the magnetic structure, the more responsive the magnetic actuator will be to an applied magnetic field [24]. The ability to make certain features thicker than others allows for more magnetic material to be included in the device, without major changes in the geometry of the design. While SU-8 is capable of high aspect ratio moulds when the dimensions are small, it is generally limited to ~500 µm of thickness using photolithography [63] and then must have long (up to 3 min) exposure times and must have careful temperature control during baking to prevent cracking of the thick SU-8 [64]. The diaphragms designed in this thesis have central magnetic features of 1mm thickness, which is not optimal for SU-8. While some researchers have demonstrated SU-8 designs that are 0.5 mm to 2 mm thick [64], [65], the casting method used to obtain thick SU-8 has issues with inconsistent thickness and persistent air bubbles which distort the structures [65]. The design process can also be quite lengthy using SU-8: each re-design requires a new mask to be printed. Each new design also requires more SU-8, which is a relatively costly polymer and requires a mask aligner and wafer spinner (which are usually only found in a cleanroom facility) to be processed correctly. The cost of obtaining the necessary equipment and materials can be a barrier to prototyping microfluidics for many in academia and industry.

As an alternative, we investigate inexpensive PMMA moulds. Moulds made from PMMA have been extensively explored by our lab [62]. PMMA moulds can be made using laser ablation and are capable of large features as well as features as small as 100µm, with an aspect ratio of seven [66]. While not sufficient for fine resolution features, it makes an excellent mould for M-CP structures, since it can make deep features which allows more magnetic material which enables a stronger magnetic response. PMMA moulds improve upon SU-8 in terms of cost, mould depth, and fabrication speed as a new mould
can be made in less than an hour. However, PMMA moulds are weak in smallest feature size, channel geometry (they have a kerf which is V-shape), and surface roughness [62]. Because the moulds are ablated by a laser, the surface is quite rough. A rough mould surface results in a rough surface on the PDMS device made by the mould. A smooth surface is essential for PDMS to PDMS bonding (via corona treatment) and for adhesive bonding using transfer adhesives, as the adhesive layer is very thin and thus unable to fill in the rough surface during bonding to make a fluid-tight seal.

Treatments with various solvents such as acetone can improve the surface roughness [67] of ablated PMMA, but the solvent treatment can also deform the features of the mould. Generally, PMMA moulds are best suited to making the active components of the microfluidic device, where surface smoothness is less essential, or for making the components where the laser ablation does not interfere with any bonding points on the device. Due to the surface roughness caused by the laser ablation, PMMA moulds are not a good choice to fabricate the channel layers of the device, as the roughness is passed on to the surface of the PDMS.

This leaves a gap for a microfluidic mould that is fast to fabricate, low cost, and has good surface characteristics. The sticker mould described in this chapter is low cost, takes less than 10 minutes to fabricate, and is able to consistently mould channels as small as 150 µm wide and 50-200µm deep, with excellent surface properties. Section 3.1 describes the design of the sticker moulds, Section 3.2 details the fabrication procedure of sticker moulds, Section 3.3 characterizes the PDMS structures that can be obtained using sticker moulds, and Section 3.4 discusses the possibilities and limitations of the sticker mould method.

The sticker moulding method characterized in this chapter is also used in Section 5.2 to fabricate the normally closed valves based on the M-CP membrane actuator.

3.1. Design

The sticker mould is based on a simple idea: what if you could cut out a shape and turn that into microfluidic channel? The sticker mould designed for ultra-rapid prototyping, is comprised of a glass substrate with layers of tape attached that have been patterned by a laser cutter. Uline Industrial Tape S-423 is used in this thesis, although initial
investigation shows that PET films should also be effective. Initially, the sticker moulds were made by cutting out the desired shape, making a “sticker”, then applying that sticker to a substrate such as glass or PMMA. This has the obstacle of aligning all the pieces by hand, which is tedious and inexact.

The sticker moulds are made by applying layers of tape to a glass substrate, patterning the tape with a laser cutter, and then removing the excess tape. A substrate such as glass is essential; a CO₂ laser is used to cut the tape and CO₂ does not ablate glass. This leaves the substrate unpatterned by the laser so that its surface is still smooth. It also has the benefit of requiring no alignment to be done by hand, so that each mould is consistent with the original design. Different mould thicknesses can be achieved by applying multiple layers of tape and multi-height moulds are possible with this method. The tape used in this thesis is 51 µm thick, which was the thickness that was readily available. Thus, mould thicknesses can be fabricated that increase in increments of 51 µm (51µm, 102µm, 153µm, etc.).

3.2. Procedure

This section describes the steps of making a sticker mould. Both single and multi-thickness moulds can be realized using this method and are described in Section 3.2.1 and Section 3.2.2, respectively.

3.2.1. Single thickness moulds

Figure 8 shows the fabrication flow of a single-thickness sticker mould. The first step is to begin with a clean glass substrate (Figure 8a). Glass microscope slides are a low cost option and make easy disposable moulds. Next, a layer of 51 µm thick Uline Industrial Tape S-423 is applied to the glass substrate and the excess removed (Figure 8b). The tape is smoothed to remove as many bubbles as possible. Layers of tape are added and then smoothed until the desired thickness of the mould is reached (Figure 8c).

The laser cutter must be calibrated such that the laser cuts only the tape and not the glass. This can be determined by inspection of the glass substrate under the microscope. If there is any cracking or damage to the glass, the laser setting is too powerful. If the tape doesn’t cut away cleanly, the laser setting is too weak. For example,
on a mould with two layers of tape (102 um depth), a Universal 360 CO₂ laser cutter set at 100% speed and 15% power was used to cut the tape and leave the glass slide undamaged.

Once the laser is calibrated, it can be used to cut out the pattern of a mould (Figure 8d). Peel away the excess tape, leaving only the desired pattern (Figure 8e). The sticker mould can then immediately be used for micromoulding, with no need for cleaning or annealing.

![Figure 8: Process flow diagram of sticker moulds: a) Glass substrate. b) Application of tape to glass substrate. c) Additional layers of tape can be added and smoothed until the desired thickness is reached. d) Laser ablation of the pattern. e) Removal of excess tape.]

3.2.2. Multi-thickness moulds

Multi-thickness moulds are made in a similar way to the single thickness moulds, with slightly different laser calibration. In these moulds, the laser should be calibrated such that it only cuts one layer of tape. In this way the layers of tape can be cut to different patterns without affecting the layers beneath.
The process begins with a clean glass substrate (Figure 9a) and applying 51 µm thick Uline Industrial Tape S-423 to the glass substrate (Figure 9b), then laser cutting a pattern into that layer of tape (Figure 9c). The patterned tape is left on the substrate then an additional layer of tape is applied, leaving some excess tape on one edge (Figure 9d). The excess tape will help with removing the tape one layer at a time in later steps. The top layer of tape is then ablated with the laser without ablating the layer below (Figure 9e). On a Universal 360 CO₂ Laser Cutter the settings of 100% speed and 10% power can cut a single layer of 51 µm thick Uline Industrial Tape S-423 without ablating the layer below. Layers of tape and patterning it are repeated as needed to obtain the final structure. Once the tape layers have all been patterned, the excess tape is peeled off layer by layer (Figure 9f) until all the excess tape has been removed and the multilayer pattern is revealed (Figure 9g). The sticker mould can then immediately be used for micromoulding processes.
Figure 9: Process flow of multi-thickness sticker moulds: a) Glass substrate; b) Application of tape to glass substrate; c) Laser ablation of the pattern for that layer of tape; d) Application of a second layer of tape; e) Laser ablation of the pattern for that layer of tape; f) Removal of the top layer of excess tape; g) Removal of the bottom layer of excess tape.

Note that this process is not suitable for overhanging features, as the tape will not peel away properly and may deform as shown in Figure 10.
3.3. Results

3.3.1. Calibration and Thickness

To make the sticker moulds, the laser cutter was calibrated for each thickness needed. This ensures that the tape is cut completely and that the substrate is not damaged by the laser. If a mould with a damaged substrate is used for polymer micromoulding, the pattern of the damage will also be patterned onto the polymer. The surface roughness introduced to the polymer may impede bonding at that part of the pattern. The quality used to calibrate the laser strength was the damage visible on the glass substrate upon inspection and under 40x magnification. A simple channel mould is cut at several different laser strengths, then the tape is removed and cleaned, leaving only the damaged substrate behind. Figure 11 shows the sticker mould, with half the tape removed and half the tape left on the substrate. Figure 12 the same sticker mould, but with the damage outlined and a dividing line added where the tape was cut away from the channel moulds.
Figure 11: Observing the damage of the laser. Three channel moulds were cut from 100 µm thick tape, at 100% Speed and 25%, 20%, and 15% power respectively.

Figure 12: Observing the damage of the laser on the substrate, with the damage highlighted. Three channel moulds were cut from 100 µm thick tape, at 100% Speed and 25%, 20%, and 15% power respectively.

It can be observed visually that the 25% power setting in Figure 12 causes significant damage to the glass substrate, the 20% power setting causes minor damage to the substrate, and the 15% power causes little damage to the substrate. The damage to the substrate is also readily visible under 4x magnification. Under magnification, it is apparent that the damage varies greatly based on the power applied by the laser (summarized in Table 1) and the correct power setting allows the tape to be cut with minimal damage to the substrate.
Table 1: Inspecting the damage to a glass substrate of various laser powers.

<table>
<thead>
<tr>
<th>Power setting of laser cutter (speed constant at 100%)</th>
<th>Damage under 40x magnification</th>
</tr>
</thead>
<tbody>
<tr>
<td>25%</td>
<td><img src="image1.png" alt="Image" /></td>
</tr>
<tr>
<td>20%</td>
<td><img src="image2.png" alt="Image" /></td>
</tr>
<tr>
<td>15%</td>
<td><img src="image3.png" alt="Image" /></td>
</tr>
</tbody>
</table>
Based on the lack of damage to the substrate, a guideline for laser cutting different thicknesses of sticker moulds is presented in Table 2. This guideline is valid only for the Universal 360 CO₂ laser cutter; the calibration must be performed for each laser cutter used as the power varies between laser cutters.

**Table 2: Power settings for Universal 360 CO₂ laser cutter for cutting different layers of tape, with the speed set to 100% for all samples.**

<table>
<thead>
<tr>
<th>Layers of packing tape</th>
<th>Total Thickness of Tape (µm)</th>
<th>Power setting %</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>51</td>
<td>10</td>
</tr>
<tr>
<td>2</td>
<td>102</td>
<td>15</td>
</tr>
<tr>
<td>3</td>
<td>153</td>
<td>18</td>
</tr>
<tr>
<td>4</td>
<td>204</td>
<td>20</td>
</tr>
</tbody>
</table>

### 3.3.2. Channel Dimensions

Figure 13 shows the fluid testing of channels that are designed to be 100 µm deep and vary in thickness from 100 µm to 500 µm. At a channel width of 100 µm the fluid does flow, but the channel is irregular.
Figure 13: Fluid testing of varying channel widths. From left to right, the widths are 500µm, 400µm, 300µm, 200µm, 100µm

The channel shape is affected by the laser cutter. A profilometer image characterizing the depth of a 750µm wide sticker mould of a simple channel is shown in Figure 14. The edges of the tape curl up due to the heat applied by the laser, creating an approximately 40µm high and 90µm wide lip on all cut edges. This distortion of the edge is passed from the mould to the channel, where a 25µm deep lip can be observed in Figure 15. Some variation may be due to measurements being taken in different locations; for instance, the PDMS channel shown in Figure 15 is made from the sticker mould shown in Figure 14, however, there is about a 100µm difference in width between the two channels.
Figure 14: Profilometer image of an approximately 600 µm wide sticker mould of a simple channel, with a 155µm depth. Vertical axis is in units of 10 µm, horizontal axis is in units of 1 µm. a) Profilometer data with no data labels for clarity. b) Identical profilometer data with scale bars.
Figure 15: Profilometer image of a 735\textmu m wide channel moulded using the sticker mould, with a 151 \textmu m depth. Verticle axis is in units of 10 \textmu m, horizontal axis is in units of 1 \textmu m. a) Profilometer data with no data labels for clarity. b) Identical profilometer data with scale bars.
The angle of the side walls is 44 degrees from the normal, which gives a channel geometry that is approximately V-shaped, with small 25 µm dips where the mould is distorted from the heat of the laser. Some compensation must also be made in the design, as the laser cutter will ablate approximately 60 µm of the tape, depending on the spot size of the laser. As such, a design that requires a 200 µm wide channel may need to be 320 µm wide in the design file to account for the laser ablation.

3.3.3. PDMS channels made from sticker moulds

The sticker mould, cut in the pattern of a simple channel, is shown in Figure 16 under magnification. The edges of the tape are not completely smooth due to the limited resolution of the laser ablation. There is also some adhesive remaining on the slide from where the excess tape was removed.

![Sticker mould under a magnification.](image)

**Figure 16: Sticker mould under a magnification.**

When used as a mould for PDMS soft lithography, the sticker mould from Figure 16 results in the PDMS channels shown in Figure 17. The PDMS channels have a similar
geometry to the sticker moulds and the substrate damage and excess adhesive on the mould has affected the channel dimensions.

Figure 17: PDMS channel made using the sticker mould. a) The focus is on the bottom of the channel. b) The focus is on the top of the channel.

The geometry of the channels is slightly easier to see in Figure 18 and Figure 19, where the PDMS channels were bonded to a glass slide and filled with water. In the Figure 18 channels, DI water with blue dye was run through the channels and then drained so that only air remained in the channels. This leaves the edges of the channels visible, as some of the blue dye remains, particularly at the edges. Figure 19 shows the channel filled with DI water and red dye.
3.4. Discussion

Sticker moulds are useful for rapid prototyping of designs that have features larger than 100 µm. They produce designs rapidly and have a low barrier to entry. No equipment is required except a laser cutter, which is common in many labs, and the materials are low cost. With a laser cutter, glass slides, and packing tape, each mould can be made for less than 10 cents.
There are some limitations to the channel dimensions, namely that the channels cannot be smaller than 100 μm and can only be make thicker in increments of 51 μm based on the number of layers of tape added. In theory, smaller features could be patterned using the laser cutter, as the wavelength of a CO₂ laser is 10.6 μm and due to diffraction limits, the minimum cutting size of the beam is approximately 20 μm. In the experimental results, there is not enough adhesive remaining on the tape on channel patterns below 100 μm to ensure reliable adhesion to the glass slide.

An exact calibration of the laser and the tape is required, meaning that the sticker moulds cut will vary based on the type of tape use, what laser cutter is used, the strength of the laser, and the spot size of the beam. The most successful tape, Uline Industrial Tape S-423, is described in this chapter, though forays into several different types of tape were made, such as scotch tape or the ARcare 8026. The Uline Industrial Tape S-423 was the most successful since it adhered strongly to the glass slide, unlike the scotch tape, but not so strongly that it couldn’t be readily removed from the glass slide like the ARcare 8026. It also smoothly adhered to itself, allowing multiple layers and increasing the possible depth of the sticker moulds.

The channels made from sticker moulds are V-shaped and have an irregular depth due to the heat from the laser distorting the tape of the moulds and causing the edges to curl up (as illustrated in Figure 14 and Figure 15). The width of the channels may vary through the channels, due to the low resolution of the laser causing the edges of the mould ablate unevenly, as shown in Figure 16. Sticker moulds may also have imperfections in the mould due air bubbles made while applying the tape. The imperfections can be transferred to the PDMS moulded from the sticker mould, or may cause the tape to delaminate from the substrate.

Sticker moulds would be most suitable for designs that are still in the initial design phase. Since their fabrication procedure is ultra-rapid, sticker moulds enable the rapid realization of ideas and are useful for early stages of work when many designs are being evaluated. Once the design is settled, other fabrication methods can be introduced to fine-tune the microfluidic structures. Another advantage of sticker moulds is that moulds that are cut in a laser cutter can be made as large as the laser print bed. This allows for large-array fabrication of moulds, which enables the batch fabrication of many micromoulded devices and the fabrication of large microdevices [62]. Sticker moulds are also favourable
for micromolding of nanocomposite polymers compared to SU-8 micromoulds. Some composite polymers need to be suspended in an organic solvent to ensure proper dispersion of nanoparticles in the polymer [62]. SU-8 is susceptible to swelling when exposed to organic solvents, which may lead to SU-8 moulds peeling off of their substrate when exposed to nanocomposite polymers that use organic solvents to aid dispersion [62]. Organic solvents are not used in this thesis, as the magnetic microparticles disperse evenly in the polymer without, but they are commonly used in other composites such as carbon-nanotube/PDMS composites [62].

A classroom environment is also an excellent venue for sticker moulds. The moulds are robust and easy for inexperienced students to fabricate and use. And if the moulds break, it is fast and inexpensive to make replacements. In the Fall 2018 semester, sticker moulds were introduced to SFU’s curriculum as micromoulds for a Y-mixer design (shown in Figure 20) in a lab for ENSC 859: Biomedical Microdevices. They proved to be an effective tool for teaching and learning about microfluidic devices.

Figure 20: Micromixer made using a sticker mould.
Chapter 4. M-CP membrane actuator

In the past decade, research into composite polymers has resulted in the publication of many magnetic composite polymer actuators, such as pumps, valves, and mixers (e.g., [20], [22], [23]). However, previous membrane actuators are typically characterized by deflection in air under a large (e.g., greater than 150 mT [25], [43]) external magnetic field [25], [33], [42]; however, this does not necessarily indicate that the actuator moves the same volume of liquid in microfluidic devices, with most failing to demonstrate this key capability once fluid considerations such as dead volume and water resistance are factors. Measuring the stroke volume, which is the displacement of water over the entire range of motion of the membrane, is an improved method of determining the useful function of a membrane actuator. The stroke volume and its relationship to the applied magnetic field provide insight into the external magnetic field requirements of the M-CP membrane which is valuable for system-level planning for systems requiring multiple device actuators.

This chapter presents the design for a permanently magnetized membrane actuator and characterizes the pump’s ability to displace water. The membrane actuator is interfaced with a thermoplastic PMMA test chamber, which demonstrates the fabrication flow of integrating an M-CP actuator with an industrially relevant material. The hybrid device is important as a next step in integrating M-CP structures into thermoplastic substrates and is discussed in more detail in Section 2.6.

The membrane design presented in this section is the basis for the micropump and the microvalve presented in Chapter 5. The test structure presented in Section 4.2 is the same as the diffusion pump presented in Section 5.1. The design is shared between the two for fabrication simplicity and allows the membrane actuators to be characterized as pumps in future work without any additional fabrication needed.

This chapter will begin with the design of the M-CP membrane actuator in Section 4.1 and the design of the test structure in Section 4.2. The fabrication procedure is

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described in Section 4.3, with the fabrication flow of the M-CP membrane actuator and the
test structure used to demonstrate fluid displacement shown in Section 4.3.1, and the
fabrication flow of the M-CP membranes as an array shown in Section 4.3.2. The
experimental test set up is detailed in Section 4.4 and the results are discussed in Section
4.5.

4.1. Membrane Design

In an M-CP membrane actuator, it is key to balance the flexibility of the membrane
with the relationship between magnetic material and magnetic force available. The thinner
the membrane, the more flexible it is. Yet the less magnetic material available, the less
the membrane will deflect under an applied magnetic field [24]. Further discussion of the
balance between mechanical and magnetic characteristics of a magnetic polymer
membrane is detailed in Section 2.2.

The membrane actuator designed for this thesis work is a 200µm thick, 7mm
diameter circle, with a central magnet feature called a “boss” that is 2mm in diameter and
2 mm thick (shown in Figure 21). The majority of the magnetic material is contained in the
boss and this boss is the driving force of the actuator. The M-CP membrane is embedded
in a thicker PDMS structure (shown in Figure 22); a thinner membrane allows for more
deflection as it is more flexible [24] and a thicker support structure leads to more robust
devices with a higher yield. This reduced mass of magnetic material is balanced by the
boss, which has enough magnetic material to deflect significantly in a low applied magnetic
field (e.g., 10mT). The membrane actuators have an average permanent magnetic field of
25 mT.
The chosen diameter of 7 mm for the membrane was based on work by Hilbich et al. [33] on M-CP membrane deflections and by Zhou and Amirouch [39] on a diffusion pump actuated with a thin bulk magnet. The large diameter membrane was able to deflect significantly (100 μm) under magnetic fields of less than 20mT [33]. An all-M-CP membrane design was chosen over a PDMS membrane with an M-CP boss based on previous work by Rahbar and Gray [24] because the all-M-CP membrane is less flexible and therefore deforms less under high fluidic pressures.

It was observed during initial investigations that under high fluidic pressures the M-CP membrane was prone to deflecting significantly and “bubbling” up such that it was not
as responsive to the magnetic field. The membrane would be at the mechanical limits of deflection because of the high fluidic pressure and the applied magnetic field would not be enough to overcome the fluidic pressure. To minimize this “bubbling” effect, the maximum deflection of the membrane was limited by the large 2 mm boss to add rigidity to withstand higher fluidic pressures. Then the flexibility of the membrane could be preserved by still being thin and the deflection/response of the M-CP could be maximized by the increased magnetic material. The flexibility, more limited range and more magnetic material allows it to be more responsive under an oscillating magnetic field, which is important for pumping applications.

4.2. Test Structure

To demonstrate the water displacement, a channel layer made out of rigid PMMA is bonded to the M-CP actuator embedded in PDMS, shown in Figure 23. This allows the M-CP membrane to move water under an applied magnetic field and demonstrate a strong stroke volume.

Figure 23: 3D model of the M-CP membrane actuator embedded in the PDMS support structure, prior to bonding the M-CP actuator to the PMMA fluidic channels. This image is reprinted with permission from [68], © 2020 IEEE.

The hybrid design is novel and allows for the actuator to deflect without any deformation in the substrate layer. It also shows that M-CP actuators can interface with thermoplastic PMMA, which is used in the microfluidics industry, and demonstrates a
simple fabrication flow that can be used low-volume (academic) manufacture and follows the design-to-manufacture principles discussed in Section 1.1 and Section 2.6 to allow designs to transition to high-volume commercial manufacture. The chamber is 400µm deep to allow the diaphragm full range of movement without contacting the bottom of the chamber.

4.3. Procedure

For fabrication of a single membrane actuator, the M-CP prepolymer is cured at the same time as the PDMS support structure which serves to cover the fluidic channels. The fabrication flow of the M-CP actuator and hybrid bonding to the PMMA test structure is described in Section 4.3.1 and the world-to-chip connectors used are described in Microfluidic connectors. The fabrication of arrays of M-CP membranes is described in Section 4.3.2. For array fabrication, the M-CP membrane is fully cured and magnetized before being embedded in the PDMS support structure due to size limitations of the magnetizer.

4.3.1. Actuator and Test Structure Fabrication

The fabrication of the M-CP membrane actuator and fluidic test chamber is shown in Figure 24, in three parts. Figure 24a shows the fabrication flow for the PMMA fluidic chamber, Figure 24b shows the fabrication of the M-CP membrane actuator, and Figure 24c shows the bonding between the two devices.

The fabrication steps for the PMMA fluidic chamber are summarized in Figure 24a. The fluidic chamber is made from a PMMA substrate (Figure 24ai). After cleaning the PMMA substrate with IPA, the adhesive ARclear 8026 is applied without removing the supporting PET film from the adhesive (Figure 24a(ii)). The supporting film remains in place to protect the adhesive from any PMMA dust during the laser ablation. The pattern for the channels is cut into the PMMA substrate using laser ablation, which also cuts away the adhesive to the exact pattern of the channels (Figure 24a(iii)). The substrate is thermally annealed at 90°C for one hour to prevent thermal shock and then carefully cleaned with IPA.
The fabrication of a single M-CP diaphragm begins with the fabrication of a mould. A PMMA substrate (Figure 24bi) is patterned as a mould via laser ablation, thermally annealed, cleaned, and treated with a mould release aerosol (Figure 24bii). The M-CP prepolymer is then applied to the central feature of the mould and the excess removed (Figure 24biii). A thin layer of PDMS prepolymer is applied over the entire mould and any excess removed with care such that the M-CP prepolymer is not disturbed (Figure 24biv). The sample is then cured at 90°C for one hour, allowed to cool, and then removed from the mould (Figure 24bv). The entire sample is then exposed to a 3T magnetic field to permanently magnetize the M-CP membrane.

To assemble the device as shown Figure 24c, the completed membrane actuator from Figure 24b is attached to the adhesive layer of the PMMA substrate from Figure 24a and connected to external tubing.

Figure 24: Process flow for diffusion pump design. Fabrication flow of the membrane actuator with top view and cross-section. a) Fabrication of the fluidic chamber. i) Begin with a PMMA substrate. ii) Apply the ARcare 8026 transfer adhesive to the substrate. iii) Pattern the adhesive and the PMMA using laser ablation to make the fluidic channels. b) Fabrication of the M-CP actuator. i) Begin with a PMMA substrate. ii) Pattern the substrate using laser ablation to make a mould. iii) Apply M-CP to the central feature of the mould and remove excess. iv) Apply PDMS to the entire mould and cure at 90°C for 1 hour. v) Remove the embedded actuator from the mould and place in a 3T magnetic field for 10 minutes. c) Alignment and assembly of the final device. This image is reprinted with permission from [68], © 2020 IEEE.
Microfluidic connectors

The microfluidic connectors used were designed in the Microinstrumentation Lab. They are 0.8cm tall, 1cm diameter PDMS cylinders, with a central port for tubing (shown in Figure 25). The central port has an hourglass shape that pinches the tubing to prevent backflow. The port for tubing can be fabricated in different sizes depending on the gauge of tubing desired. For the work in this thesis, tubing with an outer diameter of 0.59mm was used with connectors with ports of 0.5mm diameter (shown in Figure 25b). To fabricate the connectors, two different sizes of crafting hole punchers were used. The hourglass shape was obtained by increasing the thickness of the PDMS connector, which caused it to compress when the hole puncher was used, leading to a port with a narrower center.
Figure 25: 3D model of the connector. a) View of the entire connector. b) Cross-section of the connector.

4.3.2. Array Fabrication

For fabrication as part of an array, the design of the membrane is altered slightly so that the M-CP membrane has raised outer rim (shown in Figure 26). The raised outer rim acts as a dam so that the M-CP membrane can be embedded in the PDMS prepolymer without the viscous PDMS prepolymer flowing into the membrane (shown in Figure 27). The design changes have no effect on the membrane actuator as an actuator, as the added elements are adhered directly to the PMMA substrate and immobile.
Figure 26: Membrane actuator modified for array integration.

Figure 27: Membrane actuator embedded in a PDMS layer.

The mould used for making the array of M-CP membranes is shown in Figure 28. The mould features multiple structures to make the M-CP membrane, which allows for batch fabrication.
Figure 28: Top view and cross section of PMMA mould for array of magnetic diaphragm actuators.

The PMMA mould is cut via laser ablation, thermally annealed, cleaned, and treated with a mould release aerosol (Figure 29a). M-CP prepolymer is applied to each of the actuator structures of the mould and the excess removed (Figure 29b). The samples are then cured at 90°C for one hour, allowed to cool, and then removed from the mould (Figure 29c). The M-CP diaphragms are exposed to a 3T magnetic field to render them permanently magnetized. To make the array, a glass substrate is used. A thin layer of PDMS prepolymer is applied to the glass slide (Figure 29d), then the M-CP actuators are aligned on top into an array (Figure 29e). A layer of PDMS prepolymer is then applied to the entire substrate (Figure 29f), cured at 90°C for one hour, allowed to cool, and then removed from the glass substrate (Figure 29g).

4.4. Experimental set up

For this experiment, the volume of water displaced by the membrane actuator is measured. This stroke volume is determined by applying an external magnetic field to the membrane actuator, which causes the membrane to deflect and move de-ionized water in the device. Water displacement is measured in external tubing connected to the inlets (Figure 30). The device is primed and then some of the water is drained from the external tubing so that the displacement volume stays within the tubes in all actuator positions. With the known tubing dimensions, the volume of water displaced is calculated.

The external magnetic field is applied via permanent magnet, with magnitude changed by varying the distance between the M-CP membrane and 200mT permanent magnet by increments of 1 mm. Measurements are taken using both poles at each magnitude of the applied magnetic field, to determine the stroke volume over the membrane's attraction and repulsion to the magnet. The magnetic field strength was
verified using a Tesla meter (F.W. Bell Hall Effect 5100 Series Tesla Meter from Pacific Scientific).

Figure 30: Top view of the membrane actuator at rest. This image is reprinted with permission from [68], © 2020 IEEE.

4.5. Results and Discussion

The membrane actuator designed and characterized in this work shows a strong stroke volume under relatively low applied magnetic fields. A range of 0.2-7.4 µL stroke volume is obtained by varying the magnetic field applied from ±19mT to ±110 mT (Figure 31). The stroke volumes for the individual samples are shown in Figure 31 alongside the average stroke volume at each increment of the applied magnetic field. The individual samples are labelled in the legend by the permanent magnetic field of their M-CP actuator structures, which varied due to slight volume differences in the M-CP actuator. The membrane actuators used in this experiment have an average permanent magnetic field of 25 mT.
Figure 31: Experimentally determined relationship between the applied magnetic field and stroke volume. The stroke volume is the net water displacement due to the M-CP actuator between the attraction and repulsion states. The average magnetic field of the samples is 25mT and the stroke volume for each individual is presented alongside the calculated average stroke volume at each increment of the applied magnetic field. This graph is reprinted with permission from [68], © 2020 IEEE.

Displacement of water is observed in all samples both when the M-CP actuator is under the applied attractive magnetic field and the applied repulsive magnetic field, though the magnitude of water displacement tends to be slightly larger in the repulsive state. The stroke volume increases with the magnetic field over all samples and permanently magnetized membrane actuators with a stronger magnetic field displace more water.

Characterization via stroke volume is a more definitive method of determining the potential of a M-CP membrane for microfluidics applications than via deflection of the same membrane in air. The stroke volume directly shows that the membrane actuator is able to overcome the fluidic resistance (roughly estimated as $1.23 \times 10^{11}$ kg/s/m$^4$ in the example device) and displace water in microfluidic channels. The permanent magnetization of the M-CP actuator is key to the large stroke volumes reported, as the actuator displaces water in both the attractive and repulsive states that are used to determine the net stroke volume.

With such high stroke volume at low applied magnetic fields, the M-CP membrane is expected to make an excellent microfluidic pump and/or valve actuator. With the addition of, e.g., check valves, the current test structure should be an effective pump. Due to the low applied magnetic field required to actuate the device, the M-CP membrane actuator
could potentially be controlled by a small, integrated electromagnet such as a Helmholtz coil that requires relatively little power. Even more water displacement is expected to be obtained by increasing the height of the central boss of the M-CP membrane (similarly to [24]); the increased magnetic material should increase the magnetic response of the M-CP membrane without increasing the applied magnetic field.

The simple hybrid-material batch fabrication process is scalable and facilitates arrays of devices to be made in one integrated layer with minimal alignment needed. The simple planar structure of the M-CP membrane shows promise for integration with commercial thermoplastic microfluidic devices and systems.

The membrane actuators demonstrated an effective hybrid-material device, with the thermoplastic PMMA integrated into the M-CP and PDMS membranes. The batch fabrication process developed in this chapter shows promise for integration with commercial thermoplastic microfluidic devices and systems. With such high stroke volume at low applied magnetic fields and the hybrid fabrication process, the M-CP membrane actuator is expected to make an excellent microfluidic pump and/or valve in a thermoplastic/M-CP microfluidic system.
Chapter 5. Test Structures

In the previous chapter, an M-CP membrane actuator was demonstrated and fully characterized. In this chapter, I present two test structures designed to show the integration of the membrane actuator into the fabrication flow for example microfluidic devices: a diffusion pump and a normally closed valve. Both the pump and the valve designs use exactly the same M-CP membrane design, with modifications made to the device substrates and supporting structures to change the functionality of the devices. The diffusion pump is a hybrid design employing a PMMA substrate to fabricate the fluidic channels. The normally closed valve is fabricated as an all-PDMS device. The purpose of demonstrating the M-CP actuator in two different fabrication flows (using two different substrate materials) is to show that the M-CP can be easily integrated into academic-standard PDMS fabrication flows, as well as in commercial thermoplastic systems. The valve uses the sticker mould fabrication presented in Chapter 3 and demonstrates the viability of the sticker mould method. Despite the all-PDMS design of the valve, a simple layer-based design is still demonstrated that follows the design principles discussed in Section 2.6 and commercialization considerations are presented in Section 1.1.

Section 5.1 presents the design of the diffusion pump, which has an identical fabrication flow to the test structure fabrication presented in Section 4.3.1, as the diffusion pump devices were used to characterize the stroke volume of the M-CP membrane actuator. Section 5.2 presents the design of the microvalve and Section 5.3 discusses the two fabrication flows (from Section 4.3.1 and Section 5.2), showing that the membrane actuator may be integrated into the fabrication flow for a variety of device designs implemented in different materials.

5.1. Diffusion Pump Design

A diffusion pump operates based on a pressure differential between a nozzle and a diffuser as a membrane actuator cycles through its stroke (discussed in Section 2.4) and is suitable for applications in which backflow is acceptable. For instance, in a lab-on-a-chip system there will be valves that can coordinate and prevent backflow in the device (such as in the µRoams project discussed in Section 2.8). The pump design is a hybrid device, with the channel layer made out of rigid PMMA, bonded to a flexible
actuator layer of PDMS and M-CP, shown in Figure 32. The actuator used in this design is the M-CP diaphragm described in Section 4.1.

Figure 32: a) 3D model of the M-CP membrane actuator embedded in the PDMS support structure, prior to bonding the M-CP actuator to the PMMA fluidic channels. b) Diagram of the diffusion pump in the PMMA substrate. This image is modified with permission from [68], © 2020 IEEE.

The rigid PMMA layer allows the actuator to deflect without any deformation in the substrate layer. They hybrid design demonstrates a fabrication flow that interfaces M-CP actuators with PMMA, which is a thermoplastic favored by the microfluidics industry (discussed in Section 1.1). An all-PDMS device could also be realized for applications that require flexible pumping (e.g., wearable microfluidics and wearable labs-on-a-chip [[69]]). The chamber is 400µm deep to allow the diaphragm to deflect fully without contacting the bottom of the chamber. The nozzle and diffusor angles are at a 10° angle (shown in Figure 32), which is a typical theta value for diffusion pumps [12], [39]. Variations on this theta value and geometry are discussed in the future work (Section 7.1). The membrane actuators have an average permanent magnetic field of 25 mT, with
dimensions of 0.5 mm thickness and 7 mm diameter, and central boss of 2 mm thickness and 2 mm diameter, as discussed in Section 4.1.

Actuation of the diffusion pump is done using an alternating magnetic field. An electromagnet is an excellent option (shown in Figure 33), as it offers a low magnetic field at high frequencies. As the M-CP actuator has demonstrated a high stroke volume at low magnetic fields (Section 4.5), an alternating magnetic field of from ±19 mT to ±110 mT would be reasonable to provide.

The future work for the pump, such as demonstrating the pumping rate and frequency, is discussed in Section 7.1.

Figure 33: Electromagnet to apply an external magnetic field.

5.2. Microvalve Design

The current design of the M-CP membrane actuator as a normally closed valve is a three-layer device, with the top layer containing the embedded M-CP membrane
actuator, the middle layer containing the valve seat and fluidic chamber, and the bottom layer containing the channels (shown in Figure 34).

Figure 34: Top view of each of the layers of the microvalve. a) The membrane actuator embedded in PDMS. b) The valve seat and fluidic chamber. c) The connecting fluidic channel.

In the closed position, the membrane actuator seals against the valve seat and prevents fluid flow (shown in Figure 35). The valve should be in closed position when an applied magnetic field pulls the membrane towards the valve seat. At low pressures, the magnetic force between the membrane and valve seat is enough to overcome the fluidic pressure and allows the membrane and valve seat to latch. The closed position is the default position of the valve when under no applied pressure, which is why it is called a “normally-closed” valve. The normally-closed state is advantageous as it prevents leakage in a system, as opening the valve requires actively controlling the system (assuming the system remains within its operating pressure and the valves do not burst). In a multi-chamber microfluidic system that requires mixing, or complicated routing in which the valves remain closed for the majority of the operating time, a valve that remains closed with no energy applied to the system can save power.
Figure 35: Microvalve in closed position.

When an applied magnetic field deflects the membrane actuator away from the valve seat and allows fluid to flow, the valve is in the open position (Figure 36).

Figure 36: Microvalve in open position.

The valve presented in this section has a design based on a simple, scalable fabrication procedure, with easily aligned and assembled layers that enable array fabrication. The channel layer could be switched out for a PMMA substrate for integration in hybrid devices.

Because of the geometry of the valves, the membrane actuators make excellent normally closed valves when the system is in a negative pressure situation (i.e. a syringe pump withdrawing water from the system). This is entirely due to the geometry of the design; an undoped PDMS membrane replacing the M-CP would still form a tight seal in that fluid condition. The M-CP valves can overcome the pressure to close and open under an applied magnetic field, but the magnetic field must be strong or the pressure low. The pressure that the valves can overcome has not been fully characterized and this future work will be discussed in Section 7.2

5.2.1. Fabrication flow

The fabrication flow for the microvalve is presented in Figure 37. Figure 37a shows the fabrication of the channel layer, Figure 37b shows the fabrication flow of the chamber layer, Figure 37c shows the fabrication of the actuator layer, and Figure 37d shows the
assembly of the layers into one device. The fabrication flow shown in Figure 37 a-d is broken down into parts and is explored in detail in Figure 38, Figure 39, Figure 40, and Figure 41.

**Figure 37:** Overview of the fabrication flow. A) The fabrication of the channel layer. B) Fabrication of the chamber layer. C) Fabrication of the actuator layer. D) Assembly of the device.
The sticker mouds presented in Chapter 3 are used extensively in this fabrication flow. Figure 38 shows the fabrication of the chamber layer, using the sticker mould to pattern the PDMS.

Figure 38: Detailed fabrication view of the channel layer. i) Glass substrate. ii) Apply tape to glass substrate. iii) Cut a mould pattern into the tape using laser ablation, then remove excess tape. iv) Apply PDMS to mould and cure. v) Demould PDMS channel layer.

Figure 39 shows the fabrication of the middle chamber layer and features the use of both M-CP and PDMS in the same mould to create a magnetic valve seat.
Figure 39: Detailed fabrication view of the channel layer. i) Glass substrate. ii) Apply tape to glass substrate. iii) Cut a mould pattern into the tape using laser ablation, then remove excess tape. iv) Apply M-CP to the valve seat. v) Apply PDMS to the mould and cure. vi) Demould PDMS/M-CP chamber layer. vii) Cut holes for fluidic connections.

Figure 40 shows the fabrication of the M-CP membrane with an added adhesive layer and holes cut for global fluidic connections.
Figure 40: Detailed fabrication view of the actuator layer. i) PMMA substrate. ii) Cut a mould pattern into the PMMA using laser ablation iii) Apply M-CP to the central actuator part of the mould iv) Apply PDMS to the mould and cure. v) Demould PDMS/M-CP actuator layer. vi) Cut holes for fluidic connections and apply ARcare 8026 transfer adhesive.

The final assembly of the device is shown in Figure 41. The channel layer from Figure 38 and the chamber layer from Figure 39 are bonded together using oxygen plasma treatment, then the actuator layer from Figure 40 is bonded to the other layers using adhesive bonding. Adhesive bonding was chosen over oxygen plasma bonding because oxygen plasma is not specific. The use of oxygen plasma treatment in this geometry bonds the membrane actuator closed against the valve seat and prevents all fluid flow. The adhesive was patterned so that no adhesive was applied to the M-CP membrane and the membrane was free to deflect and permit fluid flow.
5.3. Discussion

The two fabrication flows presented in this chapter demonstrate the M-CP membrane actuators presented in Chapter 4 integrated into an all-PDMS fabrication flow, as well as a hybrid PMMA/M-CP fabrication flow. Integration into an all-PDMS fabrication shows that the M-CP material can be easily developed in an academic setting, as well as transition to hybrid devices for more commercial applications. This illustrates the potential of the material, which can be further developed in an academic lab setting and then transition to a commercial mass-manufacture if sufficient care is taken with ensuring the designs meet the volume manufacture guidelines (discussed in Section 1.1).

As a test design, a diffusion pump was chosen because of its simple structure and test parameters. Under an external applied magnetic field the membrane actuator oscillates and moves water effectively, but does not pump under a wide variety of diffuser
and nozzle valve designs. The water simply moves up and down in the external tubing without any net displacement. Based on this observation, a check valve pump would be much more effective than the current design and should be pursued in future work.

The microvalve design is three layers, which is more complex and requires more alignment than a two-layer device would, but the geometry is essential to the valve behaviour. The microvalve also features a chamber with some dead volume, which is also essential to the valve behavior. Ideally, the valve should be made in two layers to decrease parts and alignment issues. There was some investigation into maintaining the geometry but reducing the layers by making a double-sided PDMS channel structure, which would combine the bottom two fluidic layers into one double-sided PDMS channel with a via connecting the two. Fabricating a double sided PDMS structure is possible, but not practical. Double-sided PDMS devices were not consistent in thickness, took longer to fabricate, and were more prone to breaking when demoulded from the double-sided mould.

The all-PDMS workflow for the micro-valves also showed that the sticker moulds presented in Chapter 3 are reliable and effective. Over 100 samples were made using that fabrication flow and the moulds were easy to fabricate and repeatable. Depending on the geometry of the mould, each mould can only be used 3-10 times, though each sticker mould was reliably multi-use. Features such as long, narrow channels (less than 200 microns wide) were prone to delaminating, as there was less adhesive to bond the tape to the glass slide. Making new moulds was not an issue though, as the entire process took only 10 minutes and the moulds were repeatable and low cost to the point of being disposable.

While both the microvalve and the micropump show promising initial results as new working devices, they have not been fully characterized due to a mixture of challenges. The challenges and future work proposed to make working devices are discussed in Chapter 7. However, the membrane actuators are able to be integrated into the pump and valve fabrication flows and displace water, showing successful integration of working membrane actuators into a potentially wide range of different devices.
Chapter 6. Magnetic OSTE+

Microfluidic devices developed in academic labs are, for the most part, made from polydimethylsiloxane (PDMS), which does not have a scalable fabrication process. Cutting-edge devices fabricated in academic labs do not generally transition into mass production because they are limited by the materials used [11]. Rigid elements of microfluidic devices can be made in thermoplastic materials, which can easily be mass manufactured via hot embossing or injection molding [7]. Flexible elements of microfluidic arrays, which tend to be actuators, may not have sufficient stroke volume if made using rigid thermoplastics and therefore do not easily translate to industry (as discussed in Section 1.1 and Section 2.6).

One avenue of investigation explored in this thesis is a hybrid device using flexible PDMS and M-CP with rigid thermoplastic PMMA. While a scalable fabrication method to bond PDMS/M-CP and PMMA structures using transfer adhesive is demonstrated (in Chapter 4), transitioning away from PDMS based devices could allow the use of mass fabrication methods such as injection moulding to fabricate entire microfluidic devices. A major barrier to material change is that the majority of the research into nanocomposite polymer actuators in microfluidics uses PDMS as the matrix polymer [19], [25], [36], [70]. This eases the integration of nanocomposite PDMS polymer structures into PDMS microfluidic devices (e.g. [22], [30], [33], [62]), but does not enable the use of injection mouldable polymer substrates. This means that many commercial thermoplastic microfluidic devices and systems may not be able to fully take advantage of advanced M-CP structures such as cilia mixers [22], unless a new injection mouldable matrix polymer for the magnetic polymer can be developed. This new material would need to be similar to PDMS in ease of fabrication and flexibility for high stroke actuation, so that existing flexible M-CP designs could be reproduced, but also easy to mass produce and integrate with thermoplastics.

A recently developed polymer, an off-stoichiometry thiol-ene-epoxy polymer (OSTE+) [59], is hydrophilic, can be manufactured in a way similar to PDMS soft lithography, and can be injection moulded [18]. The background on OSTE+ polymer resins is discussed in Section 2.10. This chapter will demonstrate a method to integrate rare earth ((Nd$_{0.7}$Ce$_{0.3}$)$_{10.5}$Fe$_{83.9}$B$_{5.6}$) magnetic particles into an OSTE+ matrix, which facilitates
the transition of M-CPs into mass manufactured injection moulded devices. In particular, we develop a new M-CP based on OSTE 324 Flex to microfabricate flexible magnetic elements and demonstrate that they remain permanently magnetized for over one year.

Section 6.1 describes the design of the experiments performed on the Magnetic OSTE+ (M-OSTE+). Section 6.2 outlines the procedure of the testing, Section 6.3 describes the results and discusses the findings of this chapter.

6.1. Design

One of the major limitations for OTSE+ as a matrix polymer is its need for a UV cure. The majority of micro- and nanoparticles that could be integrated into the polymer may render the OSTE+ opaque. Thus integrating magnetic particles into the OSTE+ to form magnetic OSTE+ (M-OSTE+) may impede the UV light from penetrating the sample and may limit the thickness of the device. If the light is able to cure the M-OSTE+, the UV exposure time required for M-OSTE+ to complete the first cure may be longer than that of regular OSTE+ (1min/mm at 12mW/cm²), as less light will penetrate the material. Another consideration is that there is a maximum dosage of UV light that the OSTE+ sample can be exposed to before the second cure begins. It is essential that the second cure of the M-OSTE+ is not triggered by the heat of the UV light, because once this cure is complete, the M-OSTE+ will no longer have epoxy groups available, removing its ability to bond to a wide variety of materials on contact [32].

The dosage of UV light is tested by first dosing the M-OSTE+ with UV light in short dosages steps (20 seconds at a time) and demoulding the sample once the first cure is completed. If the sample does not demould, the UV exposure is considered insufficient and more UV light can be applied. After exposing the samples to a 3mT magnetic field for 10 min, the single-cured M-OSTE+ is heated to 110°C for 1 hour to complete the second cure.

Moulds with different depths, 150 µm, 250 µm, and 350 µm, were utilized to make simple 5 mm by 5 mm squares. The different depths allow for an analysis of how the M-OSTE+ reacts to the same dosage of UV light at different thicknesses. This test helps determine the relationship between thickness and UV light dosage.
PDMS micromoulds are recommended for the OSTE+ polymers OSTEMER 322 Crystal Clear and OSTEMER 324 Flex [59], though in initial investigations into OSTE+ show that PMMA moulds and the sticker moulds presented in Chapter 3 are effective moulds. Since the the samples are only treated on one side with UV light, PMMA is an appropriate mould material for tests where the UV dosage needs to be measured exactly. It is fast to fabricate, with good thickness, and it absorbs light in the UV frequency. In contrast, some mould materials, such as SU-8, are reflective and can reflect the UV light back into the sample, increasing the dosage in unpredictable ways.

The essential components tested in this chapter are whether the M-OSTE+ can be moulded with sufficient thickness to be useful as a magnetic polymer actuator, whether it can be permanently magnetized, and whether it remains permanently magnetized over time. The simple 5 mm by 5 mm test squares of varying thickness are fabricated and the UV light dosage required to make cure them is measured. The samples are exposed to a strong magnetic field to render them permanently magnetized, their magnetic field measured, and then they are left alone for over one year. After a year has elapsed, the magnetic field of the samples is once again measured and is compared to when they were newly magnetized. Due to the small magnetic fields of the samples and the resolution limit of the F.W. Bell Hall Effect 5100 Series Tesla Meter from Pacific Scientific used, the measured magnetic field data is used to confirm the qualitative observation that the samples have magnetic poles of their own. The magnetic field data cannot be used to confirm the magnitude of this pole or whether the magnetic field remains identical after 15 months has elapsed, as the magnetic fields are small (near the resolution limit of the Tesla meter).

6.2. Procedure

The OSTE 324 Flex comes as two components that need to be mixed at a ratio of 1.24:1. The \((\text{Nd}_{0.7}\text{Ce}_{0.3})_{10.5}\text{Fe}_{83.9}\text{B}_{5.6}\) magnetic particles are mixed into component A, as it is the less reactive of the two components. The particles are mixed manually and then the solution is agitated with an ultrasonic mixer to ensure a uniform density. This “pre-mixing” allows the magnetic polymer to be made faster and in a more replicable way, since the component A doped with the magnetic particles can be kept for a few weeks and used for subsequent devices.
Once the components A and B are combined, the prepolymer has a final magnetic particle doping of 25% magnetic particles with respect to weight. Only a small amount of the prepolymer should be mixed at any one time, as the mixture is prone to seizing after several minutes. The prepolymer is still viable at this time, it just becomes more viscous and will not spread into the mould as easily.

PDMS micromoulds (shown in Figure 42a) are made by patterning PDMS using the micromoulding technique described in Section 4.3.1. To micromould the M-OSTE+, a small amount of the magnetic prepolymer is applied to a PDMS micromould and the excess is scraped away. Using a mask aligner with a 365nm mercury lamp, the sample is exposed for 20 seconds at a time with a PMMA sheet under the sample to prevent reflected light from affecting the dosage as PMMA blocks UV light (shown in Figure 42b). Between exposures, the samples are probed with metal tweezers to determine if the first cure is complete and removed once the cure is complete (shown in Figure 42c). Once the exposures are finished, the sample will be firm and can be removed from the mould.

The samples are placed on a PET film and are not applied to any substrate as they will form irreversible epoxy bonds during this phase and are heated to 110°C for 1 hour to complete the second cure (shown in Figure 42d). Once the samples have finished the second cure, they are removed from the PET film and placed in a 3 mT magnetic field for 10 minutes to allow the samples to become permanently magnetized (shown in Figure 42e).
Figure 42: Process flow of M-OSTE+ test samples. a) The PDMS mould. b) M-OSTE+ applied to the mould, with the excess scraped away and a PET film applied to aid in demoulding. c) Exposure to the UV light, then each sample demoulded when cured, repeating 20s exposures as needed. d) All M-OSTE+ samples completed the first cure and placed on a PET film, then cured at 110°C for 1 hour to complete the second cure. e) Samples removed from PET film and magnetized under a 3T magnetizer.

Using a Tesla meter, the magnetic field of the sample is measured and the thickness is measured using a micrometer. The samples are then placed in a petri dish and left unmodified for 15 months, when a second magnetic field measurement is taken. The data obtained from this test is the thickness of the samples and the magnetic field at two different dates.

6.3. Results and Discussion

The exposure time required for M-OSTE+ is approximately triple the cure time of undoped OSTE+, which is 1mm/minute. This relationship is summarized in Table 3. Because the UV exposure was done in steps of 20 seconds, the successful dosage may not be the minimum dosage required for the sample to cure, but is a close approximation of this value.
### Table 3: Change in dosage when OSTE+ is doped with magnetic particles.

<table>
<thead>
<tr>
<th>Sample theoretical thickness (µm)</th>
<th>Actual thickness (µm)</th>
<th>Exposure time (s)</th>
<th>Dosage (mm/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>150</td>
<td>156</td>
<td>40</td>
<td>0.234</td>
</tr>
<tr>
<td></td>
<td>206</td>
<td>40</td>
<td>0.309</td>
</tr>
<tr>
<td></td>
<td>227</td>
<td>40</td>
<td>0.3405</td>
</tr>
<tr>
<td></td>
<td>233</td>
<td>40</td>
<td>0.3495</td>
</tr>
<tr>
<td>250</td>
<td>244</td>
<td>60</td>
<td>0.244</td>
</tr>
<tr>
<td></td>
<td>245</td>
<td>60</td>
<td>0.245</td>
</tr>
<tr>
<td></td>
<td>303</td>
<td>60</td>
<td>0.303</td>
</tr>
<tr>
<td></td>
<td>322</td>
<td>60</td>
<td>0.322</td>
</tr>
<tr>
<td>350</td>
<td>344</td>
<td>60</td>
<td>0.344</td>
</tr>
<tr>
<td></td>
<td>347</td>
<td>60</td>
<td>0.347</td>
</tr>
<tr>
<td></td>
<td>349</td>
<td>60</td>
<td>0.349</td>
</tr>
<tr>
<td></td>
<td>357</td>
<td>60</td>
<td>0.357</td>
</tr>
</tbody>
</table>

The flexibility of the M-OSTE+ is shown in Figure 43. This flexibility was observed qualitatively to be similar to that of the undoped OSTE+ 324.
All of the samples for the time lapse test were permanently magnetized to 1-3mT successfully and showed no appreciable change between July 4, 2018 and October 2, 2019 (data summarized in Table 4). From this, we can conclude that the M-OSTE+ can be permanently magnetized (though the data should be interpreted as a confirmation that the permanent magnetic field exists, not as a measure of the magnetic field magnitude, due to the resolution limit of the Tesla meter). The magnetic field of the M-OSTE+ is smaller than that of the PDMS-based M-CP, which is due to the doping percentages. The M-OSTE+ is 25%wrt weight, where the PDMS M-CP is 80%wrt weight. Due to limitations in time, the M-OSTE+ could not be characterized at different doping percentages, though it is clear from observations during the procedure (such as the magnetic polymer thickening when the prepolymer is left uncured for several minutes) that the OSTE 324 Flex will not be suitable for the high doping percentages that PDMS is capable of.

Figure 43: Deformation of M-OSTE+ to show flexibility.
Table 4: Summary of magnetic field strength of M-OSTE+ over an 18 month time lapse.

<table>
<thead>
<tr>
<th>Sample theoretical thickness (µm)</th>
<th>actual thickness (µm)</th>
<th>volume (mm³)</th>
<th>magnetic field July 2018 (mT)</th>
<th>magnetic field Oct 2019 (mT)</th>
<th>Permanent magnetic field retained</th>
</tr>
</thead>
<tbody>
<tr>
<td>150</td>
<td>156</td>
<td>3.9</td>
<td>1</td>
<td>1</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>206</td>
<td>5.15</td>
<td>1</td>
<td>1</td>
<td>Yes</td>
</tr>
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<td></td>
<td>227</td>
<td>5.675</td>
<td>1</td>
<td>2</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>233</td>
<td>5.825</td>
<td>1</td>
<td>1</td>
<td>Yes</td>
</tr>
<tr>
<td>250</td>
<td>244</td>
<td>6.1</td>
<td>1</td>
<td>1</td>
<td>Yes</td>
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<td>245</td>
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</tr>
<tr>
<td></td>
<td>322</td>
<td>8.05</td>
<td>2</td>
<td>2</td>
<td>Yes</td>
</tr>
<tr>
<td>350</td>
<td>344</td>
<td>8.6</td>
<td>2</td>
<td>2</td>
<td>Yes</td>
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<td>Yes</td>
</tr>
<tr>
<td></td>
<td>349</td>
<td>8.725</td>
<td>3</td>
<td>3</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>357</td>
<td>8.925</td>
<td>3</td>
<td>3</td>
<td>Yes</td>
</tr>
</tbody>
</table>
This chapter has detailed a novel magnetic composite polymer made from OSTE+. The M-OSTE+ can be permanently magnetized, and retains a permanent magnetic field for at least 15 months, which could give commercial devices made using M-OSTE+ a shelf life of at least a year. For applications such as water testing, a year-long shelf life should be adequate.

OSTE+ is a strong candidate for a matrix polymer for more microfluidic applications requiring functionalized composite polymers. Further investigation is needed to determine the maximum possible thickness of the M-OSTE+, the maximum doping percentage that the OSTE 324 Flex is capable of withstanding, and the effects of the doping percentage on the Young’s Modulus of the material. Future work should also include characterization of the performance of actuators made from M-OSTE+ in comparison to those of magnetic PDMS.
Chapter 7.  Future Work

7.1. Membrane actuator pump

While the membrane actuator micropump described in Section 5.1 has an excellent stroke volume and simple fabrication flow, a successful pumping scheme has not been demonstrated at the time of publication and would be an excellent next step to take in the project.

The design of the pump is that of a diffusion pump; the design was chosen because it was simple to fabricate and required no additional parts like check valves to make a functioning system. The nozzle and diffuser of the pump are within the typical geometries of working diffusion pumps [12], [39], but the pressure differential is incorrect. Table 5 summarizes the various evaluated in this project. The parameters varied are the angle of the nozzle/diffuser (theta), the length of the nozzle/diffuser (from chamber to inlet/outlet), and the depth of the channels.

Table 5: Summary of geometries attempted for the diffusion pump

<table>
<thead>
<tr>
<th>Geometry</th>
<th>Theta (degrees)</th>
<th>Length (mm)</th>
<th>Depth (um)</th>
</tr>
</thead>
<tbody>
<tr>
<td><img src="image1.png" alt="Image" /></td>
<td>10°</td>
<td>13</td>
<td>400</td>
</tr>
<tr>
<td><img src="image2.png" alt="Image" /></td>
<td>5°</td>
<td>13</td>
<td>400</td>
</tr>
<tr>
<td><img src="image3.png" alt="Image" /></td>
<td>10°</td>
<td>20</td>
<td>400</td>
</tr>
<tr>
<td><img src="image4.png" alt="Image" /></td>
<td>10°</td>
<td>20</td>
<td>140</td>
</tr>
</tbody>
</table>

The fluidic resistance of the nozzle and diffuser should be tuned to find the optimal pressure differential; this may involve making the nozzle and diffuser asymmetrical. In some samples, the fluid in the sample is only displaced through the nozzle, indicating that...
the fluidic resistance may be too high on the diffuser side. This could be corrected by making the diffuser with a higher theta than the nozzle, lowering the fluidic resistance. Other solutions include making the fluidic channels shallower, or making the nozzle and diffuser shallower and leaving the chamber at a greater depth.

A diffusion pump also relies on finding the correct frequency at which to actuate the membrane. The current actuation scheme uses an electromagnet that offers +/-15 mT (shown in Figure 33). In a pumping test, the sample is primed with DI water then the +/-15 mT alternating magnetic field is applied at a frequency varying from 1-50 Hz to try and find the pumping frequency. A stronger alternating magnetic field should be implemented, as the membrane has been verified in Chapter 4 to work well at applied magnetic fields of 19-110 mT. This will increase the deflection of the membrane, increase the stroke volume, and increase the pumping rate of the pump to the point where it can overcome the backflow and effectively pump water.

Finally, check valves would be an excellent and fast way to implement a pump with the membrane actuator. The high stroke volume demonstrated in Section 4.5 is evidence that the membrane would work as a check valve pump, because the check valves would prevent any backflow and stop the oscillating behaviour of the water observed during the pumping tests.

7.2. Valve

The valves are promising as flow switching valves. When an external magnetic field is applied to close one valve and open the other (via a permanent magnet), there is enough of a pressure difference to between paths to change the path of the fluid (shown in Figure 44) when the flow rate is below 5 µL/min.
Figure 44: The valves in flow switching configuration. The channels have been primed with blue DI water and then one valve is shut and red DI water is pumped in to test the flow control. A) The channels prior to the beginning of the flow switching test. B) The channels several minutes later, where the flow has switched and only flows through one path.

The valves have not been fully characterized as flow switching valves because of fabrication and testing challenges. The valves require a long time to prime and remove bubbles from the vertical channels between the layers. Generally, when fluid testing, bubbles can be removed by using a syringe pump to apply a negative pressure to quickly withdraw water through the channels. The geometry of the valve design means that the negative flow conditions provide the correct flow conditions for the valve to work as a normally-closed valve (as discussed in Section 5.2). When the valve is sealed under
negative pressure, the seal is extremely robust and no bubbles can be removed from the system as no fluid flows. Therefore, bubbles have to be removed using positive pressure, which must be delivered at lower flow rates of 5-50 µL/min, ramped up slowly from 5 µL up to 50 µL to prevent the device from bursting due to fluid pressure.

Frequently, a bubble would be stuck in the valve seat, which could not be seen through inspection due to being in an opaque layer, but would escape several hours into testing. The valves never seem to be quite equal, which was likely due to bubbles stuck in the valve seat. The bubble acts as a capacitor, so any bubbles in the valve would invalidate the testing results. Often only one of the two valves would seal completely in a flow switching test, though there were no blockages or other meaningful differences in device geometry.

Due to these testing issues, the valve has been extensively tested without delivery of significant results. The observation of the valve working as a normally closed valve is qualitative; the valve seals robustly when under the negative pressure conditions and can be opened with an applied magnetic field, but the pressure that the valve can overcome when an external magnetic field is applied must still be measured.

A scheme to integrate the valve design with the micropump presented in Section 5.1 could also be developed so that the pump and valve could be used together (as motivated in Section 2.7), which is a long term goal that fits into the future work of the µROAMS project.

7.3. M-OSTE+

Magnetic composite OSTE+ (M-OSTE+) is an exciting new material, with many avenues for characterization. As has been laid out by previously developed M-CPs [62], there are several key steps that need to be taken to characterize it fully. The M-OSTE+ must first be characterized as a material and then as an actuator. This begins with the analysis of doping percentages to determine the maximum percentage of rare earth magnetic particles that can be suspended in OSTE+ before the mechanical properties of the polymer resin (such as the Young’s Modulus) are adversely affected. The analysis of the doping percentages is essential because there is an optimal doping ratio at which the magnetic concentration is high, but the matrix polymer still has the desired properties such
as flexibility. An additional consideration for M-OSTE+ is whether the doping percentage will affect the maximum viable thickness of the M-OSTE+. The opacity of the magnetic particles suspended in the OSTE+ increases the UV exposure time required for the M-OSTE+ and therefore affect the maximum thickness attainable by M-OSTE+.

The M-OSTE+ at various doping concentrations should also be inspected under an SEM to evaluate the dispersion of magnetic particles in the matrix polymer and insure that the dispersion is uniform. If the magnetic particles in the matrix polymer is not uniform when viewed under an SEM, then the dispersion of magnetic particles can be improved by suspending the magnetic particles in a solution prior to mixing into the matrix polymer. The magnetic characteristics of the M-OSTE+ at varying doping concentrations should also be measured using a SQUID (superconducting quantum interference device) and the effect of magnetizing the M-OSTE+ after the first cure versus after the second cure should be analysed.

OSTE+ can be bonded to many different substrates. An investigation must be made into M-OSTE+’s ability to bond to different substrates and whether M-OSTE+ can be embedded into undoped OSTE+ should be made. Magnetic PDMS integrates easily into PDMS structures and can readily be embedded in PDMS; OSTE+ and M-OSTE+ should be investigated for the same properties. Doping of rigid OSTE+ (OSTEMER 322 Crystal Clear) with rare earth magnetic particles should also be investigated, as having rigid polymer magnets that integrate easily into OSTE+ microfluidic devices could result in useful geometries.

The undoped OSTE+ can be injection moulded [18], but the M-OSTE+ may face more challenges. As observed in Section 6.2, the viscosity of the M-OSTE+ increases once both components of the polymer resin have been mixed. This may prevent the M-OSTE+ from being injection moulded and the cause of the viscosity change should be determined and minimized. Once the viscosity change is minimized, a test design should be made for an actuator that could be injection moulded using M-OSTE+. This test design should incorporate all of the knowledge gained in the previously described investigations. The M-OSTE+ test actuator will then have an optimum thickness, magnetic particle concentration, and fabrication method integrates/interfaces with other polymers.
Chapter 8. Conclusion

This thesis focusses on the fabrication of magnetic composite polymer actuators for microfluidic systems and creating M-CP actuator designs that can cross the gap between microfluidics in academia and industry. The contributions of this thesis are realized through advancements in fabrication methods, actuator designs, and material development.

The fabrication methods developed enable faster fabrication of PDMS/M-CP structures and facile hybrid bonding of M-CP/PMMA devices. The new micromould method of sticker moulds presented in Chapter 3 is ultra-rapid and extremely low cost, requiring no UV-photolithography equipment for rapid soft lithography. Chapter 4 demonstrates the fabrication flow of a hybrid M-CP/PDMS/PMMA device, which shows that M-CP microfluidic actuators can be integrated with commercially-relevant thermoplastic materials in a simple, highly repeatable method.

The actuator designs for magnetic polymer membranes presented in Chapter 4 have a powerful stroke volume of 0.2-7.4 µL at an applied magnetic field of ±19mT to ±110 mT. With large volumes of water displaced at a low magnetic field, the membrane actuator has the potential to be used as a pump or a valve. Designs and initial characterizations of a diffusion pump and a normally closed valve are presented in Chapter 5, without any change in membrane geometry between the two actuators. Fabrication flows for the pump actuator are done as hybrid devices, with PMMA substrates, which demonstrates the M-CP membrane working with commercial substrate and scalable fabrication methods. The valve design is presented as an all-PDMS device to show the possibilities of developing the material in academic settings and uses the sticker moulds presented in Chapter 3, verifying their viability and reliability for fast fabrication in research applications. The future work needed to fully characterized these devices as elements of a commercial microfluidics system is described in Chapter 7.

A new magnetic polymer, an off-thiol-ene-epoxy doped with rare earth magnetic particles, is investigated in this thesis as an avenue to injection moulding flexible magnetic polymer devices. Chapter 6 shows the characterization of the physical and functional properties an off-thiol-ene-epoxy (OSTE+) polymer resin doped with rare-earth magnetic powder (Nd₀.7 Ce₀.3)₁₀.₈Fe₈₃.₉B₅.₆. Further characterization of the M-OSTE+ magnetic and
mechanical properties is needed to show its full potential as a microfluidic actuator material, as is described in Chapter 7.

This thesis makes important steps in the transition of magnetic composite polymer actuators from academic labs to commercial mass manufacture by presenting hybrid devices, scalable fabrication methods, and an alternative magnetic polymer that is injection mouldable. Addressing the gap between academic microfluidic research and commercial microfluidic products is challenging, but valuable. This thesis aids the transition of exciting new technology and materials from a small lab to wide audience.
References


A. Khosla, “MICROPATTERNABLE MULTIFUNCTIONAL NANOCOMPOSITE POLYMERS FOR FLEXIBLE SOFT MEMS APPLICATIONS.”


