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A microfluidic actuator based on thermoresponsive hydrogels

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Abstract

We have evaluated the potential use of thermoresponsive hydrogels based on *N*-isopropylacrylamide as actuators in microfluidic and lab-on-a-chip devices. This required fabrication of hydrogel actuators on the μm length scale, anisotropic swelling of the resulting materials, and control over the kinetics of the hydrogel volume phase transition. The fabrication procedure combined gel polymerization and casting techniques from the life sciences with more traditional semiconductor fabrication protocols for spin-coating, patterning, and etching. The actuator design used a PDMS membrane to separate the hydrogel actuator from the microfluidic channel and a separate reservoir for fluid to swell the actuator. As a result, the actuator could control flow for organic as well as aqueous solutions over a wide range of pH and ionic strength. The presence of a fixed substrate causes the gel swelling to be highly anisotropic, and the actuating motion is perpendicular to the substrate. The anisotropic swelling also limits the degree of swelling for the responsive hydrogel, and the total volume change is lower than the corresponding bulk materials by as much as an order of magnitude. The resulting actuators conform easily to the shape of the microfluidic channel, and the rate of the hydrogel response could be increased by using a series of semi-interpenetrating hydrogel networks. The microfluidic channels ranged in diameter from 180 to 380 μm, and the typical actuator was between 100 to 500 μm in diameter. The time scale of the actuator response was approximated by fitting with a single exponential ($\sim \exp[-t/\tau]$). The time scale (τ) varied as a function of the actuator size and composition, ranging from 10 min to less than 10 s.

Keywords: Polymer materials; Hydrogels; Actuators

1. Introduction

Hydrogels are known to respond to a variety of stimuli such as pH [1], temperature [2], and electric field [3]. They can be made to imbibe and expel aqueous solutions, causing the volume of the polymer gel to change dramatically. We have explored the use of thermoresponsive *N*-isopropylacrylamide (NIPAAm) gels as actuators in microfluidic channels. Poly(NIPAAm) has a lower critical solution temperature (LCST) in water upon heating above 32 °C [4], and much is known about these materials because of their extensive use as biomaterials [5] and in drug delivery applications [6]. However, NIPAAm gels are only responsive in the presence of aqueous solutions with relatively low ionic strength, which significantly limits the types of fluids that can be manipulated [7]. Hydrogels can swell with the process fluid to block the channel, but there is no net change in total volume (polymer plus process fluid). Therefore, valving action can be achieved through the high polymer-solvent friction [8], but hydrogels inside a fluidic channel cannot be used to displace fluid.

The volume phase transition can be described theoretically as a balance between mixing free energy, rubber elasticity free energy, and, in the case of ionizable comonomers, the osmotic contribution of the counterions [9–13]. The degree of swelling increases dramatically with the incorporation of ionizable comonomers, leading to the use of hydrogels as 'superabsorbers.' Moreover, the transition temperature increases as the ionized groups make the polymer more hydrophilic [14]. When used as an actuator, the swelling of the hydrogel is confined to one dimension, perpendicular to the substrate. Detailed studies of this anisotropic swelling behavior have shown that the degree of swelling and the phase transition temperature are distinctly different from the corresponding bulk gels [15–18]. The stress distribution in the hydrogel layer also

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changes and has been described as a function of the osmotic pressure of the ionizable groups [19].

The rate of the volume phase transition is known to scale with the square of the gel dimension and should therefore increase dramatically with microscaling [18,20], but the time scale is also a function of the gel composition, microstructure, and surface properties [21–23]. There is a difference in the response to increasing and decreasing temperature as well, with the collapse of the gel network being significantly faster than the swelling process [21]. The rate of water diffusion in the network is known to be a limiting factor, and this can be improved by creating macroporous or inhomogeneous network structures [24–27]. The network tends to collapse at the surface first, creating a hydrophobic layer that slows the diffusion of water from the core of the hydrogel [28], and approaches to increasing the rate of hydrogel response are often related to reducing the formation of the surface layer or improving transport through the layer.

Interpenetrating (IPNs) and semi-interpenetrating (semi-IPNs) polymer networks consist of both linear and crosslinked polymers that are polymerized simultaneously or sequentially. Semi-IPNs with NIPAAm as the crosslinked component [29,30] and NIPAAm hydrogels with comb-grafted oligomers [31,32] have been shown to accelerate the collapse of the gel network. The concentration and molecular weight of the comb-grafted oligomers is lower than the linear polymers in the semi-IPNs, but the mechanism of the linear polymers is thought to be similar. On the one hand, hydrophilic oligomers such as poly(ethylene glycol) and poly(acrylamide) are swollen under all system conditions and are thought to act by providing water-release channels through the hydrophobic surface layer and facilitating the diffusion of water in the network [29,33]. On the other hand, responsive oligomers such as poly(NIPAAm) and poly(*N*,*N*-diethylacrylamide) can be swollen or collapsed depending on the system conditions. They can nucleate the hydrophobic aggregation throughout the network and are thought to result in a more uniform network collapse by reducing the formation of the surface hydrophobic layer [33,34]. A more detailed analysis of the kinetics of the semi-IPN volume phase transition will be published separately [30], and here we will focus on the use of these materials in microfluidic actuators.

Microfluidic and lab-on-a-chip devices lend themselves to the manipulation of small sample volumes, which increases the rate of analysis while decreasing the amount of reagents required [35]. These systems are therefore compatible with biochemical and medical analysis, using flow control that is often based on electrokinetic principles. Mechanical valves and pumps have been designed for these applications and are usually incorporated through traditional semiconductor fabrication techniques [36,37]. Responsive hydrogels have also been incorporated in microfluidic devices as part of switchable supports and separation schemes [38,39] and as biomimetic actuators and control elements that require no external stimulus [40–43]. Existing studies of responsive hydrogels in the bulk suggest that these materials should be well-suited for applications in microfluidic systems, but it is not clear whether this can be extrapolated to μm-scale hydrogels when the volume phase transition is confined to one dimension. There has also been no direct comparison between the properties of hydrogels in the bulk and hydrogel actuators incorporated in a microfluidic channel. The motivation of the present study is to evaluate responsive hydrogels as building materials for microfluidic systems using several criteria: the ease of fabrication of the actuators, the kinetics of the volume phase transition as a function of gel size and composition, the ability of the actuators to block and displace the flow of different fluids, anisotropic swelling of the hydrogel, and the response to different stimuli.

2. Experimental section

2.1. Materials

N-Isopropylacrylamide (NIPAAm; Aldrich) was purified by recrystallization from hexane and dried in vacuum, and *N*,*N*-dimethylacrylamide (DMAAm; Aldrich) and *N*,*N*-diethylacrylamide (DEAAm; Polysciences, Inc.) were distilled under vacuum to remove inhibitor. *N*-Methylacrylamide (MAAm; ABCR), *N*-ethylacrylamide (EAAm; ABCR), acrylamide (AAm; Life Technologies), sodium acrylate (SA; Aldrich), ammonium persulfate (APS; Aldrich), *N*,*N*‘methylenebisacrylamide (BIS crosslinker; Aldrich), *N*,*N*,*N*′,*N*′-tetramethylethylenediamine (TEMED; Aldrich), and γ-methacryloxypropyltrimethoxysilane (bind silane; Sigma) were of the highest available purity and were used as received.

2.2. Swelling experiments

Responsive hydrogel layers were formed from free-radical polymerization of NIPAAm as the thermoresponsive component, SA as the ionizable comonomer, and BIS as the crosslinker, initiated by APS and TEMED. Thin films of NIPAAm gel were cast using Teflon spacers as described previously [44]. The refractive index and film thickness of the hydrogel layer were measured simultaneously, using a combination of surface plasmon resonance (SPR) and optical waveguide spectroscopy (OWS) in the Kretchmann configuration, as described previously [17,18,45]. The refractive index and film thickness results could then be used to find the swelling ratio and polymer volume fraction of the hydrogel layers. The SPR substrates were LaSFN9 glass coated with a 50 nm gold layer. A silica layer was formed on the SPR substrate, which was treated with bind silane to form a covalent bond between the hydrogel layer and the substrate [44]. The swelling behavior of the hydrogel layer was examined as a function of temperature by placing the samples in a flow-through cell that was connected to a peristaltic pump and a temperature controller.

Bulk hydrogel samples were prepared by free-radical polymerization as described above, except that the pre-gel solution was poured into a test tube containing glass capillaries (inside diameter 0.95 mm). The solution was bubbled with nitrogen, and 0.4 mg/mL APS was added to initiate the reaction. After 24 h, the gels were removed from the capillaries and allowed to equilibrate in a large volume of MilliQ water to extract any unreacted reagents. The degree of swelling was determined by measuring the diameter of the resulting cylindrical gels, using an optical microscope with a water-filled sample holder and an Instec HCS400STC200 stage to control the temperature.

2.3. Microfluidic actuators

A fluid channel was defined in glass using standard procedures [46]. Borofloat glass was coated with a 1000 Å thick layer of amorphous silicon, which was coated with photoresist, exposed with the channel pattern, and developed. A plasma etcher was used to etch through the amorphous silicon layer, and a wet etch using 49% HF was used to create the microchannels in the glass. The photoresist was stripped, and the remaining amorphous silicon layer was removed. This becomes the bottom plate, which contains all of the microfluidic channels and will be described in the following section, and a drill press was used to form access holes in each end of the microchannel.

The actuator assembly was made by first clamping the support plate to a blank plate, separated by thin Teflon spacers (Fig. 1(A)). A 1% solution of agarose in water was first melted by heating above 85 °C and then introduced as a sacrificial interlayer between the support plate and the blank plate. Capillary action filled this space, as well as the holes in the support plate, and the assembly was cooled, allowing the agarose gel to solidify (Fig. 1(B)). The PDMS membrane was spin-coated onto the surface of the support plate and cured (Fig. 1(C)). The PDMS was Sylgard 184, which can be diluted up to 40% with hexane to adjust the viscosity and therefore the thickness of the resulting PDMS membrane. The curing conditions vary as a function of the hexane concentration but typically ranged from 2 to 5 h at 80 °C. Membranes ranged from 15 to 60 μm thick with a typical thickness being 25 μm. The sacrificial interlayer was removed by immersing the assembly in 90 °C water for 5 min to melt the agarose gel. The blank plate was removed, leaving the support plate with the PDMS membrane (Fig. 1(D)). The pre-gel solution containing the APS initiator was added in the existing hole and allowed to polymerize (Fig. 1(E)). For smaller actuators, photopatterning could be used to control the size of the actuator. The glass could also be treated with bind silane so that the hydrogel slabs are covalently linked to the glass surface, but with the design outlined here, the PDMS membrane and support plate were sufficient to keep the hydrogel in place. The support plate was clamped to the top plate, again separated by thin Teflon spacers (Fig. 1(F)). The top plate could be blank, resistive heaters could be patterned onto the top plate, using standard microfabrication methods, or resistive NiCr heating wire could be incorporated into the actuator assembly. The fluid reservoir layer was filled by capillary action (Fig. 1(G)), and the assembly was mounted onto the bottom plate containing the fluid channel (Fig. 1(H)). The fluid reservoir could be filled with any porous, hydrophilic substance; in this case we used acrylamide gel. The resulting hydrogel actuator is shown in Fig. 2(A).

2.4. Kinetics experiments

The kinetics of the volume phase transition in hydrogel actuators was measured by monitoring the flow rate in the channel. The hydrogel actuator was placed on top of a heat sink, and resistive heaters were used to trigger individual actuators. Therefore, an actuator that was not actively heated was cooled along with the rest of the assembly. The temperature inside the channel could not be measured directly, but a reference experiment measuring the flow rate in the channel as a function of the current across the resistive

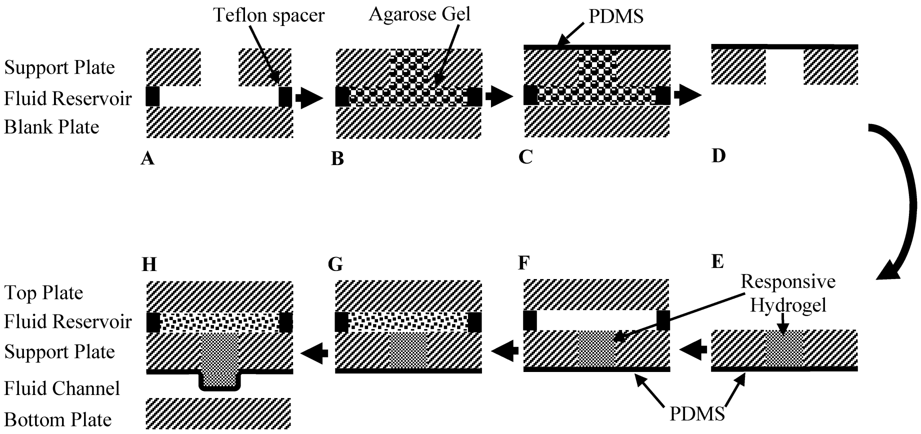


Fig. 1. An example of process flow for the fabrication of the hydrogel-PDMS actuator.

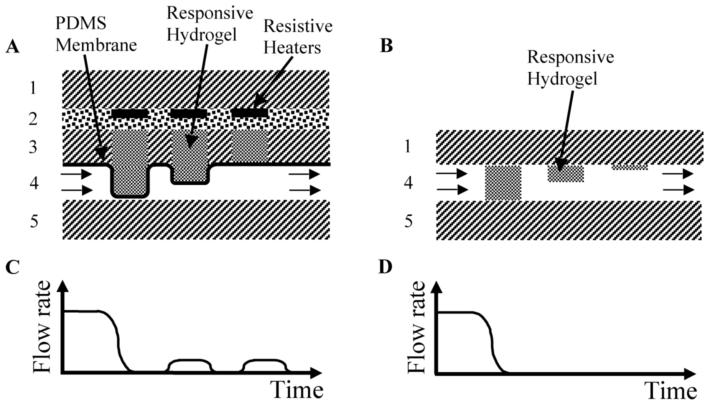


Fig. 2. Designs for hydrogel actuators (A) swollen by a separate fluid reservoir and (B) swollen by the fluid in the microfluidic channel. The assembly consists of (1) a top plate, which is either blank or contains the resistive heaters; (2) a porous reservoir layer; (3) a support plate that contains microslabs of responsive hydrogels and is coated on one side with a flexible PDMS membrane, which is in direct contact with the fluid being impelled in the channel; (4) a microchannel that contains the fluid being manipulated and is patterned on (5) the bottom plate. The hydrogels can shrink and swell in response to a small change in temperature, and the actuator design either displaces or blocks fluid as indicated (C, D) for three consecutive actuators being triggered. Resistive heaters can also be used for design B but are not necessary for proper functioning of the actuator.

heater allowed us to estimate the temperature relative to the volume phase transition temperature. Pressure-driven flow was then used with a step-change in temperature to observe the rate of swelling and deswelling for hydrogel actuators in a microfluidic channel. Swelling of the hydrogel actuators can also be observed directly by separating the actuator from the fluidic channel and imaging the swelling of the actuator, as shown in Fig. 3. The kinetics data for both bulk hydrogels and hydrogel actuators was approximated by fitting the diameter of the cylindrical bulk gel or the degree of swelling of the hydrogel actuator as a single exponential ($\sim \exp[-t/\tau]$).

The kinetics for hydrogel actuators based on semi-IPNs were compared to the corresponding bulk materials. All these semi-IPNs contain NIPAAm as the crosslinked component and different linear polymers synthesized from the *N*-alkyl acrylamides in Scheme 1. The samples are therefore named according to the corresponding linear component. This series of linear polymers allows us to gradually make the polymers more hydrophobic and, in the case of NIPAAm and DEAAm, thermoresponsive. This preparation and characterization has been described previously for bulk semi-IPNs [29], and cylindrical samples were used as described in Section 2.2. Again using the microscope and heating stage, the kinetics of the volume phase transition was observed by measuring the change in the hydrogel diameter in response to step changes in temperature. The temperature was changed from 22 to 50 °C for deswelling experiments and from 50 to 22 °C for swelling experiments. Because the time required for the step change in temperature was approximately 10 s, the first data point was taken at 10 s.

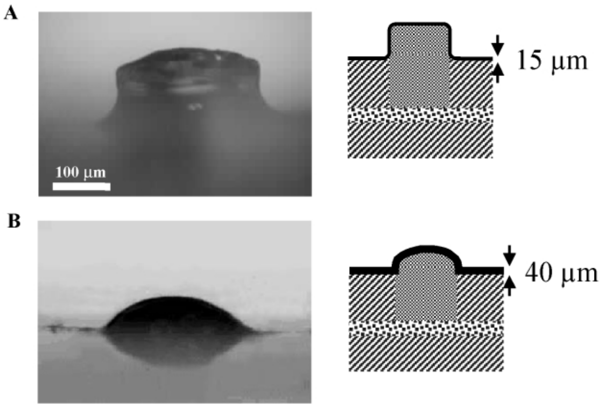
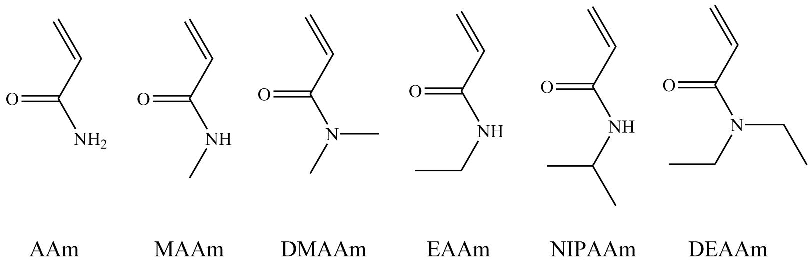


Fig. 3. The effect of PDMS thickness on the swelling of a hydrogel actuator. (A) A 15 μm PDMS membrane conforms to the shape of the hydrogel slab, while (B) a 40 μm PDMS membrane limits the swelling of the hydrogel slab to a spherical cap. The hydrogel composition is 684 mM NIPAAm, 16 mM SA, and 6.8 mM BIS and is swollen with DI water from the reservoir layer.

3. Results

3.1. Actuator design

Responsive hydrogels are attractive for use as microfluidic actuators due to their ease of fabrication, response to a wide range of stimuli, and fast response rate with microscaling. Most of the existing literature focuses on specific applications [42,43,47], while the present study will focus on how the polymeric materials affect the performance of the resulting device. NIPAAm gels are only responsive in a limited range of fluids and the actuator cannot be used to displace fluid with which it is in direct contact. We have developed a simple method of microfabricating gel actuators that avoids these problems; an example of this process flow is shown in Fig. 1. A silicone elastomer membrane is used to separate the hydrogel actuator from the fluid being impelled, and a separate reservoir provides the aqueous solution that allows the gel to swell. The hydrogel response is a function of the fluid with which it is in contact, and with a separate fluid reservoir, the



Scheme 1. Monomers used for semi-IPN linear polymer synthesis.

hydrogel actuation is independent of the fluid in the channel. As a result, it should be possible to manipulate fluid with a wide range of compositions, and the actuator can then be used to displace fluid in the microchannel as it swells with fluid from the reservoir (see Fig. 2(A)).

This approach can be compared to designs that place the hydrogel actuator in direct contact with the fluid in the channel (see Fig. 2(B)). When the hydrogel swells from the fluid in the channel, it is a direct function of the fluid composition. The advantage of this approach is that no external trigger is needed, but it also significantly limits the types of fluids that can be manipulated. There are several examples of these actuators in the literature [40,41,48,49] as well as responsive polymers incorporated into separation schemes for microfluidic devices [38,39]. There are also examples of microfluidic systems incorporating elastomeric membranes in combination with pneumatic actuators [50, 51]. However, in the case of hydrogel actuators, the actuator motion can be either passive, responding to the temperature of the fluid, or active, responding to local temperature change from a resistive heater. The differences between the two designs are highlighted in Fig. 2(C) and (D), showing the response in flow rate to three consecutive actuators. The first actuator will block flow for both examples, but only the PDMS membrane will displace fluid, acting as a microfluidic peristaltic pump [47]. Because of the ability to displace fluid, the potential applications of these actuators include pumping and metering. Also, because the hydrogel and the fluid being manipulated are not in direct contact, the response will be the same, regardless of the type of fluid being manipulated.

The material properties of the hydrogel and the PDMS affect how the actuator functions. In particular, the PDMS thickness changes the profile of the actuator, as shown in Fig. 3. A thicker PDMS membrane will give a spherical cap, while a thinner PDMS membrane conforms to the shape of the hydrogel, giving a more square profile. The membranes used for these experiments ranged from 15 to 60 μm. Thinner membranes did not withstand repeated cycling of the resulting actuator, while thicker membranes could not be displaced sufficiently by the swelling of the hydrogel slabs. We are using these actuators with hemicylindrical channels, making the spherical profile more appropriate. However, microfluidic systems made from polymeric materials are often machined or patterned using soft lithography, producing square channels, and for these applications the square profile would be a better fit. In the example shown here, the PDMS membrane covers one wall of the channel, which is compatible with polymeric microfluidic systems and pressure-driven flow. For applications using electroosmotic flow, the glass walls of the channel bear a surface charge, which determines the fluid flow properties. In this case, the PDMS membrane would disrupt the flow in other regions of the microchannel system; thus, an alternate process flow may be needed to place PDMS only at the actuator [47].

3.2. Hydrogel valve

For different sizes and shapes of channels, the fact that the PDMS-hydrogel actuator is conformable should help in blocking the channel. This was tested by intentionally making the actuator too large for the channel (see Fig. 4). A 500 μm valve was used to close channels ranging in size from 180 to 380 μm. The valve is large, which makes its rate of response too slow for most microfluidic applications. However, the 500 μm valve is able to block 90% of a 380 μm channel, and the data clearly shows that this valve is sufficiently conformable to close off a significant portion of these smaller channels. A 250 μm valve was successfully

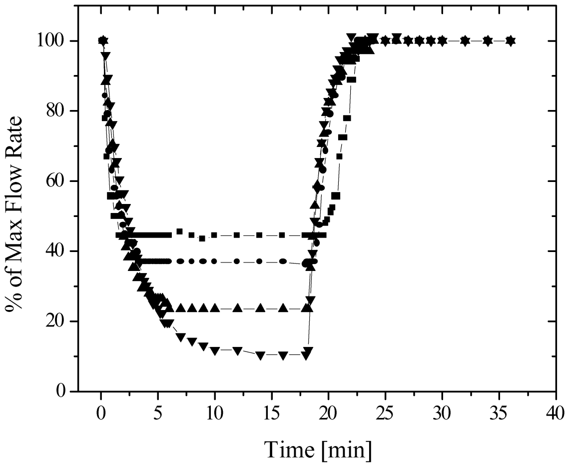


Fig. 4. An intentional mismatch of the valve and the channel dimensions to test the conformability of the hydrogel actuator. The valve is 500 μm in diameter with corresponding slow kinetics, but the actuator is able to partially block channels with a diameter of 180 μm (■), 230 μm (●), 280 μm (▲), and 380 μm (▼).

used with the 230 and 280 µm channels (see Fig. 5(A)), and these valves appear to function properly, as defined by full closure of the channel, in spite of the difference in size. This provides flexibility in the design of these actuators, but the rate of the actuator response appears to be a function of the channel diameter, which is likely due to force exerted on the hydrogel as it conforms to the shape of the channel. Again, this example uses hemicylindrical channels, but the conformable actuators should work well with square channels as well, which further simplifies the fabrication process.

The transition temperature and swelling behavior of NIPAAm hydrogels are known to change with hydrostatic pressures on the order of 1000 psi [44], and the design presented here may not be appropriate for use with high-pressure systems. Because the applied pressure could actually cause the hydrogel to collapse, a maximum back pressure was not determined for this actuator design. However, lower pressures, such as those generated by a syringe pump, do not appear to affect the performance of the valve (see Fig. 5(A)). Instead, the force generated by the PDMS membrane appears to be a limiting factor, with a thicker PDMS membrane slowing the rate of response and limiting the swelling of the hydrogel slab (see Fig. 5(B)). The actuator with the 60 µm membrane swells more slowly and is not able to fully close the 280 µm channel, indicating that the hydrogel is not able to sufficiently displace the membrane. However, the thickness of the membrane is also approaching the diameter of the channel, and it is possible that a 60 µm membrane is simply too thick to conform to a 280 µm channel. The relative size of the valve and the channel does have some effect on the rate of response (see Fig. 5(A)), and while a 250 µm actuator can be used to block both a 230 and 280 µm diameter channel, the response for the 280 µm channel is slightly faster.

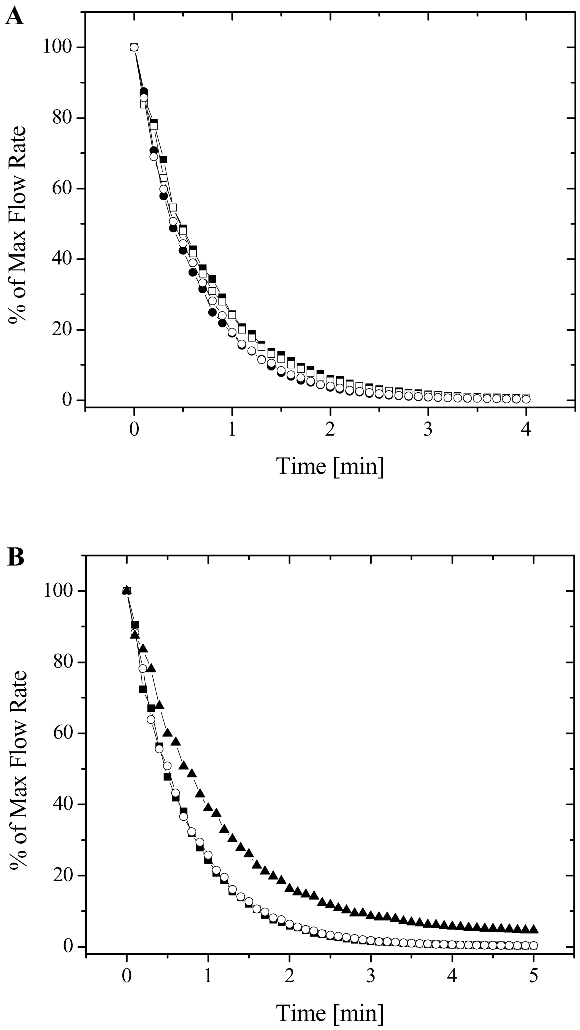


Fig. 5. (A) A 250 µm actuator is used to block channels with a diameter of 230 µm (□,■) and 280 µm (○,•) and a 25 µm PDMS membrane. The pressure-driven flow is from constant hydrostatic pressure (0.9 psi, open symbols) and a syringe pump (274 psi, closed symbols). (B) A 250 µm actuator is used to block a 280 µm channel with PDMS membrane thicknesses of 15 µm (■), 25 µm (○), and 60 µm (▲).

The PDMS-hydrogel actuator design is very versatile and has been tested with a wide range of solvents as shown in Table 1. The successful tests represent actuators that were able to function as a valve in response to repeated temperature cycling, while the failures represent actuators that were closed for all temperatures. Hexane and toluene are good solvents for PDMS, and it is likely that the swelling of the membrane by these solvents caused the channel to be blocked. Therefore, it appears that this approach excludes the use of solvents that significantly swell or degrade the silicone elastomer membrane.

3.3. Kinetics of the volume phase transition

The time scale of the volume phase transition scales with the square of the gel dimension, and this has been shown to hold for gels swelling in three dimensions over a wide range of length scales [18,20,21]. However, the response of gel layers swelling anisotropically under constraint is limited by the lateral diffusion of water and no longer obeys the expected scaling law [52]. For the geometry of the PDMS-hydrogel actuator, the time scale (τ) of swelling and deswelling was measured over a wide range of length scales and varies with the gel dimension to the 1.85 power (see Fig. 6). Water is only able to diffuse in and out of the gel through one side, and this lower exponent suggests that

Table 1
Test of different solvents with PDMS-hydrogel valve

Success	Failure
Buffer pH 3	Hexane
Buffer pH 7	Toluene
Buffer pH 10	
DI water	
Acetone	
Methanol	
Ethanol	

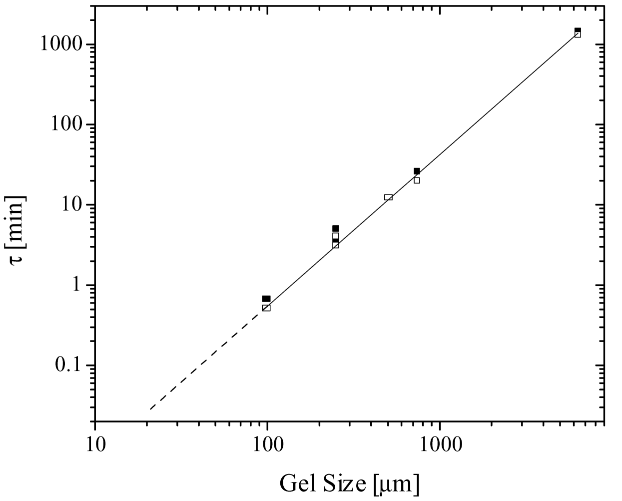


Fig. 6. The time scale (τ) of the volume phase transition as a function of the gel actuator size (d). The time scale of the actuator response is approximated by fitting with a single exponential ($\sim \exp[-t/\tau]$), and the kinetics are shown in response to a step change in temperature—increasing temperature for deswelling (□) and decreasing temperature for swelling (■). The solid line shows the scaling of the hydrogel response as $\tau \sim d^{1.85}$, and the extrapolation to even smaller actuator dimension suggests that the actuator design should be appropriate for most microfluidic applications. The hydrogel composition is 684 mM NIPAAm, 16 mM SA, and 6.8 mM BIS with a 25 μm PDMS membrane, and the swelling is with DI water from the reservoir layer.

the swelling and collapse of the actuator is diffusion-limited. Bulk hydrogels have significant hysteresis between the rates of swelling and deswelling, with the swelling typically being an order of magnitude faster than the deswelling [21]. By contrast, as shown in Fig. 6, the time scales of swelling and deswelling are similar for the gel actuator, which simplifies the application of these actuators in peristaltic pumping.

We have studied the effect of semi-IPNs on the kinetics of the volume phase transition for bulk hydrogels and how this relates to the hydrogel morphology, which was imaged using confocal microscopy [30]. Similar to comb-grafted oligomers, hydrophilic linear polymers are thought to act as water-release channels, while responsive linear polymers allow rapid hydrophobic aggregation throughout the network and a more uniform network collapse [33,34]. The same set of semi-IPNs was used to create actuators, and the kinetics of the volume phase transition was tested in a microfluidic channel. The monomers used for the semi-IPN linear component are shown in Scheme 1, and the resulting linear polymers range in properties from the most hydrophilic (AAm) to the most hydrophobic (DEAAm). The two most hydrophobic polymers have an LCST (NIPAAm 32 °C, DEAAm 25 °C), and we will refer to these polymers as 'responsive.' In addition to being relatively hydrophobic, these polymers change their solubility in water in response to changes in temperature. Some portion of the linear polymers are likely to be activated by chain transfer during the free-radical polymerization of the crosslinked component. Previous studies of this effect have shown that anywhere from 20 to 40% of the linear polymer remains trapped within the network after leaching of the linear polymers [29]. Therefore, the effect of the linear polymer is both as a porogen and as a linear component of the resulting semi-IPN. However, the leaching process does not affect the kinetics of the volume phase transition, and this suggests that the kinetics are more strongly affected by the trapped linear polymers than by the linear polymers acting as porogens [29].

For bulk hydrogels, the time scale of response clearly increases with increasing size of the hydrophobic *N*-alkyl group on the linear polymer, with the exception of NIPAAm (see Fig. 7). In this case, the linear and crosslinked components of the semi-IPN are the same, and the two components will respond simultaneously to the step change in temperature. The cooperative mobility between the linear and crosslinked components may be responsible for the fast rate of response for this particular sample [30]. However, this effect does not appear to accelerate the kinetics of the volume phase transition for hydrogels used as actuators. The more hydrophilic linear polymers act by providing a hydrophilic channel that aids in the diffusion of water out of the gel, and has a similar effect for both bulk hydrogels and gel actuators. The fabrication of the hydrogel actuators does not include a leaching step to remove unreacted linear polymer, but because leaching does not affect the kinetics of the bulk semi-IPNs, we do not expect this to be a significant factor in the comparison of the two sets of samples. Therefore, hydrogel actuators based on semi-IPNs have a fast response due to the improved diffusion of water out of the gel, which is related to the hydrophilic nature of the linear polymer, and this same mechanism appears to be present in the corresponding bulk hydrogels. However, bulk

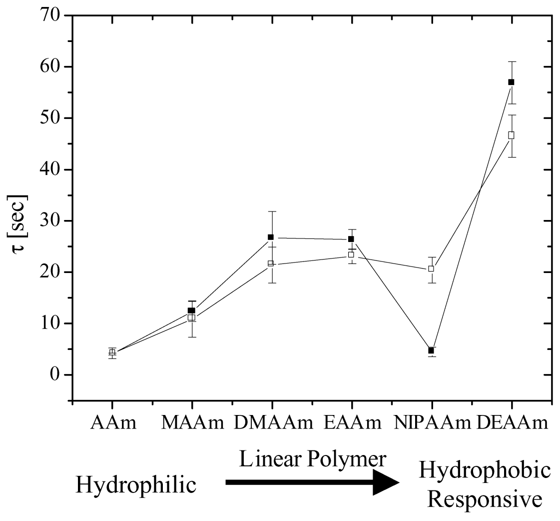


Fig. 7. Effect of semi-IPN composition on the time scale (τ) of the volume phase transition. The crosslinked network composition is 600 mM NIPAAm and 6.8 mM BIS with 100 mM of the linear polymer as labeled. The kinetics of cylindrical bulk hydrogels (■) is compared to that of the PDMS-hydrogel actuators (□).

hydrogels also have fast response due to the nucleation of collapse in the gel core, which is related to the responsive nature of the linear polymer, and this mechanism does not appear to affect the hydrogel actuators.

4. Discussion

The response of a hydrogel actuator should be on the order of seconds for a microfluidic system, and this seems possible for actuators smaller than 100 μm, as suggested by the extrapolation of the data in Fig. 6. However, the rate of response can also be controlled by forming semi-IPNs with NIPAAm as the crosslinked component, with the fastest response being for linear polymers that are either hydrophilic or responsive [29,53]. We have studied this approach in detail for bulk hydrogels [30], and the results for both bulk hydrogels and gel actuators are shown in Fig. 7. For hydrogel actuators of the same size, the time scale of response can be increased or decreased by the choice of linear polymer and crosslinked network composition. The combination of faster kinetics with smaller gel dimensions and the design of semi-IPNs should therefore allow the design of actuators with the appropriate rate of response.

An interesting feature of the valve assemblies with smaller dimensions is that the time required for the gel polymerization reaction to reach the gel point is significantly longer for the fabrication of smaller actuators. Since the gelation takes place in the presence of the hydrophobic PDMS membrane, this is likely related to a hydrophobic substrate effect [54]. Similar effects have been seen with 2-acrylamide-2-methyl-1-propanesulfonic acid (AMPS), acrylic acid, DMAAm, and NIPAAm hydrogels in the presence of substrates with low surface energies [23,55,56]. During free-radical polymerization, oxygen trapped at the hydrophobic substrate is thought to retard the polymerization reaction, and this creates an inhibition of gelation near the substrate [57]. For larger actuators, this inhibition is likely present but is limited to a small region near the PDMS membrane. As the dimensions of the hydrogel actuator become smaller, this interfacial region becomes more significant and the apparent gelation kinetics slows down.

The total volume change as a hydrogel swells is related to the osmotic pressure of the ionizable comonomers. For bulk hydrogels, this volume change can be on the order of 100 times the dry volume of the hydrogel [14]. The transition temperature also increases as the ionized groups make the polymer more hydrophilic, and this has been observed for constrained hydrogel layers as well as bulk hydrogels [44, 58]. However, the related swelling is not seen for hydrogel layers, and the total volume change is much less than the corresponding bulk gels [44]. We have calculated the total volume change in bulk hydrogels with different degrees of ionization and compared these to hydrogel layers with the same chemical composition (see Fig. 8). Both sets of samples are crosslinked in the swollen state, but the hydrogel layers were studied with a combination of SPR and OWS while the bulk hydrogels were observed with optical microscopy. Increasing the total volume change can be a significant advantage to applications in microfluidic systems. However, the addition of ionizable comonomers has no effect on the total volume change in hydrogel layers, as shown in Fig. 8.

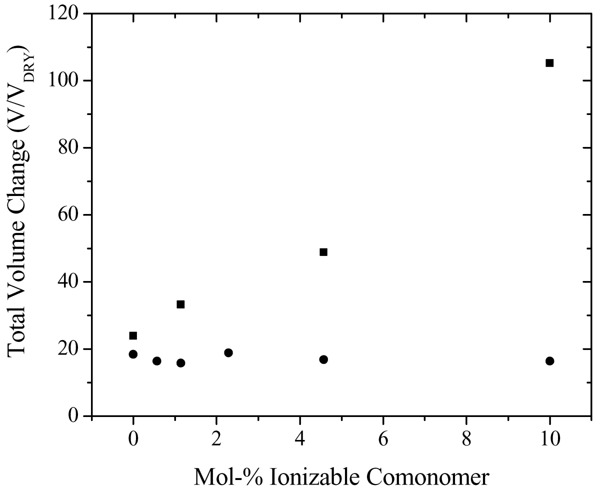


Fig. 8. The effect of ionizable comonomers on the total volume change in bulk hydrogels and hydrogel layers. The bulk hydrogels (■) and the corresponding hydrogel layers (•) consist of NIPAAm with different mol% of SA. The total volume change of the volume phase transition is estimated by the ratio of the gel volume (V) at temperatures below the volume phase transition to the volume in the dry state (V_{DRY}).

Previous measurements on hydrogel layers that are photocrosslinked in the dry state also show no effect of ionizable comonomers on the total volume change, which is even lower than the corresponding hydrogel layers that are crosslinked in the swollen state [58]. Instead, the total volume change is a function of reference state (crosslinking in the dry or swollen state) and the crosslinking density [18, 44]. The anisotropic swelling of hydrogel layers has been modeled, and the degree of swelling is limited by the lateral stresses in the hydrogel layer [19,59,60]. For thicker layers of hydrogels based on strong polyelectrolytes, the lateral stresses can cause the hydrogel layer to buckle, and in this case the degree of swelling can approach that of bulk hydrogels [19]. However, the hydrogel layers in this study have a dry film thickness less than 10 μm, the degree of ionization is at most 10%, and therefore, the total volume change is much lower than the corresponding bulk hydrogels.

5. Conclusion

We have introduced a design for microfluidic hydrogel actuators that incorporates a PDMS membrane and swells the hydrogel with fluid from a separate reservoir located above the microfluidic channel. This design allows us to manipulate a wide range of fluids, excluding only those that might significantly swell or degrade the PDMS membrane. The volume phase transition of responsive hydrogels is well understood in the bulk, but it is not clear if this can be applied to the anisotropic swelling of the hydrogel actuator in an actual device. We have used this design to study the behavior of responsive hydrogel actuators, focusing on the scaling behavior of the kinetics, the effect of semi-IPNs, and the degree of swelling. The scaling behavior is similar to bulk gels with the added advantage that the swelling and deswelling kinetics are very similar. The effect of semi-IPNs containing hydrophilic linear polymers is similar to that in the bulk, but the effect of responsive linear polymers is significantly reduced. Finally, the effect of ionizable comonomers is to increase the transition temperature as in the bulk, but the associated swelling due to the osmotic pressure of the counterions is not seen.

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