01 Jun 2009

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Recommended Citation

K. H. Henthorn, "A Rapid Method for Flow-ready Cylindrical Microchannel Fabrication," *Review of Scientific Instruments*, American Institute of Physics (AIP), Jun 2009.
The definitive version is available at https://doi.org/10.1063/1.3152365

A rapid method for flow-ready cylindrical microchannel fabrication

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Most current methods of microchannel fabrication are expensive and time consuming or result in noncylindrical channels, which is undesirable for many applications. A rapid inexpensive method for the fabrication of flow-ready cylindrical polymer microchannels from polydimethylsiloxane (PDMS) is presented here. In this method, cylindrical needles are used as molds for the channels and also function as flow delivery devices after fabrication. Validation of channel function and smoothness can be accomplished by comparing experimental data to theoretical models. One model was previously developed by the author to predict the incipient motion of a glass sphere in contact with a perfectly smooth PDMS surface and deviations from the model are expected to be a function of channel roughness. The data collected in the present channels show fairly good agreement with the theoretical model, indicating a relatively smooth and consistent surface. In addition, detailed SEM images of the channel showed that the internal surface was qualitatively very smooth. © *2009 American Institute of Physics.* [DOI: 10.1063/1.3152365]

The need for microchannel fabrication has grown tremendously over the past several years due to the rapid increase in miniaturized devices and interest in microscale flows. Microchannels with circular cross sections are gaining popularity; their applications are particularly relevant in biomedical engineering, including microcirculation studies, flow cytometry, microsensor development, and drug delivery processes. Many methods of rapid microchannel fabrication exist, but due to the inherent nature of these methods (such as soft lithography), most result in noncylindrical channels. To create cylindrical microchannels using these techniques, it is typically necessary to join two halves to create a complete circular cross section. This method requires careful alignment of the upper and lower halves and may result in fluid leakage and flow disturbances due to the existence of seams or edges. Other fabrication methods, including dual-beam laser etching,[1] allow for the production of cylindrical microchannels, but these methods are very time consuming and extremely expensive.

One rapid, inexpensive method of cylindrical microchannel fabrication was proposed by Perry *et al.*,[2] where glass rods were used as a channel mold for polydimethylsiloxane (PDMS) polymer. They were able to fabricate uniform cylindrical channels ranging from 57 to 250 μm in diameter and demonstrated their functionality by flowing suspensions of red blood cells through them. However, once the glass rods were removed during fabrication, it was necessary to attach the resulting microchannels to a flow system in order to introduce fluids into the channels. The insertion of syringe needles or tubing into a small opening without damaging the polymer at the inlet or outlet is a tedious task and may result in leakage if the two connecting parts vary significantly in size.

The method described in this paper allows for the rapid, inexpensive fabrication of cylindrical flow-ready PDMS microchannels. The use of fluid delivery needles as the microchannel mold eliminates potential leakage problems associated with gaps due to mismatched sizes of microchannels and connecting tubing or needles. In addition, damage to these connecting sites is minimized because the microchannels are immediately flow ready.

The method discussed in this article involves molding a polymer around needles that ultimately serve as the fluid delivery device. Although many polymers are suitable for this application, PDMS and a polystyrene-(polyethylene-polybutylene)-polystyrene triblock copolymer (SEBS) were used here. PDMS is widely used in microchannel applications due to its desirable optical properties, biocompatibility, and low cost. The PDMS mixture used in this work consisted of a SYLGARD 184 silicone elastomer base and curing agent (World Precision Instruments, Sarasota, FL) that were mixed in a ratio of ten parts base to one part curing agent by mass. The uncured PDMS mixture was allowed to solidify in a vacuum chamber for 24 h. SEBS is an inexpensive thermoplastic elastomer gel that has the ability to be heated and reformed.[3] For this material, a 30 wt % SEBS resin (Candle Factory Co., Hialeah Gardens, FL) and 70 wt % mineral oil (white, light, Sigma-Aldrich) mixture was created and held under a vacuum for approximately 12–14 h to ensure an even distribution of oil on the solid resin surface. The mixture was then heated to approximately 150 °C until the polymer was completely melted (approximately 2–4 h) and then brought to room temperature to solidify. Both PDMS and SEBS are hydrophobic as prepared.

The microchannel diameter is determined by the outside diameter of the fluid delivery needle. For this work, a 3 in. long needle (110 μm i.d. and 230 μm o.d., Popper and Sons, Lincoln, RI) was partially inserted into a 1 in. long needle (260 μm i.d. and 460 μm o.d., Hamilton Co., Reno, NV) through two small holes on opposite ends of a smooth rectangular metal mold [Fig. 1(a)]. For applications involving particle injections, such as studies of detachment mechanisms or flow around obstructions, the shorter needle is used to create an injection port. If an injection port is not desired,
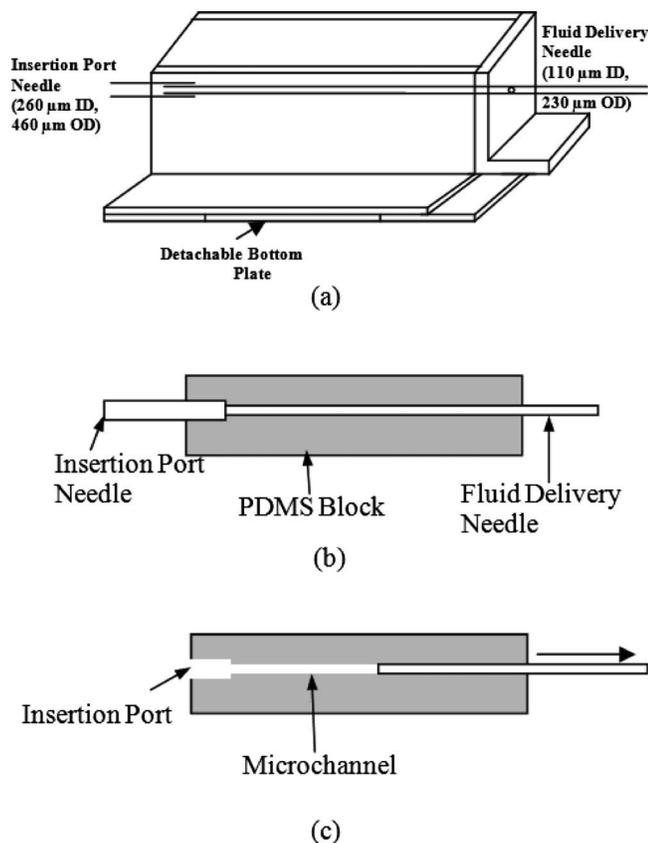
FIG. 1. Fabrication method. (a) Configuration of the metal mold and needles; (b) uncured polymer is poured around the needles and cured, and (c) the insertion port needle is removed and the delivery needle is partially removed (the metal mold is omitted here for clarity).

FIG. 2. Junction between the microchannel and the insertion port (a) SEBS with polymer residue blockage (b) PDMS with no blockage.

a single long needle can be threaded through both openings. Uncured polymer was poured into the mold so that the needles were completely submerged [Fig. 1(b)]. Once the polymer was allowed to cure, a detachable bottom plate was removed to leave an opening which served as a viewing window for the inverted microscope. It was critical to remove the bottom plate first to avoid misalignment of the microchannel with respect to the outlet hole. The shorter needle was then completely removed from the solidified polymer, leaving a 460 μm diameter microchannel to serve as an insertion port. The longer needle was partially removed, leaving a 230 μm i.d. flow-ready microchannel [Fig. 1(c)].

Channels fabricated from SEBS were highly irreproducible, mainly due to polymer residue remaining in the channels after the removal of the needles. This residue was found near the transition region between the insertion port and microchannel, and was likely due to seepage of the uncured polymer into the 30 μm annular space between the two needles during fabrication. This polymer residue dislodged when the needles were pulled out and would often permanently block the channel. This residue is clearly seen in Fig. 2(a), which is an SEM image of the transition region for the cross section of an SEBS channel. Though the solution to this problem may be as simple as sealing the space between the two needles with a material such as petroleum jelly before introducing the polymer, it was a concern that a secondary substance might introduce additional complications by reacting with the polymer or insufficiently blocking the
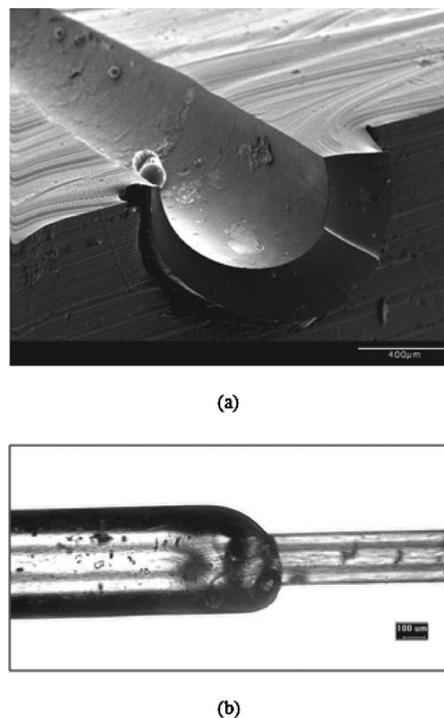
clearance due to probable changes in rheological properties at higher temperatures. Therefore, it was concluded that SEBS was not a suitable choice of polymer for the current method of fabrication.

Like the SEBS microchannels, most of the PDMS channels fabricated with this method retained some residual polymer at the junction point. However, in most cases the residue was easily removed by flushing water inside the channel at a relatively high flow rate [Fig. 2(b)]. This offered a considerable advantage over the SEBS channels where it was practically impossible to remove the residual polymer deposited inside the channel. The analyses that follow all pertain to PDMS microchannels.

Figure 3 shows SEM images of PDMS microchannel cross sections at two different magnifications to give a qualitative measure of the surface roughness characteristics of the inside wall of the microchannel. This characterization was important since roughness affects both the flow profile of the fluid and particle-wall interactions. These images indicate that the extent of roughness was minimal and it was assumed that these small features would not affect the flow characteristics or particle adhesion. The stainless steel needles used to mold the microchannels were very smooth, and the process of removing them from the cured polymer did not appear to introduce imperfections from tearing or rubbing.

The fluid velocity required to detach a single glass microsphere from the surface of the microchannel was experimentally measured to further validate the channel smoothness and test the performance of the microchannels fabricated by the proposed process.[4] An aqueous 0.15 g/L Tween 80 solution was flowed through the empty microchannel by means of a microinfusion pump connected to the fluid delivery needle in order to purge air from the system. A single glass sphere (Mo-Sci Corp., Rolla, MO) was intro-
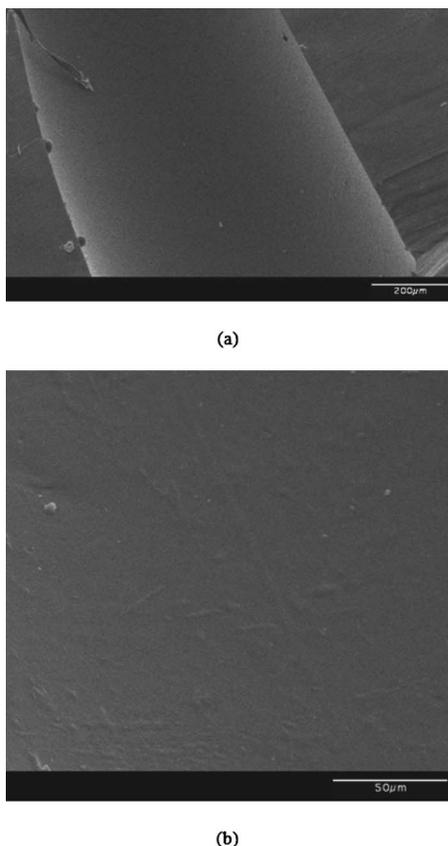
(a)

(b)

FIG. 3. SEM images of PDMS microchannel cross sections at two magnifications.

duced into the channel via the insertion port. The fluid was slowly withdrawn and pumped until the particle was positioned near the middle of the channel to nullify any entrance and end effects (Fig. 4). The fluid was pumped into the channel at a very low flowrate ($\sim$0.01 μL/min) to avoid initial entrainment of the particle by the fluid. The initial position of the particle was noted and the velocity of the fluid was increased gradually in a step-wise manner until some displace-
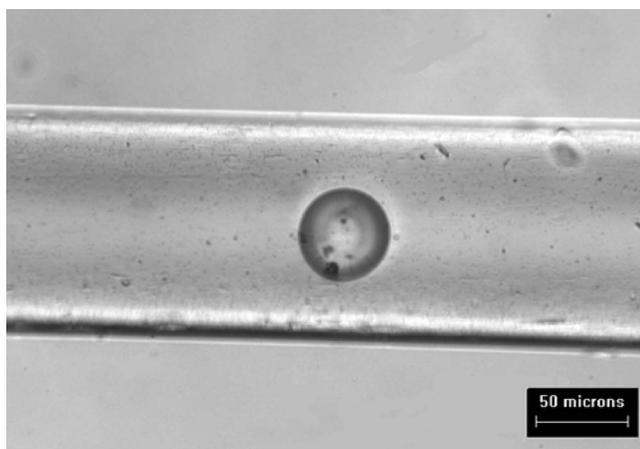

FIG. 4. Single glass sphere ($\sim$50 μm) positioned inside the 230 μm microchannel.

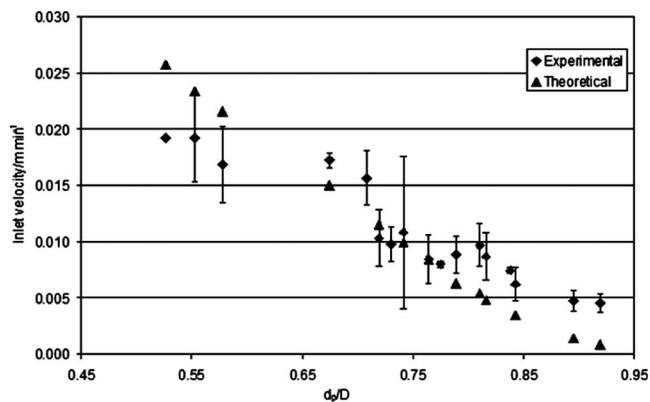FIG. 5. Effect of relative particle diameter on inlet velocity at incipient motion in a 230 μm PDMS microchannel (Ref. 4).

ment of the particle was observed. Particle movement was monitored using a microscope coupled with a charge coupled device camera integrated to a computer monitor. The minimum fluid velocity that caused the particle displacement was taken as the detachment velocity for the particle.

The experiment was repeated with glass spheres of various sizes by varying the ratio of the particle diameter to the channel diameter d_p/D (Fig. 5). These data were compared to a theoretical model previously developed by the author; this model was based on a simple moment balance on a single, large sphere resting on the bottom of a perfectly smooth cylindrical microchannel.[4] As can be seen in Fig. 5, the magnitude of the experimental data is in fairly good agreement with the values predicted by the model, indicating a relatively smooth and consistent surface. Deviations from the model are expected to be a moderate function of channel roughness, because the contact area between the particle and the channel surface increases as channel roughness increases, directly affecting the particle-wall adhesion term.

A new method to rapidly produce inexpensive, flow-ready cylindrical microchannels has been described here. This method eliminates some of the problems associated with other cylindrical microchannel fabrication techniques, including misalignment and seaming of joined half-channels. In addition, the integration of fluid delivery needles into the mold reduces damage to the channel that can result from the insertion and removal of tubing during use. SEM images of microchannels fabricated with the new technique show that surface roughness is qualitatively minimal. Experimental measurements of particle detachment behavior agreed well with predicted values, further supporting the claim that the effect of channel roughness is negligible.

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