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We explore the three-dimensional capabilities, resolution, and optical clarity of microfluidic devices fabricated by stereolithography using a mail-order service, and we compare the cost and prototyping speed of soft lithography with those of stereolithography.
493x134mm (96 x 96 DPI)

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ARTICLE

Mail-Order Microfluidics: Evaluation of Stereolithography for the Production of Microfluidic Devices

Cite this: DOI: 10.1039/x0xx00000x

Received 6th December 2013,
Accepted 00th January 2012

DOI: 10.1039/x0xx00000x

www.rsc.org/

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The vast majority of microfluidic devices are developed in PDMS by molding ("soft lithography") because PDMS is an inexpensive material, has physicochemical properties that are well suited for biomedical and physical sciences applications, and design cycle lengths are generally adequate for prototype development. However, PDMS molding is tediously slow and thus cannot provide the high- or medium-volume production required for the commercialization of devices. While high-throughput plastic molding techniques (e.g. injection molding) exist, the exorbitant cost of the molds and/or the equipment can be a serious obstacle for device commercialization, especially for small startups. High-volume production is not required to reach niche markets such as clinical trials, biomedical research supplies, customized research equipment, and classroom projects. Crucially, both PDMS and plastic molding are layer-by-layer techniques where each layer is produced as a result of physicochemical processes not specified in the initial photomask(s) and where the final device requires assembly by bonding, all resulting in a cost that is very hard to predict at the start of the project. By contrast, stereolithography (SL) is an automated fabrication technique that allows for the production of quasi-arbitrary 3D shapes in a single polymeric material at medium-volume throughputs (ranging from a single part to hundreds of parts). Importantly, SL devices can be designed between several groups using CAD tools, conveniently ordered by mail, and their cost precisely predicted via a web interface. Here we evaluate the resolution of an SL mail-order service and the main causes of resolution loss; the optical clarity of the devices and how to address the lack of clarity for imaging in the channels; and the future role that SL could play in the commercialization of microfluidic devices.

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Electronic Supplementary Information (ESI) available: [details of any supplementary information available should be included here]. See DOI: 10.1039/b000000x/

Introduction

Understandably, most microfluidic researchers – including our group – are under a "PDMS spell": PDMS is inexpensive, optically clear, biocompatible, and can be molded using methods that can be safely used even by kindergarteners. PDMS has its Achilles' heel, though: PDMS molding is a manual process that is difficult to automate and is slow – too slow for producing large numbers of copies of a device per day, as required of any commercialization process.

For that reason, most commercial microfluidic devices are developed in plastic (a very inexpensive, transparent material) by injection molding, a technique that requires a large expenditure for producing metal molds capable of withstanding high pressures. (Glass etching is still used to make many glass microfluidic devices that pump fluids through their channels by electroosmotic flow because the surface charges on plastic are not as favorable to electroosmosis as those on glass; however, glass etching is comparably more expensive than plastic molding as a fabrication method.) The general procedure for producing a PDMS or plastic device is schematized in Fig. 1a, starting with the master mold (usually fabricated via photolithography or CNC milling), the molding process, and a bonding step to close the channels. Advanced fluid routing and functionalities often require multiple PDMS layers to be molded and aligned with one another, a manual process which suffers from dependence on individual skill and poor reproducibility[1]. The European microfluidic plastic molding industry has made great advances in reducing costs by converging to a common mold format based on the microscope slide (which can be very limiting for certain applications); even so, the cost of a very simple mold is in the range of $15,000[2]

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and an intricate three-layer device can easily reach $100,000. Alternatives to injection molding for microfluidics have been implemented successfully, such as thermoforming[3] (which heats up a thin layer of plastic until it conforms to a master mold with vacuum application) or hot embossing[4] (which presses a hot mold onto the surface of the plastic), but compared to injection molding these are limited by the range of features that can be created and by much lower throughputs.

In any case, a startup company that is interested in producing the commercial form of a PDMS or glass prototype will have to develop a plastic-molding process that is entirely different than the original soft lithographic one, as depicted schematically in Fig. 1b. This problem is not extremely critical for simple one-layer devices but it can be a major challenge for multi-layer devices featuring vias and interconnects, as with many designs featuring a mixer module, and it can become a practical issue for laboratories that design several devices a year (e.g., the Principal Investigator does not have time to start several startups a year). Even when a novel microfluidic assay outperforms an existing clinical assay, potential investors are not easily convinced that the profits obtained from the future sales of the devices can recover the initial investment in a reasonable time. In addition, microfluidic devices are dramatically absent from markets that demand small production volumes, such as in biomedical research instrumentation, clinical trials, classroom projects, and customized devices. Not surprisingly, despite the enormous hopes that researchers and the public have placed on microfluidics as an enabling technology for biotechnology, only a small number of companies are actually producing microfluidic devices. Additional factors limiting the commercialization of microfluidic devices include successful integration of device components and obtaining regulatory approval[5, 6].

Stereolithography (SL) is an established technique for producing 3D polymer structures from a liquid photopolymer resin by means of a focused laser or LED light source[7] (Fig. 1c). (SL is a form of 3D printing or rapid prototyping; other forms of 3D printing, such as laser sintering, inkjet printing, and thermoplastic extrusion[8], are not as adequate for microfluidic fabrication.) Microchannels are defined by polymerizing the walls of the channel cavities and subsequently draining the uncured photopolymer precursor[9]. Note that no alignment or bonding is necessary to produce 3D structures, which substantially simplifies the processing with respect to soft lithography (especially for complex devices). Many mail-order 3D printing companies offer SL services. As opposed to plastic microfluidic devices produced by injection molding or other methods (which often began development as PDMS devices and were re-designed into plastic), with SL services the design process is directly initiated in plastic, optimized in plastic, and can seamlessly be transferred to a (startup) company once it is suitable for production (in plastic), as shown schematically in Fig. 1d. For this paper we built all of our devices in Somos® WaterShed XC 11122, a hard, biocompatible transparent plastic. Here we address resolution, materials, optical clarity and cost concerns that have until now prevented many microfluidic researchers from adopting SL as a route to facilitate the commercialization of their devices. We also discuss the salient features of SL compared to soft lithography as a rapid prototyping tool.

Methods

3D CAD designs of the devices were drafted in Autodesk Inventor (San Rafael, California) and converted to .stl format. For our device-integrated female Luer connectors, an open-source "Fitting Luer Lock" design was downloaded from GrabCAD[10] and modified such that a constant 5.842 mm outer diameter extended 5.842 mm below the threaded portion of the connector. The bottoms of the connectors were then fused with the inlet and outlet microchannels of the devices to create the final CAD designs.

The devices were built by FineLine Prototyping, Inc. (Raleigh, North Carolina) on a 3D Systems Viper SL system (Rock Hill, South Carolina) in High-Resolution Mode with Natural finish, corresponding to 100 µm XY resolution determined by the laser beam diameter and 50-µm Z-steps. DSM Somos® WaterShed XC 11122 resin (Heerlen, Netherlands) was used as the resin due to its properties of being nearly colorless, not swelling in water, and meeting biocompatibility standards ISO 10993-5 Cytotoxicity, ISO 10993-10 Sensitization, ISO 10993-10 Irritation, and USP Class VI. For some devices, a proprietary process called "Substrate Build Style Option", in which the FineLine Prototyping devices are built on top of a flat smooth substrate, was used to enhance the optical clarity of the bottom surface of the devices.

Surface roughness measurements were taken using a KLA Tencor P-15 stylus profilometer (Milpitas, California). Microchannel width measurements were taken using a Nikon Eclipse Ti inverted microscope (Melville, New York).

Results

Three-dimensional capabilities of stereolithography

By convention the bottom surface of the vat of fluid resin defines the XY plane and the orthogonal direction defines the Z direction in SL. Amongst the greatest advantages of SL over soft lithography for device prototyping is the ability to arbitrarily define structures in a fully 3D space, with no significant increase in fabrication complexity and time for prototyping. In soft lithography, pseudo-3D devices are fabricated by stacking and assembling multiple individual PDMS molds together, each of which may only contain features on one or two Z-planes, depending on whether the mold is created against only a bottom master or both a top and bottom master. Additionally, a given master may only contain a small number of feature heights, and each feature height must be defined by individual photoresist spin and exposure steps. Furthermore, non-converging overhanging features cannot be prototyped using PDMS soft lithography[7]. In contrast, SL devices are built layer by layer in a single piece, with feature geometries only limited by machine resolution. With SL, a device with fluid routing through multiple Z-planes (Fig. 2a) can be built with the same relative ease and cost as a device with fluid routing through only a single Z-plane (Fig. 2b). Using integrated female Luer connectors, tubing can easily be reversibly interfaced with devices using barbed adapters (Fig. 2b). This also allows syringes to be interfaced directly with the devices and a wide variety of commercially-available Luer components including check valves and on/off valves can be connected to the device inlets and outlets for added functionality. Furthermore, with SL we are able to easily build void areas in a device's footprint to reduce printing costs and increase optical clarity or mechanical accessibility (Fig. 2c), design channels and chambers featuring optical access through

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one of its sidewalls (Fig. 2b-d), overlap channel features in separate Z-planes (Fig. 2a and e), and place Luer connectors in lateral, non-moldable configurations (Fig. 2f).

Resolution

Although commercial SL systems have laser beam diameters on the order of 100 µm and Z-step sizes on the order of 50 µm, during the SL fabrication process the microchannel walls trap uncured liquid resin which must be drained out of the device prior to a final UV curing step[11]. This "hydrodynamic" limitation ends up being more important than the laser's limitations. The SL service we used, which advertises experience in building microfluidic devices, only commits to clearing microchannels that are 500 µm × 500 µm or 635 µm x 635 µm depending on overall device complexity.

The female Luer connectors built into the devices not only simplify the interfacing of external fluid sources with the devices but also facilitate the draining of liquid resin from the microchannels and rinsing of the microchannels with solvents. Taking advantage of the integrated Luer connectors during the cleaning process, a complete seal can be formed between a syringe or reservoir of cleaning solvent and the Luer inlets; thus, the pressure applied to the source of cleaning solvent is fully transmitted through the Luer inlets into the microchannels for efficient clearing of uncured resin, regardless of microchannel geometry.

In order to assess the resolution limits of the SL service, we printed and evaluated a resolution test device which contains recessed trenches, recessed and raised patterns (a circle, a square, a star, and a set of parallel lines of various spacings, all 500 µm-deep), and closed microchannels of varying widths (Fig. 3). The microchannels, measuring 500 µm in height and between 50 and 1000 µm in width were arranged radially on the device originating from a shared female Luer connector, such that all of the microchannels could be filled with fluid from a single input, which could also be used to flush solvent through the device to clear resin from the microchannels. In this device the microchannels from 50 to 200 µm wide (Fig. 3d) could not be cleared of resin and thus sealed shut during post-curing. The 300 µm-wide channel (Fig. 3e) partially cleared of resin, with the majority of the channel having a final width of ~270 µm and a small constriction near the outlet having a width of ~180 µm. Finally, the microchannels from 400 to 1000 µm wide (Fig. 3f-i) exhibited a deviation from the expected width of less than 30 µm each. Three copies were built of a similar device, in which the 50, 100, and 200 µm-wide channels were removed. In these devices the measured channel widths matched the designed channel widths up to the limits of our measurement method using microscopy (Fig. 3j).

We stress that, in many microfluidic phenomena, the most important parameter is not the size of the channel but the Reynolds number (Re), which dictates the hydrodynamic regime. For example, fluid mixing in a (PDMS-molded) 100 µm-wide microchannel will be identical to fluid mixing in a (SL-printed) 500 µm-wide microchannel as long as the Re is kept constant in both. Since $Re = \rho v L/\mu$ where ρ and μ are the density and the kinematic viscosity of the fluid, respectively, v is the average fluid velocity, and L is the characteristic dimension (e.g. width of the channel), re-designing a PDMS-molded 100 µm-wide microchannel into a 500 µm-wide SL-printed microchannel would require that the average fluid velocity be slowed by a factor of 5. Therefore, it might be possible to print many PDMS devices at larger sizes by SL and still retain the same functionality.

Optical clarity

Laser over-curing. In SL, when the laser beam fabricates an overhanging feature in the Z-direction (such as a microchannel roof) it exhibits a well-documented roughness due to the laser beam over-curing beyond the plane of the desired feature[12, 13]. This roughness can be a hindrance for microscopy and microfluidic device operation. An example is shown in Fig. 3d-i. In order to assess the roughness of microchannel roofs, we measured the roughness of a 50 µm-deep trench built along the bottom of the resolution test devices. An average roughness of 2.54 µm was measured (see Fig. S1 in Supplementary Info). Given that the microchannels we have successfully built with SL are approximately 500 µm in height, this roughness corresponds to deviations of about 0.5% from the total channel heights. Thus, for these large channels, the roughness created by laser over-curing is mostly a concern for imaging rather than for device performance. (See discussion below regarding the disturbance of laminar flow in the microchannels.)

Refractive index-matching. While anisole (refractive index of 1.516-1.519) has previously been used to match the refractive index of WaterShed XC (1.512-1.515) and thus allow undesired features to be hidden during imaging[14], we found that the organic compound started dissolving and softening WaterShed XC noticeably over a 24 hour period. As an alternative, we found that Wacker AP 150 silicone oil (Sigma-Aldrich, St. Louis, Missouri), which has a refractive index of 1.510, also acts as a suitable match for WaterShed XC during imaging and does not dissolve WaterShed XC. By coating the outer surfaces of the devices during imaging, we are able to reduce the prominence of the roughness of the outer surfaces and enhance the visibility of the internal microchannels. This index-matching strategy is very effective in reducing the opacity of the rough features at the bottom of devices built without the Substrate Build Style Option (Fig. 4) and can also be used to reduce the prominence of ripples visible on device sidewalls (see below).

Build orientation. In order to remove the over-cured roof from the light path during microscopy, it is possible to build devices on-edge with SL (Fig. 5), which then places the over-cured features along the sidewall of a microchannel rather than on its roof. The roughness from over-curing is thus no longer visible during microscopy (Fig. 5e and f). However, as a trade-off, the layering of the outer surfaces of the device and of the effective microchannel roofs and ceilings, built as vertical sidewalls in the on-edge orientation, is visible. Because the region of photopolymer cured by the laser beam has a Gaussian or parabolic profile[15], vertical sidewalls built with SL are actually formed as a series of stacked parabolas. The visibility of the

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outer sidewall ripples can be reduced using the refractive index-matching strategy discussed previously (Fig. 5f). Additionally, features built at an angle to the build platform exhibit stair-stepping[16, 17] as the SL software must attempt to approximate a smooth surface despite the build process occurring in discrete Z-steps. As seen in Fig. 5b and c, laser over-curing affects the imaging of the flat build device while sidewall roughness affects the imaging and flow profile of the on-edge build device (Fig. 5e and f).

Cell imaging. In order to assess the feasibility of imaging cells within microchannels built with SL, CHO-K1 cells were stained with Calcein AM and seeded into the flat-built three-inlet device (Fig. 6a-c) and on a glass microscope slide (Fig. 6d-f). Discrete cells are clearly visible in the SL device at 20x magnification under both phase contrast (Fig. 6b) and fluorescence (Fig. 6c). Some amount of autofluorescence can be seen in the WaterShed XC material, which is present both above and below the microchannel containing cells in the optical path of the microscope. Due to the shallower depth of focus in inverted microscopy at higher magnifications, the roughness of the microchannel roofs is less visible at 20x magnification under an inverted microscope compared to under a stereoscope or an inverted microscope at lower magnification. WaterShed XC is a promising material for cell culture due to its existing biocompatibility certifications. However, long-term cell cultures in closed microchannels built using SL will require strategies to overcome the poor gas permeability of Watershed XC.

Discussion

Design modularity in soft lithography versus stereolithography

Sophisticated microelectronic devices such as smartphones are designed modularly by very large teams of engineers, whereas the vast majority of microfluidic devices are non-modular, being designed essentially by a single person. This lack of modularity is not by choice but a subordinate feature of the microfabrication process: in traditional microfluidic fabrication, the engineer must design a photomask and a series of physicochemical steps (PDMS molding or glass etching, after a photolithographic step) that are not well suited for packaging the device into a user-friendly, industry-standard unit (a "module", with standard connectors, etc.) because of the complex 3D geometries of these peripheral components. In SL, the design of the device is fully specified in CAD software. Very large CAD libraries of 3D digital objects – contributed by a community of designers – are available online for free[18]. Plastic modular 3D circuits with packaged connectors are trivially built with SL[19]. Importantly, the designs can be inspected and improved upon by another group or groups before printing. With SL, the microfluidic designer has no other choice but to fully specify the packaging (interface, connectors, etc.) prior to printing. We forecast that teams formed by mechanical engineers, industrial designers and interaction/experience designers will increasingly contribute to microfluidics to improve the functionality of the devices. Hence we envision a future dominated by SL printing where microfluidic designs will be available online and it will be possible to download them and integrate them as a module that will be connected to other devices or modules.

Rapid prototyping and cost in soft lithography versus stereolithography

The costs of a technique are rarely analyzed in engineering journals. An old tenet of the microfluidics field holds that soft lithography is an "inexpensive rapid prototyping technique". However, here we argue that these assertions might need to be revisited in light of what SL has to offer.

We start by noting that manufacturing cost and speed are, in general, inversely coupled (the faster the fabrication, the cheaper the product), so processes that are laborious and manual tend to be more expensive while processes that can be automated tend to be cheaper. Despite having been invented about 20 years ago, soft lithography remains a manual technique that requires long PDMS curing times. If the devices are multilayered, the alignment and bonding steps require great skill. An expert student or postdoc typically takes at least one day to finish a single-layer device, counting the photolithography steps.

Assuming an average base salary (or salary + tuition) cost of $40,000/year, the costs of a simple, very small soft lithographic device made in one day can be approximated, at a minimum (not counting clean room fees), by $160 (salary) + $10 (silicon wafer) + $30 (high-resolution photomask) + $5 (50 g of PDMS) + $10 (10 mL of SU-8 and 50 mL of SU-8 developer) = $215. These numbers, of course, vary geographically but the total always stays within a factor of 1.5 or less. Note that the materials and labor cost about 25% and 75% of the total, respectively; some of these costs can be spread over several devices if a single wafer is designed to contain several devices. For a multilayer device, the fabrication costs would typically double if two days are needed to complete the fabrication.

Generally, the time and cost requirements for producing a single SU-8 master mold and a single SL device are roughly equivalent. (Most of the devices shown in this paper cost ~$200.) For multilayer PDMS devices, additional master molds are needed which increase costs and production times in a roughly linear manner. However, many PDMS replicas can be molded from a single set of masters. With SL, additional devices cost ~20% of the first copy due to the ability to print many copies simultaneously on the same machine build area. While this cost per device remains relatively high, it is likely that it will decline rapidly in the near future with increasing competition as more SL patents expire over the next few years and SL desktop systems become widely available. More critically, while complex multi-layer PDMS devices require careful alignment from a trained engineer and are susceptible to poor reproducibility, SL devices maintain roughly the same costs and time requirements regardless of complexity and do not suffer from issues of device-to-device variability. Furthermore, for very complex devices with intricate architectures (which in PDMS would have to be built in multiple layers), it is much simpler and faster to troubleshoot the design with CAD software than to develop it in PDMS (where errors can arise that require re-design cycles).

Importantly, since SL is an automated fabrication technique, the costs of fabricating a given device can be predicted using automated algorithms. This feature is in contrast with traditional technologies for developing a plastic device, typically initiated with soft lithographic prototypes and later transferred to an injection molding, hot embossing or thermoforming process. These molding technologies do not allow for predicting cost, which can cause an escalation of debt in startups during their initial pre-commercialization phase. SL

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is an inherently cost-efficient technology because it is additive – materials are not etched so material waste is minimal – and because it allows for orders as small as one – it is even possible to interrupt production of a batch and update the design at no significant cost. Hence we believe that cost prediction with SL will prove to be a very valuable feature for controlling expenditures in plastic microdevice startups.

SL will become most attractive in low-volume production markets where plastic molding technologies are not able to reach due to the required enormous initial investments. First and foremost, SL will enable the production of devices for many clinical trials that would otherwise not have been produced. Second, the market of research supplies is presently dominated by large firms (Corning, BD, etc.) that design commercially-viable devices in plastic (petri dishes, bottles, etc.), but it need not be that way; with SL, the small designer will be able to compete with these large firms and commercialize all sorts of imaginative designs (e.g., gradient generators, connectors, blood filters, etc.) via a simple website. (Fashion and industrial designers are already using this model with other 3D-printing technologies[20-23].) Third, SL is very attractive as a microfluidics teaching tool because the design can be troubleshooted in the virtual space of a computer at little cost, printed inexpensively with a desktop SL system, and more time can be invested in the observation of an actual experiment compared to the lengthy safety and manual training of photolithography and soft lithography.

Desktop stereolithography versus mail-order stereolithography

The recent commercial availability of affordable desktop SL printers presents an interesting cost-saving alternative to the service-based model that we have evaluated in this paper. A comparison of desktop SL systems is shown in Table 1. The majority of desktop SL systems utilize Digital Light Processing (DLP)-based projection systems with a discrete pixel resolution. Thus, the build area of these systems is inversely proportional to the minimum feature size of the build. Despite this limitation on build area, these systems offer attractive cost-saving opportunities. Given the significantly lower costs of both projection and laser-based desktop SL systems, we are optimistic about their potential for microfluidics, but we have not tested their ability to print small channels. (Recall from above that "resolution" in SL, defined as the smallest printable feature, is not equivalent to the size of the smallest printable channel.) A clear advantage of owning a desktop SL machine is the possibility of printing on pre-processed surfaces (e.g. biochemically or nanopatterned surfaces, surfaces containing microelectrode circuits, etc.), which might be crucial for labs developing tissue engineering and biosensor applications.

However, here we argue that the choice between a desktop SL printer and a mail-order SL service is not only about cost saving but also about ease of access to the commercialization path. We note that a laboratory that uses taxpayer funds to buy an SL desktop system for the development of their own devices would not be able (by law) to sell the devices that have been printed with that desktop system. Microfluidic designers should learn from fashion and interior designers, who have long been using 3D printing services to "set up shop" on the web because, in their trade, they could not afford production and marketing technologies that would force them to a) produce large numbers of copies of an original; and b) sell through physical outlets. With a mail-order SL service, the microfluidic designer can focus on what (s)he does best (i.e. design), and leave all the fabrication, production and distribution to the SL service. In the near future, the microfluidic designer will not need to start up a company to commercialize his/her product(s). In this model, the microfluidic designer will use the SL service to optimize the device design and only when the device design is deemed optimal, then it will be made publicly available for sale with the click of a mouse. The microfluidic designer will be freed of the hassles of starting a new company for the fabrication and commercialization of each of his/her devices – including the process of fund raising, recruiting CEOs and attracting investors: the SL service is already set up for producing and selling the devices[20-23]. Although the option of "setting up shop" (pioneered by Shapeways) is still a minority, it seems to be increasing in popularity among various types of designers and it seems likely that it will become the norm in the 3D printing industry. We forecast that, just as fashion designers did in the recent past, microfluidic designers will also embrace it enthusiastically as soon as it becomes available to them.

Conclusions

Stereolithography (SL) is a rapid prototyping technique that can print in transparent biocompatible polymers at acceptable resolution for many microfluidics applications. Compared to soft lithography, SL is more convenient, faster, more cost-efficient, allows for producing 3D architectures that are not possible with PDMS molding, and can print at throughputs that are attractive to various commercialization routes (ranging from single-part buyers to large clinical trials). We clearly see from Fig. 1 that SL offers both a more user-friendly technology and a simpler commercialization path compared to plastic/PDMS molding. On the other hand, commercial SL systems have laser beam diameters of approximately 100 μm; therefore, devices with features smaller than this laser beam diameter cannot be readily fabricated using SL. Furthermore, the existing wide variety of PDMS microvalve and micropump designs might be difficult to replicate in plastic, so soft lithography may continue to be a dominant technique in microfluidic automation. Importantly, we believe that SL's automated fabrication will stimulate mail-order SL services to offer "set up shop" services (as offered already by several 3D printing services) that will provide a very facile route for the commercialization of plastic microdevices. There are two key advantages to this new commercialization process: 1) the designer does not need to order expensive molds and set up a company for launching production of his/her devices; and 2) the customer can order as few as one print (no "minimum quantities"). Since the resolution and the cost are both likely to improve (and some forms of valving might be possible with SL), we conclude that SL has the potential to displace soft lithography as the technique of choice for the fabrication of microfluidic devices that do not require extensive, high-density automation. Compared to injection molding, SL cannot yet achieve the very high throughputs and high volume production (with a few cents per part), so it is likely that injection molding will continue to dominate the later phases of commercialization.

For biomedical scientists in particular, microfluidics is a well-recognized enabling technology because it allows for the automated manipulation of sub-microliter volumes of biofluids. However, biologists and clinicians do not have routine access to the engineering expertise and costly equipment that is necessary to fabricate or use microfluidic devices. For most biology or medical laboratories, hiring an engineer is often not enough to cross over the "threshold of activation". Stereolithography, a

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"skill-less" fabrication technique, will allow biomedical scientists to have direct access to the "immediate manufacturing" of microfluidic devices. If batch fabrication techniques provided access to mass manufacturing, stereolithography will facilitate mass access to manufacturing. We envision a web-based microfluidics marketplace driven by a large number of designers where all designs are available for sale and produced by SL for a wide variety of low-volume applications.

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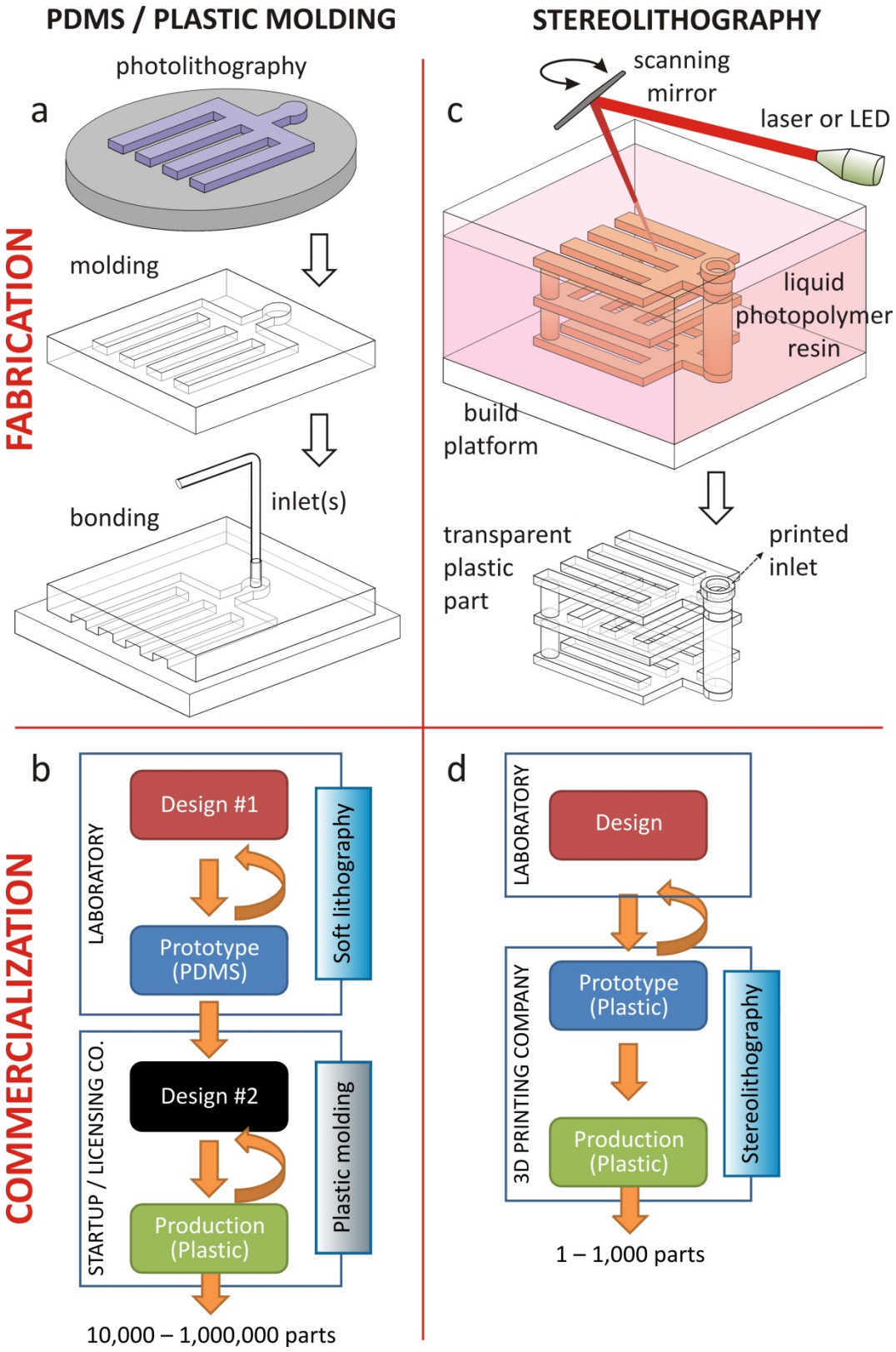

Fig. 1. Comparison between **(a,b)** PDMS/plastic molding and **(c, d)** stereolithography showing geometric design constraints and the path to commercialization.

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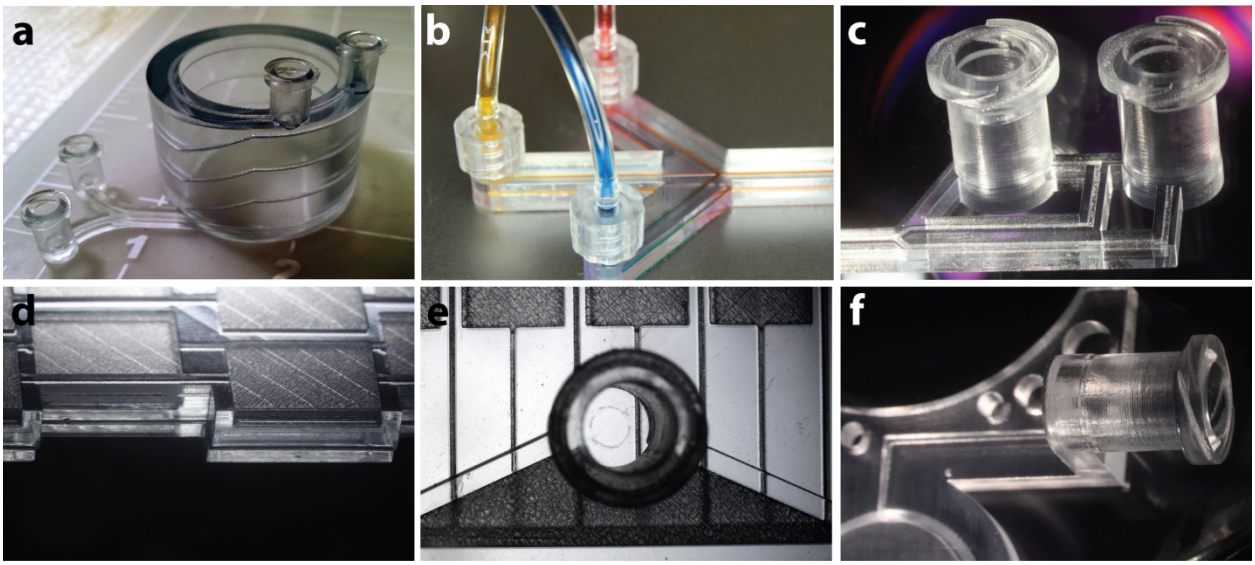

Fig. 2. Photographs of **(a)** a coil-shaped microchannel device and **(b)** the interfacing of tubing with integrated Luer connectors. Micrographs demonstrating **(c)** how void areas can be removed from the footprint of SL devices, **(d)** how closed chambers can be optically accessed from precisely-defined sidewalls of a device, **(e)** an inlet distributor with a folded geometry, and **(f)** a Luer inlet oriented parallel to the plane of the chip.

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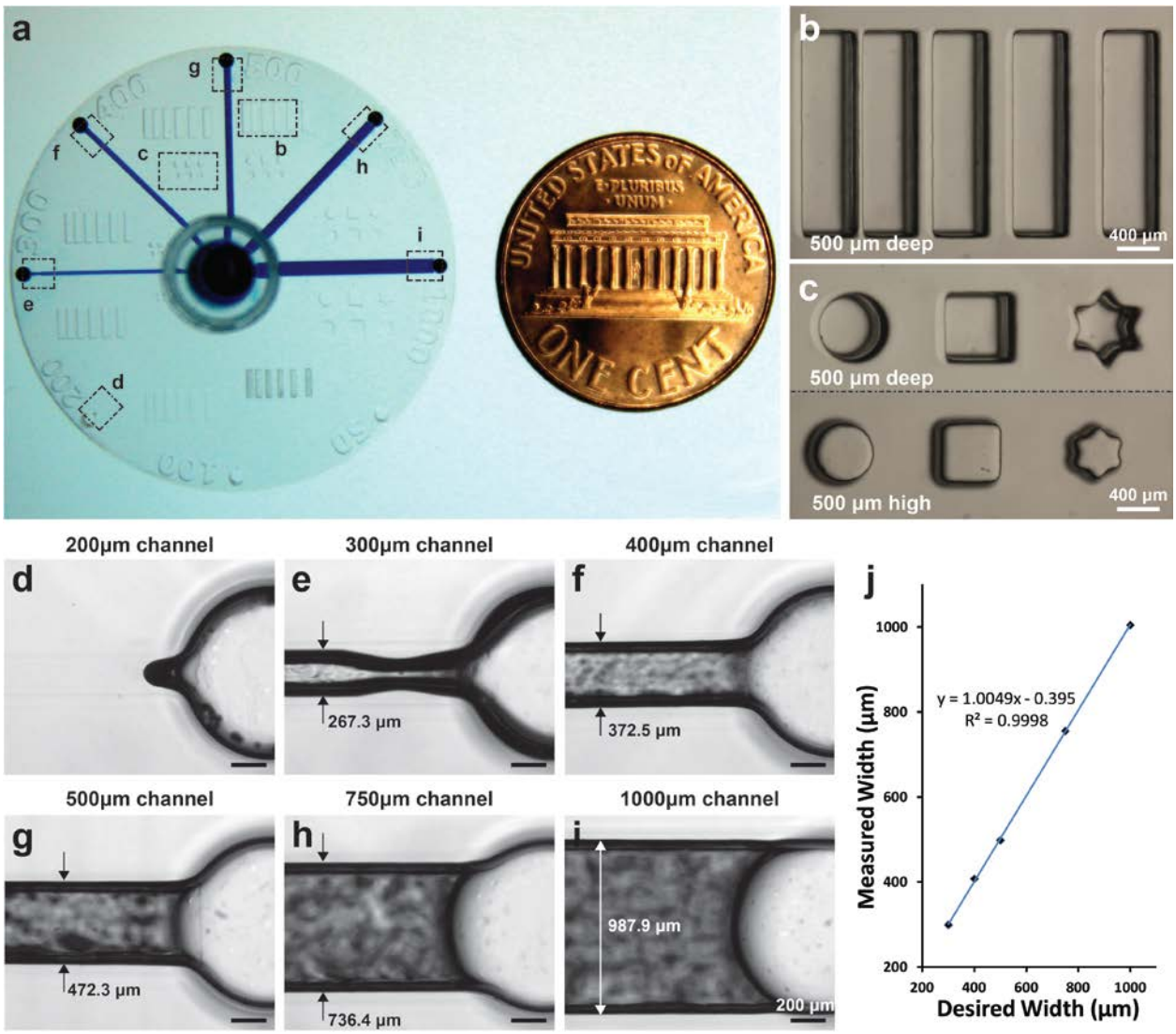

Fig. 3. (a) Photograph of the resolution test device with the cleared channels filled with blue dye. **(b-c)** Micrographs of **(b)** trenches with varying gap distance and **(c)** raised and lowered shapes**. (d-i)** Micrographs of 500 µm-tall channels built in a range of widths. The 300 µm-wide channel could only be partially cleared of resin, while the smaller channels did not clear at all and larger channels were fully cleared. **(j)** A plot comparing the desired and measured widths of channels 300 to 1000 µm wide in similar devices.

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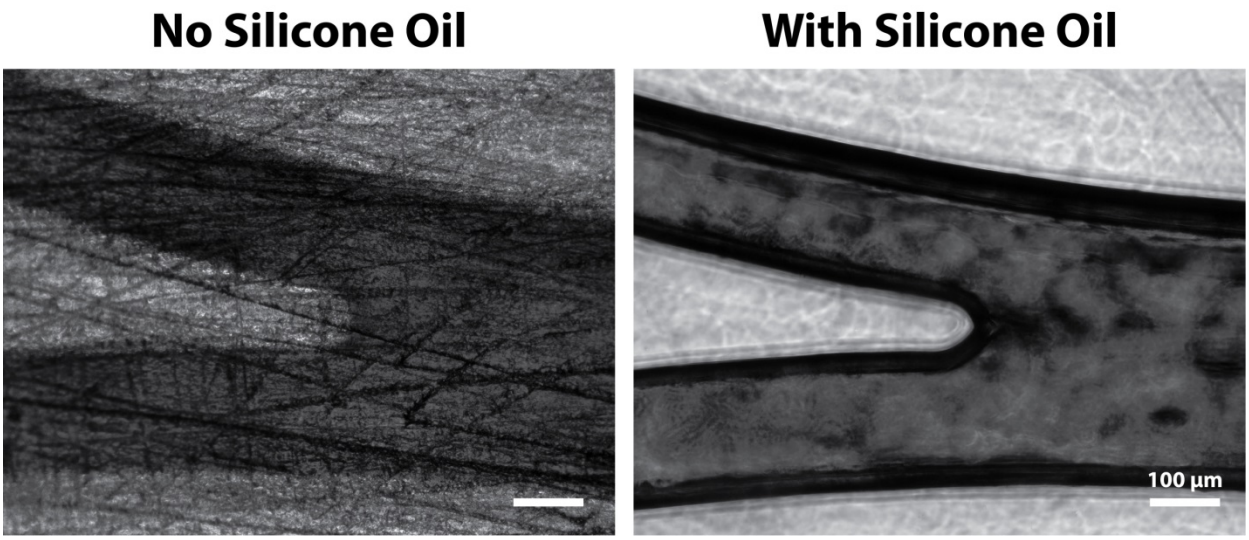

Fig. 4. Micrographs comparing the optical clarity of a device lacking the Substrate Build Style Option, with and without silicone oil coating the bottom surface of the device.

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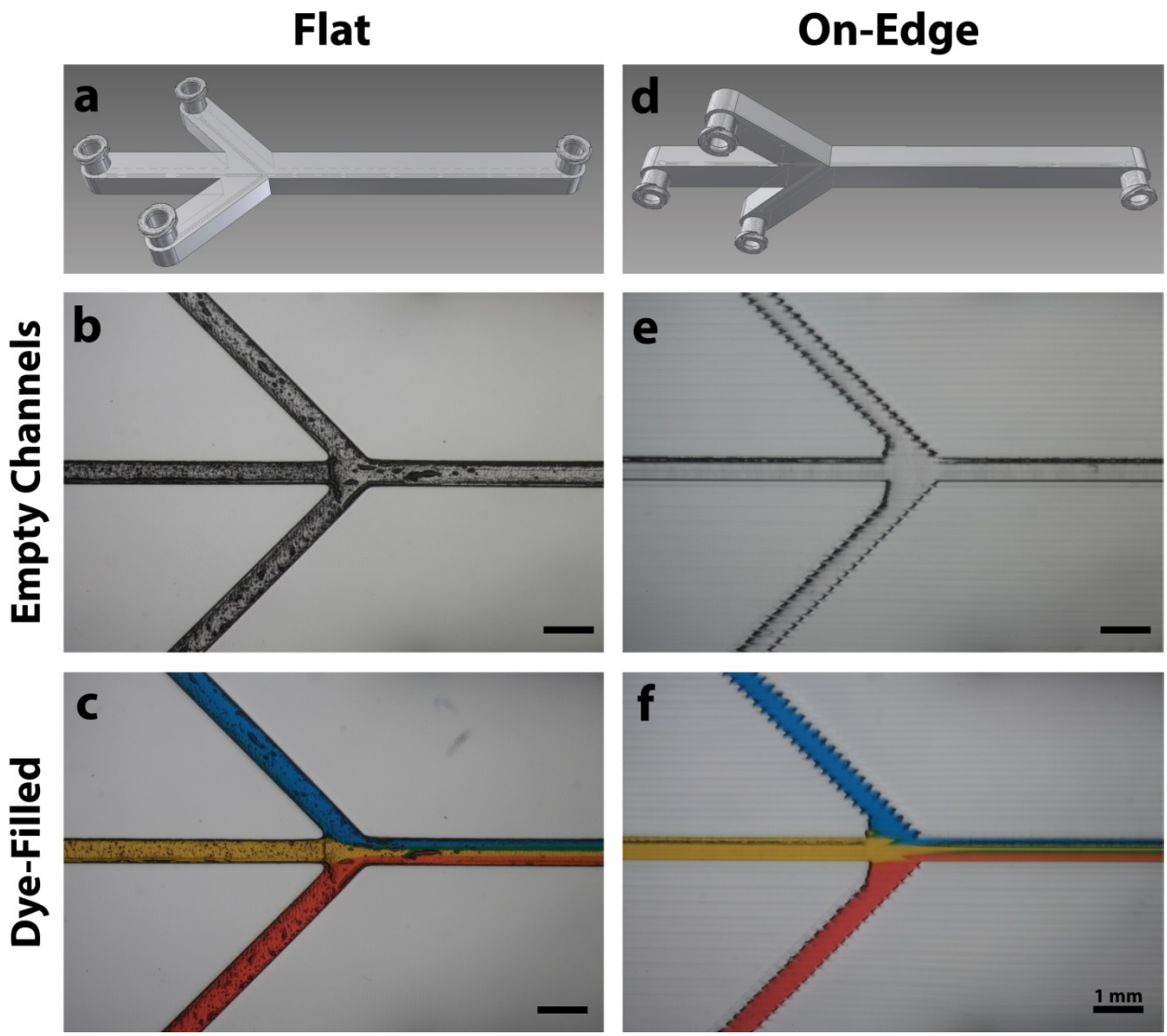

Fig. 5. Comparison of microchannels built **(a-c)** flat and **(d-f)** on-edge filled with air and food-coloring dyes. A silicone oil of similar refractive index to WaterShed XC was applied to the top and bottom surfaces of the devices to reduce the prominence of outer sidewall ripples in the on-edge device.

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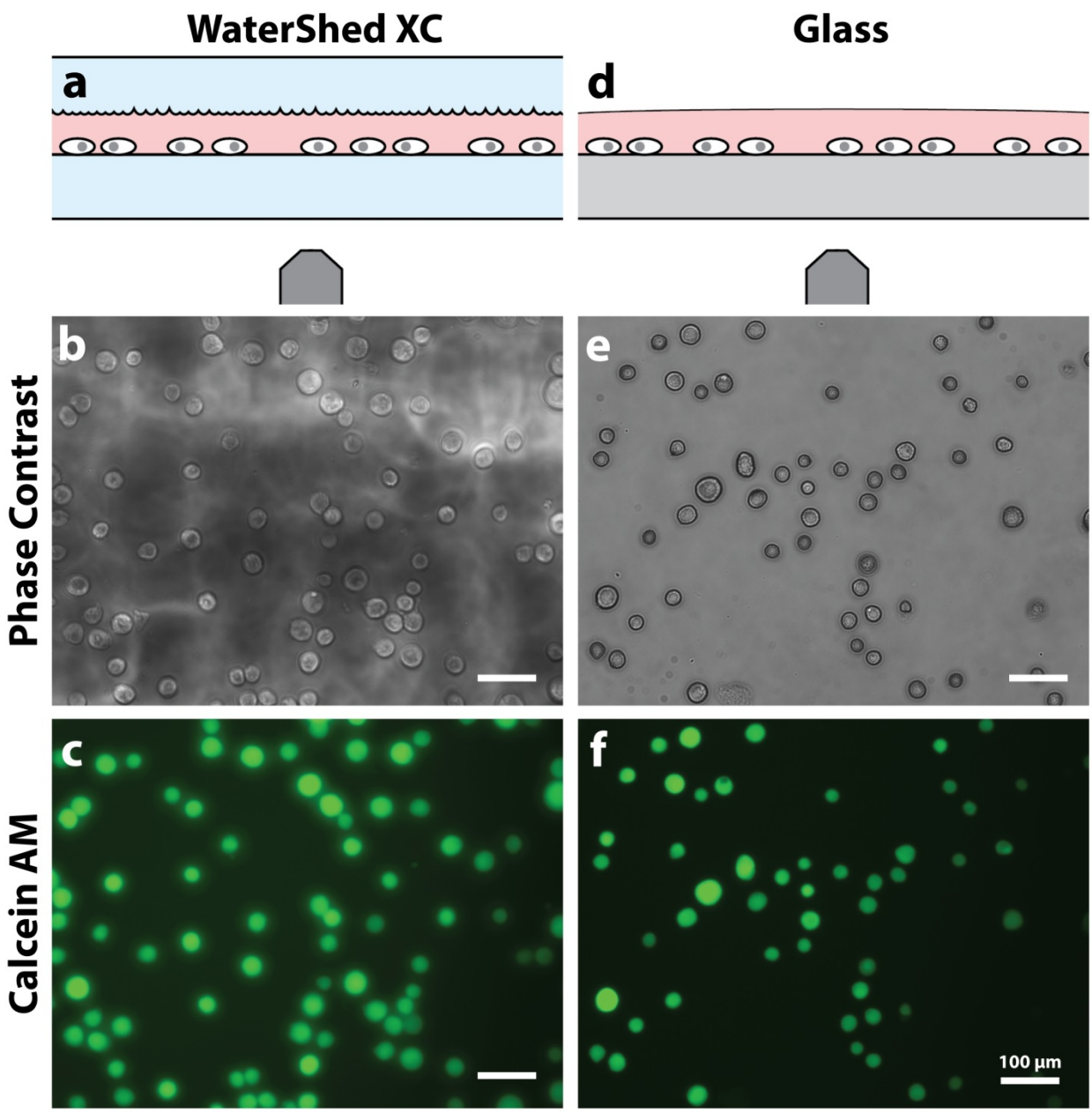

Fig. 6. Comparison of optical imaging through **(a-c)** a microchannel built in Watershed XC and **(d-f)** a glass microscope slide.

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Table 1. Comparison of desktop stereolithography systems.

	Technology	X-Y Resolution	Price	Open-Source
Asiga Pico systems	DLP	27-39 µm	$6,990-$8,990	no
EnvisionTEC Micro	DLP	31 µm	€12,999	no
B9Creator	DLP	50-100 µm	$2,990-$4,995	yes
MiiCraft	DLP	56 µm	$1,999	yes
3D Systems ProJet 1200	DLP	56 µm	$4,900	no
Ilios HD	DLP or laser	variable	€2,805+	yes
DigitalWax systems	DLP or laser	many options	many options	no
formlabs Form 1	laser	300 µm	$3,299	no

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