

COMPARISON OF RAPID PROTOTYPING POLYMERS FOR HIGH PRESSURE INJECTIONS

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ABSTRACT

Here we assess several polymer alternatives to PDMS for rapid prototyping, especially in view of high pressure injections: Thermoset Polyester (TPE), Polyurethane Methacrylate (PUMA) and Norland Adhesive 81 (NOA81). We provide a systematic analysis of these materials with side-by-side experiments conducted in our lab, to similarly assess their magnitudes of deformation and dynamic characteristics, but also other properties such as biocompatibility, solvent compatibility, and ease of fabrication. We emphasize that TPE, PUMA and NOA have some considerable strengths for rapid prototyping when bond strength, predictable operation at high pressure, or transitioning to commercialization are considered important for the application.

KEYWORDS: Rapid prototyping, material deformability, high-pressure, alternative fabrication materials.

INTRODUCTION

Multiple microfluidic device fabrication methods exist, each with different advantages for implementation in mass production or academic research. In the research community, polydimethylsiloxane (PDMS) has been the standard rapid prototyping material due to its low cost, ease and speed of fabrication, which are particularly advantageous in the exploratory stages of research [1]. However, among other problems the high material compliance of PDMS creates difficulties in moderate to high pressure applications, yielding (i) large feature distortion, (ii) non-linear pressure to flow rate relations, and (iii) difficulties to accurately predict the flow rates in complex microfluidic networks [2]. Recently, several UV-curing polymers have been developed with similar rapid fabrication capabilities as PDMS but with higher rigidity and improved chemical compatibility: Thermoset Polyester (TPE) [3], Polyurethane Methacrylate (PUMA) [4] and Norland Adhesive 81 (NOA81) [5]. Here we compare these polymer alternatives to PDMS for rapid prototyping, especially for high pressure microfluidic applications. Other properties of these materials, such as ease of fabrication, biocompatibility and solvent compatibility are also compared, yielding a complete and systematic characterization of these substrates as a useful guide for the microfluidics community.

EXPERIMENTAL

Our test device consists of a simple straight rectangular channel (5 cm long, 60 x 52 μm). PDMS, TPE, PUMA and NOA chips were fabricated as indicated in Figure 1, following protocols published elsewhere [1, 3, 4, 5].

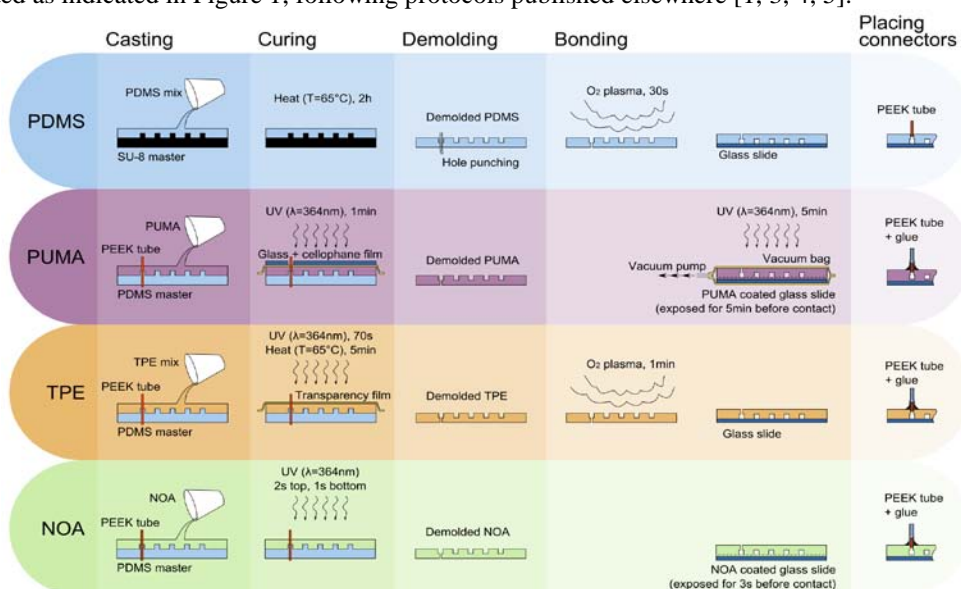


Figure 1: Protocols for fabrication of PDMS, PUMA, TPE and NOA chips. The main steps were similar but with UV-curing for TPE, PUMA and NOA, instead of thermal-curing for PDMS. TPE fabrication also involved additional plasma treatment before bonding.

Fluorescent particles ($\varnothing=1, 4.8$ and $9.9 \mu\text{m}$, 1.05 g/mL , Duke Scientific) or buffer (Fluorescein 1 mM) were injected into the devices using non-deformable glass syringes (Hamilton), with a syringe pump (Harvard Apparatus PHD 2000) or a pressure control system. The maximum pressure (P_{max}) of each device is evaluated by measuring the pressure at chip leakage. Based on the images recorded from fluorescein injections, the intensity profile of a given channel cross-section was plotted as a quantitative representation of cross-section shape and extension for each flow rate. Relative changes in depth (ΔH) and width (ΔW) were defined as the normalized expansions in the X and Y axis respectively, compared to the $10 \mu\text{L/min}$ “no deformation-baseline”. As other characterization, the Young’s modulus is measured with an Instron tensile tester and identical rectangular specimens ($2 \times 26 \times 78 \text{ mm}$). Biocompatibility is characterized with HeLa cells, incubated in square chambers ($1 \times 1 \text{ cm}$), and observed after 2 and 24 hours, with live/dead staining. Also for polymer/solvent pairs - (i) 70% ethanol in water, (ii) PDMS 200 fluid, 1.0 cSt -, a failure flow rate ($Q_{Failure}$) is measured corresponding to device delamination or limit of the syringe pump at increasing flow rates. Water is used as a reference. Additionally, feature integrity was characterized under a microscope, with ‘+’, ‘-’ or ‘--’ respectively indicating no swelling, swelling of features, or disappearance of features.

RESULTS AND DISCUSSION

The fabrication protocol was characterized by looking at the turnaround time and bond success rate. For PDMS, which is simple and fast (3hrs) to fabricate, bonding success rate was variable (50-100%). Contrastingly, TPE, PUMA and NOA have much higher repeatability and bond fidelity (90%), with 3hrs of fabrication for PUMA and NOA but 1hr+1day for TPE.

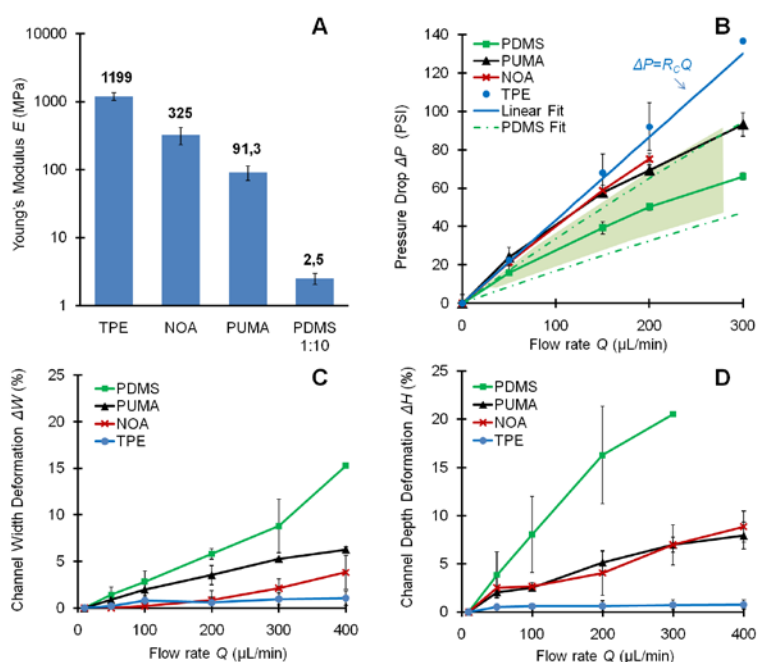


Figure 2: Mechanical properties. (A) Comparison of Young’s Modulus (MPa) for each material, measured with an Instron instrument for identical $2 \times 26 \times 78 \text{ mm}$ specimens. (B) Elasticity differences affect the pressure drop (ΔP) vs. flow (Q) characteristics. ΔP measured at the inlet is close to linear theory for rigid channels (TPE), where ΔP is linearly proportional to Q and fluidic resistance (R). ΔP measured for deformable PDMS chips is compared to a theoretical model (dotted green lines) [2] setting a 3 mm top wall thickness and $E = 2.5 \text{ MPa}$. (C, D) Microchannel width (C) and depth (D) deformation as a function of Q for each polymer, 3 chips/polymer, from 10 to $400 \mu\text{L/min}$ compared to $10 \mu\text{L/min}$ as a reference.

Furthermore the elastic modulus of TPE, NOA and PUMA was found to be two to three orders of magnitude higher than PDMS (Figure 2.A) leading to significantly less deformation than PDMS (Figure 2.B and C). This deformation in PDMS, present even at low flow rates, leads to channel cross-sections higher and wider than expected ($\Delta H=20.5\%$, $\Delta W=8.8\%$) and non-linear alterations of the flow in these channels with applied pressure. Rigid polymers exhibited linear and predictable pressure vs. flow relationships which is useful when transitioning a complex microfluidic network to commercially relevant

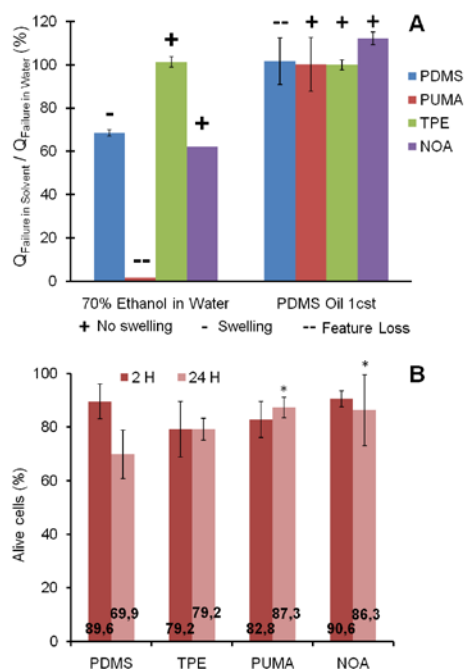


Figure 3: Other properties. (A) Material solvent compatibility. For each material, the maximum flow rate to chip failure for 70% Ethanol in water and 1 cSt PDMS oil was normalized to the maximum flow rate in water ($Q_{Failure}$ in Water). Feature integrity was evaluated with ‘+’, ‘-’ or ‘--’ respectively. (B) Material biocompatibility measured with HeLa cells incubated in microchambers made of each polymer, after 2 and 24hrs, using a live/dead assay, with 3 chambers per condition.

thermoplastics with similar operation parameters (Figure 2.C). Under dynamic flow, these polymers are also superior to PDMS yielding flow settling times on the order of a few seconds compared to minutes for PDMS. Additionally, the maximum sustainable pressure P_{max} for TPE (150 PSI), PUMA (110 PSI) and NOA (75 PSI) chips was two to three times larger than PDMS (51 PSI), indicating these alternatives are much more robust and appropriate to high pressure than PDMS.

In addition to rigidity and robustness, we experimentally confirmed that several of these polymers are tolerant to common solvents, such as ethanol and silicone oil, where PDMS is not (Figure 3.A). In general, the rigid polymers allowed successful cell culture without acute toxicity even providing an improvement over PDMS for PUMA and NOA after 24 hours (Figure 3.B). However, TPE exhibits a lower and relatively unknown biocompatibility for long-term cell culture.

CONCLUSION

We have shown that TPE, PUMA and NOA are promising alternatives to PDMS for microfluidic rapid prototyping, with considerable advantages when bond strength, predictable operation at high pressure, or transitioning to commercialization are considered important. Since these polymers all have pros and cons (Table 1), the choice of the material will depend on the application. For applications coupling microfluidics with optical analysis, such as inertial focusing and flow cytometry, for which an accurate prediction of particle alignment is required, channel deformation and resulting pulsation of channels is the main issue to consider: TPE is suggested as the material of choice because of its higher rigidity. We also believe TPE is the best alternative to PDMS for particularly high pressures or for situations demanding fast flow-stabilization like in stop-flow lithography. For applications where biocompatibility is crucial, PUMA and NOA will be both perfectly suitable, with an advantage for PUMA which has already satisfied all clinical regulations [4]. In applications that involve chemicals or surface treatment, such as droplet microfluidics, TPE and NOA are particularly adapted thanks to their excellent compatibility with most oils and solvents [3, 5].

		PDMS	TPE	PUMA	NOA
Material Properties	Hardness	-	++	+	+
	Biocompatibility	+	-	++	++
	Optical Transparency	++	+	+	+
	Solvent Compatibility	-	+	+	+
Performance for fast prototyping and high pressure injections	Cost	< \$1/mL	~\$4.5/mL	~\$2.2/mL	< \$1/mL
	Ease of Use	+	-	++	++
	Fabrication Time	1h+2h	1h+1day	1h+2h	1h+2h
	Replication Fidelity	+	+	+	+
	Channel Deformation	-	++	+	+
	Stabilization Time	-	++	++	++
	Maximum Pressure	-	++	+	+

Table 1: Overall performance of PDMS, TPE, PUMA and NOA as materials for rapid prototyping of microfluidic devices and high pressure injections.

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