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# A thin PDMS nozzle/diffuser micropump for biomedical applications

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ABSTRACT

Micropumps are finding more applications in biomedical fields. One application that holds great potential for micropumps is the treatment of glaucoma. Shunts are currently used in the treatment glaucoma. They lower intraocular pressure by passively increasing the outflow of aqueous humour from the anterior chamber in the eye. Nozzle/diffuser micropumps are an attractive alternative to shunts. They would provide both passive outflow and variable active outflow of aqueous humour as necessary. They would be able to overcome increases in flow resistance caused by biofouling and scar tissue growth. This study presents a proof of concept of a 1-mm-thick electromagnetic PDMS nozzle/diffuser micropump for biomedical applications. The pump is composed of a cast PDMS body, a spin coated PDMS membrane, and commercial silicone tubing, all bonded together with PDMS. Micromachining a pump cast enables multiple pumps to be produced quickly and reliably. An in-plane design is used to attach the inlet and outlet tubes. The pump produces a peak flow rate of 135 μL/min and a maximum backpressure of 25 mm$H_2O$ at an actuation frequency of 12 Hz and a duty cycle of 25%. The membrane thickness, actuator type, and duty cycle were shown to have a significant impact on the performance of the pump.



## 1. Introduction

Micropumps have been used for a number of years in industrial applications, such as in inkjet printers and microelectronic cooling [1]. With recent advancements, micropumps have started being used in biomedical applications [1,2]. One particular biomedical application that holds great potential for micropumps is the treatment of glaucoma. Glaucoma is the leading causes of permanent blindness in the world. It is a chronic condition in which damage to the optic nerve occurs. The primary method for treating glaucoma is to decrease the intraocular pressure of the eye below 21 mmHg. Current treatment methods include eyes drops [3], surgery [4,5], drug implants, and shunt devices [6]. Shunt devices are passive drainage systems that are implanted in the eye. A shunt device is a passive drainage system that is implanted in the eye. They reduce the intraocular pressure by allowing aqueous humour to flow from the anterior chamber to a discharge chamber with a lower pressure. Unfortunately, shunt devices degrade over time as biofouling and scar tissue growth increase flow resistance, leading to a return of high intraocular pressure.

Nozzle/diffuser micropumps are an attractive alternative to shunt devices. Nozzle/diffusers micropumps rely on fluidic structures that act either as a diffuser or a nozzle depending on the direction of fluid flow. Typically, they have a lower flow restriction in the direction of the diffuser, resulting in net fluid flow in that direction during pumping. A nozzle/diffuser micropump would allow the rate of aqueous humor outflow to be controlled. When the pump is off, it would provide a passive outflow of aqueous humor at a low flow rate due to its valveless structure. When it is on, it would provide an active outflow of aqueous humor at a variable flow rate, depending on the strength and frequency of pump actuation. The micropump could also overcome the increased flow resistance caused by biofouling and scar tissue growth. A suitable micropump would be wireless, thin, flexible, biocompatible, and robust.

There has been quite a bit of work on nozzle/diffuser micropumps. Amirouche et al. provided a good review of actuation schemes, flow directing elements, and chamber configuration for micropumps in general [1]. Zhou and Amirouche developed a PDMS valveless micropump that generated a maximum flow rate of 320 μl/min and a maximum back pressure 9.5 mm$H_2O$ [7]. Yang et al. investigated the use of obstacle structures such as fins for improving the performance of nozzle/diffuser micropumps [8]. Recently, Yang and al. developed a piezoelectric micropump with double chambers based on the Coanda effect [15]. They demonstrated a maximum flow rate of 408 μl/min and a maximum back pressure of 32.4 mm$H_2O$. However, most nozzle/diffuser micropump designs [9,7,10,11] are not suitable for implantation because

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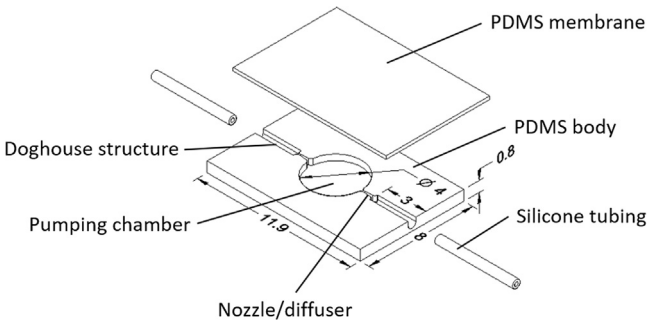


**Fig. 1.** Exploded view of the pump (dimensions in mm).

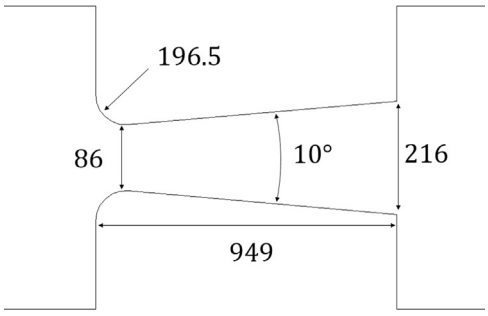


**Fig. 2.** Nozzle/diffuser geometry (dimensions in mm). The height of the pumping chamber and the nozzle/diffusers is 450 μm.

of their geometry, the design of their actuation method, or the materials they are fabricated with.

This study presents a proof of concept of an electromagnetic PDMS nozzle/diffuser micropump for biomedical applications. It is 1-mm thick, has in-plane inlet and outlet connections, and it is conveniently fabricated using a micromilled mold.

## 2. Methods and materials

### 2.1. Micropump design

The micropump design is presented in Fig. 1. It was designed to have a total thickness of 1 mm.

The nozzle/diffuser geometry (see Fig. 2) was designed based on nozzle/diffuser theory [12] and the limits of the milling machine used to produce the mold. According to the theory, sharp inside corners are desired in certain locations for ideal performance, but they could not be produced by milling the mold. Rounded corners with the smallest radius possible (imposed by available milling tools) were used instead.

An in-plane design was used for the tube attachment in the interest of keeping the pump thin. Doghouse structures (cross-section = 300 × 600 um rectangle + 300 μm diameter semicircle) were formed on both the inlet and the outlet side of the pump body allowing the tubes to be seated within.

Contrary to most micropump designs [2], both the pump body and membrane were fabricated with PDMS, a viscoelastic material. This makes the pump flexible and allows it to be mounted to non-planar surfaces.

The pump body (Fig. 1) was cast using a mold. Casting is relatively simple compared to other micropump fabrication methods such as powder blasting, lithography, laser ablation, and hot embossing. It requires no specialized facilities such as clean rooms or equipment such as embossing stamps, lasers, or powder blasters. Casting simply involves pouring degassed PDMS into the mold and curing it in an oven.

The mold was milled from a piece of brass. Brass molds are robust and reusable, allowing multiple pumps to be fabricated from the same mold. Prototyping different micropumps designs is also fast since the total milling time per mold is low (≈1 h). Due to both of these reasons, the total fabrication cost is very low.

### 2.2. Micropump fabrication

The pump was fabricated with using a six-step process (see Fig. 3).

Step 1: The mold for casting the pump body was milled with a *Microlution 363-S Horizontal Milling Machine (*XYZ positional accuracy of 1 μm, top spindle speed of 50 000 RPM, self-zeroing, and linear motors with zero backlash) using carbine miniature cuttings tools from *Harvey Tools*. The mold was modelled in *Solid Edge ST6* and the tool paths were generated using *Mastercam X5: Mill Level 3 CAD/CAM*. A piece of rectangular brass stock (6.35 mm thick, 22 mm long, and 25.4 mm wide) was secured in the milling machine using double sided tape. It was milled using the following steps:

1. Surface high-speed area roughing (3 mm square end mill) to remove the bulk of the material from the mold cavity while leaving 50 μm on walls and floors.
2. Surface contour finishing (1 mm ball end mill) to give the mold sidewalls a smooth and polished finish with a 10° taper. This facilitated the removal of the PDMS.
3. Surface high-speed rest roughing (1 mm square end mill) to finish the base of the mold and to rough the contour of the pump while leaving 50 μm on the walls of the pump features.
4. Surface contour finishing (0.381 mm square end mill) to finish the surface of the pump features
5. Parallel surface finishing (0.460 mm ball end mill) with a 90° milling angle to mill the top surface of the doghouse structures for the inlet and the outlet. A rough surface was produced to improve bonding between the PDMS and the tubing.

After machining, the mold was cleaned to remove oil and any remaining metal bits.

Step 2: The pump body was cast. PDMS (*Sylgard 184 Silicone Elastomer Kit*) was prepared by mixing the pre-polymer and curing agent together and degassing the mixture in a vacuum chamber for 45 min. This mixture is henceforth called pre-cured PDMS. The pre-cured PDMS was poured into the mold until it reached a height of x 1 mm, and then it was cured by putting the mold in an oven preheated at 65 °C for 15 min. The pump body was released from the mold using tweezers, and it was placed on a clean glass slide feature side up. A surgical scalpel was used to cut off the edges of the pump body, removing uneven sections and exposing the ends of the inlet and outlet doghouse structures.

Step 3: The tubing was bonded to the pump body. The doghouse structures for the inlet and outlet tubing were painted with a thin layer of pre-cured PDMS using the tip of a pin. Two pieces of silicone tubing (*Silastic Laboratory Tubing*, inner diameter 0.3 mm and outer diameter 0.64 mm) were cut to a length of 10 mm and the ends were inserted into the inlet and outlet doghouse structures. Care was taken to ensure no pre-cured PDMS entered the nozzle/diffusers or the inside of the tubing. The pump body and tubing were again placed in the oven at 65 °C for 15 min, curing the PDMS and firmly attaching the tubing to the pump body.

Step 4: A PDMS membrane was spin coated using a two-step process: a spreading step followed by a thinning step. Pre-cured PDMS (5 g) was poured onto a 100 mm Petri dish. The Petri dish was spun for 15 s at 500 RPM to evenly spread the pre-cured PDMS over the Petri dish, then spun for a further 45 s at a given speed to thin out the pre-cured PDMS. The thinning spin speed determined the final thickness of the membrane. Membranes of different thicknesses