

Original Article

Performance of and Pressure Elevation Formed by Small-diameter Microtubes Used in Constant-flow Sets

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Purpose: We explored the performance of and pressure elevation caused by small-diameter microtubes used to reduce overfiltration.

Methods: Using a syringe pump-driven constant-flow setting (2 μ L/min), pressures were measured for polytetrafluoroethylene (PTFE) microtubes 5 mm in length with inner diameters of 51, 64, and 76 μ m and for polyether block amide (PEBAX) microtubes with an inner diameter of 76 μ m. Experiments (using microtubes only) were initially performed in air, water, and enucleated pig eyes and were repeated under the same conditions using intraluminal 9/0 nylon stents.

Results: The pressures measured in air in 51-, 64-, and 76- μ m-diameter PTFE microtubes differed significantly (22.1, 16.9, and 12.2 mmHg, respectively; $p < 0.001$), and that of the 76- μ m-diameter PEBAX microtube was 15.8 mmHg ($p < 0.001$ compared to the 12.2 mmHg of the 76- μ m-diameter PTFE microtube). The pressures measured in water also differed significantly among the three microtubes at 3.9, 3.0, and 1.4 mmHg, respectively, while that in the PEBAX microtube was 2.6 mmHg (all $p < 0.001$). Using the intraluminal stent, the pressure in water of the three different PTFE microtubes increased to 22.6, 18.0, and 4.1 mmHg, respectively, and that in the PEBAX microtube increased to 10.5 mmHg (all $p < 0.001$). Similar trends were evident when measurements were performed in pig eyes.

Conclusions: Although microtubes of smaller diameter experienced higher pressure in air, reduction of the inner diameter to 51 μ m did not adequately increase the pressure attained in water or pig eyes. Insertion of an intraluminal stent effectively elevated the latter pressures. PEBAX microtubes created higher pressures than did PTFE microtubes.

Key Words: Intraluminal stent, Intraocular pressure, Microtube, Polyether block amide, Polytetrafluoroethylene

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Molteno introduced the modern glaucoma implant in 1969, and many different types and associated materials have since been developed. Currently, the Ahmed glaucoma valve, the Molteno implant, and the Baerveldt implant are used widely. Most currently available implants have a

silicone tube with an outer diameter of approximately 600 μm and an inner diameter of approximately 300 μm . The inner diameter of the tube is oversized for an approximate 2.0 $\mu\text{L}/\text{min}$ inflow rate of aqueous humor to the eyeball. Therefore, these current implants are vulnerable to overfiltration of the aqueous humor and resultant hypotony, choroidal effusion or detachment, or hypotony maculopathy [1-5]. To avoid overfiltration, some glaucoma implants were developed with a one-way valve mechanism [6]. Nevertheless, overfiltration remains a frequent and serious problem of implant surgery. Therefore, many surgeons temporarily occlude the drainage tube using an intraluminal stent or tube ligation [7-11]. The large outer diameter of current drainage tubes can induce conjunctival erosion over the tube [12,13]. To reduce the incidence of this complication, various types of tectorial support between the tube and overlying conjunctiva are applied during surgery. Despite this effort, tube exposure is still one of the main complications of implant surgery [5,12,13].

In contrast, a tube with a smaller inner diameter can provide more resistance to fluid flow, and a smaller outer diameter should decrease conjunctival erosion over the tube and the accompanying complications. A shunt with a small diameter, the ExPRESS implant (Alcon Laboratories, Fort Worth, TX, USA), is currently available. It has a short metallic tube with a 50 / 400 μm (inner / outer diameter) but is too short to provide resistance to aqueous flow.

In this study, we evaluated fluid flow in microtubes of various sizes (51, 64, and 76 μm in inner diameter) and materials (polytetrafluoroethylene [PTFE] and polyether block amide [PEBAX]) under different conditions (air, water, and enucleated pig eyes). In addition, we investigated the effects of an intraluminal stent.

Materials and Methods

Preparation of microtubes

Three extruded PTFE tubes with different inner diameters (51, 64, and 76 μm) and one PEBAX tube with an inner diameter of 76 μm were tested. We chose 51-, 64-, 76- μm extruded PTFE tubes, the three smallest supplied by the manufacturer (Zeus Inc., Orangeburg, SC, USA), and 76- μm PEBAX tubes. PTFE and PEBAX are significantly different. PTFE has strong lubricant qualities and the low-

est coefficient of friction of any known polymer. PEBAX is more flexible than PTFE, is more readily shaped, and can be readily mixed with other materials. All PTFE and PEBAX tubes fulfilled the biocompatibility standards of the United States Pharmacopeia Class VI plastic test. The wall thickness of each tube is 51 μm . The tolerances of the inner and outer diameters are $\pm 10 \mu\text{m}$ (provided by Zeus Inc.). Twenty of each of the four tube types were tested (80 tubes in total). Before the experiment, all of the tubes were carefully cut to a length of 5 mm to avoid distortion of the tube. Using a microscope (Optiphot-2; Nikon, Tokyo, Japan) and a measurement program (Prores CapturePro 2.8.8; Jenoptik Optical System, Jupiter, FL, USA), tubes with a less than $\pm 2 \mu\text{m}$ inner diameter tolerance at both distal cut ends were selected for the study. After trimming to a length of 5 mm, the inner diameters of the tubes were 51.58 ± 0.63 , 64.46 ± 0.58 , 76.53 ± 0.89 , and $76.75 \pm 0.69 \mu\text{m}$ (mean \pm standard deviation) for the PTFE tubes of 51, 64, and 76 μm diameter, respectively, and the 76- μm PABAX tube. Fig. 1A-1F shows light micrographs of the cut ends of each tube type and 9/0 nylon.

Device for measuring pressure

We assembled the equipment to measure pressure due to fluid flow. The definition of pressure in this manuscript is not resistance but maintenance pressure. The device consisted of three components: (1) perfusion pump and syringe, (2) pressure transducer and detector, and (3) silicone chamber and microtube. One perfusion pump and one syringe were connected to two pressure transducers, a detector, a silicone chamber, and a microtube. We ran the two pumps and syringe simultaneously. Therefore, we could simultaneously measure pressures within four microtubes. We adopted a perfusion pump with a stepper motor (Model 11; Harvard Apparatus, Holliston, MA, USA), which was connected to a 1-mL glass syringe (Hamilton, Reno, NV, USA). A three-way valve was connected to the end of the glass syringe, and a calibration reservoir was attached for pressure calibration or air bubble removal. A pressure transducer (PX260; Edwards Lifesciences, Irvine, CA, USA) and detector (8SP; ADInstruments, Colorado Springs, CO, USA) were connected to a personal computer, which monitored the pressure change in units of 0.01 mmHg using Chart v5.1 software (ADInstruments). A 10-mL syringe tip was attached to each side of a cylindrical

silicone chamber (length, 40 mm; diameter, 18 mm; silicone thickness, 400 μm ; volume, 7 mL) with silicone glue. A microtube was attached to the distal end of the silicone chamber using a 30-gauge needle hub after removing the needle shaft. The microtube was inserted 2 mm into the needle hub and fixed in place with silicone glue (Fig. 2). Experiments were conducted after the silicone glue had dried and cured for 2 weeks.

Measurement of pressure

The space within the device was completely filled with distilled water. Air bubbles that formed during the filling process were removed via the calibration reservoir. Calibration measurements were performed using a mercury blood pressure gauge connected to the calibration reservoir [14-16]. The syringe pump was set at 2 $\mu\text{L}/\text{min}$, and pressure was measured in real-time for 24 hours.

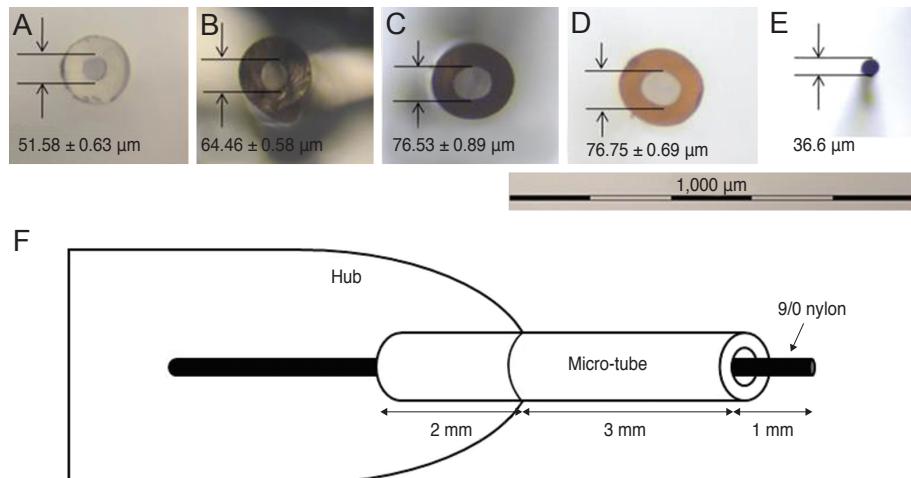


Fig. 1. Light micrographs of the cut ends, and average internal diameters of the microtubes and 9/0 nylon thread. (A) 51- μm polytetrafluoroethylene (PTFE) tube. (B) 64- μm PTFE tube. (C) 76- μm PTFE tube. (D) 76- μm polyether block amide tube. (E) 9/0 nylon. (F) Tube with a 20-mm length of intraluminal 9/0 nylon; one end of the thread was inserted into the tube 6 mm from the distal end.

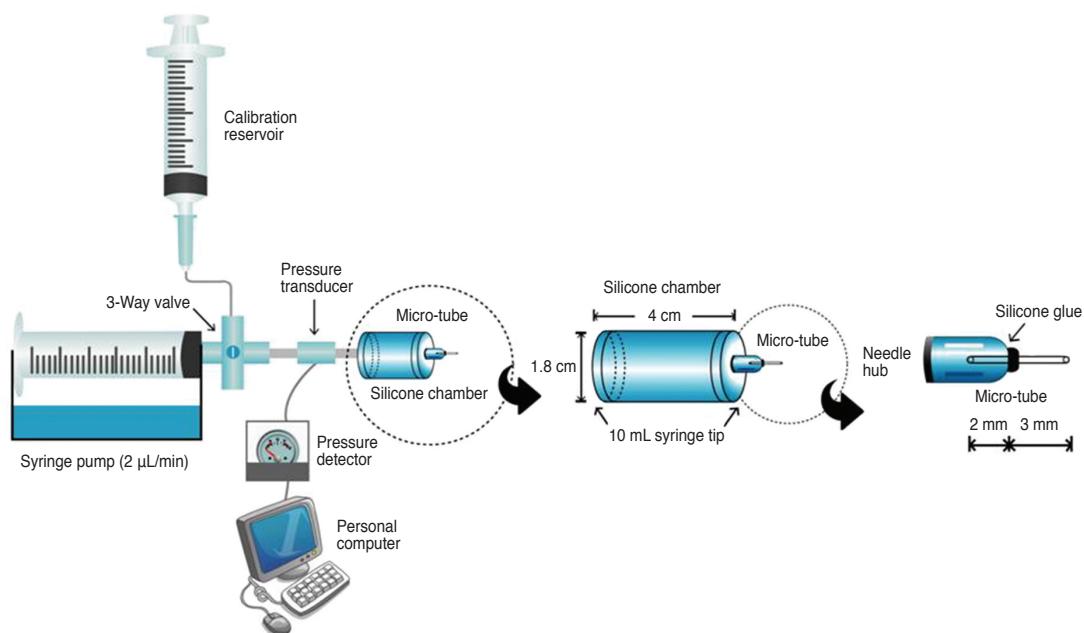


Fig. 2. Experimental setup for fluid flow in air. The setup consisted of a perfusion pump with a syringe, pressure transducer, pressure detector, silicone chamber, and microtube.

Experimental conditions

Four tube types of various diameter and material, 20 tubes of each condition, were tested under three different conditions: in air, in water, and in enucleated pig eyes. All experiments were performed at a constant room temperature of 23°C. Measurement of fluid flow was performed first in air (Fig. 2). Then, the distal end of the microtube was inserted 1 mm into a 50-mL conical centrifuge tube filled with distilled water to a level 1 mm above the distal end of the microtube. To avoid an increase in pressure from the increased height of the water surface, an outlet for the water was made 4 mm above the end of the microtube (Fig. 3). Fresh enucleated pig eyes served as an animal model. A limbal-based conjunctiva flap and a Tenon's capsule flap were created, and the distal end of a microtube was positioned on the sclera beneath the two flaps. After confirming that the distal end of the microtube was not occluded by surrounding tissue, the experiment commenced (Fig. 4). The experiments were repeated under the same conditions with a 20-mm length of intraluminal 9/0 nylon (Ethilon; Johnson and Johnson Medical, Cincinnati, OH, USA), and one end of the thread was inserted into the tube 6 mm from the distal end (Fig. 1F).

Statistical analysis

Statistical analysis was conducted using nonparametric statistics with the Kruskal-Wallis test and Mann-Whitney test with SPSS ver. 12.0 (SPSS Inc., Chicago, IL, USA). In all analyses, $p < 0.05$ indicated statistical significance. From the continuous 24-hour measurements, the pressures at 10 and 20 hours were collected and compared with regard to the following aspects: (1) the difference in pressure among PTFE tube sizes (51, 64, and 76 μm) under each condition; (2) the difference in pressure between PTFE 76 μm and PABAX 76 μm under each condition; (3) the difference in pressure in tubes of the same size and material at different conditions; and (4) the difference in pressure between tubes with and without the intraluminal 9/0 nylon stent for each tube and condition.

Results

Table 1 shows the mean pressure (mean \pm standard deviation, mmHg) measured for each of the four micro-

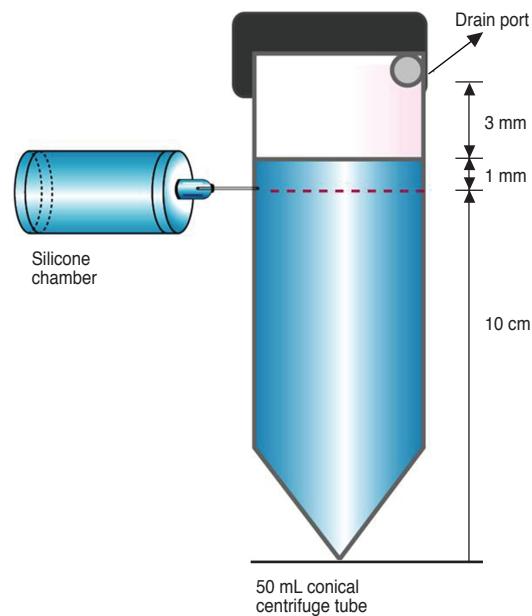


Fig. 3. Experimental setup for fluid flow in water. All settings were identical to those in Fig. 2, except that the tip of the microtube was connected to a tube filled with distilled water.

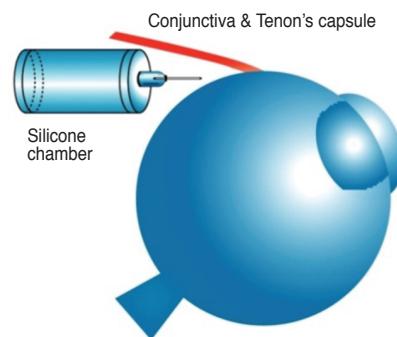


Fig. 4. Experimental setup for fluid flow in an enucleated pig's eye. All settings are identical to those in Fig. 2, except the tip of the microtube was positioned on the sclera beneath the conjunctiva and Tenon's capsule.

tube types 10 hours after beginning the flow under three conditions (in air, water, and enucleated pig eyes). Pressures measured under each condition were much different from the pressure theoretically calculated using Poiseuille's formula for a tube length of 5 mm, inflow rate of 2 $\mu\text{L}/\text{min}$, viscosity of 0.932 centipoise (cP), and room temperature of 23°C. The pressures measured under each condition differed significantly among the different PTFE tube sizes. As the diameter increased, the mean pressure decreased, with values of 14.9, 12.0, and 8.0 mmHg in air; 3.6, 2.9, and 1.2 mmHg in water; and 4.3, 3.1, and 1.8 mmHg in enucleated pig eyes, respectively (all $p < 0.001$). The pressure in

Table 1. Pressures in microtubes of various diameters and materials 10 hours after flow began

Condition	PTFE tube			<i>p</i> -value*	PEBAX tube 76 μm	<i>p</i> -value†
	51 μm	64 μm	76 μm			
Without stent (mmHg)						
In air	14.9 \pm 1.6	12.0 \pm 1.5	8.0 \pm 1.0	<0.001	10.4 \pm 1.8	<0.001
In water	3.6 \pm 0.8	2.9 \pm 0.6	1.2 \pm 0.4	<0.001	2.3 \pm 0.6	<0.001
In pig eye	4.3 \pm 1.2	3.1 \pm 1.1	1.8 \pm 0.8	<0.001	2.9 \pm 1.1	0.002
<i>p</i> -value‡	<0.001	<0.001	<0.001		<0.001	
With intraluminal 9/0 nylon stent (mmHg)						
In air	15.9 \pm 1.4§	12.1 \pm 1.2§	8.0 \pm 0.8§	<0.001	11.1 \pm 1.4§	<0.001
In water	14.6 \pm 1.3¶	7.0 \pm 0.9¶	2.6 \pm 0.6¶	<0.001	6.0 \pm 0.9¶	<0.001
In pig eye	16.3 \pm 1.5¶	8.0 \pm 1.0¶	4.4 \pm 1.1¶	<0.001	6.7 \pm 1.2¶	<0.001
<i>p</i> -value‡	<0.001	<0.001	<0.001		<0.001	

Values are presented as mean \pm standard deviation.

PTFE = polytetrafluoroethylene; PEBAK = polyether block amide.

*Among PTFE tubes (Kruskal-Wallis test); †Between PTFE 76 μm and PEBAK 76 μm (Mann-Whitney test); ‡Kruskal-Wallis test; § $p \geq 0.05$;

¶ $p < 0.001$, comparison of pressures with and without an intraluminal stent (Mann-Whitney test).

PEBAK 76 μm was 10.4 mmHg, which was significantly higher than the value of 8.0 mmHg for PTFE 76 μm in air ($p < 0.001$). The pressure was 2.3 and 1.2 mmHg, respectively, in PEBAK and PTFE in water ($p < 0.001$) and 2.9 and 1.8 mmHg in enucleated pig eyes ($p = 0.002$). The pressures measured in each tube of the same size and material in air differed significantly from those in water and pig eyes. The pressures were higher in air compared to those in water and pig eyes (all $p < 0.001$). Pressures measured in the air for the PTFE 51, 64, 76 μm , and PABAX 76 μm tubes were not significantly different with and without the intraluminal 9/0 nylon stent ($p = 0.068$, 0.698, 0.779, and 0.289, respectively). However, the pressures measured in water were 14.6, 7.0, 2.6, and 6.0 mmHg for PTFE 51, 64, 76 μm , and PABAX 76 μm tubes with the intraluminal stent, respectively, which were significantly higher than those measured in the same tubes without the intraluminal stent (3.6, 2.9, 1.2, and 2.3 mmHg, respectively; all $p < 0.001$). The pressures measured in pig eyes with PTFE tubes of 51, 64, and 76 μm diameter and PEBAK tubes of 76 μm diameter were comparable to those for intraluminal stents and were significantly higher than those measured without stents (16.3, 8.0, 4.4, 6.7 vs. 3.6, 2.9, 1.2, 2.3 mmHg, respectively; all $p < 0.001$).

Table 2 shows the mean pressure (mean \pm standard deviation, mmHg) measured for each of the four microtube types 20 hours after initiation of flow under the three

tested conditions. The pressures measured under each condition were different from those theoretically calculated using Poiseuille's formula. The pressures measured under each condition differed significantly among PTFE 51, 64, and 76 μm . As the diameter increased, the mean pressure decreased (22.1, 16.9, 12.2 mmHg in air; 3.9, 3.0, 1.4 mmHg in water; and 4.6, 3.5, 1.9 mmHg in pig eyes, respectively; all $p < 0.001$). The pressure in the PEBAK tubes of 76 μm diameter in air was 15.8 mmHg, which was significantly higher than the 12.2 mmHg within the PTFE tube of 76 μm diameter ($p < 0.001$). The values were 2.6 and 1.4 mmHg, respectively, in water and 3.1 and 1.9 mmHg in pig eyes (all $p < 0.001$). The pressures measured in each tube of the same size and material differed significantly among air, water, and pig eyes; the pressures measured in air were higher than those measured in water and pig eyes (all $p < 0.001$). The pressures measured in PTFE 76 μm and PABAX 76 μm tubes with and without intraluminal 9/0 nylon stent were not significantly different when measured in air ($p = 0.091$, 0.317, respectively), but the remaining pressures with the intraluminal stent were higher compared to those measured without the stent (all $p < 0.001$).

The cumulative pressure profiles are shown in Fig. 5A-5F. Most graphs reached a plateau at 20 hours after initiation of flow. Insertion of the intraluminal stent increased the pressure in the experiments performed in air, water, and pig eyes. When the test was performed within water

Table 2. Pressures in microtubes of various diameters and materials 20 hours after initiation of flow

Condition	PTFE tube			<i>p</i> -value*	PEBAX tube	
	51 µm	64 µm	76 µm		76 µm	<i>p</i> -value†
Without stent (mmHg)						
In air	22.1 ± 2.3	16.9 ± 1.7	12.2 ± 1.4	<0.001	15.8 ± 1.5	<0.001
In water	3.9 ± 0.8	3.0 ± 0.6	1.4 ± 0.4	<0.001	2.6 ± 0.6	<0.001
In pig eye	4.6 ± 1.1	3.5 ± 0.8	1.9 ± 0.7	<0.001	3.1 ± 0.6	<0.001
<i>p</i> -value‡	<0.001	<0.001	<0.001		<0.001	
With intraluminal 9/0 nylon stent (mmHg)						
In air	32.9 ± 3.6 ^{II}	24.0 ± 2.8 ^{II}	11.4 ± 1.5 [§]	<0.001	16.2 ± 2.2 [§]	<0.001
In water	22.6 ± 3.0 ^{II}	18.0 ± 2.2 ^{II}	4.1 ± 0.9 ^{II}	<0.001	10.5 ± 2.5 ^{II}	<0.001
In pig eye	25.6 ± 3.0 ^{II}	19.1 ± 2.7 ^{II}	5.6 ± 1.3 ^{II}	<0.001	12.4 ± 2.9 ^{II}	<0.001
<i>p</i> -value‡	<0.001	<0.001	<0.001		<0.001	

Values are presented as mean ± standard deviation.

PTFE = polytetrafluoroethylene; PEBAK = polyether block amide.

*Among PTFE tubes (Kruskal-Wallis test); †Between PTFE 76 µm and PEBAK 76 µm (Mann-Whitney test); ‡Kruskal-Wallis test; §*p* ≥ 0.05;
II *p* < 0.001, comparison of pressures with and without an intraluminal stent (Mann-Whitney test).

and pig eyes, the increase in pressure was not remarkable without the intraluminal stent, but the pressure increased markedly upon stent insertion.

Discussion

Most currently available implants in glaucoma implant surgery use a silicone tube with outer / inner diameters of approximately 600 / 300 µm. According to Poiseuille's formula, if a tube 15 mm in length with an inner diameter of 300 µm is located in an ideal fluid flow system in a given fluid flow of 2 µL/min with a viscosity of 0.932 cP (water at 23°C), the pressure difference between the ends of the tube is about 0.02 mmHg [17]. Although the actual *in vivo* fluid flows are likely to differ from this calculation, this extremely low difference in pressure indicates that the tube itself does not create significant resistance to aqueous flow, indicating the inevitability of overfiltration and hypony after glaucoma implant surgery. To reduce these complications, a valve mechanism [6], two-stage surgery [18,19], tube ligation, and intraluminal stents have been used in the clinical field [7-11].

Adopting a tube with a smaller inner diameter can provide more resistance to fluid flow due to the size of the tube itself. If it is sufficiently small, the tube can provide a stable intraocular pressure without overfiltration and hy-

potony after surgery. Recently, a metallic small-diameter aqueous draining tube (ExPRESS tube implant) was introduced. This tube has an inner diameter of 50 µm and an outer diameter of 400 µm. However, as the length of the tube is only 3 mm, it cannot produce sufficient resistance to the aqueous flow. To our knowledge, there have been no previous reports regarding whether a small tube diameter is the only factor that determines the pressure of microtubes or whether other factors such as the external environment are involved [20].

In the present study, the smallest microtubes with an inner diameter of 51 µm and length of 5 mm produced pressures less than 5 mmHg flow in the water and pig eyes, in contrast to the relatively high pressure in air. Experimental conditions cannot exactly duplicate those pressures *in vivo*. We tested the tubes under three conditions: in air, water, and pig eyes. The latter mimics the conditions *in vivo* better than the former, in that it can remove the bias in pressure formation caused by the different environment around the end of the tube. These results were similar to those of a previous report by Prata et al. [16], who reported that the maintenance pressure of glaucoma implants was not achieved in fluid (balanced salt solution and plasma) and was higher in air than in fluid. The lower pressures observed in fluids (water and pig eyes) compared to air are likely attributable to the cohesive power and capillarity surface tension of water at the ends of the micro-

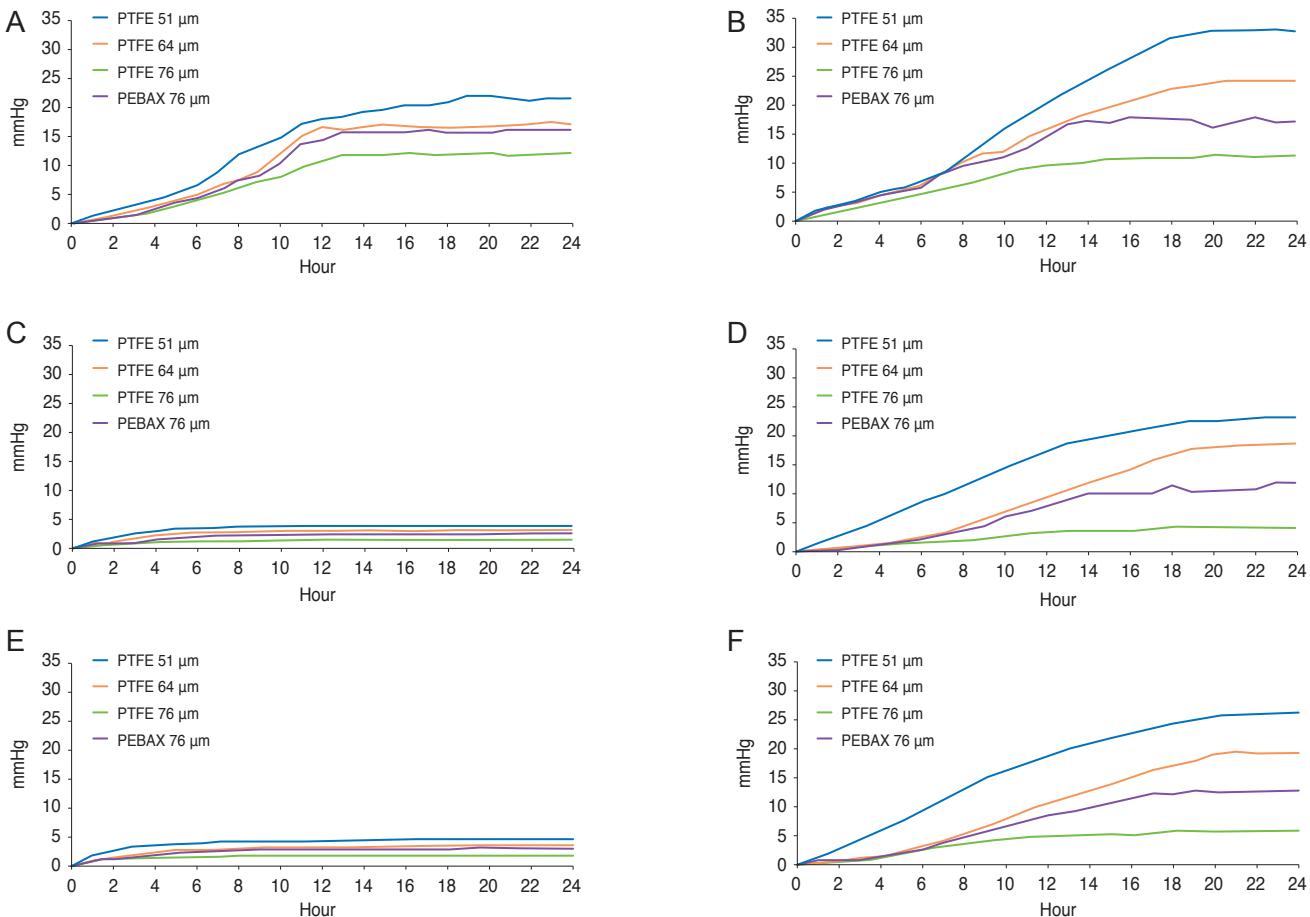


Fig. 5. Cumulative pressure profiles for microtubes of different diameters and materials. (A) Microtubes without a stent in air. (B) Microtubes with an intraluminal 9/0 nylon stent in air. (C) Microtubes without a stent, connected to a water chamber. (D) Microtubes with an intraluminal 9/0 nylon stent, connected to a water chamber. (E) Microtubes without a stent, connected to a pig eye. (F) Microtubes with an intraluminal 9/0 nylon stent, connected to a pig eye. PTFE = polytetrafluoroethylene; PEBAK = polyether block amide.

tubes [21].

It is difficult to manufacture tubes with an inner diameter smaller than 50 μm (outer diameter about 150 μm), and it is technically difficult for the surgeon to manage such small tubes. Therefore, increasing the pressure by further reducing the diameter of the tube does not fulfill the basic clinical requirements for tube use. In addition to reducing the tube diameter, additional pressure can be provided by insertion of an intraluminal stent into the microtube, as shown in this study. A 9/0 nylon (diameter about 36.6 μm , Fig. 1E) intraluminal stent reduces the cross-sectional inner area of the tube by approximately 1,000 μm^2 . Such a stent can effectively increase the pressure across the tube and can be removed if necessary to increase the aqueous flow after surgery.

Interestingly, the effect of the intraluminal stent was

minimaInterestingly, the effect of the intraluminal stent was minimal in the experiment performed in the air. At 20 hours after initiation of flow, the pressure formed by the PTFE 76 μm microtube showed an even lower pressure with than without an intraluminal stent, although the intraluminal stent increased the pressure in general. It is postulated that, in the experiment with the large-diameter tube in air, although an intraluminal stent narrowed the lumen, it reduced the frictional force and enhanced flow through the tube. In contrast to the experiments performed in air, those in water showed persistent pressure increases when an intraluminal stent was used. As the diameter of the 9/0 nylon is 36.6 μm , a 76- μm microtube with an intraluminal stent has a cross-sectional area ($4,534 - 1,052 = 3,482 \mu\text{m}^2$) similar to that of a 66- μm microtube ($3,419 \mu\text{m}^2$), and a 64- μm microtube with an intraluminal stent has a

cross-sectional area ($3,215 - 1,052 = 2,163 \mu\text{m}^2$) comparable with that of a 51- μm microtube ($2,123 \mu\text{m}^2$). However, as shown in Tables 1 and 2, the pressure measured in water and pig eyes for the 64- μm microtube with an intraluminal stent (7.0 and 8.0 mmHg at 10 hours and 18.0 and 19.0 mmHg at 20 hours, respectively) was greater than that for the 51- μm microtube without an intraluminal stent (3.6 and 4.3 mmHg at 10 hours and 3.9 and 4.6 mmHg at 20 hours, respectively), despite the larger cross-sectional area of the 64- μm tube with an intraluminal stent. Therefore, if using an intraluminal stent, factors other than the tube diameter likely influence the pressure.

The pressure also differed according to the tube material in the present study. This may have been due to differences in unevenness of the inner surface or differences in elasticity between PEBAK and PTFE. Hetsroni et al. [22] and Li [23] reported that the uneven internal wall of a tube can increase the pressure by 15% to 37%. In addition, the coefficient of friction may affect the results and changes according to both material and inner diameter of the tube. The coefficient of friction for PTFE is 0.1, whereas that of PEBAK varies between 0.22 and 0.60. In comparison, the reference coefficient range of friction for silicone tubes is 0.25 to 0.75.

A microtube with a very small diameter has some disadvantages for clinical applications, including difficulty of manipulation during surgery and the possibility of tube blockage by tissue debris or inflammatory materials. Therefore, inserting an intraluminal stent in a relatively large-diameter microtube may be useful in clinical settings, and the diameter can be larger for microtubes made of PEBAK rather than PTFE while maintaining the same pressure.

This study had some limitations. The experimental conditions differed from those *in vivo*. Distilled water was used for the study, and the experiments were performed at room temperature (23°C). The pressure is affected by viscosity, which changes with temperature. Distilled water is similar to the aqueous humor in terms of viscosity [24], but viscosity may be further altered by the inflammatory response in glaucoma patients during surgery. Because viscosity differs according to temperature (0.932 cP at 23°C and 0.691 cP at 37°C), the pressure at 37°C is 35% lower than that at 23°C [17]. Numerous unexpected factors may affect the pressure in clinical settings. In particular, the pressure at the end of the microtube located within the

filtering bleb may be a primary factor affecting fluid flow, and this resistance would be determined in part by the size, height, thickness, and elasticity of the filtering bleb. The pressure of the sub-Tenon's space and the facility of the aqueous outflow through the bleb wall can also affect fluid flow [25]. To explore such *in vivo* situations, we used freshly enucleated pig eyes. The distal end of each microtube was always placed beneath Tenon's capsule. In addition, elasticity of the eyeball and flow characteristics of the aqueous humor (laminar flow or turbulent flow) can affect the results. Despite these limitations, we compared the relative differences in pressure formed by the different sizes and materials of microtubes under identical conditions, rather than performing precise measurements of pressure. Further studies in a setting similar to the clinical setting are needed to confirm the efficacy and safety of microtubes in preventing overfiltration to avoid hypotony and tube exposure.

In summary, although the pressure can be elevated by microtubes of a smaller diameter in air, reducing the microtube inner diameter to 51 μm did not sufficiently increase the pressure measured in water or pig eyes. The external environment around the tube tip and the tube material are important determinants of flow and pressure. Insertion of an intraluminal stent effectively increased the pressure measured in water and pig eyes. PEBAK microtubes afforded higher pressures than did PTFE microtubes.

Conflict of Interest

No potential conflict of interest relevant to this article was reported.

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