A valveless capillary mixing system using a novel approach for passive flow control

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Abstract Using capillary-driven forces as a means to pump and actuate fluids is attractive for many applications including portable, lab-on-a-chip diagnostic systems. However, the lack of reliable and easy-to-integrate means of flow control remains a major challenge for implementing more complex bioassays using capillary microfluidics. This paper presents an easy-to-fabricate, valveless capillary system allowing for high-reliability flow control without using traditional capillary stop or trigger valves. Analytical calculations were used to deduce a design criterion for flow control in the valveless system. Furthermore, electrical analogy modeling was used to analyze and optimize the flow characteristics in the valveless system. Silicon-based test structures and capillary mixing systems were fabricated to check the design criterion and benchmark the electrical analogy model, respectively. Experimental results were in good agreement for the analytically derived flow control criterion. Experimental measurements of the liquid–vapor interface position with respect to time agreed well with the electrical analogy modeling results. The flow control scheme developed in the work provides designers with another means to realize complex fluid processing requirements without the need for traditional valve structures.

1 Introduction

In recent years, portable point-of-care devices have received much interest (Chin et al. 2012). Microfluidic systems are attractive for such applications due to their ability to handle small sample volumes, low reagent usage and small amounts of waste. Commonly, external pumps (Laser and Santiago 2004), flow sensors and movable valves are used to perform liquid handling in microfluidic devices (Su et al. 2015). However, such components are relatively bulky, expensive and difficult to integrate into miniaturized platforms (Junker et al. 2002). Capillary-based microfluidic devices provide an attractive solution for such applications since external equipment or movable parts are not needed for fluidic manipulation (Geravis and Delamarche 2009). However, a major challenge in passive capillary microfluidics is to provide a reliable means of flow control in a low-cost, easy-to-manufacture microsystem (Oh and Ahn 2006).

Several attempts to realize robust solutions for capillary passive valves have been made. However, these attempts either suffer from poor reliability or add extra complexity to the fabrication process. Melin et al. (2004) introduced the planar trigger valve that stops fluid flow in one channel and triggers the valve with fluid in a second channel. The first liquid stops at the valve outlet due to the presence of an abrupt lateral expansion in the fluidic channel. Once the second liquid arrives, both liquids can flow through the outlet as liquid contact will be made at the junction. However, Zimmermann et al. (2008) showed that these planar trigger valves suffer from poor reliability. Depending on the design details, success rates for stopping the fluid range from 7.7 to 84.6%. The poor reliability of the planar capillary trigger valves originates from leakage of liquid along the top and bottom hydrophilic surfaces of the valve channel. The abrupt

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lateral expansion only pins the liquid–vapor interface along the two vertical edges of the valve structure.

Safavieh and Juncker (2013) reported a two-level trigger valve design which consists of a shallow conduit intersecting a deeper one, with a hydrophobic cover. Their design allows for abrupt expansion of the valve cross section to happen in both horizontal and vertical planes, which they hypothesized would improve the reliability of the valve. No valve failures were reported with 50 experiments conducted. Majeed et al. (2015) reported a fabrication process for two-step trigger valves in silicon. Although the valve reliability improves markedly compared to the planar trigger valve, the drawback to these two-level designs is an increase in complexity of the manufacturing process.

It is worth noting that other platforms have been developed, such as droplet-based digital microfluidics (Oh and Ahn 2006), that negate the need for traditional microfluidic valve structures. For instance, electrowetting on dielectric (EWOD) can be used to manipulate microfluidic droplets between two parallel plates by controlling the wetting contact angle using electric potential (Lee et al. 2002; Pollack et al. 2002; Paik et al. 2003). EWOD was also extended from the discrete droplets to plugs of reagents stored on chip and separated from each other by air spacers (Linder et al. 2005). However, not all applications are well suited to EWOD-based systems due to the extra fabrication and design complexity added by the electrodes, more stringent requirements of the materials in contact with the liquid droplets (*i.e.*, must be a dielectric with a suitable wetting characteristics) and the necessity to work with discrete droplets rather than continuous flows of fluid.

In this paper, we propose a novel concept for controlling the liquid flow in a capillary microfluidic system without using valves. This concept allows mixing of multiple liquids in sequence without entrapping air bubbles in the microfluidic channels. The design concept was analyzed using electrical analogy modeling and verified by experimental characterization of the fabricated devices.

2 Concept

The basic concept behind the flow control method reported in the manuscript is outlined in this section. It should be noted that the means of flow control is achieved without the use of traditional valve structures. In the next section, the theoretical foundation behind the means of flow control will be expounded.

Figure 1 shows a plan view of capillary-driven flow of a liquid in a branched network of channels with wettable walls. The channels are of a uniform depth. The width of branch 1 is less than the width of branch 2, except for an expansion element, which has a width greater than the width of branch 2. The expansion does not necessarily need to be abrupt, as shown in Fig. 1, but can be gradual. For the purposes of this discussion, it is assumed that liquid–vapor interface transitions from branch 1 into the expansion without pinning of the meniscus. Figure 1a shows an instance in time when the liquid–vapor interface is in the main channel leading up to the branches. The value of the flowrate (Q_{total}) is determined by the value of the capillary pressure (P_c) and the upstream resistance (R_{up}). Once the liquid arrives to the branching point (see Fig. 1b), the liquid will flow only in the branch 1 if the capillary pressure developed in the channel of branch 2 (P_{c2}) is not sufficient to overcome the pressure loss in the main channel ($\Delta P_{\text{loss}} = Q_{\text{total}} \times R_{\text{up}}$).

The situation can be reversed by providing an expansion in branch 1 (see Fig. 1c). The width of the expansion in branch 1 can be engineered so that the capillary pressure in the expansion of the first branch (P_{c3}) is not sufficient to get over the pressure loss ($\Delta P_{\text{loss}} = Q_{\text{total}} \times R_{\text{up}}$) as the liquid flows in branch 2. Thus, the flow proceeds only in the branch 2 and stops at the expansion in branch 1.

As will be shown in later sections of this manuscript, using a system of appropriately sized channels and expansion elements, microfluidic processing functions like mixing and dilution steps can be realized without using conventional valve structures.

Fig. 1 A schematic drawing of capillary flow in a branched microchannel. The fluid moves in a channel of fluidic resistance R_{up} (a). When the flow arrives at the branching point (b), in general it is divided between the two branches, but if the condition in Eq. 9 is sat-

isfied it flows only in the first branch. If there is an expansion in the first branch (c), the flow stops and starts to flow in the second branch if the condition in Eq. 9 can be applied between P_{c2} and P_{c3}

A critical function of many lab-on-a-chip systems is the ability to store one liquid (e.g., chemical reagents) on chip until another liquid (e.g., test sample) is dispensed. How this operation can be realized in a valveless system is described as follows. Consider a branched channel network with an inlet reservoir containing a volume of fluid as shown in Fig. 2. Note that in this discussion, the inlet reservoir is vented (*i.e.*, open to atmosphere) to allow air to replace the liquid as the liquid is sucked into the branched channel network. Once the liquid is added to the inlet reservoir (Fig. 2a), it starts flowing in the capillary system. The limited amount of the liquid in the inlet reservoir will be consumed in the system (Fig. 2b). As the inlet reservoir becomes empty (Fig. 2c), the liquid will primarily be confined in the first branch and the supply channel. The liquid cannot proceed in the second branch because the capillary pressure in the second branch, P_{c2} , is less than the capillary pressure at the pinned liquid–vapor interfaces, P_{c0} and P_{c1} . Also, the liquid cannot be sucked into the expansion in the first branch as the capillary pressure in the expansion, P_{c3} , is less than the capillary pressure at all the pinned liquid–vapor interfaces. Assuming evaporation is low, the liquid remains stable in the microfluidic network until a new source of liquid is added to the inlet reservoir, which has the effect of removing the pressure, P_{c0} , originally pinning the fluid to the inlet; fluid flow can then proceed into branch 2. An implementation for on-chip liquid storage will be further detailed in Mixing System Design section.

$$P_{c0} > P_{c1} > P_{c2} > P_{c3}.$$

3 Theory

The capillary pressure arises from the surface tension at the interface between liquid and gas phases. The capillary pressure jump across the liquid–vapor interface, P_c , can be expressed as (Bruus 2007)

$$P_c = \frac{\partial G}{\partial V}, \quad (1)$$

where G is the total surface energy and V is the liquid phase volume. The capillary pressure can be computed, knowing

the surface curvature of the liquid–vapor interface, from the Young–Laplace equation (Chakraborty and Chakraborty 2010)

$$P_c = -\gamma \nabla \cdot \hat{n}, \quad (2)$$

where γ is the liquid–vapor surface tension coefficient and $\hat{n}$ is the unit normal to the liquid–vapor interface. Using geometrical manipulation of Eq. (2) for a rectangular cross section channel, the capillary pressure jump can be expressed as

$$P_c = 2\gamma \cos \theta \left(\frac{1}{w} + \frac{1}{h} \right), \quad (3)$$

where w and h are the channel width and height, respectively, and θ is the contact angle between the liquid–vapor interface and the solid surface.

For a microchannel with a pressure drop, ΔP , and a hydraulic resistance, R , the flowrate, Q , can be expressed as

$$Q = \frac{\Delta P}{R}. \quad (4)$$

Considering the liquid phase resistances only (*i.e.*, neglecting the gas phase resistances), the relationship between the capillary pressure and the flowrate in each of the channel branches is found by equating the capillary pressure jump to the system pressure drop as follows:

$$P_{c1} = Q_1 R_1 + (Q_1 + Q_2) R_{up}; \quad (5)$$

$$P_{c2} = Q_2 R_2 + (Q_1 + Q_2) R_{up}. \quad (6)$$

Setting the flowrate in branch 2 to zero results in the following conditions

$$P_{c1} = Q_1 (R_1 + R_{up}); \quad (7)$$

$$P_{c2} = Q_1 R_{up}. \quad (8)$$

Inferring from Eqs. (7) and (8), in order to have flow in the first branch only, then

$$\frac{P_{c2}}{P_{c1}} < \frac{R_{up}}{R_1 + R_{up}}. \quad (9)$$

Fig. 2 A schematic drawing of a branched microchannel with inlet reservoir. **a** A liquid is added to the inlet reservoir. **b** The liquid starts flowing in the system. Once the reservoir becomes empty (**c**), the liq-

uid–vapor interface is pinned at the restriction (P_{c0}) and cannot proceed in either of the downstream branches

The derived flow criterion, Eq. (9), defines a design rule prescribing when liquid will flow preferentially in only one of two branched channels.

Referring to Eq. (3), which defines the capillary pressure in a rectangular microchannel, the flow criterion can be reformulated as

$$\frac{\cos \theta_2 \left(\frac{1}{w_2} + \frac{1}{h_2} \right)}{\cos \theta_1 \left(\frac{1}{w_1} + \frac{1}{h_1} \right)} < \frac{R_{up}}{R_1 + R_{up}}. \quad (10)$$

Note that for a system with a uniform contact angle (*i.e.*, $\theta_1 = \theta_2$), the flow criterion is only dependent on the geometrical parameters (and the liquid viscosity if the viscosity varies within the system).

4 Mixing system design

To process a biological sample (*e.g.*, whole blood, plasma, saliva, urine) the first step in many microfluidic systems is to mix the sample with reagents. The accuracy, speed, efficiency and mix ratio of the mixing step vary based on the application and the assay being performed. In a capillary-driven system, bubble-free mixing is often a requirement. In this section, the design of a valveless system for bubble-free mixing with a given mix ratio is outlined.

Figure 3a shows a generic valveless system for performing a mixing step. The system consists of loading ports for the sample and reagents, a supply channel connecting both reservoirs through capillary retention valves (CRV), a reagent storage branch, a mixer branch, and expansion elements. The same scheme illustrated in Fig. 2 is used to

store the reagents on chip. The width of the various elements is as follows:

$$w_1 < w_6 < w_2 < w_5 < w_3 < w_4.$$

The dimensions of the channels of the system shown in Fig. 3a are presented in Table 1.

The mixing procedure in the system is as follows: First, the reagents are loaded into the inlet port and flow into the supply channel. Expansions 1 and 2 prevent the flow from entering branches 1 and 2 during filling of the supply channel due to the low value of capillary pressure at the expansions relative to the supply channel. After the supply channel is filled, the liquid flows preferentially through expansion 1 (note that the width of expansion 1 is less than that of expansion 2) and fills the storage (branch 1), as shown in Fig. 3b, as the ratio between the capillary pressure in expansion 2 and capillary pressure of expansion 1 satisfies the condition in Eq. 9. The liquid continues to flow into the reagent storage channel until the inlet reservoir is emptied. The capillary retention valve just after the inlet port prevents the liquid from proceeding once the inlet reservoir is empty. The liquid is maintained in the storage channel (branch 1) as the capillary pressure in the

Table 1 Channels' dimensions of a valveless mixing system

Channel	Symbol	Width (μm)
CRV	W_1	10
Supply channel	W_2	25
Expansion 1	W_3	150
Expansion 2	W_4	200
Branch 1	W_5	50
Branch 2	W_6	20

Fig. 3 **a** A schematic drawing of a generic valveless system. **b** Liquid interface propagation in the same system shown in **(a)**

storage is higher than the capillary pressure in expansion 2, but lower than the capillary pressure in the mixer branch (i.e., the meniscus will acquire a shape that balances the capillary pressure and results in a zero net flow condition).

When liquid is added to the sample port, flow proceeds through expansion 2. Once the liquid reaches the mixer (branch 2), it starts to suck the reagents from the storage since the capillary pressure in the branch 2 is higher than in branch 1. The reagents and sample proceed to mix in the mixer branch, as shown in Fig. 3c.

The mixing ratio (r_{mix}) between the sample and reagents can be defined as the ratio of the flowrates from the reagent storage (Q_r) to the sample port (Q_s). The flowrates can be estimated from the value of the capillary pressure at the interface in the branch and the respective hydraulic resistance to the flow in this branch (Eq. 4).

$$r_{\text{mix}} = \frac{Q_r}{Q_s}; \quad (11)$$

$$Q_r = \frac{P_j - P_{\text{cr}}}{R_r}; \quad (12)$$

$$Q_s = \frac{P_j}{R_s}; \quad (13)$$

$$P_j = P_{\text{cm}} - Q_m R_m, \quad (14)$$

where P_j is the pressure at the junction between the supply channel and the mixer channel, P_{cm} , Q_m and R_m are the capillary pressure, flowrate and hydraulic resistance in the mixer channel, respectively, P_{cr} and R_r are the capillary pressure and hydraulic resistance in the reagent storage channel, respectively, and R_s is the hydraulic resistance in the sample supply channel, as shown in Fig. 3b.

Substituting Eqs. 12, 13 and 14 into Eq. 11 yields

$$r_{\text{mix}} = \frac{(P_{\text{cm}} - Q_m R_m) - P_{\text{cr}}}{(P_{\text{cm}} - Q_m R_m)} * \frac{R_s}{R_r}. \quad (15)$$

Since the hydraulic resistances of the mixer channel and reagent storage channel, as well as the flowrate in the mixer channel, are a function of the liquid–vapor interface positions, Eq. (11) indicates that the mixing ratio value varies with time. However, the variation in the mixing ratio with time can be minimized by appropriate design of the valveless system, as will be shown later in Sect. 6.

5 Experimental setup

Figure 4 shows a representative cross section of a fabricated device. Processing started on 200-mm silicon wafers. The test devices were fabricated in silicon using deep reactive ion etching (DRIE). Details of the fabrication procedure can be found in Majeed et al. (2012). The etch depth was approximately 250 μm . To render the surfaces hydrophilic, 200 nm of thermal oxide was initially grown on the silicon surfaces. 100 nm of the thermal oxide was stripped by wet chemical etching leaving 100 nm of clean thermal oxide remaining on the silicon surfaces. A glass wafer was anodically bonded to the silicon wafer. The silicon wafer was thinned to about 400 μm in thickness by grinding the backside. Access holes and vents were then etched from the backside of the silicon wafer. The silicon/glass wafer was then diced into 34 by 28 mm dies. Figure 5 shows a photograph of the fabricated generic valveless system described in the previous section.

Fluorescently dyed water was used to visualize the fluid behavior in the fabricated devices. An inverted microscope with a high-speed CMOS camera was used to track the liquid movement through the glass cover. Figure 6 shows the sequence of filling a generic valveless system shown in Fig. 3. The reagents are loaded into the access hole at the backside of the test device. The reagents stop in the beginning of both branches until the supply channel is completely full since the capillary pressure in the supply

Fig. 4 A schematic cross section through the fabricated test devices

Fig. 5 A photograph of the fabricated valveless system shown in Fig. 3

channel is much higher than the capillary pressure in the branches (Fig. 6a). After filling the supply channel, the liquid enters the first branch only as it satisfies the condition in Eq. 9 (Fig. 6b). Hence, all the reagents are stored in the first branch and the reagent port becomes empty (Fig. 6c). Finally, the sample is added to the other port to initiate flow in the second branch. The stored reagents and sample are both sucked into the second branch where mixing occurs (Fig. 6d).

The fluid front position was tracked with time to generate benchmark results for the model (to be discussed in the next section). The images were collected at a frame rate of 200 frames per second. A MATLAB® in-house code was developed to extract the liquid–vapor interface positions along the serpentine reagent storage and mixer channels. The uncertainty of the position detection of the interface is estimated to be approximately $\pm 14 \mu\text{m}$, derived from the pixel size and thickness of the liquid–vapor interface observed in the images. The uncertainty is estimated to be $\pm 2.7 \times 10^{-11} \text{ m}^3/\text{s}$ for flowrate measurement. The results are presented in Results and Discussion section.

Fig. 6 Image sequence of fluorescently dyed water propagating in the capillary system shown in Fig. 3. **a** The reagents are loaded to the port and stop in the beginning of the branches. **b** The liquid enters the first branch only. **c** All the reagents are stored in the first branch. **d** The sample is added to the other port to initiate flow in the second branch. The stored reagents and sample are both sucked into the second branch where mixing occurs

6 Modeling

The usage of electrical circuit analysis methods in analyzing microfluidic networks is a well-established practice. Oh et al. (2012) showed concrete examples of pressure-driven microfluidic networks modeled using electric circuit analogs. Many other examples exist in the literature (Ajdari 2004; Mansouri et al. 2007; Plecis and Chen 2008). However, analyzing capillary-driven microfluidic systems with standard electrical circuit analysis tools proves more challenging since the hydraulic resistance is a function of the liquid–vapor interface position. Thus, a time-dependent solution algorithm often has to be utilized, as demonstrated in Kang and Banerjee (2011) and Safavieh et al. (2014). We adapted the model shown in Kang and Banerjee (2011) to include the different components of the valveless system. The model uses Kirchhoff's voltage and current laws to analyze the electrical circuit equivalent to our capillary microfluidic system.

Except for the initial stage of liquid filling the capillary system where velocities can be sizable, the Reynolds number is typically low in the capillary microfluidic network. Consequently, the inertial effects will be neglected and only viscous effects will be considered. We also considered fully developed flows in all our calculations since most of the channels are much longer than the hydraulic diameter [note that the hydrodynamic entrance length is approximately 0.6 times the hydraulic diameter for very low Reynolds numbers (White 2006)].

For a rectangular microchannel partially filled with liquid, as shown in Fig. 7, the position of the liquid–vapor interface, x , determines the value of hydraulic resistance as follows (White 2006):

$$R = \frac{3(\mu_l x + \mu_g(l-x))}{4wh^3} \left[1 - \frac{192h}{\pi^5 w} \sum_{i=1,3,\dots}^{\infty} \frac{\tanh\left(\frac{i\pi w}{2h}\right)}{i^5} \right]^{-1}, \quad (16)$$

where μ_l and μ_g are the dynamic viscosities of the liquid and gas phases, respectively, and w , h and l are the channel's width, height and length, respectively. The infinite series was approximated by the first five terms.

Fig. 7 A schematic drawing of capillary flow in a microchannel

The velocity of the liquid–vapor interface in the channel, u , can be estimated as

$$u = \frac{P_c}{RA}, \quad (17)$$

where P_c is the capillary pressure in the channel and A is the channel's cross-sectional area. The analogous electrical circuit to the generic valveless system is shown in Fig. 8. After each time step, the velocities are calculated and used to update the liquid–vapor interfacial positions in each of the branches for the proceeding time step.

7 Results and discussion

Five different designs were tested experimentally. The ratio of capillary pressure in expansion 1 to expansion 2 is varied (see Fig. 3). All designs have the same hydraulic resistance for the supply channel (*i.e.*, the upstream resistance, R_{up}).

Table 2 shows the experimental test results of the five designs. The first three designs satisfy the design criterion in Eq. 9. Experimental observations reveal that the liquid flows into branch 1 only and stops until additional liquid is added to the sample port (the sequence of events is shown in Fig. 6). In design number 5, liquid flow was observed in both branches 1 and 2 as it does not meet the criterion in Eq. 9. Design number 4 is borderline; therefore, it stopped the flow in only 1 out of 3 experiments.

Figure 9 shows the propagation of the liquid–vapor interface in the valveless system with time for design number 1. The solid lines were generated by solving the fluid network model indicated in Fig. 8. The contact angle between the liquid–vapor interface and the channel walls was set to 45° corresponding to the typical value of the contact angle of the test devices. The modeling results are validated against the experimental measurement of the liquid–vapor interface position with time in the supply channel and the first branch. The position of the interface is measured from the beginning of each channel.

Indeed, as observed in the experiments, the liquid propagates first in the supply channel and the inlet part of each branch and then stops at the expansions of both branches

Fig. 8 Symbolic electrical analogy representation of the valveless system shown in Fig. 3

Table 2 Experimental test results of the valveless system

Design	P_{c1}/P_{c2}		$R_{up}/(R_{up} + R_1)$	Flow status
1	0.603	<	0.833	One branch only
2	0.687	<	0.833	One branch only
3	0.694	<	0.833	One branch only
4	0.844	>	0.833	Both branches
5	0.911	>	0.833	Both branches

Fig. 9 Liquid–vapor interface propagation in the valveless system calculated using electrical analogy modeling and measured experimentally from images taken at 200 frames per second

until the supply channel is completely filled. Afterward, the liquid flows into the reagent storage channel (first branch) only. The flow stops once the entire reagents volume is transferred from the reservoir into the first branch. For the second branch (mixer channel), the liquid stops at the expansion 2 and then does not proceed further, as shown in Fig. 9.

After the sample is added to the sample port, the liquid starts to flow in the mixing channel. Figure 10 shows the flowrate in the mixing channel (branch 2) with time. The flowrate starts with a very low value at time $t = 0$ (when fluid is added

Fig. 10 Flowrate tracking with respect to time after the sample fluid is added to the sample port in the mixer part of the valveless system using electrical analogy modeling and experimental measurements

to the sample port) as the liquid–vapor interface traverses expansion 2 (see Fig. 3). Then, the flowrate increases sharply due to the sudden increase in capillary pressure in the mixer channel. As the liquid–vapor interface continues to proceed in the mixer channel, the capillary pressure remains constant since the channel cross section does not change. However, the hydraulic resistance increases with time as the liquid fills the channel leading to a decrease in flowrate with time, as shown in Fig. 10. The electrical analogy model results are in good agreement with experimental measurements of the liquid–vapor interface velocity, albeit with some scatter in the experimentally determined velocities. The velocity measurements were generated using central differencing of the interfacial position taken from successive image frames.

The relative flow fraction of sample and reagents to the total flowrate is shown in Fig. 11a. The mixing ratio (sample flowrate divided by reagents flowrate) varies considerably with time from 0.4 at the beginning of the mixing process to 4.5 at the end of the process. This change results from the variable resistances and flowrate values in Eq. 15 which governs the mixing ratio in the valveless system.

Fig. 11 Flow fraction of the sample and reagent channels out of the total mixer flowrate: **a** without optimizing the flow parameters and **b** with optimized parameters

However, by optimizing the flowrate and upstream hydraulic resistance parameters in Eq. 15, the variability of the mixing ratio with time can be reduced. The main parameters affecting the mixing ratio variability are the supply channel resistance and the reagent storage channel resistance and capillary pressure. By decreasing the supply channel length to half of its current value and doubling the width of the supply channel and the reagent storage channel, the variability of the mixing ratio can be decreased significantly, as shown in Fig. 11b. After the optimization, the mixing ratio value varies from 4 at the beginning of the mixing process to 4.2 at the end of the process. However, this decrease in mixing ratio variability comes at the cost of the filling time, which increased from 4.3 s in the original design to 145 s in the optimized case.

8 Conclusion

In this work, an easy-to-fabricate, robust capillary system implementing valveless flow control was demonstrated. A design criterion for flow control was developed analytically. The analytical condition details the link between the capillary pressure ratio of two parallel branches and the hydraulic resistances of the microchannels. Capillary forces drive the liquid flow preferentially through only one of the parallel branches as long as the condition is satisfied. Test devices were fabricated and tested experimentally. Experimental results show good agreement with the flow control design criterion. Furthermore, electrical analogy modeling was conducted to investigate the liquid–vapor interface propagation in the valveless system. The electrical analogy modeling results were validated against experimental measurements of the liquid–vapor interface position with respect to time. The experimental results showed good agreement with the electrical analogy modeling results. This derived flow control condition can aid in the design of capillary-driven microfluidic systems where sequential flow is required or for bubble-free mixing systems without using capillary trigger valves.

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