# Lab on a Chip

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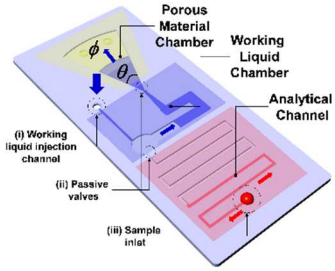
A Self-powered Imbibing Microfluidic Pump by Liquid Encapsulation (SIMPLE), a disposable pump for microfluidic POC devices.



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TECHNICAL INNOVATION

# Self-powered Imbibing Microfluidic Pump by Liquid Encapsulation: SIMPLE

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*Reliable, autonomous, internally self-powered microfluidic pumps are in critical demand for rapid point-of-care (POC) devices, integrated molecular-diagnostic platforms, and drug delivery systems. Here we report on a Self-powered Imbibing Microfluidic Pump by Liquid Encapsulation (SIMPLE), which is disposable, autonomous, easy to use and fabricate, robust, and cost efficient, as a solution for self-powered microfluidic POC devices. The imbibition pump introduces the working liquid which is sucked into a porous material (paper) upon activation. The suction of the working liquid creates a reduced pressure in the analytical channel and induces the sequential sample flow into the microfluidic circuits. It requires no external power or control and can be simply activated by a fingertip press. The flow rate can be programmed by defining the shape of utilized porous material: by using three different paper shapes with circular section angles 20°, 40° and 60°, three different volume flow rates of 0.07 µL/s, 0.12 µL/s and 0.17 µL/s are demonstrated at 200 µm x 600 µm channel cross-section. We established the SIMPLE pumping of 17 µL of sample; however, the sample volume can be increased to several hundreds of µL. To demonstrate the design, fabrication, and characterization of SIMPLE, we used a simple, robust and cheap foil-laminating fabrication technique. The SIMPLE can be integrated into hydrophilic or hydrophobic materials-based microfluidic POC devices. Since it is also applicable to large-scale manufacturing processes, we anticipate that a new chapter of a cost effective, disposable, autonomous POC diagnostic chip is addressed with this technical innovation.*

The use of point-of-care (POC) assay which integrates sample processing, fluid handling, signal amplification and detection into a single microfluidic diagnostic chip has a huge potential to influence the healthcare in both developing and developed countries[1,2]. Among other characteristics of an ideal diagnostic test as defined by the World Health Organization (WHO)[3,4] it should be affordable, user-friendly, rapid and equipment-free. Therefore, the reliance on large equipment such as syringe pumps, microscopes and computers needs to be overcome.

The external pumping mechanisms (syringe pumps, compressed air, electro-pneumatic systems, high-voltage supplies or motors) make device control more complex, cumbersome and expensive; therefore, they are to be avoided in the POC diagnostic chip, especially in a low resource environment scenario. For these reasons, a number of on-chip pump approaches have been proposed and investigated[2,5]. However, most of the on-chip pumps require an external control and electrical power supply.

Completely autonomous and internally self-powered pumping mechanisms are much sparser, but some of them have been successfully implemented and commercialized, such as the pregnancy test and some other lateral flow immunoassays[6]. A similar and very promising platform is paper microfluidics[7,8] or wicking in textiles[9]. Fluid flow in this kind of platforms requires only the capillary force with no pumping, valves or energy sources necessary. This makes such systems cheap, small and easy to dispose of; however, transport of cells in a paper matrix and decreased effectiveness of the paper assays due to biological sample interaction with cellulose fibers can be a drawback.

Another concept that is using the capillary drag is the microfluidic capillary system (CS)[10]. The liquid is dragged into the system by capillary forces which are enhanced by a branched microstructure (capillary pump –CP) at the end of the system. CSs need no external power supply or control devices. The pumping and valving elements can be integrated into the microfluidic system. This kind of systems can be implemented in silicon[11], PDMS[12] or other materials; however, these materials need surface modifications to be hydrophilic or hydrophilisized.

A different liquid transport concept is the co called "degas-driven flow"[13,14,15]. In this pumping mechanism the inherently high porosity and air solubility of the PDMS is used. The material is first evacuated in a vacuum chamber to remove air molecules from

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†Electronic Supplementary Information (ESI) available: Movie 1; SIMPLE pump operation, Movie 2; Working liquid prefilling, Supplementary; Supplementary Equations. See DOI: 10.1039/b000000x/

the bulk of PDMS. When the material is taken out of the vacuum chamber or a vacuum-sealed container the air fills the microchannels. Gas inside the microfluidic channel diffuses into the degassed PDMS which results in the lower pressure inside the microchannel. The droplet of a liquid sample is deposited onto the inlet of the microfluidic channels and gets drawn into the microfluidic chip due to the pressure difference. Since this phenomenon is limited to permeable materials Li *et al.* suggested to use it as a "place'n'play" portable pump[16].

Several other passive pumping methods have been presented and investigated: a difference in the surface curvature of differently sized droplets can be used as the flow driving force[17,18], a gas-producing chemical reaction can create a pressure difference which pushes the fluid through a microfluidic system[19], the evaporation of a liquid can induce a concentration difference as the driving force for an evaporative pump[20,21,22], also gravitation force was investigated for a gravitational pump[23,24], finger force for a finger-activated pump[25,26] and textile wicking for thread pump[27]. Among the described pumping mechanisms they all have their pros and cons; however, few of them fit most of the POC requirements that are: autonomous, cheap, disposable and robust.

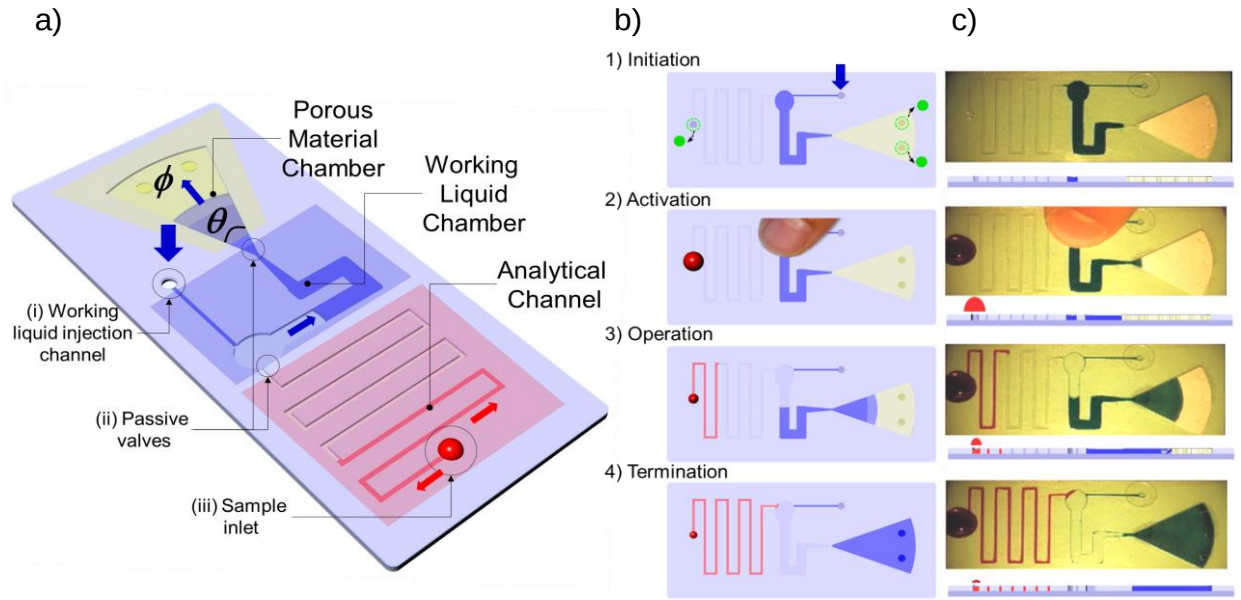


**Fig.1** The logic and design of a self-powered imbibing microfluidic pump by liquid encapsulation (SIMPLE) (a). Sequential pump operation is shown in the schematic (top view) (b), and an experimental presentation (c). Initially, the chip is prefilled with the working liquid (blue) through the inlet denoted by thick blue arrow and encapsulated by impermeable protective foil patches (green circles). Before the activation, the foil is removed, and the sample (red) is deposited over the inlet hole and the pump can be activated by a temporary finger force. When the working liquid touches the paper, the finger can be removed and the pump is activated. The pump is working until the working liquid saturates the paper or until all of the fluid has been sucked into the paper.

In order to solve the problem of the microfluidic POC pump, we propose a novel concept of a Self-powered Imbibing Microfluidic Pump by Liquid Encapsulation (SIMPLE) that will better suit the need of a POC diagnostic chip. The pump requires no external power or control. It is autonomous, robust, simple, disposable and can be used with a hydrophobic polymer microfluidic system and is therefore applicable to mass replication technologies.

The schematic of the SIMPLE design is presented in Fig.1. The chip can be divided into three functional parts. The first part is the analytical channel, the second is the working liquid chamber and the third is the hydrophilic porous material (paper) chamber (Fig.1a). Initially, the analytical channel is dry or contains a reactant liquid. The working liquid is preloaded into the container at the end of the microfluidic chip fabrication process and stored on-chip until application (Fig 1b and 1c). There is a passive valve[28] between the analytical channel and the working liquid chamber, to prevent the working liquid (blue) from spilling into the analytical channel. This passive valve is a rectangular shaped channel of 400 $\mu$m width. After the deposition of the sample droplet (red), at the opening of the inlet channel, the pump is activated by a force (i.e. with a fingertip) applied at the top of the encapsulated liquid chamber. There is another passive valve between the working liquid chamber and the porous material chamber in order to prevent a spontaneous activation of the pump. The second valve is of a trapezoidal shape with the thinnest side width of 400 $\mu$m. This passive valve and the volume of encapsulated liquid define the sufficient force for the pump activation. Force pushes the working liquid into the activation channel toward the paper. When the encapsulated liquid touches the paper, the pump is activated and the external activation pressure can be released. Now, the paper absorbs the working liquid and a reduced pressure is created in the emptying working liquid chamber. The reduced pressure results in a suction of the sample fluid into the analytical channels. The sample fluid will now fill the analytical channel. The pumping terminates when the paper is saturated by the encapsulated working liquid or all of the working liquid is sucked into the paper. This integrated imbibing microfluidic pump by liquid encapsulation requires no external power or control and can be simply activated. The flow can be modified by the shape of the porous material. The pump is autonomous, robust, simple, disposable, and can be integrated in hydrophilic or hydrophobic polymer microfluidic systems. In addition, the pump has practically no dead volume, it

does not require a high resolution microfabrication and due to its modularity it is easy to integrate into existing microfluidic designs.

The basis of the described pumping mechanism is the imbibition of the encapsulated liquid into a porous material. This phenomenon is commonplace in nature and in technological processes. The flow into two-dimensional porous material is derived from the Darcy's law. After the integration and inserting of the appropriate permeability model we arrive at an implicit equation which describes the time development of liquid penetration into infinite circular porous material from an unlimited circular source (see Appendix 1 and Fries[29] and Hyväluoma et al.[30] for more details):

$$\left(\frac{r}{r_0}\right)^2\left(ln\frac{r}{r_0}-\frac{1}{2}\right)+\frac{1}{2}=\frac{\gamma a cos\theta}{6\mu r_0^2}t, \quad (1)$$

where $r$ is the radius of the wetted area, $r_0$ is the radius of the liquid source, $a$ is an effective porous radius, $\gamma$ a surface tension, $\mu$ is the dynamic viscosity of the liquid, $\theta$ is the wetting angle of the liquid on the porous material and $t$ is time.

The wicking is the dominating force for the pumping and in our case largely exceeds the friction force and resistance force due to nonwetting of the channel walls (Appendix 2). This capillary pressure can easily reach several tens of kPa (in our case around 38 kPa), and depends on structure and moisture content of the porous material.

Our elementary proof-of-concept microfluidic device was fabricated in the foil laminating technique as described by Yuen *et al.*[31]. In our case the microfluidic network is cut into a pressure-sensitive adhesive (PSA) double sided foil (3M, 467MP 200MP adhesive, 200 $\mu m$) and sandwiched between two hydrophobic laser printer transparency foils (Ibico, Ibiclear® PVC - 200 $\mu m$). For cutting the channel structure, a Cameo Silhouette digital tabletop craft cutter was used. The channel width down to 200 $\mu$m can be fabricated with this method. In our case the width of the analytical channel was 600 $\mu$m, the "working liquid channel" was 2.6 mm in width and the activation chamber was 5 mm wide. A piece of cut filter paper (Schleicher & Schuell, grade 595) was inserted into the pumping chamber to absorb the working liquid during the pump operation. The three layer microfluidic device was laminated together as shown in Fig. 2. After lamination, the working liquid is carefully injected through a special "working liquid injection channel" into the "working liquid chamber". This was done, with sufficient precision, manually by a syringe (Video 2). The syringe needle was prolonged by silicone tubing which was pressed onto the injection channel opening for precise filling of the container (approx. 20 $\mu$L of working liquid). After filling the injection channel, the opening was closed by a PSA tape patch. Blue dyed water was used as the working liquid and red dyed water was used as the sample in our experiment. Beside the sample and the working liquid inlet holes there are two additional vent holes in the top PVC foil which allow for air outlet from the paper when air is replaced by the working liquid during the pump operation. This four inlet/outlet holes are sealed with additional impermeable protective foil to prevent contamination and evaporation during the chip storage. The chip is finally ready for storage, transportation and application.

To test the pump functionality we performed several experiments with various pumping powers. There are at least two ways to passively control the sample flow rate inside the channels. The first one is via the control of pumping power with the type, shape and size of the absorbing medium (paper) and the second is via control of flow resistance inside the channel with the shape and geometry of the channel. We tested the influence of the shape and size of the absorbing paper on the flow rate. It can be understood that if the liquid is absorbed into homogenous and isotropic material the radius-time dependence

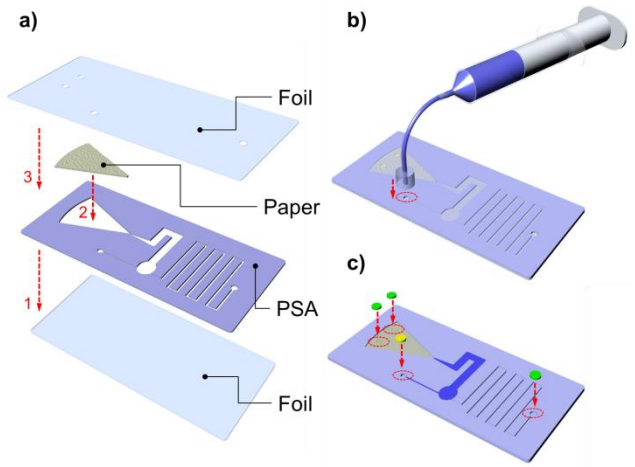


**Figure 2:** The fabrication steps of a self-powered imbibing microfluidic pump by liquid encapsulation (SIMPLE) using the lamination method. (a) Device consist of four parts: bottom PVC foil, PSA layer with microchannels, insert of filter paper and top covering PVC foil with inlet and outlet holes. Device is assembled in three steps denoted by numbers. (b) The working liquid is prefilled into the working liquid chamber and c) the holes are patched by PSA foil for the storing and application. The yellow patch closes the working liquid inlet permanently and the three green patches are removed prior the device application.

would remain the same for any circular section of this material. However, the volume flow rate will not remain the same, but will have a linear dependence on central angle $\omega$, such that

$$Q=\frac{dV}{dt}=\frac{\omega}{360°}\phi 2\pi r h\frac{dr}{dt}, \quad (2)$$

where $Q$ is the volume flow rate, $\phi$ is the porosity of material, $r$ is the wetting radius in the porous material and $h$ is the thickness of the porous material. Further, the relative importance of individual pressure terms can be estimated (Appendix 2). In our chip geometry the wicking force exceeds the Young-Laplace and Hagen-Poiseuille terms for almost two orders of magnitude and is strongly dominating the process. In this case the radius-time dependence in the porous medium can be approximated by equation (1). The volume flow rate of sample into the analytical channel will be the same as the volume flow rate of the working liquid into the porous material and the speed of the sample flow ($v=Q/S$) is proportional to the central angle $\omega$. Our absorbing paper was cut in the shapes of circular sectors with different central angles $\omega$. We used three different angles of 20°, 40° and 60°. The results of experiments are shown in Fig.3. The size of the paper surface (volume) was the same in all three cases in order to preserve the same pumping energy.

We see from the figure that the speed of the sample liquid front in the analytical channel depends on the shape of the insert paper. The larger the angle, the larger the pumping power. To compensate for the transition phenomena induced by finger activation, we start

measuring the speed at the first curve in the analytical channel and finish at the last curve of the channel (Fig.3a). In all three cases the speed slightly decreases with filling the channel, but remains fairly constant. Similar observation was reported in Mendez et al.[32] and Wang et al.[33]. In order to estimate the linearity of the curves we performed a linear fit to this curves and read out the liquid speeds of 0.58 mm/s, 1.00 mm/s and 1.42 mm/s corresponding to respective volume flow rates of *0.07µL/s, 0.12 µL/s and 0.17µL/s.* However, we encounter a relatively high measuring uncertainty which is mostly the consequence of the elementary fabrication procedure which results in rough sided walls of the channels and uneven channel widths. We can still conclude that the liquid flow speed can be modulated fairly well by the paper shape.

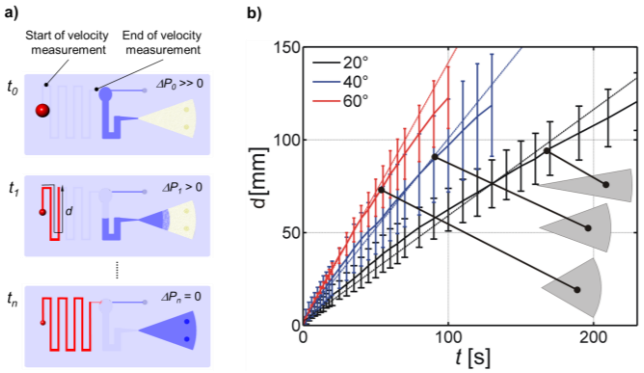


**Figure 3:** Characterizations of volume flow rate modulation (a). A passive sample speed modulation with the shape of the absorbing paper (b). Paper was cut in shapes of circular sectors with different central angles of 20° (black), 40° (blue) and 60° (red). Despite the relatively high measuring uncertainty (error bars) which is a consequence of the crude fabrication technique, we can conclude that the speed (volume flow rate) is relatively constant and can be modulated by the paper shape. Straight lines show a linear fit to experimental results.

A novel self-imbibing microfluidic pump by liquid encapsulation for autonomous, robust, easy, and disposable POC microfluidic devices is demonstrated. We utilized a simple foil lamination technique to demonstrate the simplicity of the fabrication method for the integrated imbibition microfluidic pump. Further, we show the ease of pump activation and operation and present a volume flow rate modulation via the shape of the absorbing paper. We demonstrate SIMPLE pumping of 17 µL of sample; however, the flow volume can be predefined by the volume (area, thickness) and type of paper and working liquid volume and can be increased to several hundreds of µL for various sample volume requirements of different diagnostics. Finally and most importantly, the imbibition pump is applicable to mass-production technology and therefore perfectly suited for the need of a POC diagnostic chip.

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