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TECHNICAL PAPER

Design and fabrication of micro-mixer with short turns angles for self-generated turbulent structures

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Abstract Micro mixing process is one of the most important drugs, synthesis and process technology. Many expectations toward enhanced fluids miscibility, yield and purity, improved safety, and access to new discovery are directed to the micro mixing technology. In microfluidic technology, mixer is the most important component because of its ability to operate either active or passive micro mixing. The study reports a simple inexpensive and cost effective micro mixer design/simulation/fabrication using 4.3a COMSOL Multi-physics software package and homemade photolithography equipment. The micro mixer was fabricated using PDMS and glass substrate with master mold fabricated with SU8 for rapid prototyping and reusability, the micro mixer and the whole mixer were fabricated in less than 20 min. Mixing profiles were observed for the efficiency evaluation by dropping four samples urine through the four inlets and collecting the sample at outlet. Flow rate and mixing efficiency were quantitatively measured by analyzing the recorded flow profiles and values of the image collected from the high powered microscope at inlet and outlet locations. and sample flow and mixing were compared for flowing efficiency and flow rate using different concentrations and the results indicate that the mixing efficiency of 98% was obtained at Reynolds number (Re) <2.

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1 Introduction

In recent time, fluid handing in microfluidics for biosensor has received full recognition because it allows a completely new platform for assays in the Pico-liter to Nano-liter size where miniaturization challenges such as: liquid handling, molecule adsorption and absorption had been a problem to further scale down scientific experiments. Obtaining a homogenous mixing in microfluidic system is highly challenging due to it low Reynolds number and thus dominated by laminar flow (Adam et al. 2013a, b). The mixing on this domain mostly characterized primarily on the slow diffusion process at the interface between the different fluid containing the various chemical species (Berry et al. 2012; Berthier et al. 2012). Scientist have done various strategies for increasing the mixing efficiency using approaches rely both on stretching the mixing interface, as well as promoting advection within the fluid stream mostly supported by external force which commonly called active mixing (Chee et al. 2012; Ko et al. 2014; Sackmann et al. 2014; Fiddes et al. 2010). However, the problem with such approach is that, it cannot be employed in some areas such biosensor operating based on probe-target based detection in which probe are mainly attached to sensing domain by some delicate linker and can easily be affected by small mechanical agitation (Karayiannis et al. 2010; Liu and Crooks 2011). Hence, Passive micromixers which employ specially designed geometries to increase the contact time and interface area between the various chemical species, are particularly interesting since they do not involve moving parts and thus have a high degree of reliability and are relatively easy to fabricate (Marques and Fernandes 2011; Novo et al. 2013; Amadi et al. 2010; Miller et al. 2011). The strategy employed in this micro mixer design is lamination splitting through zig zag geometry and recombining due to short turn angle promoting

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turbulent fluid flow motion which is related to the stretching and distortion of the interface by biased angular components (Toepke and Beebe 2006; Adam et al. 2013c, d, e, f). Glass and silicon micro-machining is commonly used for microfluidic devices due to its fidelity in achieving small feature resolution, surface stability and solvent compatibility (Gibb et al. 2014; Thom et al. 2012). Especially, glass is a material of choice in microfluidics due to its beneficial optical properties, and the strength of anodic bonding which allows an excellent resistance to high pressures (Vezy et al. 2011; Yang et al. 2011; Yeo et al. 2011; Young et al. 2011, 2013). Glass as a well-known material with minimum chemical reaction to the body is used in this paper. Several types of glass are used in mF devices, such as soda lime, quartz, and borosilicate. Due to the low cost and simplicity, commercially available microscopic slides have been used. Various masking techniques have been implemented to make micro channels using glass substrate. Different mask layers for chemically glass etching have been reported using different materials such as Cr, Cr/Au, polysilicon to deposit a layer as a mask on glass in order to make an open region for wet chemical etchant by different deposition methods such as CVD, LPCVD, sputtering or other methods which needs special clean room instruments. However, making a mold and allow this shape to be transferred to produced desired shape is highly challenging as the size becoming smaller are highly difficult to create complex structure. Hence, In this study we designed and fabricated by employing strategy in design that can promote lamination splitting through zig zag geometry and recombining due to short turn angle promoting turbulent fluid flow motion.

2 Channel simulation

The convection-diffusion equations is used for transport of fluid contained species across 45° short turn of zig-zag within micro channels. A comparison of the relative diffusive and convective transport time scales in the x-direction along the channel, can be predicted by Peclet number $Pe = u_0 L/D$, which presumed the molecular transport of the solutes in the $x$direction (along the length of the channel) is dominated by convection due to short angles. For typical values of $L(\theta)10\mu(45^{\circ}))$, u_0 (1×10^{-2} mm/s -1 mm/s) and D ($1-30 \times 10-6$ mm^2/s), values of Pe are between 11 and 1×10^4. This design shows convection dominates since $Pe > 10$.

From the above assumption, the equations governing diffusion and convection of solute in the several different regions that are identical in form, and can simplified to the following:

$$u_i \frac{\partial C_i}{\partial x} = D_1 \frac{\partial^2 C_1}{\partial y^2} \cdots \mathrm{D_i} \frac{\partial^2 C_i}{\partial y^2} \tag{1}$$

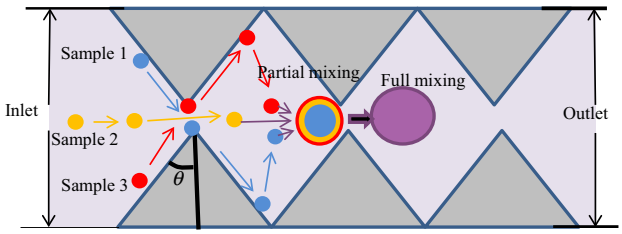


Fig. 1 Illustration of the mixing within grooved micro channel

where the sample concentrations in the channel 1 and other channels are C_1 and C_i respectively, and u_i is the velocity of the fluid in the horizontal (x) direction. The Poiseuille velocity profile in terms of the device geometric parameters is given by

$$u_i(y) = \frac{w_i}{2\mu_i}\left(-\frac{dp}{dx}\right)_i \frac{y}{w_i}\left(1 - \frac{y}{w_i}\right) \tag{2}$$

The collective contribution for degree of mixing initia on the walls by this flow field given by $\tau_i = \mu_i \left[\partial u_i(y)/\partial y\right]_{y=0.w_i}$. The corresponding flow rates are $Q_i = w \int_{y=0}^{y_i} u_i dy$, where the width of the channel is. Fluid enters each channel ($x = 0$) at a defined concentration, C_0, and there is no flux of solute across the upper wall of the top chamber. The above assumption does not strictly hold true for devices made from PDMS where gaseous substances' are employed, due to the high permeability of PDMS; it is a reasonable assumption for alternate less-permeable materials now being employed for devices. Samples consist of fine solute along the lower wall in the channel create a diffusive flux of mixing across the central mixing zone combining the two channels. Defining θ as the effective diffusion coefficient of the solute in the channel, and t the stiffness as illustrated in Fig. 1.

2.1 Device specifications

The microfluidic devices consist of two distinct designs to convey the fluids in the channels. The curvy design A is a structure that has a length from the device's opening to the next opening of 12.0 mm, device's width of 4.2 mm, and a channel width of 70 nm. The device is designed to have groovy shape with minor angle of 11° and major angle of 45° and stiff height (t) of 1 mm this done in order to enhance the liquid flow and mixing capability in the channel device. The design B consists of three microfluidic devices stacking vertically as shown. The device's design has a length of 12.0 mm, device's width of 5.75 mm, and a channel width of 70 mm. However, the second design has wider openings, measuring 0.7 mm for the for conveyers only. The device has 1.5 mm wide chambers The design structures and specifications can be seen as in Fig. 2a, b.

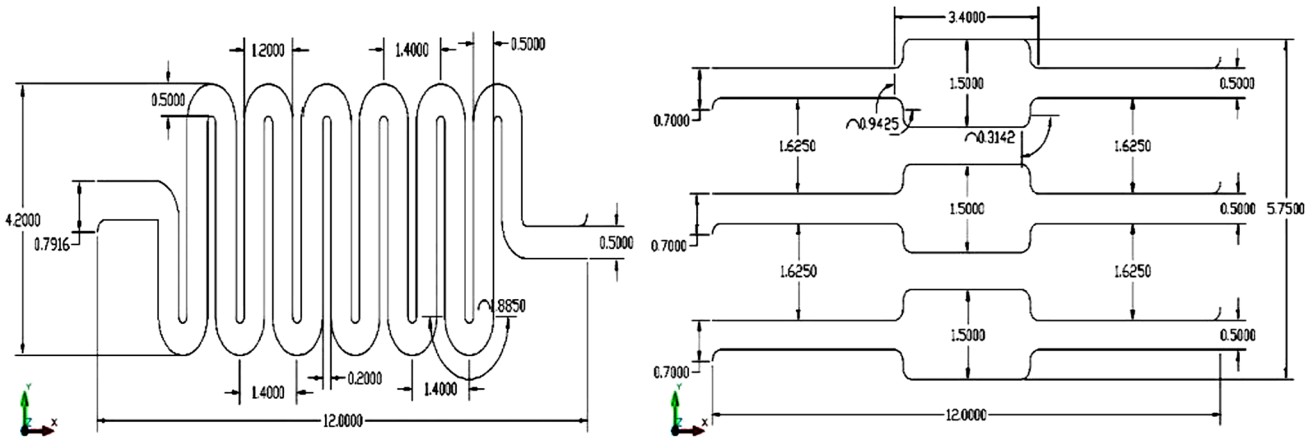


Fig. 2 The design structure and specifications of the micro channel device **a** channel design, **b** chamber design

3 Materials and methods

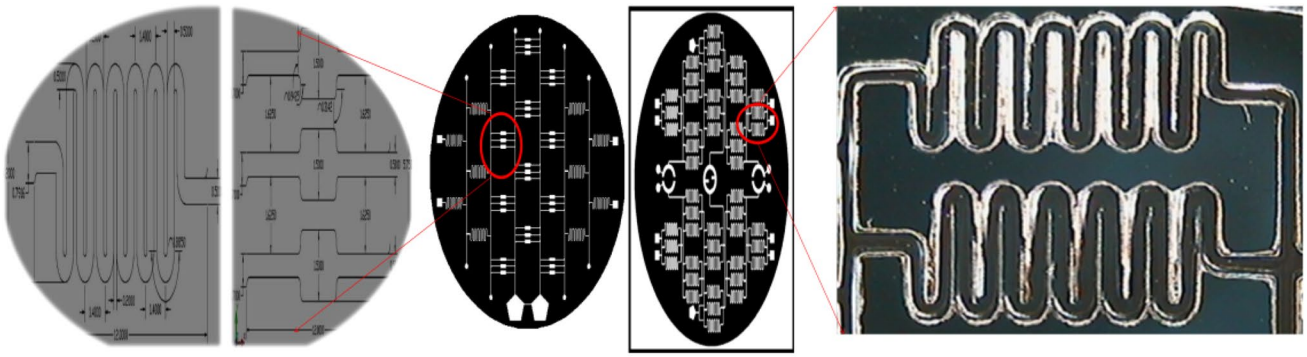
3.1 Materials

The study target device with cost efficient and portable, therefore, affordable materials are necessary, this novel approach tried to use as little materials and equipment as possible, especially for device the device fabrication. To fabricate the mold, we need glass substrate, SU8 negative resist, isopropyl alcohol (IPA) and petri dish.

3.2 Device fabrication

The micro channel mold preparation, it produced through transparent mask on which geometry are printed on. The most needs to allow the pilling up after fabrication, to achieve this, we lubricated the mold and it the entire curing dish with grease lubricant, the geometry of the channel are grove which have barriers that can control the flow and promote mixing. There are multiple ways to create this including rotating design and multi-junction design. However, each technique has positive and negative aspects about it but in general the approach with the least amount of steps and easiest to mass-produce is the 45° grove geometry is the best. There are several material being for microfluidic fabrication, how PDMS based micro fluidic device is able to succeed because of PDMS is very affordable, non toxic and easy to work with. Mold preparation, prior the, the glass substrates were cleaned using ethanol in ultrasonic bath to remove residues accumulated on the glass surfaces during storage. Surface residues could prevent adhesion of a photoresist onto the glass surface. Then, in order to create a pattern that will ultimately create a molding master for the microfluidic devices, a negative-tone photoresist SU-8 50 layer was spin-coated onto the cleaned glass substrate. This process was performed at 500 rpm for 15 s spin, followed by 30 s spin at 1,250 rpm. The spin acceleration is 100 rpm/s throughout. Then, the coated substrate was prebake at 90 °C for 75 min. After cooling down to the room temperature, the coated substrate was exposed to an ultraviolet light (300–400 nm) (KARL SUSS) under the prepared mask to impress the channel geometry onto the photoresist layer. The mask as shown in Fig. 1b was drawn in CAD software and professionally printed on transparencies. The exposure dose depended on the thickness of the SU-8 layer. Post-exposure bake of the substrate is followed at 90 °C for 15 min and then was allowed to cool to the room temperature. The pattern was then developed in Su-8 developer solution (MicroChem, Newton, MA) in the photolithography developing station. Once all excess photoresist has been lifted, the patterned master was rinsed in isopropyl alcohol (IPA) and blown dried in pure nitrogen stream. The master could be used multiple times to fabricate PDMS channels if carefully handled and stored in a sealed container away from light and humidity. An optical microscope is used to inspect the photoresist pattern that has now been printed on the glass substrate and the height of the pattern is measured using a surface profilometer.

After successful exposure and development process through masks as shown in Fig. 3, the curing process follows. The micro channels were formed in PDMS by replica molding technique. Replica molding is a process of casting the prepolymer against a master mold and generating a negative replica of the mold in PDMS. The Sylgard 184 silicon elastomer and its curing agent (Dow Corning, Midland MI) as shown in Fig. 2b were mixed at a 10:1 mass ratio, and it was stirred vigorously for 5–10 min until well mixed. The mixture was then placed in a vacuumed desiccator for degassing. The PDMS mixture was slowly poured onto the master mold and together cured at 75 °C on a hotplate for 45 min. Once the PDMS had solidified and cooled, it was peeled off from the master mold, with the micro channels structure engraved.



(a) **(b)**

Fig. 3 Mask for the design **a** micro chamber channel design, **b** micro channels

Sealing/bonding: Microfluidic devices made from PDMS typically need to be bonded to a glass slide before being used. The bonding process creates a tight seal between the PDMS and the glass, so that fluids and cells remain confined to the channel. This process is facilitated by a technique known as plasma bonding. In essence, exposing both bonding surfaces to oxygen plasma makes them extremely clean, which facilitates the bonding process. Oxygen plasma treatment can be used to covalently bond PDMS to itself or to glass. The mechanism is thought to be related to the breaking of bonds on each surface during treatment followed by the formation of Si–O–Si bonds when the two surfaces are brought into contact. Details of the process are poorly understood, but numerous groups have reported processing parameters that work, and a few have published systematic optimizations of bonding. In general, clean PDMS surfaces, low plasma powers, and short treatment times lead to the highest bond strengths. For the purpose of this study, PDMS were treated under oxygen plasma through plasma preen system machine and then taken into contact with glass substrate to complete micro channel structure glass substrate and the PDMS channel were cleaned with isopropanol, rinsed with deionized water, and dried with a stream of nitrogen. They were kept free from oils and dust. The two pieces are exposed to low-power oxygen plasma either in a plasma asher or in a PECVD and plasma-oxidized for 1 min. 12) The PDMS and glass are brought in contact within 10 min after oxygen plasma exposure for irreversible bonding. The molded PDMS pattern is then sealed against glass substrates (wafer, coverslips or slides) to provide confinement of the channels.

3.3 Devices testing

The micro channel thickness and surface morphology of micro fabricated mold were characterized by surface profiler. With a proper preparation steps, a well-developed master mold as shown in Fig. 3b. From the results, the mold has no deformity on the channel's and the surface analysis performed in 2 µm × 10 µm area on the micro channel, from the observation we found out that the average height of the micro chamber is between 20 and 70 micron there is slight surface roughness and this is due to the transparency of SU-8 mold on the wafer surface. However, this is and advantages for need as this will promote the turbulence phenomena. In overall, the surface of the mold is flat and is capable to produce good patterns of micro channels on PDMS elastomer. After the fabrication, the next step is preparing the micro channel for the flow characterization. The four urine samples were obtained. The samples were the urine before breakfast, after breakfast, after lunch and after dinner. The device is prepared with both inlet and outlet. To test for the amount of urine, the urine is drop in the inlet point and the urine flows into the branches and without any blockage. For the mixer testing, the same procedure is followed with four different sample samples, the urine mix with the each other and we obtained a uniform mixing with the channel (Table 1).

Table 1 Urine sample taken from three volunteer

Volunteer	Sample 1 Before breakfast (ml)	Sample 3 After breakfast (ml)	Sample 3 After lunch (ml)	Sample 4 After dinner (ml)
#1	10	10	10	10
#2	10	10	10	10
#3	10	10	10	10

4 Results and discussion

The fabrication has eliminated the need for spinner-coater system to achieve the desired 300 microns of SU-8 coating

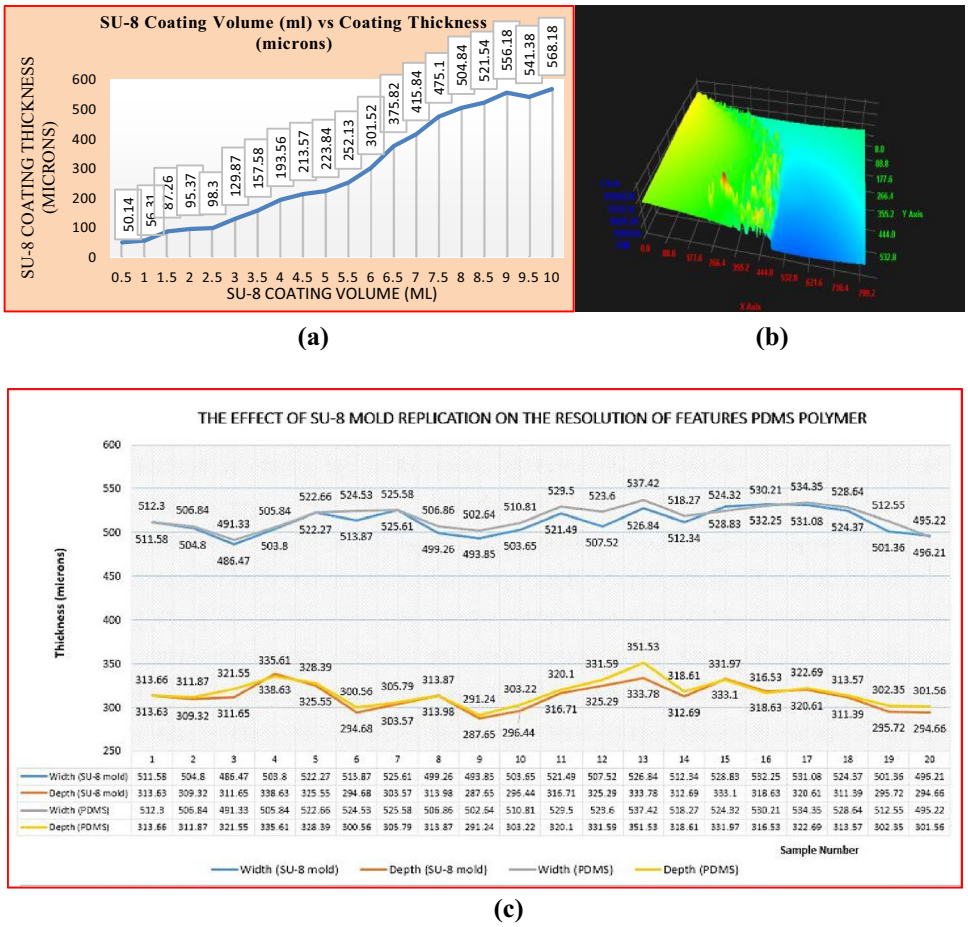


Fig. 4 **a** The graph of the relationship between the SU-8 resist coating volume (ml) and the thickness of the SU-8 resist (microns), **b** surface profiler image showing 70 µm with slight surface roughness, **c** the deviation in feature sizes (channels width and depth) of the PDMS polymer from the SU-8 master mold after the replication process

thickness for efficient fluid flow. Figure 4a below shows that by manually layering the substrate surface with 6 ml of SU-8 2010 (viscosity: 380 cSt) from a dropper, a desirable uniform layer of the resist of 71.52 microns thick could be achieved. The SU-8 resist layered was left for 120 min on a perfectly flat surface for the solution to self-spread on the substrate through gravitationally-driven inertial flow. At coating volume of 8–10 ml, the thicker SU-8 coating of 500.84 to 568.18 microns produced have exhibited severe non-uniformity due to the hill-like coating formation on the surface of the substrate this because as the SU-8 thickness increase, it need a higher setting to obtain a smooth surface thus, this show that one can produce perfect SU-8 based micro mold manually without using costly spin coater.

The characterization was conducted using surface profiler to measure the channels width, while channels depth was measured by implementing a surface profiler (Hawk 3D Surface Analysis) (Fig. 4a, b). The characterization result has found a slight deviation in the feature size of the PDMS (relatively larger feature dimensions) as compared to SU-8 master mold which was probably caused by the expansion of the polymer due to moisture effect. The issue could be rectified by increasing the duration of hard bake to 45–60 min to evaporate the remaining solvents this could be seen in Fig. 4c.

In Fig. 5, The model results for urine concentration at the each channel for the both layers fluid dragging capabilities and since the stiffness is corresponding to angular short turn. The least stringent set of mixing conditions for maintaining uniform concentrations is demonstrated down the length of the devices involves sparse mixing convolved at the lower end of the reported ranges of consumption rates through the three sample. For definite flow rates $Q_1 = 20$ *nL/s* (control by maximum allowable shear on the channel layers) with $Q_2 = 10Q_1 = 50$ *nL/s* in the upper channel angle as indicated in Fig. 1, the urine concentration profile is almost invariant along the length of the channel flow area except for a small zone of depletion in the initial region, due to the assumption of zero concentration in the inlet that show single sample. However, in the experimental channel device, where the angle is controlling fluid flow from high shear flow required to deliver fluid by diffusion, fluid were dropped in a single channel configuration with a low flow rate given by $Q = 10$ *nL/s* to protect them from shear for overriding at each other instead of the full interwoving to achive uniform concentration falls rapidly along the length

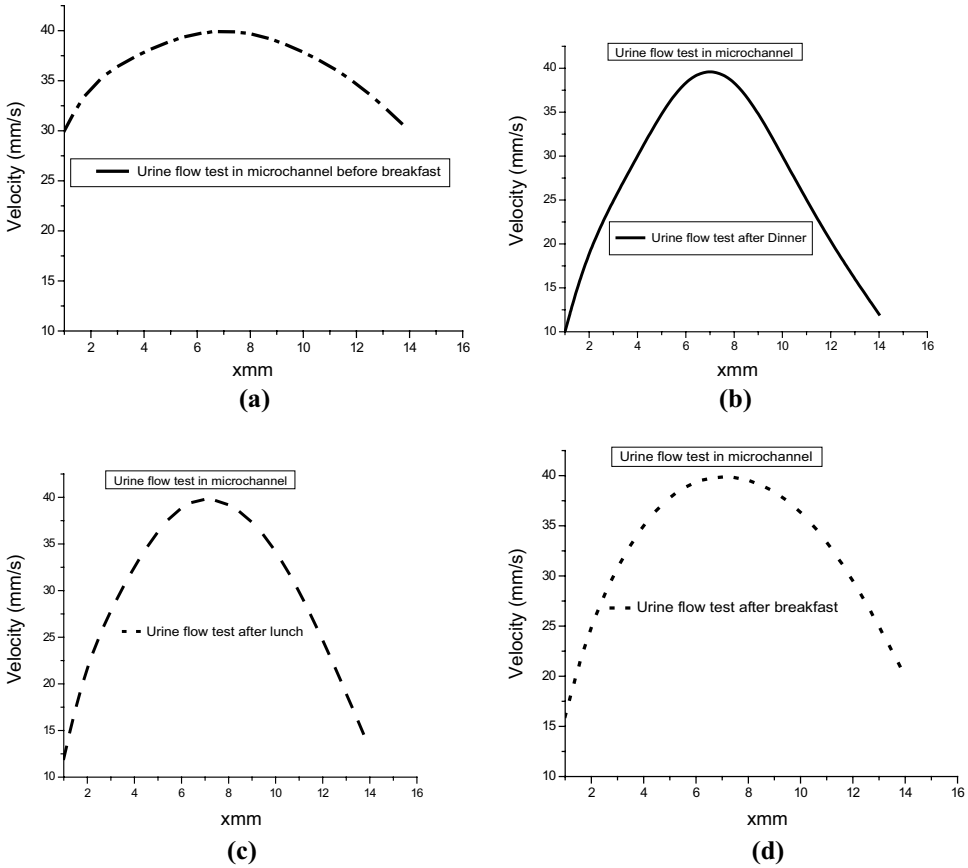


Fig. 5 **a** The results of the sample of the urine before breakfast, **b** sample of the urine after breakfast, **c** sample of the urine after lunch, **d** sample of the urine after dinner

of the channel where two appex meet as shown in Fig. 1. This futher confirmed through the experimental result velocity profile for increasing flow to 30 mm/s → 40 mm/s Fig. 4a comparable to the flow in the upper channel in the two devices in the numerical within the single channel. However, this comparatively diminishes in experimental result as indicated above and this drop is quite interesting because the slow the rate give higher chances for homogenous mixing and further notices as the concentration is increase slightly in Fig. 4b, c, d.

The results presented in Fig. 5 indicate the range of operating parameters that enable a two channel microfluidic device to provide constant flow and uniform mixing Fig. 6, combine levels of fluid to various densities at different point of assay could achieve, promoted by combine effort from shear stresses induced by the volumetric flow rates necessary to deliver the fluid. Moreover, based on capillary effect, it is possible to achieve controlled levels of delivery sample without an unacceptable increase in the pressure increase or drop to which the solute within the channel are exposed. Significantly, we see in Fig. 5 that the concentration and delivery profiles are seriously depend on the sample composition observing the Fig. 5 again; it can be observed that all sample have different concentration and upon mixing, the concentration is harmonies with nearly uniform and slightly dropped.

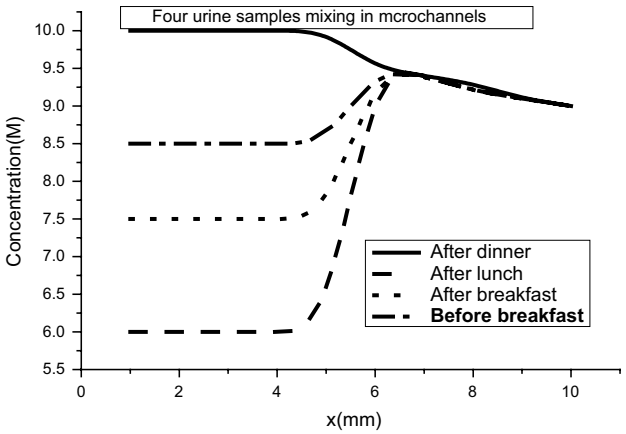


Fig. 6 Mixing profile of the urine sample tested for molecular flow behavior and mixing trend

In the results presented in Fig. 6, an extreme discrepancy in the concentration in fluid flow in channel are noticed, although the sample show wide different but the fluid show uniform mixing before the total length in around 60 % the inlet sample concentration could be tailored to whatever value is desired. This use of these minor and major angles, however, provides an total the length scale

over which effects of the lower inlet sample concentration persist before diffusion from the upper chamber reaches a fully homogenous state. Intuitively, it is anticipated that the entry zone effects for the extreme mixing condition will be diminished for lower flow rates in the both channels. In case, if is decreased to <10 mm l/s with the flow channel flow rate Q_2 reduced to <40 mm l/s.

5 Conclusion

The study has demonstrated the ability using fluidic to distinguish various fluid concentrations and obtaining homogenous mixing profiles and one aspect that has been come to light in this analysis is the fully developed nature of the fluid flow self generated short turn. Normally, it is often useful to ensure that the fluid flow within the operational portions of the device is both laminar and fully developed in micro size domain. With fully developed flow, the velocity field is assumed to be the same along the operating length of the device and in other introduced other functionality such mixing, a disturbance must be introduced. Other device configurations designed to shield cells from stress, such as those featuring grooves or complex channel geometries, may introduce mixing and other effects that could interfere with the dynamic measurement of quantities of interest, such as oxygen concentration, and which would introduce complex flow and concentration fields under the high flow conditions required to maintain a relatively constant concentration field along the length of the device. In order to characterize the entry length for the fluid, we may refer to one of the several correlations that describe the entry length for a fluid. The study has demonstrated a novel and yet simple fabrication PDMS micro mixer with controlled pattern transfer to critical dimensions employing a conventional photolithography approach. A simple microfluidic channel was fabricated. This approach will make fabrication simpler, easier, quicker and inexpensive and can be exploited for a wide range of application. The simplicity of device structure coupled with easy fabrication approach makes it feasible to miniaturize it for the development of portable sensors facilitating its use in both clinical and non-clinical environments. With its simple geometric structure and potential for mass commercial fabrication, the device can be developed to become a portable organic biosensor that can be use for both environmental and diagnostic application.

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