Design of a hybrid advective-diffusive microfluidic system with ellipsometric detection for studying adsorption

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Establishing and maintaining concentration gradients that are stable in space and time is critical for applications that require screening the adsorption behavior of organic or inorganic species onto solid surfaces for wide ranges of fluid compositions. In this work, we present a design of a simple and compact microfluidic device based on steady-state diffusion of the analyte, between two control channels where liquid is pumped through. The device generates a near-linear distribution of concentrations. We demonstrate this via experiments with dye solutions and comparison to finite-element numerical simulations. In a subsequent step, the device is combined with total internal reflection ellipsometry to study the adsorption of (cat)ions on silica surfaces from CsCl solutions at variable pH. Such a combined setup permits a fast determination of an adsorption isotherm. The measured optical thickness is compared to calculations from a triple layer model for the ion distribution, where surface complexation reactions of the silica are taken into account. Our results show a clear enhancement of the ion adsorption with increasing pH, which can be well described with reasonable values for the equilibrium constants of the surface reactions. *Published by AIP Publishing.* [<http://dx.doi.org/10.1063/1.4953105>]

I. INTRODUCTION

Chemical gradients are important factors for guiding the growth and migration of biomolecules,^{1–3} the differentiation of cells⁴ in living tissues, and for controlling adsorption behaviors at solid-liquid or liquid-liquid interfaces in various technical applications.^{5–8} The interest in elucidating these phenomena has led to the development of numerous microfluidic gradient generators. Generally, these devices can be categorized into two types depending on the methods used: convective-based and diffusion-based microfluidics.

Convection-based gradient generation methods have, for example, been used to study chemotaxis of cells.^{9,10} A unique advantage of convection-based methods over diffusion-based ones is that they allow an immediate *in situ* control over concentration profiles via the flow rates.^{11,12} A successful example of this kind is the “Christmas tree” microfluidic device first presented by Whitesides and coworkers,^{13–15} and recently modeled using Lattice-Boltzmann simulations by Hu *et al.*¹⁶ In this scheme, two laminar streams carrying different chemical agents are repeatedly split, mixed, and recombined. However, adverse effects of the involved flow on the process under study (e.g., the secretion of molecules by biological cells¹⁷) might occur. The viscous shear stress introduced by the flow can also interfere with the migration of biomolecules.¹⁸ Therefore, for some studies, diffusion-based microfluidics is preferred.

Great efforts have been spent on developing the diffusion-based gradient microfluidics in the past years. In these methods, membranes^{19,20} or hydrogels^{18,21} are widely utilized, as these materials can minimize the convection, while they still allow the diffusion of small molecules. Additionally, hydrogels are bio-compatible which is important for their use in biological

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studies. To integrate the membrane or hydrogel into the microfluidic device, usually these devices contain two additional layers, to “sandwich” the membrane or hydrogel. This complex fabrication process limits a wide exploitation by biologists and chemists.

Considering the aforementioned drawbacks of the existing convection- and diffusion-based microfluidic devices, we developed a simple diffusion-based concentration gradient generator which is easier to fabricate and operate. This device can generate a large number of (well spread) discrete concentrations. Besides a mold and a polydimethylsiloxane (PDMS) precursor, no additional parts (i.e., hydrogels or membranes) are required once the design is made. The principle of the device is simple: a concentration gradient is generated across flow-free channels which connect the “source” and “sink”-channels.²² A similar design has been used by Saadi and coworkers.²³ Their device, termed as Ladder Chamber, creates a continuous linear concentration profile (or a constant concentration gradient) along a uniform flow-free channel. In comparison, our flow-free channel design consists of large wells linked by small connectors; this ensures well-defined concentrations in discrete areas that are easy to sample. In addition, our method allows for designing arbitrary distributions of concentration amongst the wells by adapting the design of the mass transfer resistances between the wells.

In this work, we demonstrate a microfluidic device which generates a near linear discretized concentration profile. The measured concentrations are compared with a full finite-element numerical simulation as well as a simple circuit model. The good agreement between the experiments and the two theories validates our concept of device design. As an application demonstration, we combined this microfluidic device with total internal reflection ellipsometry to achieve a high throughput measurement of cesium ion adsorption from CsCl solutions at various pH conditions. The relation between the optical signal and the adsorbed amounts allows us to get quantitative results. Finally, the experimental results are compared with theoretical calculations for the ion distribution; here, we assume a triple layer model and combine this with surface complexation reactions.

II. MATERIALS AND METHODS

A. Device design and fabrication

Fig. 1(b) shows a viewgraph of one of our actual devices. Additional details of the design are given in Fig. 1(d). The key feature of the device is an array of wells flanked by two parallel inlet feeding channels. There are 4 rows of wells in the device, and each row has 3 wells which are connected by small channels. When the two feeding channels are supplied with two different concentrations of one analyte of the same flow rate (this keeps pressure balance between two feeding channels, thus eliminating convective flow between the wells), the diffusion takes place across the device. In steady state, a set of well spread discrete concentrations could be established in 12 wells by proper design of dimensions of the small connections between the wells and the feeding channels. This number of wells should typically be appropriate for measuring adsorption isotherms. To ensure homogenous concentrations in wells and minimize convection in the wells, the height of small connections ($12\text{ }\mu\text{m}$) is designed to be much smaller than that of wells and feeding channels ($118\text{ }\mu\text{m}$). The diameter of the wells was chosen to be $300\text{ }\mu\text{m}$ in order to facilitate easy alignment during ellipsometric measurements at variable angle of incidence. With the field of view of our ellipsometer, 4×3 cells of the given diameter could be studied simultaneously.

The fabrication of a microfluidic device is shown in Fig. 2. It was prepared using the standard SU-8 negative resist photo-lithography technique. A two-layer fabrication method was used to create the two parallel flow channels and the 12 circular sample wells (depth: $118\text{ }\mu\text{m}$), and the perpendicular running connecting channels (depth: $12\text{ }\mu\text{m}$). First, the SU-8 (MicroChem, SU-8 2010) is spun onto a silicon wafer to create the $12\text{ }\mu\text{m}$ layer (Fig. 2(A1)). Then, it is exposed to UV light via a high resolution quartz photo-mask with the complete design of the microfluidic network (Fig. 2(A2)). After UV exposure and post-baking (95°C , 3 min), the second layer of SU-8 (MicroChem, SU-8 50) is spun (Fig. 2(A3)). This layer of $106\text{ }\mu\text{m}$ is exposed with a second mask, which contains only the pattern of the flow channels and wells (Fig. 2(A4)). Both patterns are

FIG. 1. Microfluidic platform and design. (a) Image of the microfluidic chip studied with imaging ellipsometry. The channel structure in PDMS is sandwiched between a gold coated glass slide (top side, under the prism) and normal glass slide (bottom side). A layer of silica coated onto the gold is in contact with the solutions; (b) microfluidic chip loaded with dye; (c) steady state concentration distribution, where the top channel is loaded with 0.1 M KMnO_4 , and the bottom channel with DI water; (d) top and side view of the channel design. Each cylindrical well has a diameter of $300\ \mu\text{m}$ and a depth of $118\ \mu\text{m}$. The length of the connecting channels is $\sim 20\ \mu\text{m}$, the width varies from 10 to $100\ \mu\text{m}$, while the depth is fixed at $12\ \mu\text{m}$. (e) Serial arrangement of (coded) elements as used in the circuit model for concentration distribution (see text for details).

superimposed with the large gap alignment option on the mask aligner (EVG620). After the same treatment (UV exposure and post-baking) as done on the first layer, the two-layer SU-8 is developed into a master template,²⁴ as shown in Fig. 2(A5). Then, the mold is treated with 1H,1H,2H,2H-Perfluoro-octyltrichlorosilane (FOTS), to prevent irreversible bonding of PDMS. Liquid PDMS with a 1:10 ratio of curing agent to base polymer is poured onto the mold (Fig. 2(B1)). Next the PDMS is degassed and put in oven at 70°C for 1 h for curing. At last, the PDMS bearing the microstructure is peeled off from the mold (Fig. 2(B2)), and small holes for connecting the channels with tubings are punched with sharp tips.

The silica coated substrate for ellipsometry experiments is prepared with e-beam evaporation. SF10 glass is coated with the following layers: titanium, gold, titanium, and silica. The thickness of the titanium adhesion layers is controlled at $\sim 1\ \text{nm}$; those of gold and silica are 30 nm and 20 nm, respectively. After the preparation, the slide is rinsed with a mixture of ethanol and isopropanol, and dried with nitrogen. Thereafter, all layer thicknesses are (more precisely) measured with ellipsometry.

The PDMS slab and substrate slide are then ready, but their surfaces still need to be made adhesive, to enable assembly of a chip without any leakage or trapping of air bubbles. Exposure to air-plasma for 30 s leads to oxidation of the $-\text{Si}(\text{CH}_3)_2-$ groups of PDMS to Si-OH , making the surface more hydrophilic, and also removes any organic contaminants—which

FIG. 2. Sketch of the fabrication of the microfluidic master mold (A1–A5) and the preparation of the microfluidic device (B1–B3).

might otherwise interfere with the binding. Then, the PDMS slab and the silica coated glass slide are bonded, as shown in Fig. 2(B3).

B. Concentration gradient characterization and validation

To validate the microfluidic device, a solution of 0.1 M KMnO_4 in de-ionized (DI) water was used. Before each experiment, the microfluidic device was filled with DI water by fully submerging it in degassed water inside a vacuum chamber. This prevents the formation of bubbles in the microfluidic device. Then, the device was connected with tubings of equal length and dimension (to keep identical conditions on both inlet sides and hence maintain a balanced pressure). Next, the two liquids were loaded in the separate inlet channels. A syringe (60 ml, BD) together with a syringe pump (PHD 2000, Harvard) was connected at the outlet. The syringe pump was kept running (i.e., sucking fluid) during the entire measurement at a flow rate of 0.1 ml/min. Observations of the time and location dependent image-brightness were made using a Nikon TE2000 inverted microscope. Images were grabbed every 5 min and post-processed with ImageJ (shareware from National Institute of Health), using the Lambert-Beer law to relate extinction to concentration, and hence extract the concentration distribution, after subtracting the background.

The design of the device was optimized with the help of numerical calculations using COMSOL Multiphysics 5.0 software (COMSOL, Inc.). The flow and concentration fields in the device were modeled by solving the Navier-Stokes (NS) equation and convection-diffusion equation, respectively,

$$\rho \frac{\partial \mathbf{u}}{\partial t} = -\nabla p + \mu \nabla^2 \mathbf{u}, \quad (1)$$

$$\frac{\partial c}{\partial t} + \mathbf{u} \cdot \nabla c = D \nabla^2 c, \quad (2)$$

where ρ is the liquid mass density, p is the pressure, μ is the dynamic viscosity, c is the analyte concentration, and D is diffusion coefficient. The vector $\mathbf{u}$ represents the velocity of the fluid.

The flow in the microfluidic device is dominated by viscous effects, i.e., very small Reynolds number, so the convective term has been neglected in the NS equation (1). Eqs. (1) and (2) were solved within COMSOL. It was noted that the momentum transport reaches steady state much faster than the mass transport, since the momentum diffusivity ($\frac{\mu}{\rho}$, $O(10^{-6} \text{ m}^2/\text{s})$) was 3 orders of magnitude higher than the mass diffusivity, $O(10^{-9} \text{ m}^2/\text{s})$. In the simulation, we could safely treat the flow problem to be pseudo-steady, by neglecting the left hand side of Eq. (1). Further, for Eq. (1), a no-slip boundary condition was used on all channel walls, while the outlet was given the flow rate as generated by the syringe pump, and the pressure of the two inlets was set to zero as a reference. For Eq. (2), a no-flux boundary condition was specified on all channel walls, two inlets were prescribed with the respective concentrations, and a convection-only boundary was used at the channel outlet.

According to the simulation results, it typically takes ~ 40 min to establish the steady-state concentration gradient. This agrees with the rough estimation based on the diffusion time scale $O(L^2/D)$, wherein L is the separation between the “source” and “sink” channels. This characteristic time constant can obviously be reduced by reducing the dimensions of the microfluidic device, in particular, the diameter of the wells. This, however, entails additional effort during the optical calibration of the substrate, which involves measurements at variable angle of incidence (see below). At the present flow rate (0.1 ml/min) and syringe volume (60 ml) exploited, the stability of the established concentration should last for around 5 h.

C. Ion adsorption

The ion adsorption experiment was conducted using a commercial imaging ellipsometer (Accurion, Germany). In our previous study,²⁵ we found that at pH 3, no ion adsorption occurs at a silica surface either from CaCl_2 or NaCl solutions. Under these conditions, changes in the ellipsometric angles Ψ and Δ can then be totally ascribed to the change in bulk refractive index, which shows a linear dependence on salt concentration. A measurement was performed with a 0.5 M CsCl solution at pH 3 to verify whether Cs^+ ion adsorption occurs at this condition. The dependences of the ellipsometric angles Ψ and Δ on the incident angle agreed well with the results of a layer-stack model, as shown in Fig. 3. Here, the optical properties of layers I, IV, and the composite layer II were taken from previous characterizations, while an optimization was made for the properties of (the adsorbed) layer III. (Comparison between the experimental and modeling results is provided in the [supplementary material](#), Fig. SI1.) From these calibration measurements, we conclude that the adsorption of CsCl onto the silica surface is indeed

FIG. 3. Layer stack model for ellipsometry data analysis. (i) To determine the salt concentration inside each well, a zero-adsorption condition is created (pH set to 3) and layers I, II, and IV are included in the modeling. Refractive indices in II are loaded from the database in Accurion’s EP 4 software, while the thicknesses in II are characterized in absence of salt. (ii) In the ion adsorption studies, all layers are included in the model. n and d are the refractive index and thickness.

negligible at pH 3. Hence, this pH condition can be used as a reference to determine the salt concentration inside each well.

These calibration measurements were performed as follows: initially, an aqueous solution of 1 M CsCl (pH 3) and water (DI, with pH adjusted to 3 with HCl) were loaded into the separate inlet channels (as displayed in Fig. 1(c)). Then, the incident angle was fixed, and Ψ and Δ angles were recorded for each well simultaneously as a function of time. Once both recorded angles had reached their respective plateaus in each well (concentration gradient established), they were measured as a function of the incident angle (this improves the accuracy of the measurement). Refractive indices of the salt solutions inside the wells were extracted by fitting the ellipsometry data with the stack of layers I, II, and IV as discussed in Fig. 3. Concentrations of CsCl were then calculated from their linear relation with refractive index.

In the ion adsorption studies, experimental procedures were the same as the reference measurement with pH 3 solutions. However, we now explored pH values of 4.5, 6, 7, and 8. In the case of ion adsorption, an additional layer must be taken into account in the stack model. The model then includes the following layers: silica coated chip—adsorbed layer (n_a , d_a)—bulk solution (n_{bulk}), all four parts shown in Fig. 3. The optical parameters for the silica-coated chip, as well as the refractive indices of the bulk solutions, were taken from the previous characterizations. Since the adsorbed layer is thin, only a combined quantity, called the optical thickness, could be obtained. It is defined as $l = (n_a - n_{bulk}) \times d_a$. Fixing the thickness of the adsorbed layer d_a to 0.4 nm, the refractive index of the adsorbed layer was extracted. More details regarding this analysis approach were described in our previous work.²⁵

III. RESULTS AND DISCUSSION

A. Experimental setup, device characterization

To validate the microfluidic design, we first performed an experiment with dye solution; in this case, also the top slide was taken from glass (see Fig. 1(b)). Dye solution and DI water were supplied through the two inlet feeding channels, as demonstrated in Fig. 1(c).

To measure the concentrations in the wells, 0.1 M KMnO_4 and DI water were sucked into the separate inlet channels, and a sequence of images was captured at different instants to track the transient buildup of concentration in the wells. Fig. 1(c) shows the concentration distribution in the microfluidic device after 60 min of pumping. The concentration field in the device at this instant does not change with time anymore, signifying a steady state. To make a quantitative comparison, we plotted the evolutions of the concentrations in wells 1, 5, and 9 for both the experiment and the numerical simulations in Fig. 4(a). It is seen that the numerical model presented in Section II B predicts the transient development of the concentrations in the device rather well. Here, the diffusion coefficient of the MnO_4^- ion was taken to be $1.632 \times 10^{-9} \text{ m}^2/\text{s}$ (Ref. 26)). Generally, it takes about 50 min to reach the steady state. Fig. 4(b) demonstrates the concentrations in the 12 wells at the steady state (after flowing solutions for more than 60 min).

FIG. 4. (a) Evolution of the concentration in three wells (numbered in Fig. 1(c)) during the first 50 min after loading: well 1 (black squares), well 5 (red circles), and well 9 (blue triangles); solid symbols are experiments, while open symbols correspond to COMSOL simulations. (b) Comparison of experiments and simulations in the steady state.

These data are the average values obtained from five different devices and are mostly consistent with the numerical calculations. The simulation results (Simulation 0) correspond to the dimensions of the device as designed. The notable discrepancy with the experiments for wells 11 and 12 is likely attributable to deviations (i.e., fabrication errors) in the dimensions of the small connections. To obtain an estimate of the extent of these possible deviations, we carried out additional simulations (simulation 1) with slightly modified dimensions of small connections to wells 11 and 12. The observed deviations can be compensated by reducing the width by $5\ \mu\text{m}$ as compared to original design, which is somewhat large but overall conceivable, given the experimental accuracy of device manufacturing.

Albeit less accurate than that obtained with the detailed numerical model, the steady state concentrations in the 12 wells can also be predicted fairly well with a simple circuit model, in which each constriction is represented by an Ohmic resistor. Fig. 1(e) sketches the circuit model for one row of wells, where C_i is the concentration inside the indicated well (corresponding to the electrical potential), and L_i , w_i , and h_i are the length, width, and height of the individual small connecting channels. Using the mass conservation law, we have the following relations:

$$D \frac{C_{i+1} - C_i}{L_i} (w_i h_i) = D \frac{C_{i+2} - C_{i+1}}{L_{i+1}} (w_{i+1} h_{i+1}), \quad i = 1, 2, 3, 4, \quad (3)$$

where D is the diffusion coefficient. In analogy with an electric circuit, Eq. (3) can be transformed to

$$\frac{C_{i+1} - C_i}{R_i} = \frac{C_{i+2} - C_{i+1}}{R_{i+1}}, \quad (4)$$

where R_i is the resistance for mass transfer, which can be related to the dimensions of the connection i by $R_i = L_i / (w_i h_i)$. Eq. (4) defines three equations from which we can solve the concentrations in the three wells (C_2 , C_3 , and C_4) as

$$C_j = C_1 + \frac{C_5 - C_1}{\sum_{i=1}^4 R_i} \sum_{i=1}^{j-1} R_i, \quad j = 2, 3, \text{ and } 4. \quad (5)$$

Since the concentrations of the solutions in the two feed channels (C_1 and C_5) as well as the resistances of all connections (R_i), are known, we can readily calculate the concentrations in each well in our device. It should be mentioned that the validity of the circuit model requires that the concentration field has reached the steady state, and also that the concentrations in the wells are uniform, or equivalently the mass transfer resistance of the wells is vanishingly small. These assumptions are justified by Figure 5 which shows the transient evolution of the concentration along the centerline of the leftmost row of wells from the full numerical simulation. The concentration profiles within each of the three wells (wells 1, 5, and 9) show good homogeneity in the steady state, and the major mass-transfer resistance is contributed by the small connections in which the concentrations change drastically.

The results from the circuit model are also included in Fig. 4(b). One notes that the circuit model agrees well with both the experiments and the full numerical simulation. In comparison to the COMSOL model, the circuit model has a distinctive advantage: for a given set of predefined concentrations, the circuit model allows us to determine the required dimensions of small connections very efficiently and conveniently during the course of device design. The circuit model slightly disagrees with the full numerical model, probably because of the finite resistance of the wells, which is omitted in the circuit model.

With the previous analytical expression Eq. (5), we can calculate the concentration profile. However, the microfluidic device allows the establishment of a 3D concentration gradient

FIG. 5. Numerical prediction of the time evolution of concentration along the centerline of the leftmost row of wells (1, 5, and 9 as numbered in Fig. 1(c)) after loading DI water and KMnO_4 solution into two feeding channels. The 6 instant concentration profiles are at 5, 10, 20, 30, 40, and 50 min, as is chosen in Fig. 4(a).

which also involves the flow and diffusion; thus, simulation with COMSOL provides both transient and steady state results more accurately.

We emphasize that the stability of the concentration gradient depends on the flow. As long as the flow is properly balanced between the two inlets, and maintained, the gradient remains stable. The flow rates and the volume of the syringe determine the lifetime of the stable gradient. With the flow rate 0.1 ml/min used in our work, the gradient can be maintained for more than 5 h if the syringe is selected large enough (i.e., 60 ml).

In practice, it turns out that it is difficult to establish the balance between the two flow channels for the design of the microfluidic channels presented here. Minor differences in the connection between the device and the two inlet reservoirs can give rise to appreciable pressure differences that generate an undesired convective cross-flow. They need to be compensated, e.g., by a small additional hydrostatic pressure head at the inlets that is separately adjusted for each inlet. These challenges can be mitigated by incorporating additional flow resistors (e.g., meandering channels) upstream of the diffusion device. With this modification in place, maintaining zero advective cross-flow is straightforward (data not shown).

B. Ion adsorption studies

In the adsorption studies, we combined the microfluidic device with ellipsometry working in the total internal reflection mode. This mode offers several advantages compared to standard ellipsometry, such as higher sensitivity^{27,28} and applicability to opaque media.²⁸ Here, we exploited the combined setup to study ion adsorption at silica-water interfaces.

Clearly, the “bulk” concentration of salt inside each well (as shown in Fig. 1(c)) now needs to be known precisely. Considering that the point of zero charge of silica is close to 2,²⁹ we tuned the salt solution at pH 3. Adsorption of common ions (e.g., Na^+ , Ca^{2+} , Cl^-) can then be safely neglected, which allows to ascribe the recorded ellipsometric signals fully to the changes in the “bulk” refractive index of the solution. Measurements (with all concentrations in the steady state) of Ψ and Δ as a function of angle of incidence were fitted with the optical model: glass $\rightarrow$ gold, silica, and titanium layer $\rightarrow$ bulk solution, as I, II, and IV shown in Fig. 3. Only the refractive index of the bulk solution was taken as an unknown parameter, and from its fitted local values, the salt concentration was extracted per well. Generally, these concentrations are consistent with the simulation results and the circuit model, as shown in Fig. 6(b). A discrepancy with experiments is observed at well number 12 for the first simulation (Simulation 0) with the dimensions of devices taken from the original design. Again, this could be the result of slightly deviating channel dimensions in the fabrication process. The second simulation (Simulation 1) with slightly modified dimensions (reduce

FIG. 6. (a) Linear relation between refractive index and concentration of CsCl solution (at pH 3), symbols are the experimental data measured with refractometer, and the line is the linear fit; (b) concentration determination with CsCl solution at pH 3. Experiments and simulations correspond to the steady state.

the width of the small connection between well 12 and the feeding channel by $5\ \mu\text{m}$) predicts the measurements very accurately in all wells. The simple circuit model apparently also works quite well. We emphasize that the discrepancy between experiment and predictions does not create any uncertainty in the concentrations: in the adsorption study, the salt concentration inside each well is extracted from the layer stack model without comparison with any other measurements.

Silica surface groups get significantly deprotonated when the pH is increased to (well) above 2. The ensuing negative surface charge needs to be compensated by the charge distribution in the electric double layer (see Fig. 7). The optical layer stack as used for modeling the data at pH 3 now has to be extended to include the electric double layer. Clearly, in reality, the refractive index of this adsorbed layer cannot be constant, since the concentrations of both cations and anions change gradually from the silica surface to the bulk solution. In the optical layer stack model, the ion adsorption is therefore taken into account via a single layer with an *effective thickness* and refractive index. For thin layers, only the product of the two can be extracted from the measurement.⁷ We therefore assume a thickness d_a of 0.4 nm, corresponding to the radius of the hydrated Cs^+ ions, and fit for the refractive index change. The refractive index of the bulk solution is set equal to the one measured at pH 3. We then obtain the optical thickness l

$$l = \Delta n \times d_a = (n_a - n_{\text{bulk}}) \times d_a, \quad (6)$$

where n_a and n_{bulk} are the refractive index of the adsorbed layer and bulk solution, respectively. Following de Feijter,³⁰ the optical thickness can also be written as

FIG. 7. Sketch of the triple layer model of an Electric Double Layer (EDL) on silica surface (a), ion distribution in the EDL (b), and optical layer stack model (c) used for the ellipsometry data analysis. Adapted with permission from Wang *et al.*, Sens. Actuators, B 210, 651 (2015). Copyright 2015 Elsevier B. V.

FIG. 8. Optical thickness of CsCl adsorption at silica water interfaces at different pH conditions: pH 4.5 (square), pH 6 (circle), pH 7 (up triangle), and pH 8 (down triangle); these experimental results are extracted from the ellipsometry data. Solid lines are the theoretical calculation using surface complexation triple layer model.

$$\Delta n \times d_a = \alpha_+ \Gamma_+ + \alpha_- \Gamma_-, \quad (7)$$

where $\alpha_{+/-}$ and $\Gamma_{+/-}$ denote the refractive index increment and excess of the cations and anions, respectively. For CsCl, $(\frac{dn}{dc})_{CsCl} = 12.66 \text{ ml/mol}$ and $\alpha_- = 3 \text{ ml/mol}$,^{25,31} so $\alpha_+ = 9.66 \text{ ml/mol}$ was used in this work. The right hand side of Eq. (7) shows that we measure an optically weighted average, where the individual adsorbed amounts per ion can be positive or negative. In the present case (a negative substrate and absence of specific adsorption by the anions), not only α but also Γ should be larger for the cation; this means that the adsorption signal will be dominated by the Cs^+ ions.

As shown in Fig. 8, the optical thickness increases with pH, and the maximum adsorption at pH 8 is almost twice the one measured at pH 6. One can also find that the adsorption increases with salt concentration at a fixed pH. From pH 6 to pH 8, the optical thickness increases steeply at a low salt concentration and approaches a plateau at a high salt concentration. At pH 4.5, the optical thickness increases gradually with increasing salt concentration. Such effects are due to the progressive dissociation of the silanol groups, which require an increasing amount of cations to neutralize the charge.

To corroborate these ellipsometry results, we performed theoretical calculations with the standard triple layer model of the electric double layer^{32,33} (see Fig. 7(a)). In this model, the deprotonation of silanol group takes place at 0 plane, β plane is where cations bind with deprotonated surface sites, and d plane defines the closest approach of fully hydrated ions to the surface. Clearly, β and d planes divide the entire double layer into three sublayers, two electroneutral inner sublayers which can be described by capacitance C_1 and C_2 , and one charged diffuse layer (beyond d plane) which is described by the Poisson-Boltzmann (PB) equation. The surface charge (boundary condition for PB equation) is calculated from surface complexation reactions at 0 and β planes, which are expressed by equilibrium constants (see Table I). This allows to obtain the surface excess and from it, to calculate the optical thickness (using Eq. (7)). We

TABLE I. Surface complexation reactions and parameters used in triple layer model.

	Surface complexation reactions	Equilibrium constant, K_i (mol/l)	Capacitance (F/m^2) ^{32,35}	
			C_1	C_2
H	$-\text{SiOH} \leftrightarrow \text{H}^+ + -\text{SiO}^-$	$10^{-6.9}$ (Refs. 33 and 34)		
CsCl	$-\text{SiO}^- \text{Cs}^+ \leftrightarrow -\text{SiO}^- + \text{Cs}^+$	$10^{-1.69a}$	1.0	0.2
Site density (nm^{-2}) ³⁴	8.0			

^aOptimized value by fitting all the data at various pH conditions.

emphasize that the total surface excess includes the positive excess of counter-ions in Stern layer and diffuse layer, as well as the depletion of co-ions in the diffuse layer.²⁵

Generally, the theoretical calculations and experimental results show agreement; the dependencies on pH and concentration are well reproduced. This corroborates the validity of the measurements. However, one can still find some discrepancies between the numerical calculation and experiments. In the modeling, there are many parameters such as capacitance C_1 and C_2 , the equilibrium constant K_H , etc. (see Table I). These parameters are not unique.³⁶ A better agreement between the optical thickness and the theoretical calculations might be obtained with a more elaborate parameter estimation.

Besides the electrostatic interaction, the ion specificity also has effects on cation adsorption. Sivan and coworkers have studied cation adsorption on solid surfaces with AFM, where they pointed out the significance of the surface hydration for cation adsorption.³⁷ To further explore the adsorption mechanism of these ions, the combination of several complementary detection techniques could provide additional insights.

IV. CONCLUSIONS

We developed a simple microfluidic device which can generate a dilution series of an injected solution by making use of steady state diffusion. The concentration profiles can be generated as desired via appropriate design of the mass transfer resistances. Combined with the total internal reflection ellipsometry, the device can be used for studying adsorption phenomena at the solid-liquid interfaces. As a demonstration, we studied cesium ion adsorption from their chloride aqueous solutions onto silica surfaces at various pH conditions. Comparison of the ellipsometry measurements with the calculations from the surface complexation triple layer model corroborates the utility of the combined setup for adsorption studies. The application of this device can be easily extended to other chemicals or biomolecules. Therefore, we believe this device will be a very straightforward and useful tool both for chemical and biological scientists to do studies with concentration gradients.

SUPPLEMENTARY MATERIAL

See [supplementary material](#) for the total internal reflection ellipsometry measurements: 0.5 M CsCl at pH 3, comparison between the experimental and modeling results.

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¹D. T. Chiu, *Curr. Opin. Chem. Biol.* **5**, 609–612 (2001).

²E. Cimetia, D. Sirabella, K. Yeager, K. Davidson, J. Simon, R. T. Moon, and G. Vunjak-Novakovic, *Lab Chip* **13**, 355–364 (2013).

³D. A. Lauffenburger and A. F. Horwitz, *Cell* **84**, 359–369 (1996).

⁴S. García, R. Sunyer, A. Olivares, J. Noailly, J. Atencia, and X. Trepas, *Lab Chip* **15**, 2606 (2015).

⁵J. Lyklema, *Solid-Liquid Interfaces*, Fundamentals of Interface and Colloid Science Vol. 2 (Elsevier, 1995), pp. 4.1–4.135, ISBN: 978-0-12-460524-4.

⁶B. C. Donose, I. U. Vakarelski, and K. Higashitani, *Langmuir* **21**, 1834–1839 (2005).

⁷H. Tompkins and E. A. Irene, *Handbook of Ellipsometry* (William Andrew, 2005).

⁸L. Wang, S. Liu, T. Wang, and D. Sun, *Colloids Surf., A* **381**, 41–47 (2011).

⁹N. L. Jeon, H. Baskaran, S. K. W. Dertinger, G. M. Whitesides, L. Van De Water, and M. Toner, *Nat. Biotechnol.* **20**, 826–830 (2002).

¹⁰D. Irimia, S.-Y. Liu, W. G. Tharp, A. Samadani, M. Toner, and M. C. Poznansky, *Lab Chip* **6**, 191–198 (2006).

¹¹D. Amarie, J. A. Glazier, and S. C. Jacobson, *Anal. Chem.* **79**, 9471–9477 (2007).

¹²Y.-H. Jang, M. J. Hancock, S. B. Kim, S. Selimovic, W. Y. Sim, H. Bae, and A. Khademhosseini, *Lab Chip* **11**, 3277–3286 (2011).

¹³N. L. Jeon, S. K. W. Dertinger, D. T. Chiu, I. S. Choi, A. D. Stroock, and G. M. Whitesides, *Langmuir* **16**, 8311–8316 (2000).

¹⁴S. K. W. Dertinger, D. T. Chiu, N. L. Jeon, and G. M. Whitesides, *Anal. Chem.* **73**, 1240–1246 (2001).

- ¹⁵S. K. Dertinger, X. Jiang, Z. Li, V. N. Murthy, and G. M. Whitesides, *Proc. Natl. Acad. Sci.* **99**, 12542–12547 (2002).
- ¹⁶Y. Hu, X. Zhang, and W. Wang, *Ind. Eng. Chem. Res.* **50**, 13932–13939 (2011).
- ¹⁷J. El-Ali, P. K. Sorger, and K. F. Jensen, *Nature* **442**, 403–411 (2006).
- ¹⁸M. Kim and T. Kim, *Anal. Chem.* **82**, 9401–9409 (2010).
- ¹⁹T. Kim, M. Pinelis, and M. Maharbiz, *Biomed. Microdevices* **11**, 65–73 (2009).
- ²⁰V. V. Abhyankar, M. A. Lokuta, A. Huttenlocher, and D. J. Beebe, *Lab Chip* **6**, 389–393 (2006).
- ²¹H. Wu, B. Huang, and R. N. Zare, *J. Am. Chem. Soc.* **128**, 4194–4195 (2006).
- ²²N. Strelnikova, F. Herren, C.-A. Schoenenberger, and T. Pfohl, *Front. Mater.* **3**, 20 (2016).
- ²³W. Saadi, S. Rhee, F. Lin, B. Vahidi, B. Chung, and N. Jeon, *Biomed. Microdevices* **9**, 627–635 (2007).
- ²⁴J. R. Anderson, D. T. Chiu, H. Wu, O. Schueller, and G. M. Whitesides, *Electrophoresis* **21**, 27–40 (2000).
- ²⁵L. Wang, C. Zhao, M. H. G. Duits, F. Mugele, and I. Siretanu, *Sens. Actuators, B* **210**, 649–655 (2015).
- ²⁶O. Levenspiel, *Tracer Technology: Modeling the Flow of Fluids* (Springer Science & Business Media, 2011).
- ²⁷H. Arwin, M. Poksinski, and K. Johansen, *Phys. Status Solidi A* **205**, 817–820 (2008).
- ²⁸H. Arwin, M. Poksinski, and K. Johansen, *Appl. Opt.* **43**, 3028–3036 (2004).
- ²⁹M. Kosmulski, *J. Colloid Interface Sci.* **253**, 77–87 (2002).
- ³⁰J. A. De Feijter, J. Benjamins, and F. A. Veer, *Biopolymers* **17**, 1759–1772 (1978).
- ³¹M. Porus, C. Labbez, P. Maroni, and M. Borkovec, *J. Chem. Phys.* **135**, 064701 (2011).
- ³²J. A. Davis and J. O. Leckie, *J. Colloid Interface Sci.* **67**, 90–107 (1978).
- ³³J. A. Davis, M. H. Weisman, and D. H. Dail, *Arch. Intern. Med.* **138**, 480–483 (1978).
- ³⁴T. Hiemstra, W. Van Riemsdijk, and G. Bolt, *J. Colloid Interface Sci.* **133**, 91–104 (1989).
- ³⁵D. A. Sverjensky, *Geochim. Cosmochim. Acta* **65**, 3643–3655 (2001).
- ³⁶J. Westall and H. Hohl, *Adv. Colloid Interface Sci.* **12**, 265–294 (1980).
- ³⁷J. Morag, M. Dishon, and U. Sivan, *Langmuir* **29**, 6317–6322 (2013).