Contactless microfluidic pumping using microchannel-integrated carbon black composite membranes

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The ability to pump and manipulate fluid at the micron-scale is a basic requirement for microfluidic platforms. Many current manipulation methods, however, require expensive and bulky external supporting equipment, which are not typically compatible for portable applications. We have developed a contactless metal electro-osmotic micropump capable of pumping conductive buffers. The pump operates using two pairs of gallium metal electrodes, which are activated using an external voltage source and separated from a main flow channel by a thin micron-scale polydimethylsiloxane (PDMS) membrane. The thin contactless membrane allows for field penetration and electro-osmotic flow within the microchannel, but eliminates electrode damage and sample contamination commonly associated with traditional DC electro-osmotic pumps that utilize electrodes in direct contact with the working fluid. Our previous work has demonstrated the effectiveness of this method in pumping deionized water. However, due to the high resistivity of PDMS, this method proved difficult to apply towards manipulating conductive buffers. To overcome this limitation, we fabricated conductive carbon black (CB) powder directly into the contactless PDMS membranes. The increased electrical conductivity of the contactless PDMS membrane significantly increased micropump performance. Using a microfluidic T-channel device and an electro-osmotic flow model, we determined the influence that CB has on pump pressure for CB weight percents varying between 0 and 20. The results demonstrate that the CB increases pump pressure by two orders of magnitude and enables effective operations with conductive buffers. © 2015 AIP Publishing LLC. [<http://dx.doi.org/10.1063/1.4933349>]

I. INTRODUCTION

Microfluidic pumps, which route and manipulate fluid flow in microchannels, are essential components for lab-on-a-chip systems.¹ On-chip, they are typically integrated with other critical components, such as microvalves, mixers, separators, sensors, and detectors, to create complete micrototal analysis systems (μ TAS) for processing small volumes of fluid in applications including drug screening, medical diagnostics, chemical analysis, and environmental monitoring. To carry out complex on-chip experiments like these, microfluidic platforms are used to control on-chip fluidic pumping, valving, mixing, and routing for specific desired on-chip function.^{2,3} For example, microfluidic large scale integration (LSI) uses microchannel networks with integrated pressure-actuated elastomeric pumps and valves to control, route, and mix fluid in channels for on-chip genetic analysis of single cells.⁴ Others have used centrifugal forces,⁵ surface acoustic waves (SAW),⁶ and electrical forces to perform fluidic metering mixing and aliquoting for applications including DNA extraction, plasma separation,⁷ cell lysis,⁸ protein crystallization,⁹ immunoassays,¹⁰ and single cell analysis.¹¹

The majority of these microfluidic platforms, however, require large and often expensive external actuation equipment for operation, which can confine their usefulness to the laboratory

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bench-top. The development of more portable platforms is therefore an important area of microfluidic research. LSI, for example, utilizes thousands of elastomeric pump and valve components for precise and local fluid manipulation, but requires bulky external computer controlled vacuum or pressure sources for component actuation. Centrifugal-based microfluidics⁵ can drive fluidic routing, mixing, and metering on rotating substrates with on-chip valves and channels, but the motor required for actuation is expensive, and this platform lacks flexibility; the spinning chip cannot be interfaced with external electrical readouts or sensors. Platforms based on SAW¹² and electrical forces¹³ offer an effective means to transport liquid on-chip, but require expensive and often large electrical equipment. In terms of developing microfluidic systems for point-of-care diagnostics, personalized medicine, and environmental monitoring, developing more portable fluidic manipulation technology is an important step for reaching this goal.

Improved methods for microfluidic pumping and valving are helping to increase the portability of these devices. In general, microfluidic pumps can be categorized based on the mechanism for driving fluid flow. Displacement pumps utilize moving parts such as membranes or valves to drive fluid motion.¹⁴ At the microscale, however, they require complex fabrication processes and can suffer from mechanical failure. Kinetic pumps, on the other hand, do not require moving parts, and operate by converting kinetic sources into fluid momentum to drive flow. Common kinetic actuation methods include electrokinetic (EK),¹⁵ magnetohydrodynamic,¹⁶ electrochemical,¹⁷ electrothermal,^{18,19} and electrowetting.²⁰

EK flows are a popular class of low-volume, low-power micropumps, and can serve as a useful alternative to mechanical micropumps.¹³ EK pumps, however, typically require direct contact between electric field-generating electrodes and a working fluid. When a direct current (DC) voltage is applied, Faradaic reactions occur at each electrode surface, which can lead to uncontrolled electrothermal flow, electrode damage, and sample damage from pH alteration and Joule heating. When a DC voltage is applied, Faradaic reactions occur at each electrode surface, which can lead to uncontrolled electrothermal fluid flow, electrode damage, and sample damage from pH generation and Joule heating. As opposed to DC driven EO flow, alternating current (AC) electro-osmotic pumps can alleviate some of these drawbacks. ACEO pumps do not suffer from Faradaic reactions, and are capable of pumping low conductivity aqueous electrolytes in microchannels.^{21,22} However, they require large microelectrode arrays, which can be expensive and time consuming to fabricate.

Recently, new types of contactless EK (cEK) devices have been developed by Sano *et al.*²³ In cEK, two conductive electrolyte-filled microchannels serve as liquid electrodes, which are physically isolated from a main flow channel by a thin polydimethylsiloxane (PDMS) membrane. An AC or DC voltage source can penetrate through the membrane and into the main channel without causing Faradaic reactions or buffer contamination issues associated with traditional non-contactless electrode systems. However, cEK liquid based designs lack necessary features for portable operations because the liquid electrodes can suffer from evaporation and cannot be reliably transported or stored long term. Moreover, it is difficult to integrate many electrode devices on a single chip for complex processing; electrodes require fluid injection ports with pipette tips and have large on-chip footprints. In previous work, we have presented a novel solid metal-based approach for contactless on-chip liquid pumping, routing, and metering. Instead of using liquid electrodes, our micropump design is based on 3D gallium solid metal electrodes fabricated directly into the sidewalls of microfluidic flow channels, and separated from the active flow region by a thin micron-scale PDMS polymer membrane.²⁴ In terms of portability, this metal-based design was a significant improvement over previous cEK devices that utilized liquid electrodes. We demonstrated that multiple micropumps could be fabricated and independently operated on a single microfluidic chip with a small external DC power supply and a computer controller for fluid pumping, mixing, and metering.

One limitation of this contactless design, however, is the high electrical resistance of the PDMS membranes; pumping requires a large kilovolt potential and is limited to low conductivity fluids. In order to drive an electro-osmotic (EO) flow, a large kilovolt potential is dropped

across the main microchannel. However, due to the insulating PDMS membrane, only 0.2% of the applied potential was actually delivered into the main channel. This low efficiency limited pumping applications to electrolytes with conductivities less than 0.1 mS/cm. Electrolytes with greater conductivity could not be pumped because the electric field was not large enough to overcome counterion screening at the PDMS membrane wall.

In this work, we present a novel method for integrating conductive PDMS composites directly in specific regions of microchannel sidewalls for significantly improved pumping efficiency. PDMS is an inert polymer that inherently lacks significant conductive properties. However, conductivity can be increased by combining the elastomer with a conducting powder, such as carbon black (CB) or silver nanoparticles, to create conductive polymer composites.^{25,26} When combined with standard soft lithographic techniques, composites can be patterned into specific structures that are suitable for on-chip applications.²⁷ For example, Li²⁸ created an array of conductive PDMS composite posts using PDMS/nickel powder composites to create a sensitive on-chip pressure sensor, and Niu *et al.*²⁹ utilized soft lithography and PDMS/silver nanoparticles to fabricate arrays of conductive composite-based on-chip circuits. One benefit of this approach is that PDMS composites inherit increased electrical conductivity, but still retain the elastomeric characteristics of PDMS, which allow them to be integrated into microfluidic systems using soft lithography.

Using this approach, we demonstrate a new type of contactless DC micro-pump with integrated conductive nanocomposite membranes for significantly improved operation with higher conductive electrolyte buffers. Our micro-pump design integrates carbon black doped conductive PDMS fabricated into the sidewalls of microfluidic flow channels with gallium metal electrodes (Fig. 1(a)). Due to the electrical properties of CB, the conductivity of a PDMS membrane can be controlled by mixing PDMS polymer with different weight percentages of CB. This technique allows us to customize our micro-pump membrane conductivity depending on the working buffer sample's conductivity.

In the first part of this paper, we present a theoretical model which combines the governing electrokinetic equations that describe electro-osmotic flow (EOF) with an electrostatic model for the applied voltage within the main microchannel. We then apply this model to predict the electro-osmotic pressure as a function of applied voltage in a microfluidic “T-shape” device, a

FIG. 1. Schematic of a gallium metal contactless micropump with integrated PDMS carbon black membranes. (a) A top view of a single pumping device. Each micropump contains four gallium electrodes separated from the main channel by a 50 μm -thick PDMS-carbon black nanocomposite membrane. (b) A micrograph of a pair of electrodes surrounding a microchannel. The nanocomposite is directly integrated into the microchannel sidewalls. Scale bar, 200 μm .

design we previously developed and presented to characterize the maximum output pressure of our micropumps. Relationship between the pumping pressure and CB weight % are then investigated in the second part of this work. Finally, using this pump data, we demonstrate the impact that conductive PDMS composite has on improving pump performance.

II. THEORY

A. Electro-osmotic micropump pressure and flowrate

In this section, we derive a theoretical model for relating electro-osmotic flow within a microchannel to the applied voltage drop and the pressure gradient produced by the electro-osmotic pump.²⁴

Electro-osmosis is an EK technique for driving flows in microfluidic channels. Solid surfaces tend to develop surface charge when in contact with aqueous liquids. A diffuse electric double layer (EDL) forms to screen the surface charge, where counter-ions of opposite charge are attracted to the surface and co-ions of similar charge are repelled. The resulting EDL possesses a non-zero charge density and its thickness depends on the ionic strength of the electrolyte; typical EDL thicknesses range from nanometers for mM electrolytes to microns for low conductive μM electrolytes. Because the EDL possesses a non-zero charge density, an external electric field applied tangential to the charged surface forces the EDL into EOF. Assuming that the electric field is uniform, the flow is unidirectional, and there is no applied pressure gradient, the Stokes equations can be written as

$$0 = \mu \frac{\partial^2 u}{\partial y^2} - \varepsilon \frac{\partial^2 \phi}{\partial y^2} E_{\text{wall}}, \quad (1)$$

where μ , ε , and E_{wall} are the fluid viscosity, electrolyte permittivity, and applied tangential electric field at the channel wall, respectively. Eq. (1) can be integrated with a no slip condition at the channel wall ($y=0$) and a far-field condition that $\phi \rightarrow 0$ as $y \rightarrow \infty$ to give

$$u_{\text{slip}} = \frac{\varepsilon E_{\text{wall}}}{\mu} (\phi - \phi_{\text{wall}}). \quad (2)$$

Here, the wall potential, $\phi_{\text{wall}} = \zeta e^{-y/\lambda_D}$, exponentially decays over the Debye length, λ_D , and approaches a Smoluchowski slip velocity

$$u_{EO} = \frac{\varepsilon E_{\text{wall}} \zeta}{\mu}, \quad (3)$$

where ζ is the zeta potential at the surface of the microchannel. For the case where a micropump must drive a flow against a hydrodynamic load, a backpressure is generated and the net fluid flow is reduced. For a slot-shaped microchannel where the width is much larger than the height, the net flow rate is a linear combination of EO and pressure driven back flow

$$Q = \frac{2wd^3}{3\mu} \left[\frac{\partial P}{\partial x} \right] - 2wd \frac{\varepsilon E_{\text{wall}} \zeta}{\mu}. \quad (4)$$

For an applied voltage, ΔV , dropped over an electrode separation length L_E , the electric field at the wall is $\Delta V/L_E$. The back pressure gradient varies linearly and can be written as $\Delta P/L_E$. The maximum pressure produced by the EO pump (P_{max}) occurs when $Q=0$, which is the case for a closed microchannel, and is

$$\Delta P_{\text{max}} = - \frac{3\mu u_{EO}}{d^2} \Delta V. \quad (5)$$

B. Voltage drop across a carbon black PDMS composite membrane

Contactless micropumps follow the same electro-osmotic principles as traditional EO pumps with external metal electrodes; however, the high electrically resistive nature of the insulating PDMS membranes reduces the effective voltage (ΔV_{eff}) within the microchannel from that of the total potential applied (ΔV_{total}). We apply a simple Ohmic current model to determine this relationship. The effective voltage drop across the working fluid is thus written as

$$\Delta V_{eff} = \Delta V_{total} \frac{R_{fluid}}{R_{total}}, \quad (6)$$

where $R_{total} = R_{CB} + R_{fluid}$.

When a DC voltage is dropped across the contactless PDMS membrane, counterions in the working electrolyte will get accumulated at the PDMS-liquid interface, and create an induced screening potential (ϕ_s), which will reduce the magnitude of the electrostatic field in the main channel. The magnitude of this screening effect is a function to the applied electric field and electrolyte counterion concentration. If the ion concentration is large enough, the applied field will be completely screened from the microchannel, and no electro-osmotic flow will be observed. If the applied voltage is large enough and able to exceed the screening potential, however, an electric field will penetrate into the bulk and impart an electro-osmotic slip velocity at the PDMS membrane surface.²⁴

A more general relationship between effective and total potential applied must take into account this screening potential,

$$\Delta V_{eff} = \Delta V_{total} \frac{R_{fluid}}{R_{total}} - \phi_s. \quad (7)$$

Under conditions where the net pump flowrate is zero, Eq. (7) can be combined with Eq. (5) to quantify the maximum pump pressure. In this work, we experimentally determine the screening potential from our experimental pump curves.

III. MATERIALS AND METHODS

Here, we describe the fabrication and operation of a contactless CB micropump and present a novel process for patterning conductive PDMS composite membranes within microchannel sidewalls, and integrating these composites with 3D gallium metal electrodes. We then demonstrate how to use a microfluidic “T-shape” device for measuring the pressure generated by the pump.

A. Microfluidic chip

Each microfluidic pump, as shown in Fig. 1(a), consists of a main flow channel and four separate electrode-containing channels. The electrode channels are filled with gallium metal, each of which are fabricated directly against a patterned region of a micron-thick PDMS CB composite membrane [Fig. 1(b)]. To fabricate the composite PDMS membranes at specific regions within a microfluidic channel sidewall, we employed a novel multistage soft lithographic process, shown in the process flow diagram in Figs. 2(a)–2(d). First, a negative photoresist, SU-8 3050 (Microchip Corp.), was used to fabricate a soft lithographic mold with a main flow channel (300 μm wide and 50 μm high) and four gallium electrode channel (900 μm wide) [Fig. 2(a)] separated from the main flow channel by 50 μm -thick gap. To create conductive PDMS composite membranes within these gaps, carbon nanopowder (Sigma-Aldrich, 633100) was combined with a 1:10 weight ratio of PDMS elastomer and curing agent and mechanically mixed in a centrifugal mixer (Thinky, ARM 310) at 2000 rpm for 30 s. The carbon powder was added at different wt. %, ranging between 5% and 20%, in order to study the influence of PDMS composite resistivity on device performance. The nanocomposite PDMS gel was injected into each gap between the electrode channel and the main flow channel using a 1 ml

FIG. 2. Process flow diagram of fabrication process for integrating patterned CB-PDMS membranes into the microchannel sidewalls. (a) Electrode channels and flow channel are defined with a thick film SU-8 photoresist mold. (b) A mixture of CB-PDMS is carefully injected into the gaps between the electrodes and main flow channel. (c) PDMS elastomer is poured atop the SU-8 mold and allowed to cure. (d) The PDMS device is sealed against a glass substrate, and hot liquid gallium metal is injected into the side channels and allowed to cool. (e) Completed device. Scale bar = 200 μm .

plastic syringe [Fig. 2(b)]. Excess gel was immediately removed from the SU-8 mold using a razor blade, and a 1:10 mixture of PDMS elastomer and curing agent was poured atop the mold and allowed to cure for an hour at 80 °C [Fig. 2(c)]. The cured PDMS with integrated composite membranes was peeled off the mold. Fluid ports were punched into each channel inlet and outlet using a 0.75 mm biopsy punch (Ted Pella, Inc.). The microchannel and a glass coverslip (Fisher Scientific, 12-548-5R) were exposed to oxygen plasma (Jelight, Model 42A), and immediately aligned and sealed under an inverted microscope. The microfluidic chip was baked for 24 h at 80 °C. This post-bond baking step improved the PDMS bond strength and prevented membrane leakage and rupture during pump operation.

To fabricate each 3D metal electrode for a low resistivity electrical connection to the composite membrane, solid gallium metal was melted and injected into the electrode channels of the microdevice [Fig. 2(d)]. First, solid gallium metal (Sigma-Aldrich, 263265) and the PDMS chip were heated to 40 °C on a hot plate. The melting temperature of gallium is 29.7 °C, and the newly melted liquid gallium was loaded into a 1 ml plastic syringe and immediately injected into each electrode side channel. Liquid gallium has a low viscosity (1.37 mPa s), and the PDMS did not require surface treatment prior to injection. Finally, electrical connections for an external power source were created by inserting 0.75 mm diameter copper wire leads into each electrode injection hole. Figure 2(e) shows the micrograph of the final complete micropump after fabrication. An enlarged view highlighting the PDMS nanocomposite membranes and gallium electrode areas is illustrated in Fig. 1(b).

B. Experimental setup: Measuring pump pressure

The final micropump device, shown in Fig. 2(e), is composed of four gallium electrodes, separated from the main microchannel by a thin nanocomposite PDMS membrane. To operate the pump, a high voltage power supply (PS325, Stanford Research System, CA) was wired to

each electrode pair to create the necessary electric field in the main flow channel; a positive voltage (V1) was delivered to one pair and the other electrode pair was connected to ground (V2). Voltages were applied between 0.80 and 2.5 kV. When the power supply was turned on, the resulting potential difference induced an electro-osmotic flow down the main channel in the direction of the applied electric field.

Using this pump design, we fabricated a microfluidic T-channel device to measure the pump pressure [Fig. 3(a)]. A micropump was integrated within each of the channel inlets. Two solutions of 2.5 mM phosphate buffered saline (PBS) were labelled with different colored fluorescent dyes, and each was then driven into separate fluid inlets. Experiments were performed by precisely varying the flow rate of each T-junction inlet and observing the impact of the micropump on the flow profile at different applied voltages.

For pressure driven channel flow, fluid velocity can be controlled by regulating either the applied pressure drop or the fluid flow rate. Experiments were performed at the channel junction. We used both methods simultaneously to obtain a precise measurement of the flow rate in the main pump channel. We used a precision syringe pump (Cole-Parmer, 78-8200C) to deliver fluid into one channel inlet a known flow rate. A constant pressure source was then applied to drive fluid flow into the second inlet. The pressure was slowly increased, and the position of the liquid interface was monitored. When the inlet flow rates were equal, a liquid interface formed at the centerline within the T-junction, as shown in Fig. 3(b). With each fluid inlet flow rate known, a single micropump was activated within the pressure driven inlet in the direction opposing fluid flow. The resulting EO-induced counter flow acted against the external flow field and served to reduce the total flow rate within the microchannel. At a critical voltage, the micropump flow rate completely balanced the externally applied flow rate, and flow from this fluidic inlet ceased [Fig. 3(c)]. The pump flow rate was combined with the hydrodynamic resistance across a rectangular microchannel length to calculate the induced pressure drop between the pumping area and T-channel outlet, which is equal to the pressure drop created by the micropump. This process was repeated over a range of fluid flow rates and applied voltages.

FIG. 3. (a) Microfluidic T-channel device used to quantify pump pressure. (b) Two fluid streams are driven into the device at a known flowrate. The left-most stream is driven using a constant pressure source, while the right-most is delivered using a constant flowrate source. When no pump is active and the flowrate of each stream is equal, the interface between the two phases is centered within the main flow channel. (c) The left-most pump is activated to completely cancel the pressure driven flow. The flow of this stream ceases and the interface position shifts.

TABLE I. PDMS composite resistivity and pump efficiency for different values of CB wt. %.

CB weight percent	0%	5%	10%	20%
Resistivity (Ω cm)	5×10^{10}	2.5×10^{10}	3.5×10^9	8.5×10^8
Efficiency (%)	0.3	0.5	3	13

Therefore, using the microfluidic T-junction in combination with an external constant pressure and flow rate source, we are able to experimentally measure the maximum output pressure of the micropump as a function of applied voltage.

C. Reagents and electrical parameters

The micropump experiments were conducted using diluted 10 $\times$ PBS and deionized (DI) water as working solutions. Fluorescent dyes were used as convective tracers to the induced EO flow using fluorescent solutions of Alexa Fluor 488 (green) and 594 (red). Because fluorescent dyes are ionic and can influence the electrical conductivity of the fluid, we based our pumping theory on the experimentally measured electrical conductivity of the fluorescent-labelled 2.5 mM PBS (113 μ S/cm) and DI water (2 μ S/cm) solution.

D. PDMS composite resistivity

The electrical resistivity of the PDMS composite membrane is needed to calculate the effective voltage drop across the main flow channel (Eq. (7)), and is an important parameter in determining the influence of carbon nanopowder wt. % on our EO pump performance. We measured the PDMS resistivity for each device using a high resistance ohmmeter (Alpha Labs, Inc., Model HR2). The PDMS resistivity in our T-channel device was 5.32×10^{10} Ω cm. The electrical resistivity of PDMS has been previously reported to be on the order of 10×10^{13} Ω cm. It is important to note, however, that we have previously observed that PDMS post-bond baking time influences this property; baking time increases from 1 h to 48 h, and the resistivity of PDMS ascends from 10^9 to 10^{13} Ω cm. Table I reports the electrical resistivity of the PDMS composite membrane at different carbon nanopowder weight percentages (wt. %).

IV. RESULTS AND DISCUSSION

A. Nanocomposite resistivity on pump performance

Figure 4 shows the micropump pressure response plotted as a function of voltage for different CB wt. %. At 20 wt. %, the PDMS nanocomposite produced a 40-fold increase in the pump pressure over the PDMS membrane without CB, leading to pressure results as high as 1200 Pa. Using Eq. (4), we predicted the maximum pump pressure as a function of the applied voltage. The theory, plotted in Figure 4, shows good agreement with our experimental data. For this microchannel geometry, the threshold operation voltage (screening potential) was 675 $\pm$ 50 V. In terms of pump performance, CB wt. % had a very little influence on the micropump screening potential, but delivered significant improvements in pump pressure and efficiency. The pump efficiency, defined here as the ability for an electric potential to penetrate the PDMS membrane ($\Delta V_{eff}/\Delta V_{total}$), is summarized for different CB wt. % in Table I. For a micropump without nanocomposite (a weight ratio of 0%), less than 1% of the total potential applied penetrates into the main flow channel. The addition of 20% CB increases the pump efficiency to 40-fold, from 0.33% to 13%.

B. Pump pressure and conductive buffer

Because contactless micropumps do not require direct contact with working buffers, they have several advantages over traditional DC EK pumps since they do not suffer from Faradaic

FIG. 4. The microfluidic T-junction was used to measure the maximum pressure generated by the micropump for different CB wt. % for DI water. The output pressure varies linearly with the applied voltage and matches the theoretical pump curve given by Eq. (5).

reactions and electrode decomposition. However, the required high resistivity PDMS membrane leads to low working efficiency, and a high kilovolt voltage is required for successful operation. One significant drawback of this feature is the inability of high resistive designs to effectively pump conductive buffer. In previous work, we demonstrated successful pumping, routing, and metering with contactless pumps without CB additives, but due to the low efficiency, the working fluids were limited to low conductivity buffers. Many microfluidic applications, however, are suited for biological purposes and require the ability to process commonly used physiological buffers. Hence, in this section, we demonstrate how integrated CB PDMS membranes extend the contactless micropump ability to pump buffers with higher electrical conductivity.

Microfluidic “T junction” experiments were repeated with a buffer with dilute PBS. Micropump maximum pressure was calculated and plotted against the applied voltage, as shown in Figure 4. The theoretical model [Eq. (4)] was applied to the dataset, which agrees well for each working fluid. When chip geometry is fixed, the slopes of output pressure curve are solely determined by the pump efficiency that calculated by $\Delta V_{eff}/\Delta V_{total}$. Higher buffer conductivity (113 $\mu\text{S}/\text{cm}$) reduces pump efficiency from 13% to 0.28% with 20 wt. % CB. Therefore, a much higher potential is required to achieve the same output pressure when operating with more conductive buffers. While CB significantly enhances pump performance for manipulating more conductive buffers, interestingly the screening potential is not significantly influenced by the buffer electrical conductivity when we switch working fluids. Similar screening potentials were observed even when increasing electrical conductivity by 50-fold. Based on this observation, we speculate that the steric hindrances limit the concentration of counterions that can electro-migrate and accumulate at the microchannel wall, which leads to a finite screening potential.

The semilog plot in Fig. 5 illustrates the influence of CB weight ratio and buffer conductivity on maximum micropump pressure. The voltage was fixed at 2000 V for all four datasets to give a direct comparison between the buffer conductivity, the CB wt. %, and the resulting pump pressure. First, the high conductivity buffer combined with the high resistivity PDMS membranes undermines the pump performance due to the extremely low efficiency of non-CB PDMS membranes; no fluid motion was observed for dilute PBS buffer. Much higher pressures are achieved with the addition of CB to the PDMS membrane. Shown in the figure, pump output pressure increases from 70 Pa to 1600 Pa when CB weight ratio increases from 5% to 20% when DI water is used as a working buffer. The addition of CB enables the successful pumping of conductive PBS buffer. However, the maximum pump pressure decreases when compared to

FIG. 5. Semilog plot for pump pressure at different values of CB wt. % for DI water and 2.5 mM PBS buffer.

DI water from 1600 Pa to 80 Pa, 480 Pa to 30 Pa, and 70 Pa to 20 Pa, for a CB wt. % of 20%, 10%, and 5%, respectively.

V. CONCLUSIONS

In this work, we have demonstrated the ability to pump conductive buffers using a contactless DC electro-osmotic micropump with conductive carbon black PDMS membranes. The PDMS membranes are used to separate the flow channel from solid gallium electrodes and contain patterned regions of conductive composite. CB membranes with weight ratios of 5%, 10%, and 20% were integrated directly into the side wall of the flow channel using soft lithography. Using a microfluidic “T” junction shaped geometry, we measured the maximum output pumping pressure for different values of CB wt. % and investigated this influence on the ability to pump conductive buffers. Under an applied external potential range (800–2500 V), the resulting pressure curves agree well with our electro-osmotic flow model. This work demonstrates that the CB conductive membranes greatly enhance contactless micropump performance for conductivity buffer processing. Although the conductivity criterion reported here still falls below common values for physiological buffer, with further optimization and more conductive additives, the proposed contactless micropump is a promising tool for portable microfluidic biological buffer processing. Future work will focus on optimizing pump geometry and membrane additives to increase pump efficiency without sacrificing electric field stability and buffer electrolysis and further applying this work to other electrokinetic applications including dielectrophoresis and electrophoresis.

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