Analysis of micromixing of non-Newtonian fluids driven by alternating current electrothermal flow

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ABSTRACT

Biochemical applications pertaining to chip scale mixing often deal with fluids which exhibit non-Newtonian behavior. This paper presents numerical investigations on the characterization of the effect of shear dependent apparent viscosity of non-Newtonian fluids on the mixing efficiency and volume flow rate in an alternating current electrothermal micromixer driven by electrothermal micropump. The micromixer consists of thin film asymmetric pairs of electrodes on a microgrooved channel floor and an array of electrode pairs on the top wall. The results show that mixing quality and flow rate have a strong dependence on shear dependent viscosity of the non-Newtonian fluid. Using a power law based constitutive model, it is found that for specific design parameters, more uniform and homogeneous mixing is achieved with increasing flow behavior index. Thus, electrothermal mixing in a microfluidic device has higher efficiency and effectiveness in the regime of dilatant fluids compared to Newtonian and pseudo-plastic fluids. Our results also demonstrate that an optimal voltage for maximum mixing efficiency progressively reduces, while the flow rate continues to increase monotonically, with enhancements in the applied AC potential. These results hold practical implications in several emerging applications, including chemical analysis of biological fluids in general, and biomedical diagnostics in particular.

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1. Introduction

Microscale mixing of fluids is an important part in biochemical analysis [1], and has widespread applications in biochemical reactions [2], clinical diagnostics [3], chemical synthesis [4] and drug delivery [5]. Most of the microfluidic devices deal with low velocity in the region of laminar flow [6]. At low Reynolds number, the mixing process is diffusion dominated. Due to larger size of biomolecules, diffusion coefficient is low, leading to large retention time ($t \propto 1/D$, $t \sim$ diffusion time, $D \sim$ diffusion coefficient). On the top of this, a long mixing channel is required to achieve uniform and homogeneous mixing. Therefore effective mixing in microfluidic devices poses a major challenge.

Several investigations have been reported in the literature in an effort to address this issue, introducing new methods for enhancement of mixing. In the studies of microfluidic mixing, mixing processes and mixing devices are categorized under passive [7] and active [8]. Passive mixers leverage the geometrical advantage, where large numbers of obstacles are mounted along the flow path to increase interfacial area between two confluent streams of fluids [9–11]. Although simpler implementation is a significant benefit, the requirement of a long channel for operation neces-

sitates large pumping power [12], thereby making them less attractive. Active micromixers, in contrast, involve a mixing process that is driven by an external driving force, thus facilitating continuous flow and efficient mixing over smaller flow lengths. The external driving force, depending on its actuation mechanism, can be classified under magnetic [13], electrohydrodynamic [14], acoustic [15], ultrasonic [16] or electrokinetic [17]. Of these, electrokinetic technique has received extensive attention as it can enable rapid, uniform and homogeneous mixing within a short time and flow length. It also offers the advantages of ease of operation, simplicity in design and manufacturing and high mixing efficiency. Early efforts on electrokinetic mixers used high voltage (up to 4 kV) at low frequency (4–5 Hz) [18]. At this high voltage, Faradaic current conjunction with generation of bubble became a drawback. This can, however, be avoided at low voltages (1–20 V), thereby rendering AC electrokinetics as an attractive option in this range.

When an AC voltage is applied on electrodes connected over the microchannel, two types of AC phenomena may arise depending on the range of applied frequency (f) and electrical conductivity (σ) of the electrolyte, namely AC electroosmosis (ACEO) and AC electrothermal (ACET). Hansen [19] reported that ACEO forces are insignificant above frequency of 100 kHz. Johnson et al. [20] used a T-shaped micromixer with a pulsed UV excimer laser to generate slanted wells. Under electroosmotic flow, slanted wells imposed transverse motion on the two confluent streams resulting in rapid mixing. At high electrical conductivity, net charges are

suppressed leading to generation of low ACEO forces. It was reported that ACEO effects fall drastically above a fluid conductivity of 0.084 S/m [21]. Biochemical applications, on the other hand, deal with samples having conductivity higher than this threshold value. ACET, operating due to variation of conductivity and permittivity induced by thermal gradient [22], therefore form an attractive alternative in such applications [23].

Over the last few years, researchers are focusing on development of ACET micromixers through theoretical modeling, experimental investigations and numerical studies. Feng et al. [24] derived a theoretical model and showed chaotic nature of mixing in a T-shaped micromixer operated by ACET. Cao et al. [25] numerically obtained a mixing efficiency of 98% in a narrow mixer with an optimum electrodes arrangement. Sasaki et al. [26] used sinusoidal inter-electrode gap to enhance mixing. Huang et al. [27] investigated mixing performance of an ACET driven mixer comparing with similar passive mixer and obtained almost complete mixing for a specific design of the mixer. Recently, Zhang et al. [28] conducted mixing process with thin film resistive heaters.

MEMS-fluidic system often deals with fluids which exhibit non-Newtonian behavior. Furthermore, many real-world biochemical applications use non-Newtonian fluids. While several existing investigations on ACET driven micromixers consider Newtonian fluids, the best of our knowledge, electrothermal based mixing of non-Newtonian fluids have not been studied so far. This paper attempts to characterize electrothermal based micromixing of non-Newtonian fluids in a microgrooved channel using a 2-D numerical approach. The fluid pumping is accomplished through ACET driven microfluidic system in the microchannel. The constitutive behavior of non-Newtonian fluid is investigated using power-law model [29]. The focus of this study is to investigate ACET driven micromixing of non-Newtonian behavior of power-law fluids.

2. ACET mixing

2.1. Theory

ACET based mixing and transport of biofluids deal with simultaneous coupling of electric field, electrothermal forces, temperature field, flow field and concentration field. Under the assumptions of time averaged forces and negligible magnetic field effect [30,31] the Laplace equation in a AC electric field can be written as

$$\nabla^2 V = 0, \tag{1}$$

$$\mathbf{E} = -\nabla V, \tag{2}$$

where V is electric potential and $\mathbf{E}$ is the electric field strength. Influence of non-uniform AC electric field invokes Joule heating on the electrolyte solution leading to non-uniform temperature which causes permittivity and conductivity variations. As a result, a body force [30,32] arises given by:

$$\mathbf{F_E} = \frac{1}{2} \varepsilon \text{Re} \left[(\alpha - \beta) \frac{(\nabla T \cdot \mathbf{E})}{1 + (\omega \tau)^2} \mathbf{E} - \frac{1}{2} \alpha \nabla T |\mathbf{E}|^2 \right], \tag{3}$$

The Coulomb (first term of the right hand side) and dielectric force (second term of the right hand side) are strong function of local gradient of the fluid properties which arises as a consequences of temperature gradient [24,33–35]. The constants α and β are $\frac{1}{\varepsilon} \frac{\partial \varepsilon}{\partial T} \approx -0.004\,°\text{C}^{-1}$ and $\frac{1}{\sigma} \frac{\partial \sigma}{\partial T} \approx 0.02\,°\text{C}^{-1}$ respectively, T is the temperature, $\omega = 2\pi f$ is the angular frequency of the AC potential and $\tau = \varepsilon/\sigma$ is charge relaxation time of the electrolyte solution.

Temperature distribution arises from applied AC electric field is written by the following energy equation:

$$\rho C_p \mathbf{U} \cdot \nabla T = k \nabla^2 T + \sigma |\mathbf{E}|^2, \tag{4}$$

where $\rho, C_p, \mathbf{U}$ and k are the density, specific heat, flow velocity and thermal conductivity of the fluid respectively, and $\sigma |\mathbf{E}|^2$ is the source of heat caused by Joule heating.

Under the assumptions of steady, laminar, incompressible flow, the continuity and momentum equations are expressed by:

$$\nabla \cdot \mathbf{U} = 0, \tag{5}$$

$$\rho (\mathbf{U} \cdot \nabla \mathbf{U}) - \nabla \cdot \mu_{app} \left[\nabla \mathbf{U} + (\nabla \mathbf{U})^T \right] + \nabla p = \mathbf{F_E}, \tag{6}$$

where (μ_{app}) is the apparent viscosity of the fluid and p is the pressure.

In a power law model [36], the apparent viscosity may be related to the local rate of deformation as:

$$\mu_{app} = m \dot{\gamma}^{n-1} \tag{7}$$

where m is the flow consistency index, n is the flow behavior index, and $\dot{\gamma}$ $(= ((\partial u/\partial y + \partial v/\partial x)^2 + 2(\partial v/\partial y)^2 + 2(\partial u/\partial x)^2)^{(n-1)/2})$ is the shear rate. Here, u is the longitudinal component of the velocity and v is the transverse component of the velocity. For shear-thinning (pseudoplastic) fluids $n < 1$ while for shear-thickening (dilatant) fluids $n > 1$. For Newtonian fluid, flow behavior index is $n = 1$.

Mass transport of the sample particle is governed by convection and diffusion mechanism under steady flow condition given by the Fick's second law:

$$\mathbf{U} \cdot \nabla C - D \nabla^2 C = 0, \tag{8}$$

where C and D are the concentration and diffusion coefficient of the sample.

Mixing quality is defined by mixing efficiency at any cross section by index γ [37]:

$$\gamma(x) = 1 - \frac{\int_0^H |C - C_\infty| \mathrm{d}y}{\int_0^H |C_0 - C_\infty| \mathrm{d}y}, \tag{9}$$

where C is the concentration profile at a distance y from the inlet of the channel across the channel height. C_0 and C_∞ are profiles of concentration at completely unmixed and completely mixed states, respectively. The value of $\gamma(x)$ varies in between 0 and 1, with the conditions completely unmixed and completely mixed states.

In addition to the mixing efficiency, another important parameter which characterises the mixing performance of the micromixer is the volume flow rate at exit of the microchannel. Flow rate of the electrolyte solution per unit width of the microchannel is evaluated from the following numerical integration of horizontal component of the fluid velocity (u) at the outlet of the channel:

$$Q = \int_0^H u \mathrm{d}y. \tag{10}$$

A scale-balance is performed to evaluate the order of applied potential on which the velocity varies. Electrothermal based microfluidic devices are involved with low velocity/Reynolds number. Low thermal Peclet number implies higher contribution of conduction heat transfer over convection heat transfer. Under this circumstance, the balance between conduction part and Joule heat generated by ACET technique gives:

$$k \nabla^2 T \sim \sigma |\mathbf{E}|^2 \text{ or } \Delta T \sim |\mathbf{E}|^2, \tag{11}$$

Similarly, balance between viscous force and ACET body force in the momentum equation follows $\nabla \cdot \mu \nabla \mathbf{U} \sim \mathbf{F_E}$. In our study, we seek a situation where Coulomb force is dominating over dielectric force. Thus, a counter balance between viscous and Coulomb force gives:

$$\nabla \cdot \mu \nabla \mathbf{U} \sim \frac{1}{2} \varepsilon (\alpha - \beta) \frac{(\nabla T \cdot \mathbf{E})}{1 + (\omega \tau)^2} \mathbf{E}, \text{ or } nmu^n \sim (\nabla T \cdot \mathbf{E}) \mathbf{E}, \tag{12}$$