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Effects of polydimethylsiloxane (PDMS) microchannels on surface acoustic wave-based microfluidic devices

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ABSTRACT

Polymers are ideal microfluidic channel materials due to their chemical and biological compatibility, optical characteristics, fast and easy prototyping capability, and low fabrication cost. Especially, polydimethylsiloxane (PDMS) is the most widely used polymer as a microfluidic channel material due to many prominent features. However, this elastomer material severely attenuates Rayleigh surface acoustic waves (SAW) when they propagate toward the sample fluid within the microfluidic channel. The acoustic energy absorption by the microchannel affects the capabilities and efficiencies of the Rayleigh SAW-based microfluidic devices. In this paper, we investigate the effects of the PDMS channel wall thickness on the insertion loss and the particle migration to the pressure node due to acoustic radiation forces induced by Rayleigh SAW. Our results indicate that as the PDMS channel wall thickness decreased, the SAW insertion loss is reduced as well as the velocity of the particle migration due to acoustic forces increased significantly. As an example, reducing the side wall thickness of the PDMS channel from 8 to 2 mm in the design results in 31.2% decrease and 186% increase in the insertion loss and the particle migration velocity, respectively.

1. Introduction

Microfluidics represents the science and technology of fluidic flow in microchannels with dimensions ranging from just few micrometers to hundreds of micrometers. It is a crucial technology in a variety of miniaturized analysis systems for chemical, biological, and biomedical applications including drug delivery [1,2], DNA sequencing [3,4], clinical and veterinary diagnostics [5,6], point of care testing [7,8], and chemical synthesis [9,10]. The choice of materials is critical for microfluidic devices and important considerations are chemical and biological compatibility, optical characteristics, fast prototyping, and low fabrication cost.

To date, three materials (namely silicon, glass, and polymers) have been used in microfluidic device designs. Recent studies reported the comparison of these materials for chemical and physical operation in a microfluidic device [11,12]. The properties of silicon are well-established and the fabrication techniques in silicon are well-developed. However, as silicon is not transparent, it conflicts with optical detection in most bioanalysis. Moreover, the silicon-based fabrication process is expensive, is limited to planar surfaces, and needs specialized facilities. Glass has a good chemical resistance and excellent optical transparency. However, it has disadvantages such as high-cost, complex fabrication process and limited integration capability with other materials used in the design. Compared to silicon and glass, polymers are more attractive materials for microfluidic channels due to the ease of fabrication such as replica molding and rapid prototyping which are simple, low cost, time saving, and highly reproducible. In addition, polymers feature a good chemical resistance, optical transparency, and no UV absorption [12].

Different polymers have been investigated as microfluidic channel materials including polystyrene [13], polyimides [14], polycarbonate [15], polymethymethacrylate (PMMA) [16,17] and polydimethylsiloxane (PDMS) [18–20]. Among them, PDMS is the most widely used as well as actively developed polymer for the microfluidics because it has prominent features compared to other polymers, including very low electromagnetic energy dissipation, high dielectric strength, wide temperature range of use, elastomeric properties, biocompatibility, non-toxicity, high optical transparency, gas permeability, reversible and irreversible bonding, easy molding, and low chemical reactivity [21–23].

Recently, PDMS-based microfluidic channels have been integrated with surface acoustic wave (typically known as Rayleigh wave) devices for wide variety of applications including focusing [24,25] and separation [26–28] of particles/cells as well as pumping [29–31] and mixing [32–34] of fluids in the microfluidic channel. Shear-horizontal surface acoustic wave has been employed for sensing applications because of reduced damping losses [35,36]. Elastomer materials, such as PDMS, severely attenuate the SAW

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propagation, thus the transmission of the acoustic energy is reduced by the channel wall thickness when the SAW propagates toward the sample liquid inside the microfluidic channel [37]. As this acoustic energy loss affects the capabilities and efficiencies of the SAW-based microfluidic devices, there is a need to investigate the effects of PDMS channel wall thicknesses on the SAW-based microfluidic device design. In this paper, we present the effects of the side and top wall microchannel thicknesses on the insertion loss of SAW and the particle migration to the pressure nodes due to acoustic radiation forces induced by SAW.

2. Materials and methods

2.1. Working mechanism

Fig. 1 shows a schematic illustration of a typical SAW-based PDMS microfluidic device. It consists of interdigitated transducer (IDT) fabricated on a piezoelectric substrate and PDMS microfluidic channel. When the IDT is excited with an RF signals, SAW propagates toward the PDMS microfluidic channel. When the SAW reach the sample fluid inside the PDMS microfluidic channel, it is converted to leakage wave resulting in pressure fluctuations and the acoustic energy radiates longitudinal pressure waves into the fluid. The pressure fluctuations induce acoustic streaming and acoustic radiation forces [28,34]. These forces have been used for many applications such as pumping, mixing, focusing, and separation. However, a portion of the acoustic energy is absorbed by the PDMS channel walls when the SAW reach the sample fluid inside the PDMS microfluidic channel. The capability and efficiency of the device are affected by this acoustic energy loss. Thus, in this study the effects of the side and top wall thicknesses (as shown in Fig. 1) of the PDMS microchannel on the transmission characteristic of SAW and the particle trajectories due to acoustic radiation forces were investigated experimentally.

2.2. Device design and fabrication

The fabrication process of the SAW-based microfluidic device used in this study consists of three steps including patterning of the IDT on a substrate, fabrication of the PDMS microfluidic channel, and bonding of the microfluidic channel to the substrate. Fig. 2(a and b) show the fabrication process steps for the A–A′ cross-section as indicated in Fig. 2(c). The detailed fabrication procedures including specific parameters, such as baking times, temperatures, and spinner speeds, can be obtained from our previous publication [26]. The IDTs were fabricated using standard photolithography. A double-side polished Y + 128° X-propagation lithium niobate ($LiNbO_3$) was selected as the piezoelectric substrate. A 100 nm chrome layer was sputtered on the $LiNbO_3$ substrate, and then S1813 positive photoresist (Shipley, Marlborough, MA) was spun on the substrate. Once the substrate was exposed to UV light, it was developed in MF 319 developer (Shipley, Marlborough, MA). The IDT features were formed by the CR-7S chrome etchant (Cyantek, Fremont, CA), and finally the photoresist was removed by AZ400T stripper (AZ Electronic Materials, Branchburg, NJ). IDT was designed to operate at a resonance frequency of 13.3 MHz, corresponding SAW wavelength of 300 μm. The width and period of the IDT finger were 75 and 300 μm, respectively.

PDMS microfluidic channel was fabricated using soft lithography replica molding technique. First, a channel mold was fabricated with SU-8 2075 negative photoresist (MicroChem, Newton, MA) spun on a silicon wafer by the photolithography. The height of the fabricated mold, corresponding to the microfluidic channel depth, was characterized with a surface profilometer. The PDMS pre-polymer base (Sylgard™ 184 kit, Dow Corning, Midland, MI) was cross-linked with the curing agent by the weight ratio of 10:1. Once both parts were well mixed, the mixture was poured onto the fabricated channel mold, and then cured at room temperature for 24 h to prevent the shrinking of PDMS. After the PDMS replica was peeled off from the channel mold, it was bonded to the substrate containing IDTs with oxygen plasma treatment. The width and depth of the fabricated PDMS microfluidic channel were 150 and 100 μm, respectively. To investigate the effects of the PDMS microfluidic channel wall, the side and top wall thicknesses (shown in Fig. 1) varied from 2 to 8 mm for each wall. Fig. 2(c) shows one of the complete SAW-based microfluidic devices used in this study.

2.3. Experimental setup

For the insertion loss measurements, the fabricated integrated device was connected to a RF network analyzer (E5061A, Agilent), and the insertion losses (S_{21}) due to the existence of PDMS microchannel in the design were measured (Fig. 3(a)). For the investigation of particle migration to the pressure node due to acoustic radiation forces induced by SAW, a solution of fluorescent polystyrene (PS) particles with 10 μm diameter (Bangs Laboratories, Fishers, IN) was injected to the PDMS microfluidic channel using the microliter syringe (Hamilton, Reno, Nevada) and the syringe pump (KD Scientific, Holliston, MA). The concentration of the polystyrene particles in the sample solution was 1% by volume. For the propagation of the SAW toward the sample solution in the PDMS microfluidic channel, AC signal generated by an arbitrary signal generator (AFG3022B, Tektronix) at the resonance frequency was amplified using a RF power amplifier (325LA, ENI), and then the signal was supplied to the IDT. The paths of the particles migration

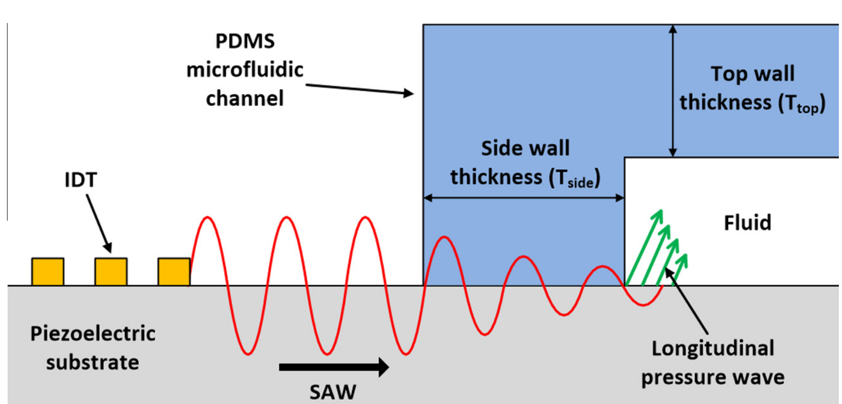

Fig. 1. Schematic illustration of working concept of a typical SAW-based PDMS microfluidic device.