Miniaturized ultrasound scanner by electrowetting

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An ultrasound imaging technology based on electrowetting has been developed and integrated in a miniaturized ultrasound scanner. The feasibility of scanning the ultrasound beam of a single-piston transducer in a three-dimensional space by using electrowetting is demonstrated. The technology has a high potential to be embedded in devices where size restrictions do not allow the use of traditional ultrasound phased-array transducers.
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Ultrasonic imaging is one of the most important diagnostic tools in healthcare, enabling the visualization of soft tissue and blood flow velocities in real-time.[1] Other important application of the ultrasound is the detection of material irregularities and inhomogeneities in nondestructive testing.[2] Generally, the transducers used in external applications, such as imaging of organs from outside of the body, are based on a phased array configuration. However, for endoscopic or catheter applications within the body, the size of the instruments is restricted.

Systems have been developed in the past for focusing ultrasound by hydraulic principles[3] for underwater operations, where by varying the pressure of a certain liquid within a rubber sealed container, the focus of the ultrasound beam could be changed. We have developed a miniaturized ultrasound scanner which uses single-element transducer and beam deflection by controlling the shape of the interface between two liquids electrostatically. The interface shape, which is not only variation of curvature but also tilt, refracts the ultrasound, thus allowing for imaging within a well defined sector either in a two- or a three-dimensional space.

Electrowetting is a phenomenon that occurs when charge is stored at the interface between a conducting liquid and a conducting solid, separated by a certain dielectric layer.[4] When the initial contact angle, θ_0, is smaller than 180°, the relationship between the contact angle, θ, and the applied voltage, V, is described by

$$\cos\,\theta = \cos\,\theta_0 + \frac{\varepsilon}{2\gamma d}V^2, \tag{1}$$

where ε is the dielectric constant, d is the thickness of the dielectric layer, and γ is the interfacial tension between the conducting liquid and the surrounding medium.

A combination of two fluids is suitable for electrowetting-based devices when the following conditions are fulfilled: (i) the two fluids are immiscible in order to form an interface; (ii) one of the fluids is conducting, thus reacts to the electric field, and the other one is insulating, (iii) the viscosity of the fluids is below 500 cS, which is essential for the dynamics of the electrowetting.

Since the electrowetting principle is used for refracting ultrasound beams, the fluids have to fulfill additional criteria that are related to acoustics as follows: (iv) their acoustic impedances must be substantially similar in order to avoid undesired reflectance: $|\rho_1 v_1 - \rho_2 v_2| \ll \rho_2 v_2$, where ρ and v denote density and speed of sound, respectively; (v) the speed of sound difference of the two media must be large $v_1/v_2 > 1.5$ in order to obtain sufficiently large refraction angles; (vi) the fluids must support the propagation of ultrasonic waves, preferably in the frequency range of 0.5–50 MHz.

From criteria (iv) and (v) it results that the ratio of the density of the two media should be inversely proportional to their speed of sound ratio, therefore $\rho_2/\rho_1 > 1.5$. Ultrasound propagation is not supported by gaseous media. Therefore from criterion (vi), it results that the fluids must be in liquid state at least at the temperature at which the device is used.

The deflection angle β of the ultrasound beam in Fig. 1(a) from the longitudinal axis of the ultrasound transducer is determined by the refraction at the interface, $\beta = \theta_t - \theta_i$, where θ_i and θ_t are the angle of incidence and transmission, respectively. Note, that in case of very thin acoustically transparent window (ATW), the refractions at the two interfaces of the material will cancel out when the liquid inside the lens and the media in which the ultrasound is coupled out have similar speed of sound. The result will be a negligible lateral shift of the ultrasound beam in case of a 50 μm thin ATW.

In case that large deflection angles are desired, the speed of sound ratio of the two media should be maximized by taking into account that their density ratio must vary inversely proportional. The intensity reflection of ultrasound at an interface in the general case with random incidence[5] is given by

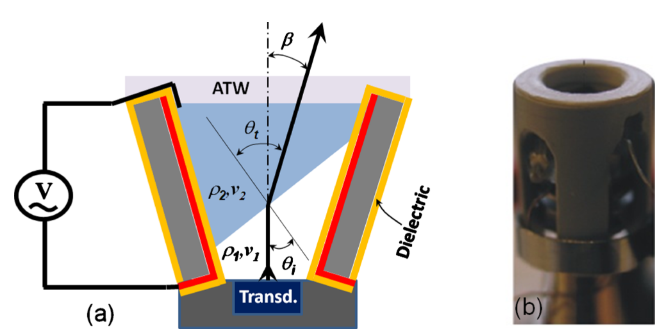

FIG. 1. (Color online) (a) Liquid lens refracted ultrasound. The transducer transmits an ultrasound beam that is refracted by the interface between two immiscible liquids and passes through the ATW; (b) ultrasound scanner prototype.

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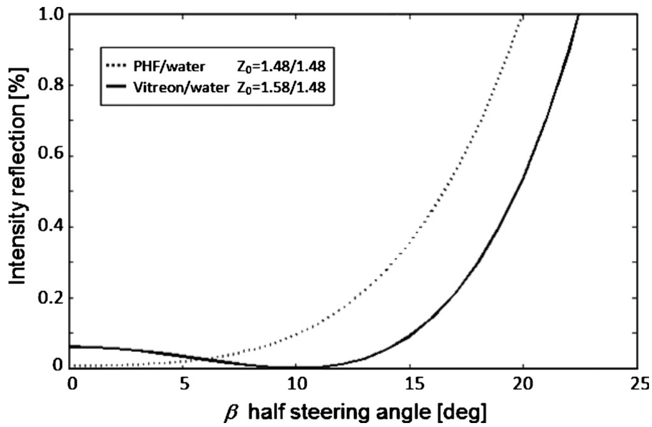

FIG. 2. Intensity reflections form the interface of two different liquid combinations. One of the combinations fulfills the condition for intromission angle, and gives lower ultrasound reflections for increasing incidence angles.

$$R = \left(\frac{Z_2/\cos\ \theta_t - Z_1/\cos\ \theta_i}{Z_2/\cos\ \theta_t + Z_1/\cos\ \theta_i} \right)^2. \tag{2}$$

A combination of two fluids may or may not have an incidence angle, different from zero, at which the ultrasound reflection is minimum. This is determined by the physical parameters (density, speed of sound) of the two media. The demand for decreasing intensity reflection for increasing incidence angle is $dR/d\theta < 0$, which results in

$$\frac{\rho_2 v_2^3 (\rho_1 v_1 - \rho_2 v_2)(\rho_1 v_1 + \rho_2 v_2)}{\rho_2^2 v_2^4 - \rho_1^2 v_1^4} > 0. \tag{3}$$

Investigations led to a couple of liquid combinations that are fulfilling criteria (i)–(vi). Figure 2 illustrates the intensity reflection of the ultrasound for two different liquid combinations. For water/perfluoroperhydrofluorene ($C_{13}F_{22}$, also denoted as PHF) the intensity reflection increases starting from normal incidence. The reflection exceeds already 1% for about 15° half steering angle β of the ultrasound beam. The combination of water/perfluoroperhydrophenanthrene ($C_{14}F_{24}$ also denoted as Vitreon) shows a different behavior. The reflection intensity decreases toward zero at an incident angle of about 10°, which is called the angle of intromission (similar to Brewster angle in optics), after which it increases again. This liquid combination is the most advantageous for ultrasound refraction, because it gives the smallest reflection from the interface for a large range of steering angles. Note that the best choice of liquids in scanning applications is not necessarily given by the perfect match of the acoustic impedances at normal incidence to the interface.

At high frequencies PHF has a lower ultrasound attenuation with respect to Vitreon (3.3 versus 3.7 dB/mm at 25 MHz), which is important for high resolution applications such as cardiac or blood vessel imaging, since the ultrasound has to travel twice in the liquids before being received by the piezoelectric element.

An inverted pyramidal frustum shape with a total diverging angle of 10° has been chosen for avoiding the refracted ultrasound colliding with the internal faces of the structure, causing specular reflection within the liquid lens. The height of the frustum was 3.5 mm, while the size of the small base of 2 mm was chosen to host an ultrasound transducer with a 1.5 mm round active element. With such a liquid lens configuration an ultrasound imaging sector of 50° can be achieved with both liquid combinations, without dramatically increasing the ultrasound intensity reflection from the liquid interface (<2.5%). To enable electrowetting, a dielectric layer of 5 μm parylene-C (SCS Coating Services) was deposited onto a 150 nm thin aluminum electrode, covering the inner faces of the frustum. The electrode is partitioned along the four edges of the frustum to form four separate electrical domains.

Increasing the contact angle of the liquid interface on the dielectric coating is important for enhancing the range of angles for which the interface can be operated by electrowetting. The hydrophobic coating AF1600 (by DuPont) that is used for optical beam refraction in liquid lenses[6] is not suitable for ultrasound liquid lenses, due to its solubility in the selected fluorinated dielectric liquids. Therefore, a different hydrophobic coating, M581 (research sample provided by Promerus LLC) at 1 wt % in hexane, has been chosen. This coating proved to be chemically inert to the liquid combinations. The advancing and receding water contact angles in air were 100° and 85°, respectively. The difference in advancing and receding contact angles did not negatively influence the operation of the liquid lens, since the tilt angle of the interface was determined by capacitive measurements of the individual walls on the frustum[7] and taken into account in ultrasound image construction. In order to increase the electric conduction of the polar liquid, a solution of 1 mM KCl in demineralized H_2O was used.

The ultrasound transparent window that seals the liquid lens is a 50 μm thick polymethylpentene sheet (TPX) (from Mitsui Chemicals) covered by a 20 nm thick silicon oxide layer on the side facing the container, for enhancing its hydrophilicity and serving as diffusion barrier for liquids. The acoustic impedance of the TPX sheet is 1.73 MRayls, which makes it a well suited material for external ultrasound imaging application of tissue, or internal ultrasound organ inspection in body fluids, since most tissue and blood impedances are in the same range.[5]

The liquid lens was operated in feed-back mode, which eliminates the interface tilt and shape irregularities due to contact angle hysteresis and temperature dependence. A frequency multiplexing scheme was applied to allow the measurement of the electric current through all four capacitors simultaneously. The driving voltages $V(f_i)$ (f_i=1, 2, 4, and 8 kHz, chosen to minimize crosstalk) generate a current i through the polar liquid comprising four different spectral components. These current components can be demodulated separately by four multipliers to which reference signals f_{1-4} were applied with the same frequency and phase as the current component of interest. A comprehensive description of the operation scheme of a liquid lens is given in Ref. 7 for an optical beam deflector. The differences between the optical beam deflector and the ultrasound beam deflector are the liquid combination, materials used to enclose the container and to enhance the hydrophobicity on the interior wall of the frustum. Snell's law is more favorable for achieving large deflection angles in acoustics compared to optics, as it is theoretically possible to have a strong acoustic refraction-lens system with simultaneous near-zero reflection at the interface. However, the necessary density difference between the liquids in an acoustic lens makes it more susceptible to gravity and acceleration forces. This limits the use of such

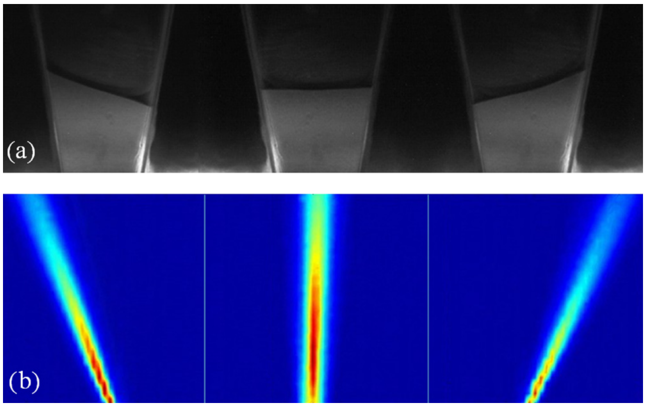

FIG. 3. (Color online) (a) Three positions of the interface between the immiscible liquids. The horizontal position of the interface represents normal incidence of the beam. (b) Measurement of ultrasound intensity for liquid lens refracted ultrasound beam (scale $20\times 20\ \mathrm{mm}^2$).

lenses to small sizes, where the surface effects become more important and start compensating for gravity.

Various positions of the interface between the two liquids are presented in Fig. 3(a). The horizontal position represents the normal incidence of the beam to the interface. Maximum deflection of the ultrasound beam is achieved for the extreme positions of the interface, as the field mapping measurements illustrate in Fig. 3(b).

Preliminary experiments have been performed to investigate the suitability of the system for medical ultrasound imaging, initially on porcine heart immersed in water. The A-lines from which the B-plane sector image has been constructed are presented in Fig. 4 with dotted lines. The front and back walls of the 6 mm thick tissue are evidently visible in the ultrasound image. The B-plane image has been obtained at 1 Hz frame rate.

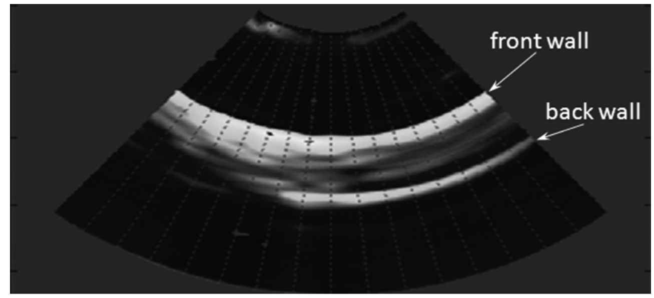

FIG. 4. Demonstration of the ultrasound sector scanning on a 6 mm thick porcine heart tissue immersed in degassed water. The dotted stripes represent the A-lines from which the real-time ultrasound image has been constructed at a frame rate of 1 Hz.

The experimental results show a promising technology for low cost small ultrasound scanners. The liquid lens refracted ultrasound configuration is one of the potential candidates for minimally invasive applications, where scanning of the ultrasound is performed by tilting an interface between two liquids in front of a transducer, which refracts the ultrasound, allowing for imaging within a well defined sector. The implementation simplicity in the minimally invasive instruments and their much lower costs are the biggest advantages, since there is no need for beam forming technology or for a multiplexing.

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