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Designing single-beam multitrapping acoustical tweezers

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ABSTRACT

The concept of a single-beam acoustical tweezer device which can simultaneously trap microparticles at different points is proposed and demonstrated through computational simulations. The device employs an ultrasound beam produced by a circular focused transducer operating at 1 MHz in water medium. The ultrasound beam exerts a radiation force that may tweeze suspended microparticles in the medium. Simulations show that the acoustical tweezer can simultaneously trap microparticles in the pre-focal zone along the beam axis, i.e. between the transducer surface and its geometric focus. As acoustical tweezers are fast becoming a key instrument in microparticle handling, the development of acoustic multitrapping concept may turn into a useful tool in engineering these devices.

1. Introduction

Noncontact particle handling methods based on optical [1] and acoustic [2] radiation forces are promoting a revolution in biotechnology and biomedical applications [3–7]. Techniques for particle trapping which employ a laser and an ultrasound beam are known, respectively, as optical tweezer [8] and acoustical tweezer [9]. Methods based on acoustical tweezers are potentially useful in applications for which optical tweezers can hardly operate. For instance, handling particles in opaque medium to electromagnetic radiation cannot be performed by optical tweezers. Furthermore, a laser beam may damage biological structures by heating and also by a process called photodamage, which is related to the formation of singlet oxygen when photon absorption occurs [10].

Two different approaches have been employed to acoustically trap microparticles, namely standing waves and single-beam methods. The first acoustical tweezer device used two counter-propagating focused ultrasound beams to form a standing wave at 3.5 MHz, which was used to trap 270 μm-diameter latex particles and frog eggs in water [9]. In other arrangement, an ultrasound standing wave generated between a 2.1 MHz linear array and a reflector trapped alumina microparticles with 16 μm mean diameter [11]. Devices in the acoustofluidics realm are, in general, based on standing waves [12,13]. Standing surface acoustic waves (SSAW) have been used to trap particles with diameter smaller than 8 μm suspended in microfluidic channels [14,15]. Standing Bessel waves were generated by a circular 64-element ultrasonic array to manipulate 90 μm-diameter polystyrene microparticles in 2D [16]. Tilted standing waves produced by a three PZT transducer setup which operated at 1.67 MHz were used to trap and transport a 100 μm-diameter silica bead [17]. Furthermore, the standing wave method has been also employed to levitate particles and droplets in air [18,19].

On the other hand, single-beam acoustical tweezers utilize tightly focused transducers and linear arrays for trapping microparticles in the device focus point. In particular, a 30 MHz-transducer with an f-number of 0.75 tweezed a 126 μm-diameter lipid microdroplet [20]. Higher frequency transducers operating at 200 MHz have been designed to handle a 10 μm-diameter leukemia cell [21] and microsphere [22]. A PZT transducer equipped with a multi-foci Fresnel lens generated a 17.9 MHz zeroth-order Bessel beam, which was employed to trap microspheres ranging in diameter from 70 to 90 μm [23]. A 57.5 MHz needle hydrophone with an f-number of 1 produced an ultrasound beam which tweezed a 30 μm diameter lipid microdroplet [24]. Also, a 64-element linear phased array operating at 26 MHz was able to trap a 45 μm-diameter polystyrene microparticle [25].

The key aspect in designing single-beam acoustical tweezers is how to form a beam to trap a particle with an specific size. Different schemes of acoustical tweezer beamforming have been previously investigated for a circular focused [26–28] and a linear array [29] transducer. In these studies, only one trapping point located in the transducer focal zone was considered. Recently, the possibility of trapping a particle in the nearfield of piezoelectric disk was discussed [30]. Perhaps the most serious disadvantage of this method is that the yielded ultrasound beam behaves like a plane progressive wave in the vicinity of the beam axis at the

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nearfield [31]. Hence, the transverse radiation force associated to the beam may not be strong enough to hold a particle in 3D. In this work, a method to form multiple microparticle traps in the pre-focal region of a piezoelectric focused transducer is proposed. The method's ability to tweeze microparticles is theoretically demonstrated through computational simulations. In so doing, the radiation force produced by a 1 MHz piezoelectric transducer, with focus at 50 mm and an f-number of 1.25, which is readily available commercially, is computed on a microdroplet, made of either benzene or peanut oil, suspended in water. For this transducer configuration, trapping points arise in the pre-focal zone. After obtaining the radiation force field, the dynamics of the microparticle trapping is simulated considering effects of gravity, buoyancy and Stokes' drag. The results show that microparticle entrapment occurs in points as close as one third of the transducer focal distance.

2. Ultrasound beamforming

Before calculating the acoustic radiation force exerted on a particle, we need to establish the ultrasound beamforming. Hence, consider a circular focused transducer, with aperture $2b$ and curvature radius z_0, mounted on a compliant baffled at $z = 0$ (see Fig. 1). The transducer is immersed in a inviscid fluid of density ρ_0 and speed of sound c_0 and is uniformly excited with a sinusoidal signal of angular frequency ω. Thus, the ultrasound beam produced by the transducer is a time-harmonic wave described by its pressure $p_{\text{in}}(\boldsymbol{r})e^{-i\omega t}$ and velocity of a fluid parcel $\boldsymbol{v}_{\text{in}}(\boldsymbol{r})e^{-i\omega t}$, both observed at the time t in the position $\boldsymbol{r}$ in a fixed coordinate system. Hereafter, the time-harmonic term $e^{-i\omega t}$ will suppressed for the sake simplicity. A useful pressure–velocity relation in first-order approximation is obtained from the momentum conservation equation as follows [32]

$$\boldsymbol{v}_{\text{in}} = -\frac{i}{\rho_0 c_0 k}\nabla p_{\text{in}}, \tag{1}$$

where $k = \omega/c_0$.

It is further assumed that the transducer concavity is fairly small and the wavelength is much smaller than its radius. Hence the ultrasound beam can be described in the paraxial approximation. In terms of the transducer parameter, these assumptions read [34]

$$\frac{1}{8N^2} \ll 1 \text{ and } 1 \ll kb \ll 128\pi N^3, \tag{2}$$

where $N = z_0/(2b)$ is the transducer f-number. The model discussed here may not be suitable for transducers with $N < 1$. On that account a different beamforming model might be considered [35].

By assuming that the transducer generates an axisymmetric paraxial beam, the pressure field can be expressed in cylindrical coordinates (ϱ, z) as

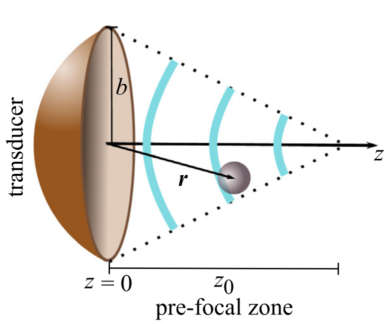

Fig. 1. Circular focused ultrasound transducer with aperture $2b$ and focus z_0 actuating on a spherical particle of radius a located at $\boldsymbol{r}$ in the medium.

$$p_{\text{in}}(\varrho, z) = -i\rho_0 c_0 k e^{ikz} q(\varrho, z), \tag{3}$$

where q is the velocity potential characteristic function. For a circular focused transducer mounted on a perfectly compliant baffle at $z = 0$, one can show that the characteristic velocity potential is given by [34]

$$q(\varrho, z) = \frac{v_0}{z}\exp\left(\frac{ik\varrho^2}{2z}\right)\int_0^b \exp\left[\frac{ik\varrho'^2}{2}\left(\frac{1}{z}-\frac{1}{z_0}\right)\right] J_0\left(\frac{k\varrho\varrho'}{z}\right)\varrho' d\varrho'. \tag{4}$$

where v_0 is the magnitude of the normal vibration velocity on the transducer surface, and J_m is the mth-order Bessel function. In the focal plane $z = z_0$, Eq. (4) becomes

$$q(\varrho, z_0) = -b v_0 \frac{J_1(kb\varrho/z_0)}{k\varrho}. \tag{5}$$

In the paraxial approximation, it follows from Eqs. (1) and (3) that the fluid velocity is

$$\boldsymbol{v}_{\text{in}} \approx -[(\partial_\varrho q)\boldsymbol{e}_\varrho + ikq\boldsymbol{e}_z]e^{ikz}, \tag{6}$$

where $\boldsymbol{e}_\varrho$ and $\boldsymbol{e}_z$ are the unit-vectors along the radial direction and the z-axis, respectively. Here we are using the shorthand notation $\partial_\varrho = \partial/\partial\varrho$ and $\partial_z = \partial/\partial z$. Note that in deriving Eq. (6), we have neglected the term $\partial_z q$, because $|\partial_z q| \ll k|q|$. This assumption is valid in a region not very near to the transducer surface, say $z > 0.3z_0$.

3. Radiation force on a Rayleigh particle

Our attention now turns to the acoustic radiation force exerted by the ultrasound beam on a particle much smaller than the wavelength, i.e. the so-called Rayleigh regime. The particle has radius a, density ρ_1, speed of sound c_1, and its position is denoted by $\boldsymbol{r}$. Viscous effects of the host fluid in the radiation force analysis are neglected. This hypothesis holds when the external viscous boundary layer of the particle [32] $\delta_0 = \sqrt{2\nu_0/\omega}$ is much smaller than the particle radius, where ν_0 is the kinematic viscosity of the host fluid.

Using the radiation force formulas in Cartesian coordinates given in Ref. [36], one can show that the acoustic radiation force $\boldsymbol{F}^{\text{rad}}$ on the Rayleigh particle is the sum of three components [37], namely the gradient $\boldsymbol{F}^{\text{grad}}$, the scattering $\boldsymbol{F}^{\text{sca}}$, and the absorption $\boldsymbol{F}^{\text{abs}}$ radiation forces. Thus, we have

$$\boldsymbol{F}^{\text{rad}} = \boldsymbol{F}^{\text{grad}} + \boldsymbol{F}^{\text{sca}} + \boldsymbol{F}^{\text{abs}}. \tag{7}$$

The gradient radiation force is given by [38]

$$\boldsymbol{F}^{\text{grad}}(\boldsymbol{r}) = -\nabla U^{\text{rad}}(\boldsymbol{r}), \tag{8}$$

where

$$U^{\text{rad}} = \pi a^3\left(f_0\frac{|p_{\text{in}}|^2}{3\rho_0 c_0^2} - f_1\frac{\rho_0|\boldsymbol{v}_{\text{in}}|^2}{2}\right) \tag{9}$$

is the radiation force potential energy. The quantities $f_0 = 1 - \rho_0 c_0^2/(\rho_1 c_1^2)$ and $f_1 = 2(\rho_1 - \rho_0)/(2\rho_1 + \rho_0)$ are the compressibility and density contrast factors of the particle, respectively. The gradient radiation force appears due to the interference between the incident and the scattered waves. Moreover, this force is responsible for trapping particles at the minima of the potential energy U^{rad}.

The scattering radiation force is given by [37]

$$\boldsymbol{F}^{\text{sca}}(\boldsymbol{r}) = \pi a^2(ka)^4\left[\frac{4}{9}\left(f_0^2 + f_0 f_1 + \frac{3f_1^2}{4}\right)\frac{\bar{\boldsymbol{I}}(\boldsymbol{r})}{c_0} - \frac{f_1^2}{6k}\text{Im}[\nabla\cdot\rho_0\boldsymbol{v}_{\text{in}}\boldsymbol{v}_{\text{in}}^*(\boldsymbol{r})]\right], \tag{10}$$

where $\bar{\boldsymbol{I}} = (1/2)\text{Re}[p_{\text{in}}\boldsymbol{v}_{\text{in}}^*]$ is the incident intensity averaged in time and $\rho_0\boldsymbol{v}_{\text{in}}\boldsymbol{v}_{\text{in}}^*$ is a dyadic (second-rank tensor).