Optical Manipulation of Micro / Nanoparticles Using Fiber-Based Optical Tweezers

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Abstract: Over forty years, the optical tweezers have been used to facilitate new scientific findings and benchmark single-molecule studies. One of the next steps, as far as applications is concerned, is expected to be their use as medical diagnostic tools, where in parallel, manipulation, sorting, and diagnosis of large number of biological particles, using the optical trapping technique will make the current clinical procedures quicker and cheaper. However, most of the current optical tweezers are based on objective lenses, which are bulky, expensive, and hard to integrate. Optical tweezers based on optical fibers have great potential to solve the abovementioned limitations. This paper attempts to provide solutions in order to overcome the limitations of current conventional optical tweezers with the objective of achieving fundamental understanding and improving the performance of fiber-based optical tweezers. Thus, we develop an optical fiber tweezers using a continuous wave diode GaAlAs laser operating at 659-nm for trapping dielectric nanoparticles (900-nm diameter and 3-μm diameter), suspended in deionized water. We systematically measure the optical trapping force and the effective trapping quality factor. We also investigate the dependence of the trapping force on both the insertion angle of the fiber into the sample chamber and the size of the trapped particle.

Keywords: Microparticles, nanoparticles, fiber-based optical tweezers, optical trapping efficiency, diagnostic tool, microfluidic chamber.

1. INTRODUCTION

Since first proposed by Ashkin [1], the optical tweezers showed a rapid progress in various fields of science and engineering [2-9]. Most of the fascinating results are achieved with conventional optical tweezers based on standard microscopes, by tightly focusing a laser trapping beam, using high-Numerical Aperture (high-NA) objective. Thus, specimens such as proteins [10], DNA molecules [11] and polystyrene particles [12] can be trapped and can be manipulated with high precision. However, most optical tweezers require high power microscope objectives and carefully adjusted optical path, thereby many device assembly restrictions are introduced, such as cost, bulk and hard to integrate set-up, sensitivity to environmental fluctuations, limitations in terms of working distances from the substrate, as well as strict requirements on the transparency of the substrate. Additionally, the optical tweezers are applied with difficulty in turbid biological media [13], since it is hard to attain the tightly focused laser beam that is necessary for optical trapping. These abovementioned limitations of conventional optical tweezers prevent them from being miniaturized.

With the advantages of easy fabrication and high flexibility, the development of fiber-based optical tweezers [14-17] can provide a solution for cost reduction and miniaturization. Moreover, the potential use of optical tweezers in microfluidic systems requires avoiding the use of high-NA objectives. Optical fiber trapping was presented for the first time in 1993 by Constable *et al.* [18]. In 1995 a pioneering system was described by Lyons and Sonek [19] which was based on lensed optical fiber tweezers. It consisted of two counter propagating beams originated from two fibers with spherical tips, which reinforced the tweezing effect [19]. Few years later, Taguchi *et al.* developed an optical fiber system and verified that the manipulation of microparticles was achieved by a focused laser beam emerging from an optical fiber, inserted into the sample chamber, at an angle of 35 degrees [20]. They also calibrated their system with the escape velocity method versus transverse offset, in a later work [21]. In 2005, Hu *et al.* [14, 22] used a similar system to demonstrate manipulation and arrangement of yeast cells. Experimental calibration of the trapping efficiency, using both the static and dynamic method, was carried out. In 2007, Abedin *et al.* [23] used a bismuth fiber, instead of a common silica fiber, with a tapered end to manipulate and rotate liquid crystal drops. Since then, more and more sophisticated optical fiber tweezers configurations have been reported exploring a range of implementations. However, the fiber optical tweezers are typically used to trap and manipulate micrometer-sized particles, but are faced with the difficulty to trap and manipulate sub-micron sized particles, due to the Brownian motion. Thus, in spite of great progress, challenges remain in fulfilling

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the combination of fiber optics versatility with more advanced trapping implementations and manipulation functions, in the context of biological applications [24].

We developed and optimized a simple optical fiber trapping system using a single optical fiber with core diameter 200-μm. The fiber trap was formed by using a continuous wave (CW) diode GaAlAs laser operating at 659-nm, with its visible beam guided to the sample from the bottom of the microscope stage. We used also a commercial microscope only to monitor the experimental procedure and to record the results with the CCD camera in video form. We verified that optical trapping and manipulation of micro and nano-objects was easily achieved in two-dimensions by a laser beam emerging from the optical fiber inclined at several angles into the sample. In addition, we investigated the improvement of the optical trapping force and the trapping quality factor with respect to different fiber insertion angles into the sample chamber, as well as with the various trapped bead diameters. Micro/nano-objects are usually immersed in the fluid of a human body in several biomedical applications, for example, in the treatment of diseases using drug delivery nanoparticles [25] or in diagnosis based on fluorescence particles for labeling molecules [26]. In all these circumstances, it is substantial to both trapping and monitoring the trapped micro and nano-objects through an optical fiber-based optical manipulation set-up, offering a bench to bedside approach, outside of the physics research lab. Thus, the manipulation of nano-objects in fluid with a non-contact and non-invasive way is of great importance for *in vivo* biomedical theranostics applications. Potential applications of the proposed fiber tweezers, for example, may be their use on sensing the interaction and communication between cells [27], especially inside an internal organ in the human body.

2. MATERIALS AND METHODS

As we mentioned in the Introduction, the optical fiber trap was formed using a temperature stabilized CW diode GaAlAs laser operating at 659-nm, with maximum output power of 60mW, pigtailed with a simple optical fiber having core diameter 200-μm. The configuration of the experimental set-up is shown in Figure **1**. Using a laser emitting at 659-nm allowed trapping of biological samples for a long period, without visible damage, due to the low absorption coefficient of the cells at this wavelength. The other end of the fiber was inserted into the sample chamber from the bottom side of the microscope stage, at several insertion angles (Figures **1b-c**). The manipulation process of the particles was observed and recorded by a conventional microscope equipped with a CCD camera, using an oil-immersion objective lens. Also, a filter in front of the CCD camera was used to block the scattering laser. The samples were Poly(methyl methacrylate) (PMMA) microspheres of mean diameter about 3-μm and polystyrene beads with 900-nm diameter which were suspended in deionized water.

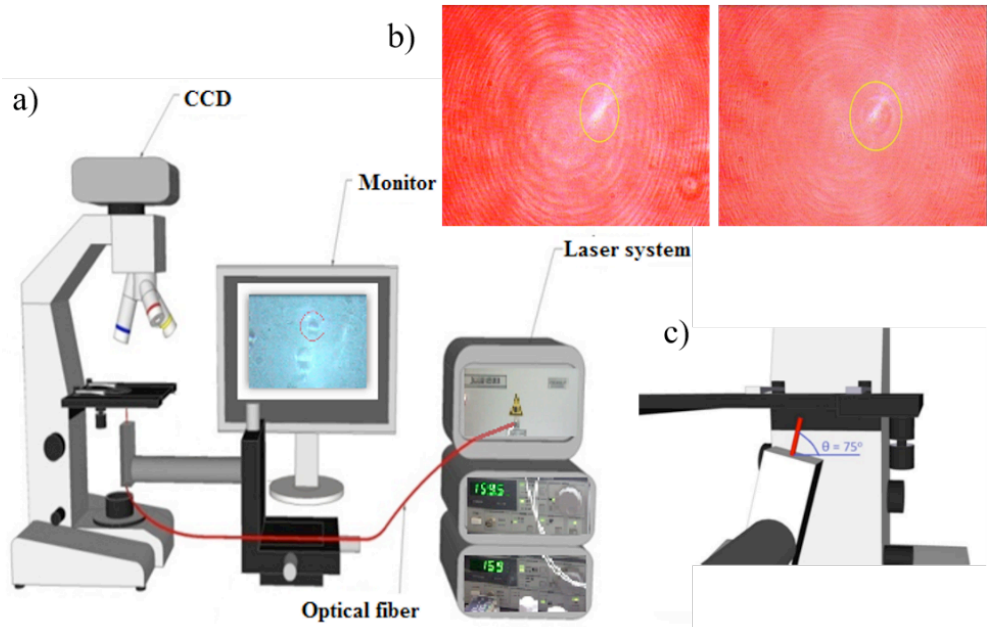

Figure 1: Experimental set up. (**a**) A schematic diagram of the fiber trapping system. (**b**) Some representative results: The beam entry was close to the cover glass and formed an aura, as can be seen in the left image. The right image shows the trapped particle by the optical forces exerted on a 3-μm PMMA bead. (**c**) A schematic diagram of the insertion angle measurements.

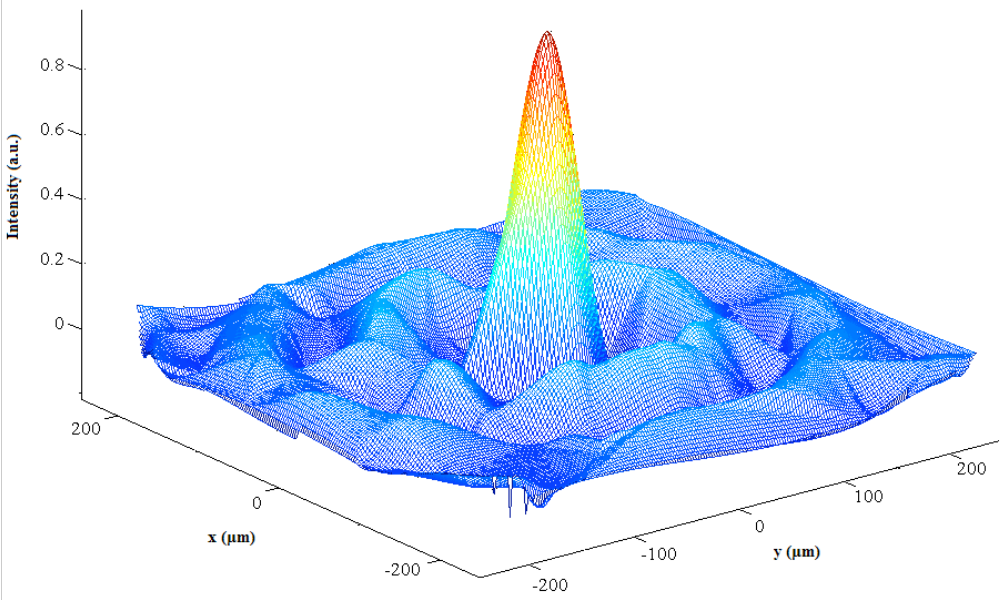

Figure 2: The theoretical output beam profile at the outlet of the optical fiber, which is used to create the optical fiber tweezers.

We evaluated the optical trapping efficiency, *Q* [1], of the fiber optical tweezers configuration employing the dielectrophoresis calibration method, which has been described in details in our previous works [28, 29]. Briefly, the specimens were placed on a microfluidic chamber, consisting of thin film electrodes, of thickness of 6000 Å, placed at a distance of 100-μm, fabricated by photolithography on the top of a microscope slide [28, 29]. Sine wave excitation of 1MHz frequency at voltage of 12 V was applied from a signal generator. Time periodic in-homogenous electric fields induced polarization and subsequent movement of the dielectric beads [28, 29]. In order to evaluate the optical trapping force and the trapping efficiency, we used the fiber optical tweezers to trap the specimens in the space between the two thin electrodes, when no voltage was applied. Then, by increasing the voltage, the specimens experienced a dielectrophoresis force, which could be calculated from the determined electric field, using analytical methods based on Green's theorem [28, 29].

We also simulated the output beam profile of the laser radiation through the optical fiber by the geometrical ray model proposed by Croitoru *et al.* [30]. The laser beam with a Gaussian profile was considered to be the input light source. As shown in Figure **2**, a small ring of higher order modes appears as the beam propagates through the optical fiber. These modes did not affect the optical trapping stability and allow trapping particles in two-dimensions.

3. RESULTS AND DISCUSSION

To demonstrate the trapping ability, 3-μm PMMA beads were suspended in deionized water and were trapped using the fiber optical tweezers, as shown in Figure **3**. When a bead was trapped, the cover glass could be moved freely along the *x* or *y* direction (Figure **3b-c**), and another bead could be moved without interfering with the trapped one. The same procedure was followed at various insertion angle degrees. Then, we evaluated the optical fiber tweezers for force sensing, by measuring the optical forces exerted on trapped beads. During optical trapping force

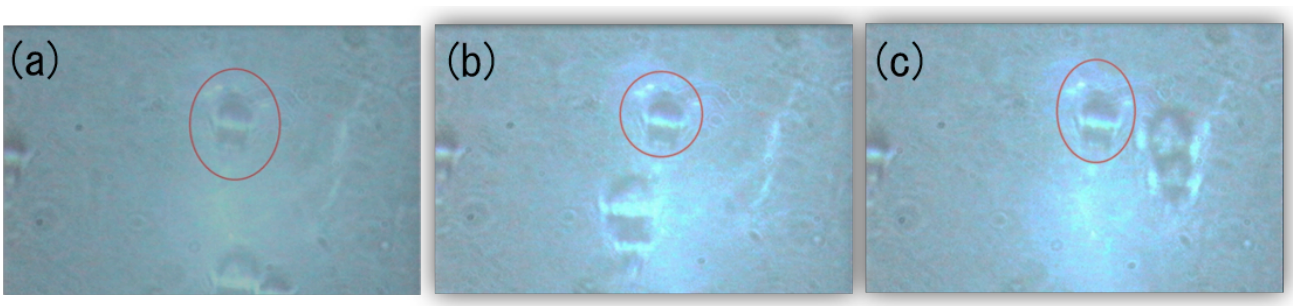

Figure 3: A series of single frames of trapped 3-μm PMMA bead using the fiber optical tweezers. The red cycle represents the position of the trapped bead. The optical fiber was inclined at 90 degrees to the sample chamber.

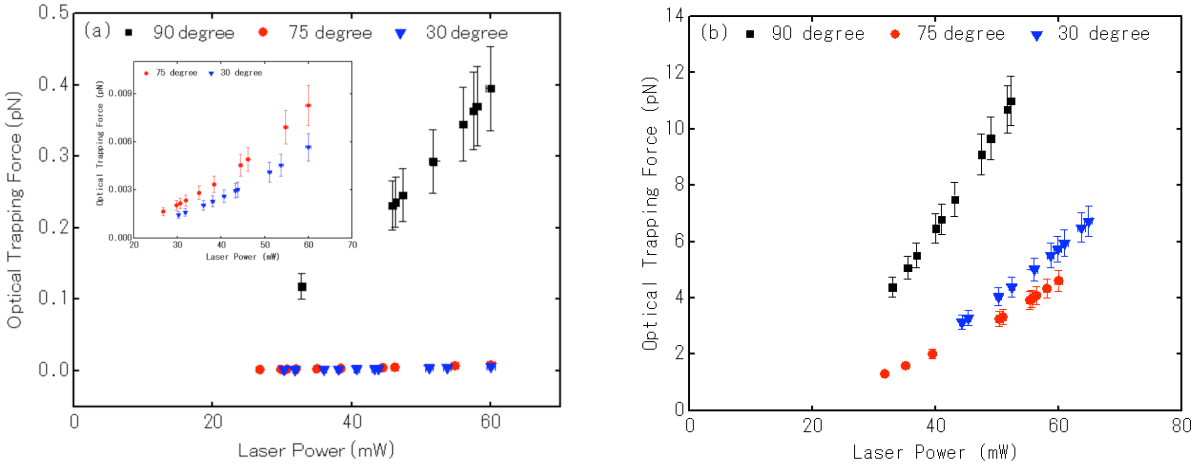

Figure 4: Optical trapping force as a function of the trapping laser power for, (**a**) 900-nm polystyrene beads and (**b**) 3-μm PMMA beads, for various insertion angle degrees into the sample chamber. Inset: the trapping force for insertion angle 75 and 30 degrees, for the 900-nm polystyrene beads. The x-error and the y-error correspond to the standard deviation of the trapping force measurement.

measurements, we systematically verified that the trapped beads were not stacked on the bottom of the cover glass, as they were released when switching off the trapping laser.

Figure **4** shows the optical trapping force exerted on trapped beads, for various insertion angles of the optical fiber into the sample chamber, on 900-nm polystyrene beads (Figure **4a**) and on 3-μm PMMA beads (Figure **4b**), as a function of the trapping laser power. In both cases, the optical trapping force increases linearly to the optical trapping laser power. Every single data point in Figure **4** resulted from ten independent measurements. The inset in Figure **4a** shows the trapping force for insertion angle 75 and 30 degrees, plotted on a finer scale. In Figure **4a**, we notice that, for angles lower than 90 degrees, the optical forces exerted on 900-nm polystyrene beads decreases abruptly. A plausible explanation for the optical trapping force decrease is that nano-sized particles intercept significantly less optical power and experience smaller optical trapping forces as the insertion angles decrease. In Figure **4b** the optical trapping force exerted on 3-μm PMMA beads shows less variation with the insertion angles than the abruptly behavior which is obtained in Figure **4a**. The comparison of Figures **4a** and **4b** indicates that lower optical trapping forces are obtained for 900-nm trapped beads, than the 3-μm PMMA beads. This indicates that the optical trapping force is highly dependent on both particle size and insertion angles. The maximum trapping force, obtained with the 900-nm polystyrene beads, is F = (0.395 ± 0.059) pN at an insertion angle of 90 degrees, for a trapping laser power of 59.7mW. For the same trapping laser power, the maximum optical trapping force, obtained with 3-μm PMMA beads, is F = (11.0 ± 0.88) pN at an insertion angle of 90 degrees. Therefore, as the trapped particle diameter increases by a factor of 3, the maximum optical trapping force increases by a factor of ~28, at an insertion angle of 90 degrees.

In order to compare the performance of both the different dimensions of trapped beads and the insertion angle employed in this work, we calculated the effective trapping quality factor, Q, according to the definition $Q = Fc/nP$ [1], where F is the optical trapping force, c is the speed of light, P is the trapping laser power, and n is the refractive index of the surrounding medium containing the trapped beads (n = 1.329 for deionized water at 659-nm [31]). The calculation is based on the slope of the trapping force *vs.* laser power plots, shown in Figures **4**. Table **1** shows the effective trapping quality factor obtained for each case. We observe that the effective quality factor for the 900-nm polystyrene beads is lower than the quality factor for 3-μm PMMA beads. Because a larger diameter's particle captures more optical laser power, it has a much larger optical trapping efficiency than the small ones. This verifies the dependence of the

Table 1: Trapping Quality Factor, *Q*, for Various Angles, in Ascending Order

Substrate	Quality Factor *Q*
3-μm PMMA beads – 90 degrees	0.0792 ± 0.0016
3-μm PMMA beads – 75 degrees	0.02706 ± 0.00078
3-μm PMMA beads – 30 degrees	0.03982 ± 0.00077
900-nm Polystrene beads – 90 degrees	(2.21 ± 0.64) x 10^{-4}
900-nm Polystyrene beads – 75 degrees	(0.401 ± 0.019) x 10^{-4}
900-nm Polystyrene beads – 30 degrees	(0.207 ± 0.012) x 10^{-4}

trapping quality with the bead diameter [1]. The quality factor obtained for the 3-μm PMMA beads at insertion angle of 90 degrees is higher by a factor of two from those obtained at insertion angles of 75 and 30 degrees. The quality factor for the 900-nm polystyrene beads obtained at insertion angle of 90 degrees is higher by a factor of five compared to insertion angle of 75 degrees and one order of magnitude higher than the quality factor at insertion angle of 30 degrees. Finally, we observe that the optical trapping efficiency values obtained for the optsical fiber tweezers are less than that of the conventional optical tweezers. For example, the trapping effective quality of conventional optical tweezers is Q~0.14, when is applied on polystyrene microsphere of 900-nm bead diameter, with a water immersion objective of NA = 1.25 [29]. Thus, the trapping ability of the fiber optical tweezers is weaker than that of a conventional optical tweezers in the case of nanometer-sized particles. However, for micrometer-sized particles, the effective optical trapping efficiency for both optical tweezers configuration is comparable [23].

Moreover, the experimental results also show that the beads can be trapped easily at larger insertion angles. On the contrary, at small insertion angles was difficult to trap the beads for a long period of time. The Eq. (6) in Ref. 22, indicates that when the insertion angle decreases, the transverse component of the optical force should increase to maintain equilibrium, which entails that the offset of the bead from the optical axis is possible [21, 22]. We applied the theoretical model proposed in Ref. 22 and we estimated the theoretical optical trapping efficiency, Q, for various insertion angles of the optical fiber into the sample chamber. Figure **5** shows that the theoretical trapping efficiency increases up to angles of 90 degrees and then gradually decreases at angles greater than these. Moreover, we notice that the quality trapping values are in agreement with the experimental values. Thus, it can be seen that the insertion angle is a crucial parameter in the single optical fiber trap systems and must be greater than a critical angle (which is less than 30 degrees in this experiment) in order to achieve stable trapping. Finally, we notice that if the output laser power is less than 30mW is hard to trap and manipulate nanometer-sized particles at small insertion angles, as the optical force exerted on particle was too small to overcome the disturbance of the environment.

CONCLUSIONS

In this work, experimental studies of an optical fiber trap has been carried out in order to obtain a fundamental understanding of the optical fiber tweezers system. The two-dimension trapping ability of the optical fiber tweezers has been demonstrated with both micrometer-sized and nanometer-sized particles. The fiber optical tweezers have been experimentally calibrated with the dielectrophoresis method. The influence of the particle size and the fiber inclination angles over the trapping performance has been investigated. We showed that the optical trapping forces exerted on nanometer-sized particles are more sensitive to the insertion fiber angles, and thus the changes in the optical trap forces become easier observable. Additionally, we also showed that the maximum quality factor is obtained at an insertion angle of 90 degrees for both particle sizes. This work paves the way for guiding several diagnostic and therapeutic procedures, especially inside an internal organ in the human body.

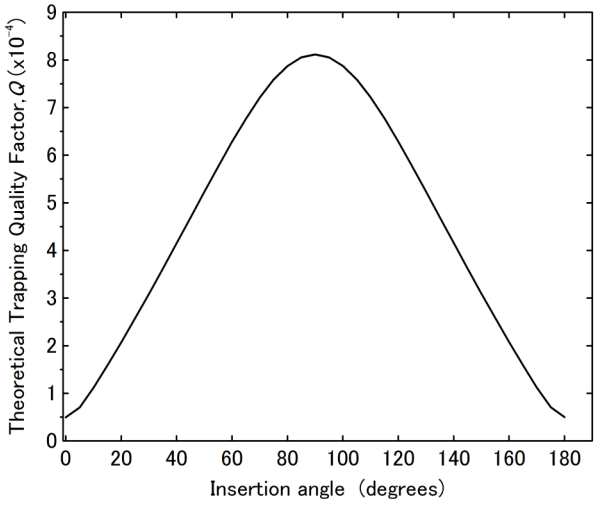

Figure 5: Theoretical trapping quality, *Q*, for various angle insertions.

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