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Optical Tweezers in Biotechnology

Zhiyong Gong and Yuchao Li

Abstract

Three-dimensional optical manipulation of microparticles, cells, and biomolecules in a noncontact and noninvasive manner is crucial for biophotonic, nanophotonic, and biomedical fields. Optical tweezers, as a standard optical manipulation technique, have some limitations in precise manipulation of micro-objects in microfluidics and in vivo because of their bulky lens system and limited penetration depth. Moreover, when applied for trapping nanoscale objects, especially with sizes smaller than 100 nm, the strength of optical tweezers becomes significantly weak due to the diffraction limit of light. The emerging near-field methods, such as plasmon tweezers and photonic crystal resonators, have enabled surpassing of the diffraction limit. However, these methods may lead to local heating effects that will damage the biological specimens and reduce the trapping stability. Furthermore, the available near-field techniques rely on complex nanostructures fixed on substrates, which are usually used for 2D manipulation. The optical tweezers are of great potential for the applications including nanostructure assembly, cancer cell sorting, targeted drug delivery, single-molecule studies, and biosensing.

Keywords: optical tweezers, optical force, optical manipulation, biophotonics, biotechnology

1. Introduction

As early as the 1970s, Maxwell revealed that electromagnetic waves can carry momentum in his famous electromagnetic field theory. When electromagnetic waves are applied to objects, they will transmit momentum. Therefore, electromagnetic waves can exert force on objects, and then the concept of electromagnetic force is proposed [1]. Until the beginning of the twentieth century, Einstein proposed the concept of photonic quantum, which believes that light is composed of a group of photons with both mass and momentum. When light is irradiated on the surface of the object, it can cause changes in the photon momentum to produce radiation pressure on the object due to the scattering and absorption of light. Subsequently, Lebedev, Nichol, and Hull first demonstrated the existence of radiation pressure experimentally. The experiment used arc lamps and torsion scales to observe the effect of light in the macroscopic physical world. However, the light produced by the arc lamp is very weak and difficult to practically apply. Until 1960, the invention of the laser provided a high-intensity optical source for studying optical force, which greatly promoted the application of optical manipulation. Arthur Ashkin, a scientist at Bell Experiments in the United States, first used the radiation pressure generated by the laser beam to push tiny particles in the liquid environment [2] and then used two opposing laser beams to capture microparticles and even atoms. However, the experimental setup used in the dual-beam capture method is too complex and can

only limit microparticles in a two-dimensional plane. Scientists hope to use a single laser to achieve three-dimensional trapping of microparticles. To this end, in 1986, Ashkin et al. used a high-numerical-aperture objective to focus a single laser to trap microparticles and named the technology “single beam gradient force trap” [3]. A year later, Ashkin et al. continued to improve this technology and achieved optical trapping and manipulation of tiny bacteria and viruses. They officially named the technology “optical tweezers” [3]. Compared with traditional macro-mechanical tweezers, the optical tweezers have the advantages of noncontact and no damage and can perform high-precision manipulation of microscopic particles. Therefore, since the birth of the optical tweezer technology, it has played an important role in the fields of biomedicine and physical chemistry.

1.1 Traditional optical tweezers

1.1.1 Basic principles

The core component of the traditional optical tweezers is a highly focused beam, as shown in **Figure 1a** [4]. When the incident laser (usually a near-infrared laser with a wavelength of 1064 nm) is focused by a high-numerical-aperture objective lens, the microparticles in the liquid environment will be exposed to optical force near the focus. This force is derived from the momentum transfer effect between light and particles. Specifically, the optical forces are divided into two components: one component along the direction of the optical gradient, called the optical gradient force, which is caused by the microparticles being in a nonuniform optical field, and the optical gradient force, which drives the particles to the area where the optical intensity is greatest; another component along the direction of optical propagation, called optical scattering force, is caused by the scattering and absorption of particles, and the optical scattering force causes the microparticles to move along the direction of optical propagation. By modulating the focused beam, the magnitude of two forces can be varied to achieve different functions such as capture, acceleration, and rotation of the microparticles. For traditional optical tweezers to construct a stable trap, it is necessary to focus the incident laser with a high-numerical-aperture (generally $NA = 1.0\sim1.4$) objective lens. The resulting

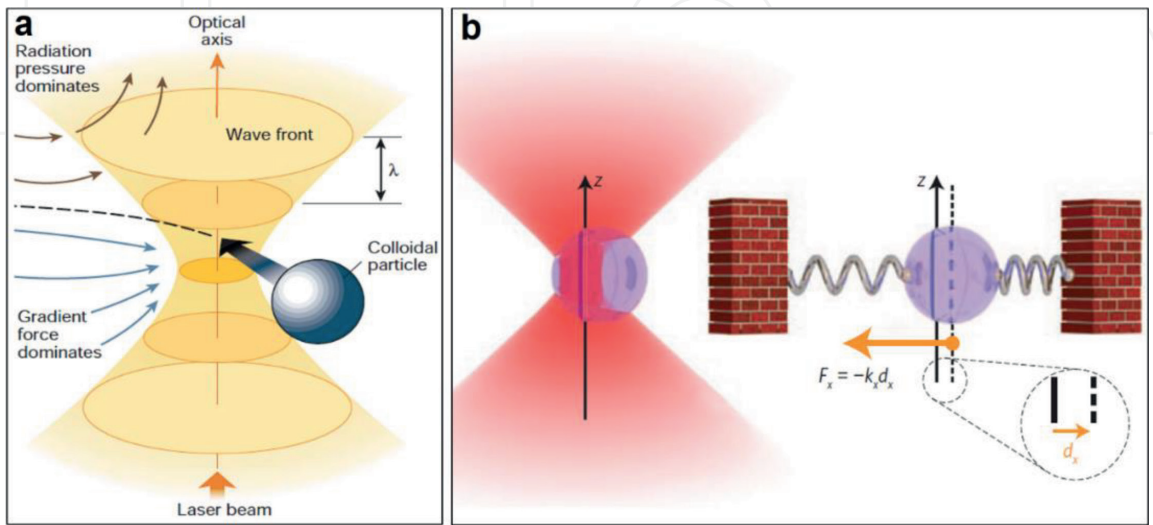


Figure 1. Schematic diagram of the traditional optical tweezers. (a) A single microparticle is trapped to the focused spot of a laser beam by gradient force and scattering force to [4]. (b) A simple harmonic oscillator model for the optical trapping of the microparticle.

optical gradient force is greater than the optical scattering force, so the microparticles or the cells can be stably trapped in the focus of light [5].

The model in which the object is trapped by the optical tweezers can be equivalent to a simple oscillator, as shown in **Figure 1b**. The magnitude of the object's received optical force (F) is proportional to the object's distance from the focus (d), which is

$$F = -kd \quad (1)$$

where the constant k represents the spring constant of the spring oscillator and the strength of the trap. Therefore, when we know the motion of an object in a trap, the magnitude of the optical force can be calculated by Formula 1. However, in the more general case, we want to quantitatively analyze the optical force when the unknown object motions and then other optical theories are needed. The theoretical analysis of optical tweezers needs to be determined according to the size of the object, specifically divided into three cases: first, when the radius (R) of the particle is much larger than the wavelength (λ) of the incident light, then a simple geometric optical method can be used to analyze the force of the object; second, if the size of the particle is much smaller than the wavelength of the incident light, the particle can be equivalent to the dipole in the electric field, and a dipole approximation model is needed; and third, if the size of the particles is close to the wavelength of the light, the situation becomes complicated, and the Maxwell equation is needed to solve the problem.

We first analyze the Rayleigh nanoparticle ($R \ll \lambda$). At this time, the nanoparticle can be regarded as a dipole in a nonuniform electromagnetic field, and the optical gradient force (F_{grad}) of the dipole in the electromagnetic field can be expressed as

$$F_{grad} = \frac{1}{2} |\alpha| \nabla \langle E^2 \rangle, \quad (2)$$

where α is the polarizability of the dipole, E is the electric field, parentheses indicate the time average, and $|E|^2$ is proportional to the intensity of the electromagnetic field. It can be seen from Formula 2 that the direction of the optical gradient force F_{grad} is along the direction of the optical intensity gradient. Thus, for a highly concentrated beam, the particles are drawn to the focus of the spot. Here, the polarizability α is a crucial parameter that directly determines the intensity of the interaction of light with object. For spherical nanoparticles, α can be expressed as [6]

$$\alpha = \alpha_0 \left(1 - \frac{ik^3 \alpha_0}{6\pi\epsilon_0} \right)^{-1}, \quad (3)$$

where $k = 2\pi n/\lambda$ is the scalar of the incident light wave vector, ϵ is the dielectric constant of the particle, ϵ_0 is the dielectric constant of the vacuum, α_0 is the quasi-static polarizability of the nanoparticle, and α_0 can be given by the Clausius-Mossotti relation [6]:

$$\alpha_0 = 4\pi\epsilon_0 R^3 \frac{\epsilon - 1}{\epsilon + 2}, \quad (4)$$

The radiation pressure (F_{rad}) is produced by the scattering and absorption of light by the surface of the particles, which can be expressed as [7]

$$F_{rad} = \frac{n \langle P \rangle}{c} \sigma, \quad (5)$$

where n is the refractive index of the surrounding environment, c is the speed of light in the vacuum, and $\langle \mathbf{P} \rangle$ is the time-averaged Poynting vector, which can be expressed as

$$\langle \mathbf{P} \rangle = \frac{1}{2} \text{Re}(\mathbf{E} \times \mathbf{H}^*). \quad (6)$$

The σ in Formula 6 reflects the characteristics of the nanoparticle, which indicates the extinction cross section of the nanoparticle, including the scattering cross section (σ_{scat}) and the absorption cross section (σ_{abs}), and σ is determined by the following formula [8]:

$$\sigma = \sigma_{\text{scat}} + \sigma_{\text{abs}} = \frac{k^4 |\alpha|^2}{4\pi} + k \alpha'', \quad (7)$$

where α'' is the imaginary part of the particle polarizability α , which represents the absorption of light by the particles. For transparent media particles, this term is approximately equal to zero and can be ignored. It can be seen from Formula 5 that the direction of the optical scattering force coincides with the direction of the glass booth vector, that is, the direction in which the optical scattering force propagates along the light. When $\mathbf{F}_{\text{grad}} > \mathbf{F}_{\text{rad}}$, the trapping of particles can be achieved.

The dipole approximation model is only applicable to spherical nanoparticles. When the shape of the captured object is irregular or the size is the same magnitude as the wavelength, it needs to be solved from the most basic Maxwell equations using simulation software. This method is based on the Maxwell stress tensor integral of the surface S of the object, as defined below:

$$\mathbf{F}_O = \oint_S (\langle \mathbf{T}_M \rangle \cdot \mathbf{n}) dS, \quad (8)$$

where \mathbf{n} is the normal vector of the surface of the object and $\langle \mathbf{T}_M \rangle$ is the time-averaged Maxwell stress tensor. The expression is

$$\langle \mathbf{T}_M \rangle = \frac{1}{2} \text{Re} \left[\epsilon \mathbf{E} \mathbf{E}^* + \mu \mathbf{H} \mathbf{H}^* - \frac{1}{2} (\epsilon |\mathbf{E}|^2 + \mu |\mathbf{H}|^2) \mathbf{I} \right], \quad (9)$$

where \mathbf{E} and \mathbf{H} are the electric field vector and the magnetic flux vector in the electromagnetic field, \mathbf{E}^* and \mathbf{H}^* are complex conjugates, \mathbf{I} is an isotropic tensor, and ϵ and μ represent the dielectric constant and magnetic permeability, respectively. After calculating the optical force, the torque of the object can also be calculated by the following formula:

$$\mathbf{T} = \int \mathbf{r}_p \times d\mathbf{F}_p, \quad (10)$$

where $d\mathbf{F}_p$ represents the unit force at the point of action p and \mathbf{r}_p is the position vector from the center of the object to the point of action p .

1.1.2 Applications of the optical tweezers

Professor Ashkin, the pioneer of optical tweezers, predicted that optical tweezers as the manipulation technology of tiny particles will be widely used in the research of molecular biology, cell biology, and mesoscopic physics, especially to promote the development of many interdisciplinary subjects [9]. As an example, we will introduce some of the applications of the optical tweezers in the following aspects:

1.1.2.1 Capture, separation, and assembly of microparticles and cells

The invention of optical tweezers was used to capture and manipulate tiny particles such as polystyrene microspheres, biological cells, viruses, and bacteria [12]. By capturing these tiny particles, the Brownian motion of particles can be overcome and fixed in the field of the microscope for the researcher to observe and detect. When the particles are stably captured, they can be moved to a specific position and arranged in a regular pattern, which is applied to the ordered assembly of particles and cell arrays (as shown in **Figure 2a**), giving it a specific function. Further, by measuring the mechanical properties of particles and cell array, the interaction between the particles or cells can be studied. In addition, since different types of particles and cells are affected by the magnitude and direction of optical force, separation and screening of particles and cells can be achieved. With the maturity of optical tweezer technology, the system of optical tweezers is gradually combined with Raman technology, fluorescence technology [13], confocal technology, and femtosecond laser technology and achieves real-time detection of captured targets, which will enrich the applications of optical tweezers in cell biology and colloidal physics.

1.1.2.2 Study of optical tweezers and single molecules

The optical technology has a high mechanical resolution (10^{-12} – 10^{-15} N), which is sufficient for the study of individual biomacromolecules. For example, the basic laws of life movement are explained by measuring the physical forces such as the tiny force of biological single molecule and the motion step size. Optical tweezer technology has become an indispensable tool for quantitatively studying life processes and transforming life activities. Since the diameter of biomolecules is generally between 1 and 10 nanometers, the optical tweezer system cannot directly observe and manipulate. In order to see a single molecule, it is necessary to combine fluorescence imaging technology; in order to manipulate a single molecule, it is necessary to connect the molecule to the microsphere and indirectly manipulate and measure by using the small microsphere as the “handle” of the manipulation. For example, the

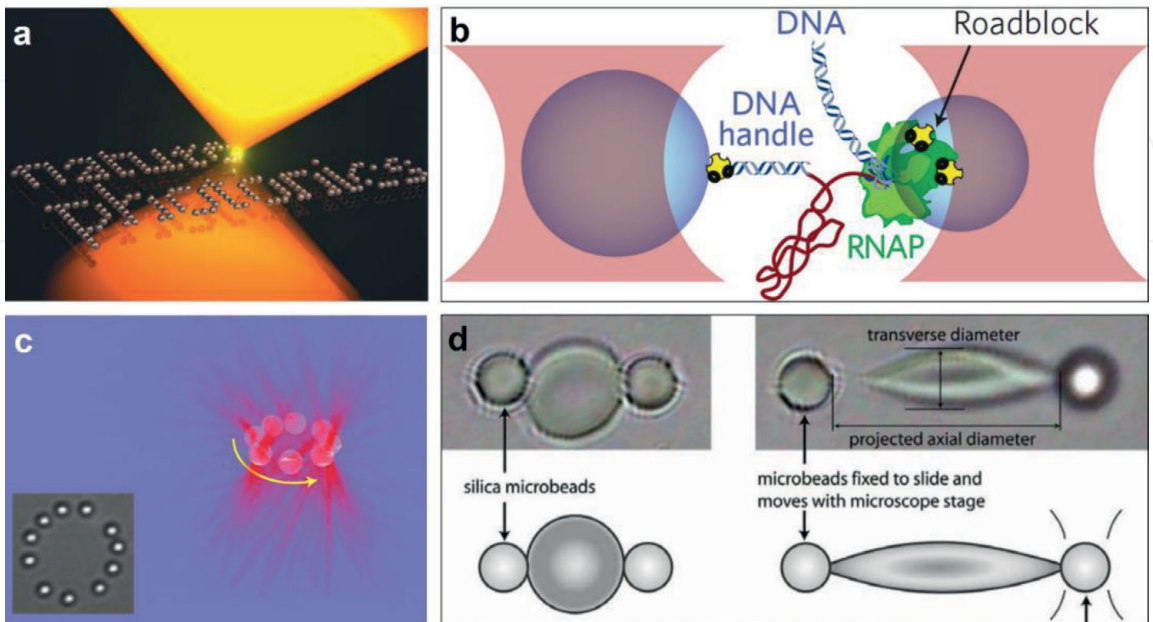


Figure 2. Several application examples of traditional optical tweezers. (a) Order and assemble microparticles and cells. (b) Study the interaction of nucleic acid molecules using micron media balls as handles [10]. (c) Rotating the microspheres using a vortex beam [11]. (d) Stretching human red blood cells using a micron media ball as a handle.

two ends of the DNA molecular chain are, respectively, connected to two microspheres, and the microspheres are manipulated by a double-beam tweezers to stretch the DNA molecular chain and measure its elastic properties (as shown in **Figure 2b**) [10]. By rotating the two microspheres in the opposite direction, the binding force of the DNA molecular chain can be calculated. Using similar methods, researchers can also study the properties of various biomacromolecules: RNA transcription, kinesin movement, the role of polymerases, etc. These are the basic processes of life activities. Its high-precision measurement can reveal the basic laws of life activities and lay the foundation for the research and application of biomedicine.

1.1.2.3 Optical rotator

The optical rotator is a branch of the optical tweezers that not only captures the microparticles but also allows the angular rotation of the microparticles as shown in **Figure 2c** [11]. This technique is based on the moment applied by the angular momentum of the light to the object. In order to achieve the rotation of the particles, the optical rotator requires a special beam of angular momentum, such as a Laguerre-Gauss beam [14]. Rotating particles or cells are used in many fields, such as rotating a tiny mechanical motor in a liquid environment to control the movement of local water flow. In addition, by rotating living cells, it can be imaged at various angles, which is beneficial to observe the full three-dimensional appearance of cells.

1.1.2.4 Optical stretchers

Stretching cells can study the elasticity of cell membranes, and the elasticity of cell membranes is closely related to many cellular diseases and can be used to reflect the activity of cells and even the health of the human body. There are many optical stretching methods based on optical tweezers, such as direct stretching of double-beam tweezers, stretching by microsphere handle, time-division multiplexed stretching, and so on. The method based on the microsphere handle-stretching method is more commonly used because of the high measurement precision. The method is shown in **Figure 2d**: two microspheres are adhered to the cell surface by chemical coupling, and then the microspheres are controlled to move in opposite directions by the tweezers. At this time, the cell membrane is stretched by shearing force. By recording the shape variables of the cells and measuring the force of stretching the microspheres, physical parameters such as the elastic modulus of the cell membrane can be calculated.

1.2 Holographic optical tweezers

1.2.1 Basic principles

Traditional optical tweezers based on single beam can only capture and manipulate one or a few particles at a time. However, researchers want to improve the efficiency of capture, such as controlling multiple particles at the same time. Based on this goal, scientists invented holographic optical tweezers. The core component of holographic optical tweezers is a hologram element: an interference pattern formed by recording the object light and reference light through the film. The wave front can be adjusted by holographic elements to construct a light field with a specific function. The holographic optical tweezers were firstly invented in 1998 by Professor Grier of the University of Chicago and his collaborators [15]. They used a holographic element (diffraction grating) to split the collimated single laser beam

into multiple independent beams, and then an array of grating is formed by focusing the lens to capture a large number of microparticles. The earliest holographic elements were prepared by coherent-optical interferometry, but the holographic elements obtained by this method have low diffraction efficiency and poor versatility, and thus this method has not been widely used. In order to improve diffraction efficiency and applicability, conventional holographic elements are often composed of spatial light modulators. The spatial light modulators include liquid crystal spatial light modulators, acousto-optic modulators, and digital microlens arrays. The spatial light modulator is controlled by a computer, and each focused beam can be individually controlled by changing the hologram element so that the formed trap well can be dynamically changed. Such holographic optical tweezers not only capture a plurality of microparticles at the same time but also control the movement of each microparticle to be arranged in different shapes, thereby achieving ordered assembly of the microparticles.

1.2.2 Applications of the holographic optical tweezers

As an emerging optical technology, holographic optical tweezers can trap and manipulate a large number of particles, showing great application prospects in the fields of particle assembly and construction of three-dimensional cell microstructure (**Figure 3**). For example, Glen R. Kirkham et al. of the United Kingdom used holographic optical tweezers to assemble one-, two-, and three-dimensional embryonic stem cell array structures (as shown in **Figure 4**) to provide a new means to study the directed differentiation of stem cells [16]. Moreover, Jesacher and his colleagues from Austria regulated the amplitude and phase of the incident light field through a liquid crystal spatial light modulator, which not only realized trapping potential wells of special shapes such as line, cross, circle, and rectangle but also controlled the microparticle movement along a specific path. In addition, holographic optical tweezer technology can also produce beams with special modes, such as Bessel beams, Laguerre-Gauss beams, and Airy beams [18]. These special-mode beams have peculiar phase distribution and propagation characteristics and can generate trapped potential wells with special functions, such as rotating particles with a Laguerre-Gauss beam, which can be used to construct micro- and nano-motors and study the transfer of orbital angular momentum; Airy beam or Bessel beam can be used to transport particles for sorting different types of particles and cells.

1.3 Fiber-based optical tweezers

1.3.1 Basic principles

Due to the low integration of conventional optical tweezer systems, it is difficult to manipulate particles located in a narrow position, such as particles inside a microfluidic channel or red blood cells in a blood vessel. The newly developed fiber-based optical tweezers are promising candidates because of its compact structure and flexible operation, which can overcome the problems of traditional optical tweezers [19]. Fiber-based optical tweezers use the output light from the end face of the fiber to achieve particle capture and manipulation, as shown in **Figure 5a**. When the laser beam passes into the fiber, it converges through the end of fiber and form a highly focused beam. The microparticles located near the tip of the fiber will be captured by the longitudinal gradient force onto the optical axis of the fiber and then captured by the lateral gradient force at the focus of the emitted light or move along the optical axis under the action of optical scattering force. For

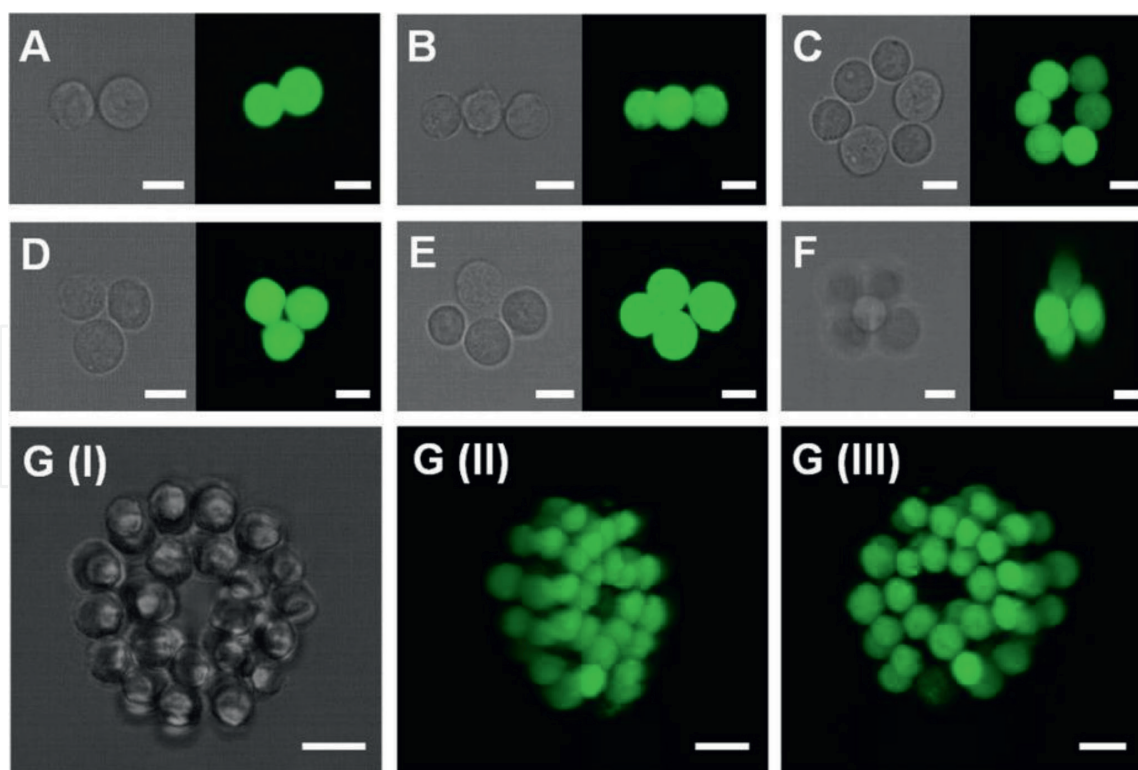


Figure 3.

Bright-field optical micrographs and confocal fluorescence micrographs of one-, two-, and three-dimensional microarray structures of embryonic stem cells assembled by holographic optical tweezers [16].

fiber-based optical tweezers, the distribution of the exiting light field depends on the shape of the fiber tip, which is a highly focused beam, to create a three-dimensional trapping potential. Currently, the tip of the fiber-based optical tweezers is generally designed as a parabolic, spherical, or conical structure. Different shapes of fiber tip can be prepared by physical polishing, heating stretching, chemical etching, and femtosecond laser processing. By changing the physical parameters of the preparation method, such as temperature, speed, time, etc., the shape and size of the fiber tip can be controlled to achieve different functions. **Figure 5b** shows the output light field distribution of a typical tapered fiber. It can be seen that the light is concentrated at the front end of fiber so that the cells can be trapped on the axis of the front end of the fiber and arranged into an ordered structure, as shown in **Figure 5c** [17].

1.3.2 Application of fiber-based optical tweezers

Since the fiber-based optical tweezers have the advantages of simple fabrication, flexible operation, compact structure, and easy integration, it has applications in many fields. For example, Xin et al. used a flame heating and melting taper to prepare a fiber-based optical tweezers with a tapered tip, which enables the capture of submicron-sized polystyrene particles and *E. coli* cells [20, 21]. Xu et al. realized the rotation of single silver nanowires using two tapered fibers, which provide a controlled and optical method for assembling plasmonic nanostructures [22]. Fiber-based optical tweezers will be developed in the direction of high integration and multifunctionality to adapt to lab-on-a-chip and in vivo requirements. In the future, the fiber-based optical tweezers may integrate multiple functions on a single-fiber probe, as shown in **Figure 6**, such as simultaneously capturing, transporting, sorting, stretching, deforming, and rotating various cells and pathogens in the microfluidics or living blood.

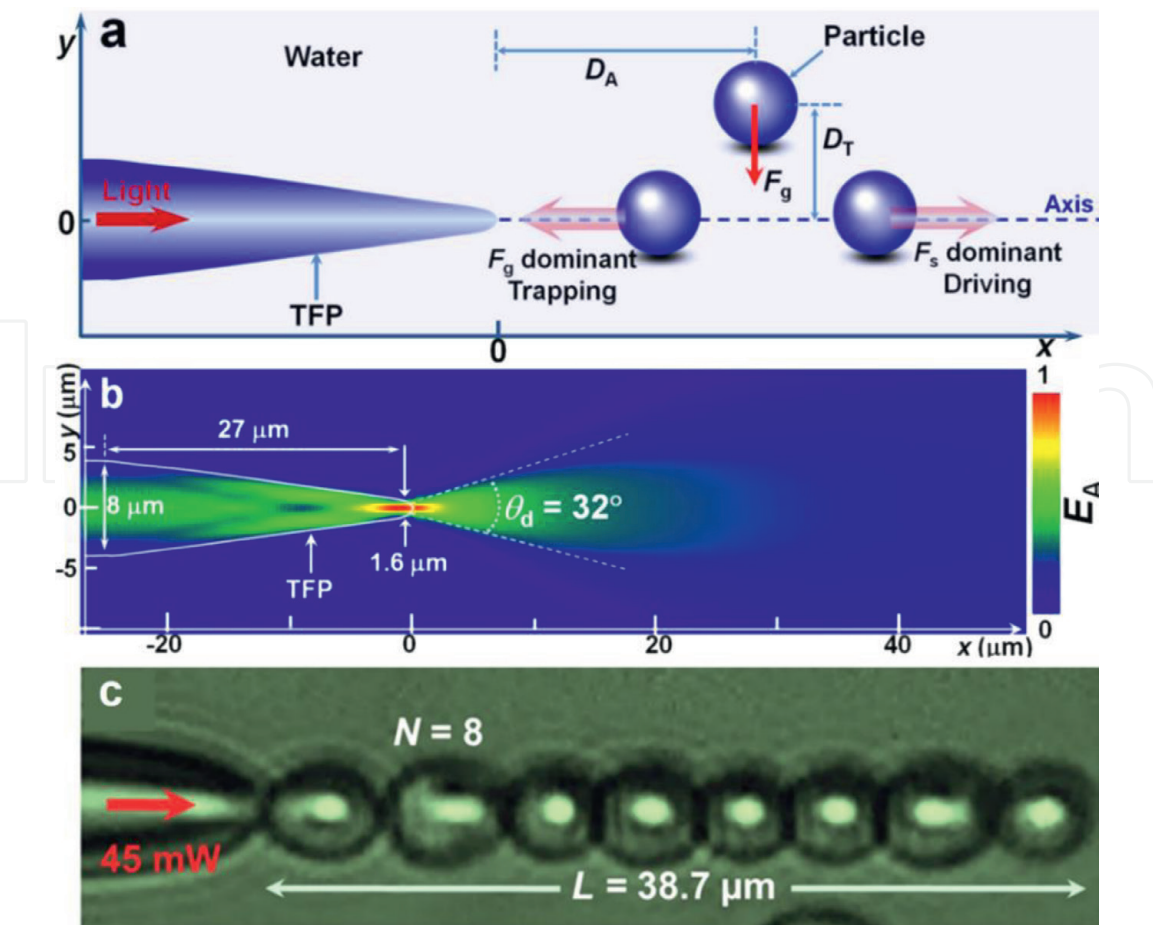


Figure 4.
The basic principle of the fiber-based optical tweezers. (a) Schematic diagram of the optical gradient force (F_g) and scattering force (F_s) applied to the microparticles by the fiber-based optical tweezers. (b) Simulation of electric field intensity distribution of the fiber-based optical tweezers. (c) A chain of yeast cells was trapped by the fiber-based optical tweezers [17].

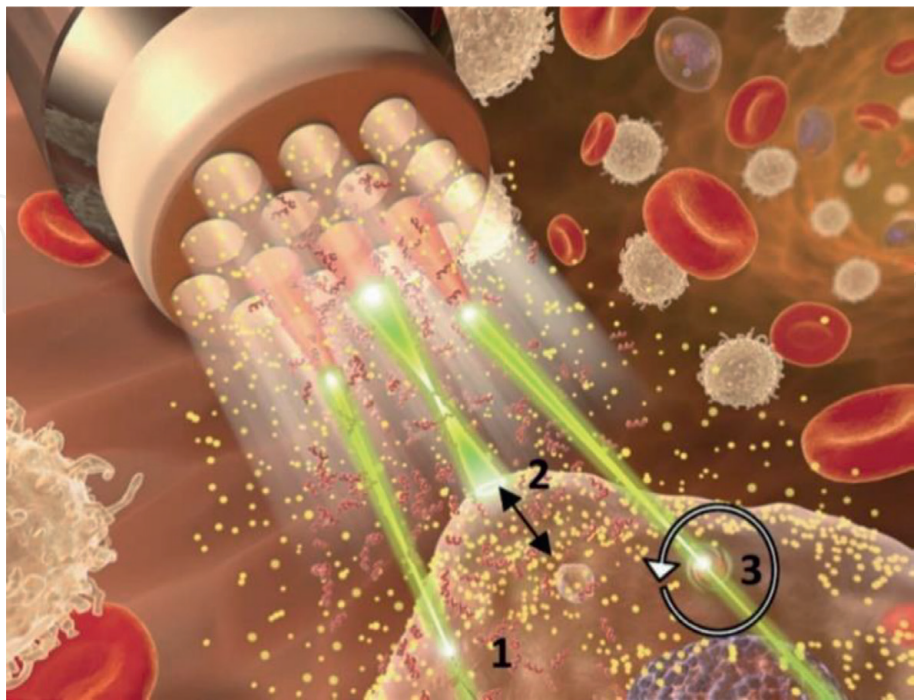


Figure 5.
This schematic shows a versatile fiber-based optical tweezers: number 1 indicates the capture, transport, and sorting of cells, number 2 indicates the optical stretching and deformation of cells, and numbers 3 indicates the optical rotation of cells.

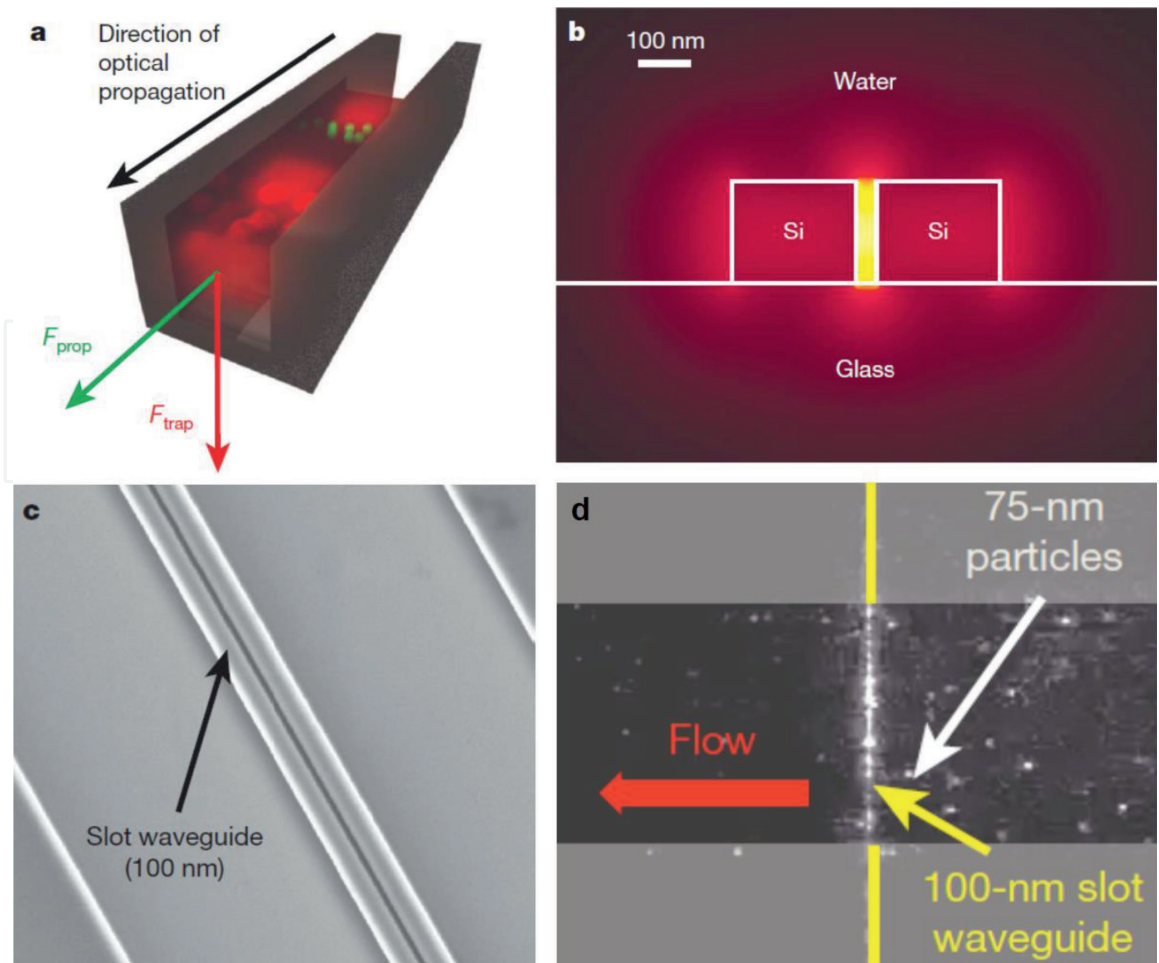


Figure 6.

Slot waveguide optical tweezers. (a) Schematic diagram of the optical gradient force and scattering force of nanoparticles in the slot waveguide. (b) A simulation result of the light intensity distribution of the slot waveguide in an aqueous environment. (c) An electron scanning micrograph of a waveguide having a slit of 100 nm. (d) The slot waveguide captures a large number of polystyrene particles of 75 nm in diameter in the water flow [23].

1.4 Nano-optical tweezers

1.4.1 Planar waveguide optical tweezers

When the light is transmitted in the waveguide, an evanescent wave is generated on the surface of the waveguide due to the total reflection. The evanescent wave is limited to a near-field range of 100 nanometers from the surface of the waveguide. When a nanoparticle enters the evanescent wave, the gradient of the light intensity changes greatly in the direction perpendicular to the waveguide, so the nanoparticles will be trapped on the surface of the waveguide by a strong optical gradient force. In the direction of light propagation, the evanescent wave can be considered to be uniformly distributed. Therefore, there is no optical gravity force in this direction. Only the optical scattering force exists. The nanoparticles move along the direction of light propagation due to the optical scattering force. Therefore, planar waveguide optical tweezers are often used for the transport of nanoparticles. Moreover, since the optical waveguide device is easily integrated into the microfluidic chip, the planar waveguide optical tweezers play an important role in the field of microfluidics. Current planar waveguide optical tweezers can be classified into three types: rectangular waveguide optical tweezers, slot waveguide optical tweezers, and nanofiber waveguide optical tweezers.

The manipulation of microparticles by a rectangular waveguide optical tweezers was first implemented by Kawata et al. [5]. They use rectangular waveguides to perform noncontact optical transport of different sizes of microparticles. This method can deliver cells or drugs over long distances. After this groundbreaking work, more and more researchers have entered this field and designed rectangular waveguides with different structures for transporting metal particles, media particles, microbial cells, etc. [5].

Since the evanescent wave of the rectangular waveguide has limited light confinement, it is challenging for the rectangular waveguide to capture particles and biomolecules below 100 nm. To solve this limitation, the researchers developed slot waveguide nanotweezers [23]. The slot waveguide is an air slit having a width of nanometers by photolithography or electron beam etching. The large refractive index contrast between low refractive index slot and high refractive index waveguide material makes the light energy highly confined in the slot region, which produces a strong optical gradient force and scattering force on the nanoparticles entering the slot. Using this property, Yang et al. achieved capture and transport of polystyrene particles and DNA molecules with sizes below 100 nanometers (as shown in **Figure 7**) [23].

A common problem with rectangular waveguide optical tweezers and slot waveguide optical tweezers is that they must be fixed on the substrate, making it difficult to operate. The emerging nanofiber waveguide optical tweezers can solve this problem. Li et al. used fibers with a diameter of 500–700 nm to achieve stable trap, bidirectional transport, optical separation, and controlled release of nanoparticles and micro-pathogens in microfluidics [26, 27]. The nanofiber waveguide optical tweezers have the advantages of low cost, production, and large control range and have important research value and application prospects in cell transportation, drug delivery, and particle collection.

1.4.2 Photonic crystal optical tweezers

Optical tweezers based on rectangular waveguides, slot waveguides, and nanofiber waveguides can only move particles along the waveguide surface but cannot be used to stably trap nanoparticles. In order to stably capture the nanoparticles, a photonic crystal optical tweezers were developed. The photonic crystal optical tweezers are based on one- or two-dimensional photonic crystal resonator structures (as shown in **Figure 8**) [24, 25]. When the laser that satisfies the wavelength matching condition is coupled into the photonic crystal resonator, static

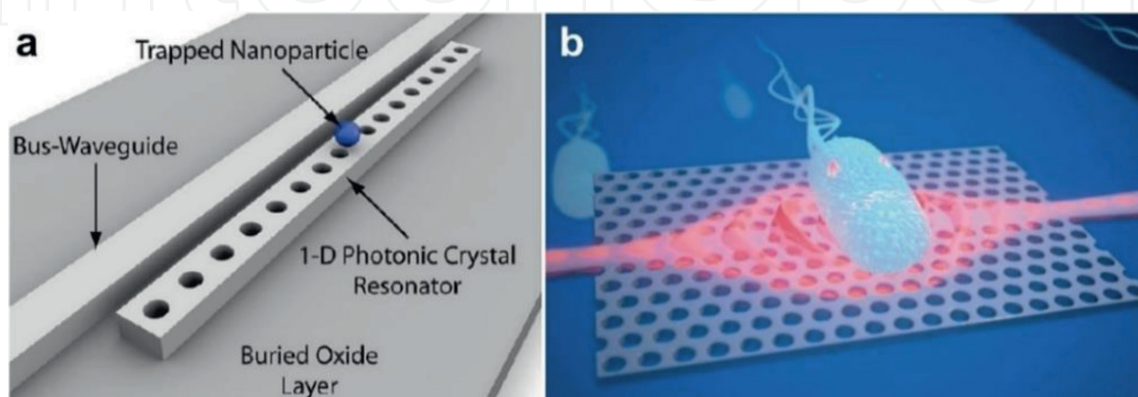


Figure 7. Photonic crystal optical tweezers. (a) Schematic representation of a single-dimensional photonic crystal resonator capturing a single nanoparticle [24]. (b) Schematic representation of a two-dimensional photonic crystal resonator capturing a single *E. coli* [25].

interference will occur in the cavity. With the resonance effect, the intensity of the light is greatly enhanced, and the size of the light spot is strongly suppressed, thereby enhancing the optical force of nanoparticles. Based on this principle, Erickson and Mandal et al. achieved stable capture and controlled release of nano-objects such as polystyrene particles, semiconductor quantum dots, and serum protein molecules in a liquid environment [30]. In addition, this method can also be used to study the angular rotation of silver nanowires or carbon nanotubes [31].

1.4.3 Plasmon optical tweezers

Plasmon is a near-field electromagnetic wave formed by the resonance of free electrons on a metal surface and incident photons. Under such resonance conditions, the energy of the electromagnetic field will be converted into the collective vibrational energy of the free electrons on the metal surface, thereby forming a special electromagnetic field: the light is confined to the sub-wavelength of the

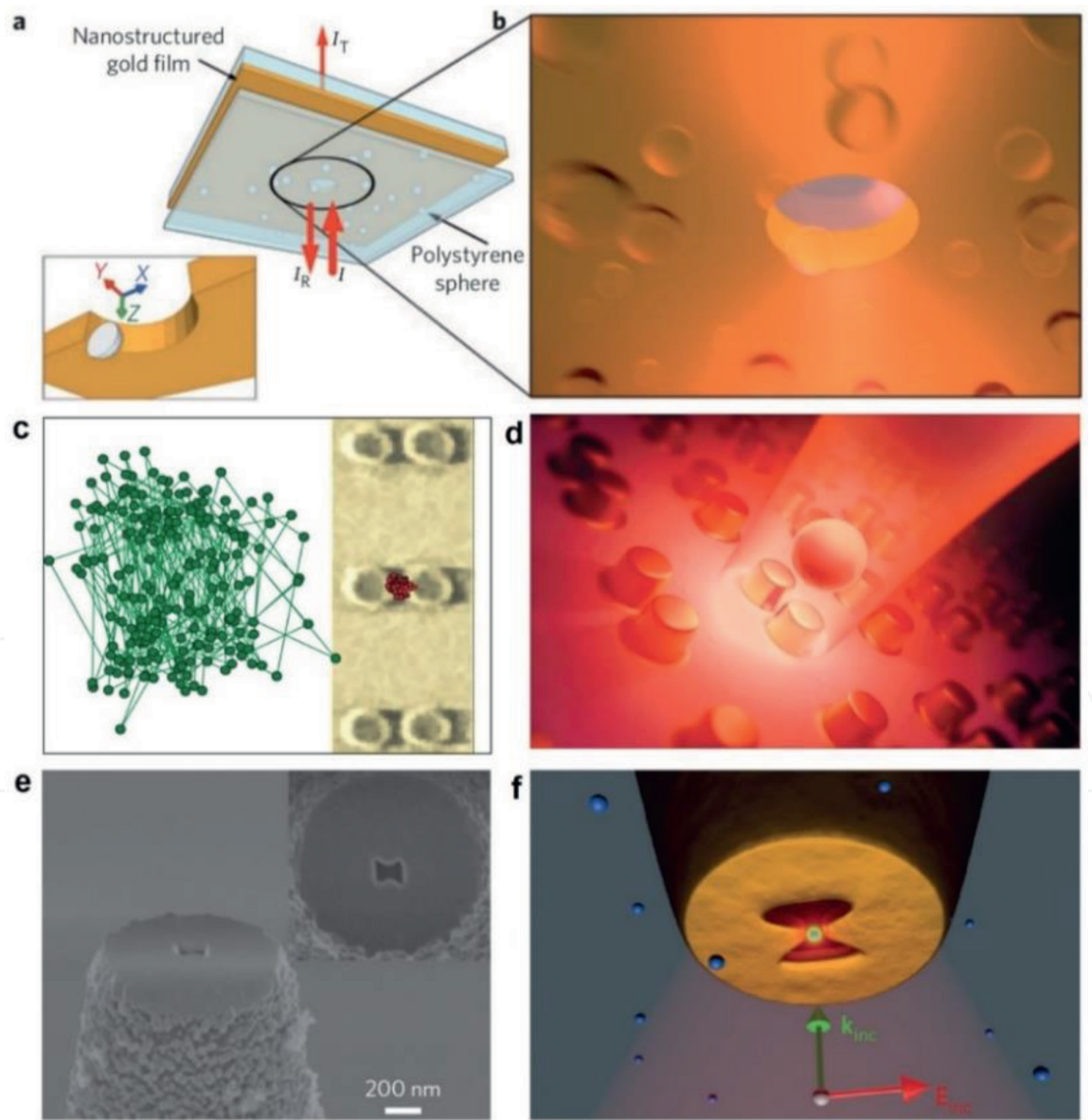


Figure 8. Plasmon optical tweezers. (a) Schematic diagram of a metal film having nanopores. (b) Schematic diagram of magnified metal nanopore capture nanoparticles. (c) SEM image of the metal nano-antenna structure and motion trajectory after the nanoparticles are captured. (d) Schematic representation of metal nano-antenna structures [28]. (e) SEM image of a metal nano-bowtie structure. (f) Schematic diagram of metal bow nanostructure [29].

metal surface and greatly enhanced. The effect is called the plasmon effect. Since the plasmon effect localizes the light in the near-field range of the nanometer order, it is widely used in the fields of fluorescence signal enhancement, near-field super-resolution imaging, high-density optical storage, integrated optical circuits, etc. [32]. In recent years, the plasmon effect has also been applied in the field of optical trapping and manipulation. The plasmon effect is divided into two types: surface plasmon resonance (SPR) and local surface plasmon resonance (LSPR), both of which can be used to enhance optical force. Researchers used a prismatic total internal reflection to couple incident light into a metal micro-disk on the substrate, which will increase the optical force of the particle by two orders of magnitude and realize the capture of the microparticle. However, the SPR-based optical tweezers can only enhance the optical force of the particle in a two-dimensional plane. Therefore, researchers have proposed an LSPR-based nano-optical tweezers to enhance the optical force of the nanoparticle in three dimensions, including metal nanopores (**Figure 8a, b**), metal nano-antennas (**Figure 8c, d**) [28], metal nano-bows (**Figure 8e, f**) [29], and metal nano-double holes [33]. By using these nano-optical tweezers to achieve trapping of various nanoparticles, such as polystyrene particles, protein molecules, gold particles, micro-pathogenic bacteria, and so on.

2. Conclusions

The noncontact and noninvasive optical trapping and manipulation of microparticles, cells, and biomolecules in liquid environments has broad application prospect in the fields of biomedicine and nanomaterial science [34–47]. Traditional optical tweezers and holographic optical tweezers play an important role in the study of microscale optical manipulation. However, in the rapid development of nanoscience, traditional optical tweezers and holographic optical tweezers are difficult to adapt integration and nano-precision requirements due to the large volume and diffraction limitations. The developed nano-optical manipulation techniques, such as planar waveguides, plasmon optical tweezers, and photonic crystal resonators, can overcome the problem of difficult integration and diffraction limitations of conventional optical tweezers and holographic optical tweezers, which hold great promise in biophotonic and biomedical applications.

Acknowledgements

This work was supported by the National Natural Science Foundation of China (No. 11774135, 11874183, and 61827822).

Conflict of interest

The authors declare no competing financial interests.

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