Table 2. Optical propulsion of RBC in different media and sera

Medium	Serum-free	HSA 5% by volume	BSA 5% by weight	FCS 10% by volume
Water	Yes	Inconclusive	No	No
Isotonic sucrose	Yes	Inconclusive	No	No
PBS	No	No	No	No
HEPES	No	No	No	No

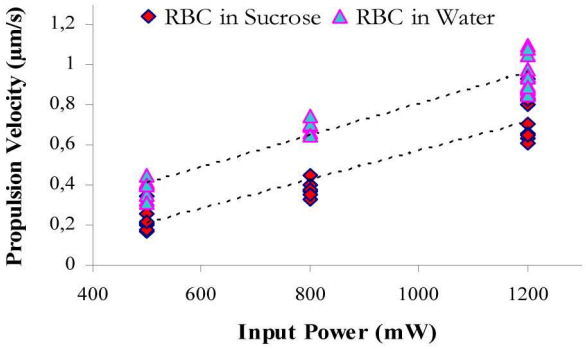

Fig. 3. Comparison of optical propulsion velocity of a cell in water and isotonic sucrose media as a function of input power. Ta_2O_5 waveguide of width ~10 µm was used.

4. Optical propulsion of Live RBCs

Live RBCs were optically trapped using the optimized waveguide material (Ta_2O_5) and medium (isotonic sucrose). We have recently shown that for a given input power, the propulsion velocity of a polystyrene micro-particle varies with the waveguide width [15]. The study found that the width of a waveguide must be carefully chosen for differently sized micro-particles to impart the largest optical force. Thus, optical propulsion velocities of cells were investigated as a function of input power and waveguide width, as shown in Fig. 4. For each width, the propulsion velocities of cells at specific input powers exhibit a wide spread. The size of RBCs varies between 6 and 8 µm, which could lead to a variation of propulsion velocity at a given input power.

As shown in Fig. 4, a 6 µm wide waveguide trapped and propelled cells with a maximum velocity of ~6 µm/s. Light is confined less tightly inside a 10 µm wide waveguide resulting in a weaker intensity near the surface and slow cell propulsion. Furthermore, propelling a cell (~7 µm diameter) over a 10 µm wide waveguide implies that the cell does not interact with the entire evanescent field present. Even though light is confined most tightly in a 3 µm wide waveguide, it has higher insertion losses (as shown in Table 1). Consequently, at a given input power less light is guided in a 3 µm wide waveguides, resulting in slower cell propulsion than on a 6 µm wide waveguide. Thus, to impart higher optical forces for efficient cell propulsion, a waveguide of optimum width must be chosen (6 µm in our case).

Optical propulsion of live RBCs on a 6 µm wide waveguide is depicted in Fig. 5. The interaction between the evanescent field and the cell generates gradient and radiation forces. The cells are laterally trapped on top of a waveguide surface by the gradient force. The radiation force propels the trapped cell along the waveguide. It is worth highlighting that the actual guided power was small, even though a high input power was employed in the experiments. Due to high coupling losses the maximum guided power was ~60 mW for a 6 µm wide waveguide when the cell propulsion velocity was 6 µm/s. From simulations it was found that ~10% of the total guided power will be present in the evanescent field for the chosen waveguide material and dimension.

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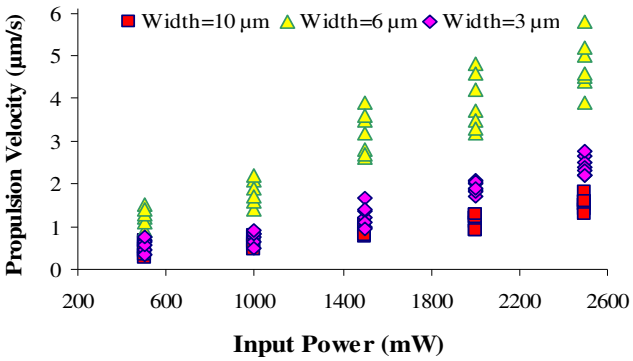

Fig. 4. Optical propulsion velocities of live RBCs as a function of input power and the width of waveguides.

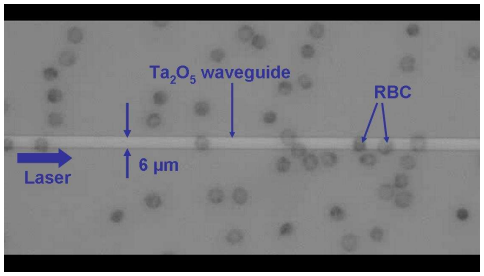

Fig. 5. (Media 1) Optical propulsion of live RBCs on a 6µm wide Ta_2O_5 waveguide.

5. Optical propulsion of cells in the presence of transverse flow

Opto-fluidics combines the advantages of micro-fluidics and optics on a common platform [28]. Integration of micro-fluidics with cell propulsion on waveguides is imperative for efficient and rapid cell manipulation, as a larger number of cells can be transported faster with flow than with optical methods. On the other hand, it is easier to precisely manipulate single cells with optical methods. The two methods can be tailored to sort cells based on optical fractionation techniques [29,30]. Optofluidic trapping on a waveguide surface has previously only been reported for polystyrene microparticles [11].

Figure 6 shows optical trapping and propulsion of cells in the presence of a transverse flow. A 6 µm wide waveguide was used, as this width provided the highest velocity based on our findings in Fig. 4. The total size of the microchip employed in our experiments was 200 µm in height and 1 mm in width. The microchip was made in PDMS (Polydimethylsiloxane), due to its optical transparent and chemically inert properties. The input power used in the experiment was 1 W. The transverse flow rate of the RBCs was ~1µm/s. The fluid flow delivers the cells close to the waveguide surface, where they are trapped and propelled along the propagation direction of light. A 4X objective lens is employed for imaging to cover a large field-of-view in this experiment. As the laser is switched-off, the trapped cells are released and are drawn away by the flow.

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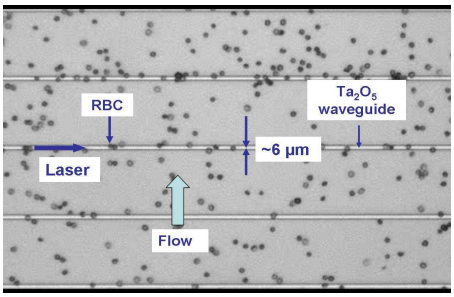

Fig. 6. (Media 2) Optical trapping and propulsion of RBC in the presence of transverse flow.

6. Conclusions

Optical trapping and propulsion of live red blood cells in the evanescent field of an optical waveguide is reported. It is shown that Ta_2O_5, due to its high refractive index contrast, provides an ideal platform for waveguide cell propulsion. Waveguides with a width of ~6 µm were found to propel cells with the highest velocities, 6 µm/s, which is significantly higher than previously reported [12]. Increasing the power in the waveguide, e.g. by decreasing coupling losses, would further increase the propulsion velocity. The velocities demonstrated are compatible with the field-of-view used in high-resolution microscopy. The method may be used to move a single cell to a given position in the field-of-view, hold the cell using counter-propagating beams [31] and study it with high-resolution techniques e.g. confocal-, fluorescence- or Raman-microscopy [32].

Waveguide trapping and propulsion of cells in the presence of a transverse flow is reported. This demonstrates that the cells are stably trapped also transversally relative to the waveguide. Microfluidics can move more cells faster than optical methods, while it is possible to manipulate single cells with optical methods. Combining the two methods in an optofluidics system is thus advantageous, e.g. for the envisioned application in microscopy and for Lab-on-a-Chip applications.

Most cell friendly media, such as sera and PBS, are found to be incompatible with optical trapping on a waveguide surface. However, cells were successfully trapped and propelled in isotonic sucrose, which stabilizes the osmotic pressure of the cells. Surface treatment of the waveguides may reduce the cell adhesion in sera and PBS, and thus make optical propulsion possible in these media. It is, however, important to note that sera, which is derived from clotted blood, is not a natural media for cells in vivo. Serum is full of ligands for scavenger receptors, which bind negatively charged proteins. Therefore, plasma may be a more appropriate media to study in future [33].

Waveguide cell propulsion is influenced by cell parameters like refractive index, cell membrane thickness, shape and size of the cells. It is feasible to sort cells based on different propulsion velocities (for cells with different parameters) and by implementing waveguide designs like a Y-junction [34].

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