Cavitation on the tip of a fibre optic : from a single bubble to a bubble cloud

Milad Mohammadzadeh

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CAVITATION ON THE TIP OF A FIBRE OPTIC:
FROM A SINGLE BUBBLE TO A BUBBLE CLOUD

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SCHOOL OF PHYSICAL AND MATHEMATICAL SCIENCES

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CAVITATION ON THE TIP OF A FIBRE OPTIC:
FROM A SINGLE BUBBLE TO A BUBBLE CLOUD

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*To Mom, Dad,
and Sadaf*

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Contents

Acknowledgements	ii
Summary	xviii
Publications	xx
1 Introduction	1
1.1 Bubbles in medicine	1
1.1.1 Non-invasive surgery and ultrasound-induced cavitation	2
1.1.2 Minimally invasive surgery and laser-induced cavitation	3
1.2 Formation and dynamics of bubbles in laser ablation	4
1.2.1 Mechanism of laser-induced cavitation	4
1.2.2 Effects of cavitation on laser ablation	6
1.3 Guide to this thesis	10
1.4 Summary	11
2 Bubble dynamics in laser lithotripsy	12
2.1 Introduction	12

2.2	Experiments: bubble dynamics and stone motion	14
2.2.1	Experimental setup	14
2.2.2	Sequence of events: bubble formation, ablation, and phantom motion	16
2.2.3	Effective parameters on the bubble shape	19
2.3	Modelling the bubble shape	23
2.4	Conclusion	27
3	Shearing flow from transient bubble oscillations in narrow gaps	29
3.1	Introduction	29
3.2	Numerical simulation	34
3.2.1	Computational domain	34
3.2.2	Numerical solver specifications	34
3.3	CFD results and analysis of the flow field	37
3.3.1	Bubble evolution and liquid velocity profile	37
3.3.2	Flow reversal due to adverse pressure gradient	39
3.3.3	Vorticity generation	42
3.3.4	Flow field contours	43
3.3.5	Effects of initial pressure, channel height, and viscosity	45
3.4	Analytical approximation for liquid velocity profile: pulsating pressure-driven flow	48
3.5	Experiments: deformation of cells	51

3.6	Discussion and Conclusion	54
4	Synthetic jet generation by high-frequency cavitation	57
4.1	Introduction	57
4.2	Experimental setup	59
4.3	Jet generation by high-frequency cavitation	61
4.3.1	Bubble formation	61
4.3.2	Jet characteristics	63
4.3.3	Mechanism of synthetic jet generation	69
4.4	Bubble formation and dynamics	71
4.4.1	Secondary bubble formation	71
4.4.2	Model for bubble growth	73
4.5	Discussion and conclusion	78
5	Photoacoustic shock wave emission and cavitation on fibre optic tips	80
5.1	Introduction	80
5.2	Experimental setup	83
5.3	Shock wave emission and cavitation on structured fibre tips . .	84
5.4	Pressure measurements	91
5.5	Photoacoustic cavitation using fibre optics with coated tips . .	93
5.6	Discussion and conclusion	96
6	Conclusions and outlook	98

6.1	Conclusions	98
6.2	Outlook	101

List of Figures

1.1	Modification of the vapour transmission channel by combining two free running lasers. The erbium laser alone generates a thin elongated bubble (a). Adding a free running holmium laser with Δt delay significantly changes the bubble shape, (b) and (c). Reprinted with permission from Pratisto et al. [1]. Copyright the Optical Society.	6
1.2	Controlling the microjet generated after bubble collapse by modifying the probe shape. The microjet induced on the convex probe shape (a) is eliminated by using a concave probe (b). Adapted with permission from [2]. Copyright AIP Publishing.	7
1.3	Cavitation on a fibre optic tip in haemoglobin solution induced by a 115 ns excimer UV laser pulse (top row). Cavitation induced deformation of a blood vessel which causes mechanical trauma during laser angioplasty (bottom row). Reprinted with permission from [3]. Copyright Wolters Kluwer Health, Inc.	8

2.1	Side and top view of the experimental set-up. The high-speed camera records the bubble dynamics and stone movement, and a photodiode measures the laser pulse profile.	15
2.2	High-speed recordings of bubble dynamics (a), transmission of energy to the stone (b), and stone motion in Ho:YAG laser lithotripsy (c). Ablation region in the phantom is marked with a red circle. The laser is operating at setting 4. The inter-frame time step for each sequence is noted as Δt	16
2.3	Photodiode recordings of the Ho:YAG laser pulse. Energy of the laser increases from setting 1 to setting 4. Although the peak power does not differ much from the lowest to the highest pulse energy, the pulse duration varies considerably (marked with arrows).	19
2.4	Evolution of the bubble shape for laser settings 1 (a), 2 (b), and 3 (c). The pulse energy increases respectively, according to the profiles shown in figure 2.3. The inter-frame time step is $\Delta t = 61.7 \mu\text{s}$	20
2.5	The effect of separation distance between the fibre tip and the ablation target. Shortening the gap to the fibre tip from (a) to (b) results in larger mass removal, evident by a large annular ring of acrylic around the fibre tip that is marked with red. The laser is operating at setting 4 and the inter-frame time step is $\Delta t = 61.7 \mu\text{s}$	21

2.6	The effect of separation distance on target movement. The stone movement is tracked after one pulse for two cases: a distance of 0.6 mm between the fibre tip and the target, and zero distance; i.e. contact mode. The laser is operating at setting 3.	22
2.7	Simulation results for laser setting 1 compared with the experiments. Bubble length (Z) is along the axis of the fibre optic, while maximum bubble width ($R = R(z = 0, t)$) is normal to the fibre axis.	24
2.8	Evolution of the bubble shape from the model (yellow) compared with high-speed recordings for laser setting 1. Using the laser pulse profile as an input allows the model to capture the pear-shaped bubble growth.	25
3.1	Sketch of the computational domain using axisymmetry and symmetry with respect to the centre-plane of the gap.	33
3.2	Evolution of the bubble shape and the radial velocity profile at $r = 50 \mu\text{m}$ from the centre of the bubble. Please note the reversal of the flow direction near the boundaries occurring before the flow in the centre of the gap.	37

3.3	Comparison of experimental cylindrical bubble dynamics [4] (dashed line with squares) with the VOF solution (filled line and dashed line for two computational grids). Additionally, the solution to the 2D Rayleigh-Plesset equation starting from the maximum bubble radius is shown with a dotted line. . . .	38
3.4	Evolution of pressure (top row) and radial velocity (bottom row) during early (a) and late expansion (b) and early (c) and late collapse (d). The times for the pressure and velocity profiles are indicated in the upper legend of each column. . . .	40
3.5	The liquid streamlines during boundary layer separation, i.e. from $t = 3$ to $5.5 \mu\text{s}$. The radial velocity profile is plotted at $r = 50 \mu\text{m}$. Two counter rotating vortices form, move upwards, and merge.	42
3.6	Sequence of contour plots of the flow field during bubble expansion and shrinkage. Pressure field in both phases, radial velocity in the liquid, and vorticity in the liquid are shown at left, centre, and right, respectively. The bubble-liquid interface is shown by the black line, i.e. contour of 0.5 liquid volume fraction. The colour bar for each figure is given underneath. Each figure is plotted with its specific colour map to better distinguish the flow field. Results are shown for a sample time instant of each different stage of the flow.	44

3.7	The effect of initial internal bubble pressure p_i . Increasing the initial pressure results in a faster flow field and a larger bubble size, but important features such as flow reversal and vorticity generation are preserved. All simulations are done with $h = 20 \mu\text{m}$ and $\nu = 1 \text{ cSt}$	46
3.8	Varying the channel height h from 20 to 30 μm . Flow reversal and vorticity generation near the walls are observed for a cylindrical bubble despite the gap height, but are more prominent in tighter confinements. Both simulations are done with $p_i = 100 \text{ bar}$ and $\nu = 1 \text{ cSt}$	46
3.9	The effect of kinematic viscosity ν . Increasing the liquid viscosity, while keeping $p_i = 100 \text{ bar}$ and $h = 20 \mu\text{m}$ constant, slows down the flow field considerably. The film thickness separating the bubble and channel walls is larger in a more viscous liquid.	47
3.10	Comparison of (<i>left</i>) the VOF simulation of the radial velocity with (<i>right</i>) the analytical approximation of the planar velocity profile during bubble shrinkage (3.2).	50
3.11	Deformation of red blood cells (RBC) due to the cavitation-induced flow in a 20 μm thin gap. RBCs are shown just before a laser generates a single cavitation bubble and after the bubble collapses. RBCs are significantly stretched in this flow. Please see [54] for details of the experiment.	52

3.12	Deformation of colon cancer cells (RKO) due to the cavitation-induced flow in a $20\text{ }\mu\text{m}$ thin gap. RKO's are much less deformed compared to RBCs while both are exposed to the same flow, see figure 3.11.	53
4.1	Experimental setup for high-speed photography and particle image velocimetry.	60
4.2	High-speed recordings of bubble formation with a high-repetition-rate pulsed laser delivered through a fibre optic. High-frequency cavitation occurs due to linear absorption of the laser energy by the liquid layer adjacent to the fibre tip, forming bubbles that expand and collapse before arrival of the subsequent laser pulse. The laser is operating at $f = 40\text{ kHz}$ repetition rate and $P = 5\text{ W}$ average power.	62
4.3	Time-averaged velocity field induced by high-frequency cavitation visualised by PIV. The time-averaged flow field resembles a continuous free jet directed away from the cavitation site. Semi-transparent photos of bubble dynamics and the fibre optic are shown to scale. The velocity magnitude, vector field, and axial velocity profiles are plotted. The laser is operating at a repetition rate of $f = 40\text{ kHz}$ with an average power of $P = 20\text{ W}$	64

4.4	Transient and time-averaged axial and radial velocity, u and v , at point $(x = 2 \text{ mm}, r = 0)$ along the jet over a few periods of the cavitation event $T = f^{-1}$. The laser is operating at a repetition rate of $f = 40 \text{ kHz}$ with an average power of $P = 20 \text{ W}$.	65
4.5	Comparison between the self-similar velocity profile of the continuous free jet (solid line) and the time-averaged cavitation-induced jet (symbols). The laser is operating at $f = 40 \text{ kHz}$ and $P = 20 \text{ W}$.	67
4.6	Decay of the jet centreline velocity $u_m = u(x, r = 0)$ for free jets (lines) and cavitation-induced jets (symbols). Origin of the free jets are fixed at the fibre tip and the cone angles are obtained from the experimental data. The laser is operating at $f = 40 \text{ kHz}$ and $P = 20 \text{ W}$.	68
4.7	Schematic illustration of a collapsing bubble that directs the flow away from a rigid boundary. Important flow features are marked in red. (a) The collapsing bubble generates a rapid microjet towards the boundary. (b) After jet impingement, remnants of the microjet compete with a strong radial inflow, forming an annular stagnation ring and a rebounding splash. (c) The splash develops into a toroidal vortex. (d) The vortex moves away from the boundary and carries the bubble fragments along the axis of symmetry.	70

4.8	The effect of increasing laser power on bubble formation with a high-repetition-rate pulsed laser operating at $f = 40$ kHz. By increasing the average laser power from left to right, the bubble size increases. Above a certain threshold, the formation of a secondary bubble at the tip of the initial hemispherical bubble is observed. This occurs when the lifetime of the first bubble is larger than the period of pulse repetition.	72
4.9	Maximum bubble radius R_{max} as a function of the laser pulse energy E . The experimental results are compared with the model presented in section 4.4.2 for three different laser repetition rates.	74
4.10	Laser pulse energy E needed for secondary bubble formation as a function of the pulse repetition rate f . The grey colour marks the region and experimental data with only hemispherical bubble formation, and white denotes the formation of secondary bubbles.	75
5.1	Schematic of the stroboscopic photography setup.	83

5.2	Acoustic transients and cavitation induced by tensile stress in front of structured fibre optics after a single laser pulse ($\tau = 6 \text{ ns}$, $H_0 < 1 \text{ J/cm}^2$, and $\mu = 355 \text{ cm}^{-1}$). Left column shows shadowgraph images and right column presents numerical solutions of the linear wave equation corresponding to the same instant of time.	85
5.3	Computational domain and boundary conditions.	88
5.4	Shock wave emission and cavitation cloud formation from a flat cleaved fibre. The shadowgraph images from the left column are compared with the numerical solution of the linear wave equation in the middle. The right column shows evolution of the cavitation cloud based on the simulation data, generated by accumulation of pressure isosurfaces at the threshold $p = -0.5p_a$	89
5.5	Shock wave emission and cavitation cloud formation from a stepped fibre tip. Similar to figure 5.4, experimental shadowgraphs are at the left, numerical solution in the middle, and prediction of cavitation clouds based on the simulation at the right column. The stepped tip emits multiple shock waves and forms two cavitation clouds.	90

5.6	Pressure measurements from a fibre optic hydrophone (blue, solid lines) compared with the numerical solution of the linear wave equation (red, dashed lines). The structured fibre tip generates multiple compressive and tensile waves on and off-axis from a single laser pulse.	92
5.7	Photoacoustic shock wave emission and cavitation on the tip of a fibre optic coated with a CNT-PDMS composite.	94

Summary

Minimally-invasive surgeries rely heavily on the use of fibre optics to steer through small incisions on the body and deliver laser pulses to remove unwanted tissue. Laser-induced heating at the fibre tip leads to vaporisation and formation of cavitation bubbles. These bubbles play two crucial roles in laser surgery: they act as transmission channels that connect the fibre tip to the ablation target and generate mechanical effects and motion in their surroundings. This thesis focuses on the dynamics of laser-generated bubbles, especially on the tip fibre optics.

The thesis begins with investigating the dynamics of elongated, pear-shaped vapour bubbles in ablation with free-running lasers. A common clinical complication is the movement of the ablation target, which we experimentally link to the bubble-induced flow. We then provide a model to estimate the bubble shape based on the laser pulse profile. Controlling the shape of the vapour transmission channel with this model helps in reducing the resulting liquid motion and mechanical trauma due to cavitation.

We then investigate the liquid motion induced by cavitation in a confined

geometry, focusing on axisymmetric flow close to an oscillating bubble. Numerical simulations reveal that the bubble induces vortical structures that generate strong shear stress near the boundaries. The shearing is probed experimentally by observing the deformation of elastic biological cells in the vicinity of the bubble.

Another cavitation-induced flow investigated in this thesis is generated by delivering a pulsed laser at a high repetition rate through a fibre optic. We observe a fast and localised liquid stream directed away from the fibre tip. Using particle image velocimetry, the stream is characterised as a synthetic jet driven by vortices generated during bubble collapse. High-speed recordings reveal the formation of double-bubbles on the fibre tip at certain laser settings, which we predict with a model that links the bubble size to the laser energy.

Finally, the formation of bubble clouds due to photoacoustic shock waves emitted from a fibre optic is studied. Interestingly, we observe that modifying the shape of the fibre tip allows us to steer the shock waves and the resulting cavitation cloud. We end with providing accurate numerical solutions to optimise the fibre tip structure for achieving a desirable acoustic field and region of ablation.

Publications

Journals

1. M. Mohammadzadeh, S.R. Gonzalez-Avila, K. Liu, Q.J. Wang, C.-D. Ohl, "Synthetic jet generation by high-frequency cavitation," *Journal of Fluid Mechanics*, 823, 2017.
2. M. Mohammadzadeh, F. Li, C.-D. Ohl, "Shearing flow from transient bubble oscillations in narrow gaps," *Physical Review Fluids*, 2 (1), 014301, 2017.
3. M. Mohammadzadeh, S.R. Gonzalez-Avila, Y.C. Wan, X. Wang, H. Zheng, C.-D. Ohl, "Photoacoustic shock wave emission and cavitation from structured optical fiber tips," *Applied Physics Letters*, 108 (2), 024101, 2016.

Conferences

1. M. Mohammadzadeh, W. Chan, S.R. Gonzalez-Avila, K. Liu, Q.J. Wang, C.-D. Ohl, "Bubble formation with a high repetition rate pulsed Tm laser," *Proceedings of The 20th Australasian Fluid Mechanics Conference*, 2017.
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Chapter 1

Introduction

1.1 Bubbles in medicine

Many patients benefit from non-invasive and minimally invasive surgeries everyday, medical procedures that require no incision, or very subtle cuts compared to conventional open surgeries. Minimally invasive procedures result in less trauma, shorter hospitalisation, and reduced recovery time for the patients. Many of these procedures involve the formation of vapour bubbles in the body, commonly through ultrasound-induced tensile stress or laser-induced heating. These bubbles could be generated intentionally to serve a therapeutic or diagnostic application, or they could be inevitably created during the procedure.

1.1.1 Non-invasive surgery and ultrasound-induced cavitation

A common mechanism of bubble formation in non-invasive surgery is ultrasound-induced cavitation. Tensile waves, if sufficiently strong, can rupture a liquid and cause a phase change to vapour, i.e. cavitation [5]. External arrays of piezoelectric transducers are widely used to generate such ultrasound waves in medical applications. For instance, extra-corporeal shock wave lithotripsy (ESWL), which has been the standard procedure for fragmentation of kidney stones for over two decades, utilises focused ultrasound waves to generate a bubble cloud that grinds the renal calculi [6]. In addition to fragmentation of hard tissue, termed "lithotripsy", ablation of soft tissue, or "histotripsy", could be achieved by using sequences of focused ultrasound waves. In this procedure, mechanical lesions and tissue liquefaction are generated by a cavitation cloud [7]. Histotripsy has already been used in-vivo, for instance to treat benign prostatic hyperplasia (BPH), and could be a novel and promising candidate for non-invasive surgery [8]. High intensity focused ultrasound, or HIFU, is another emerging non-invasive therapeutic application that involves cavitation bubbles. In HIFU, focusing of acoustic energy over millimetre-sized regions results in intense heating and thermal lesions that are clinically used for treatment of tumours [9].

The interaction between bubbles and acoustic waves can also be used for diagnostic and therapeutic applications. Injection of coated microbubbles

that act as ultrasound contrast agents in the blood stream can considerably improve diagnostic ultrasound imaging [10]. Bubbles undergo significant deformation in response to acoustic waves and generate shear stress in the surrounding liquid. Sonoporation, a novel method of drug delivery, utilises such deformations to locally enhance cell membrane permeability and mass transfer into cells [11, 12]. Coating microbubbles with cell-specific receptors allows their adhesion to a particular target cell, which could reduce side effects of chemotherapy. Recent clinical trials have shown the effectiveness of sonoporation for cancer treatment [13, 14]. The interaction of ultrasound waves and cavitation bubbles may also enhance the thrombolytic effect of drugs used for treatment of ischemic strokes [15]. Cavitation-enhanced drug delivery is comprehensively reviewed in the literature [16].

1.1.2 Minimally invasive surgery and laser-induced cavitation

While ultrasound-induced cavitation is involved in non-invasive surgeries, minimally invasive surgeries rely heavily on laser-generated bubbles [17]. Commonly, a fibre optic is inserted through a small opening or incision on the body to deliver a pulsed laser to a target. For instance, to fragment urinary stones that are located in the urethra, a fibre optic is inserted through the ureter and a laser beam is aimed at the stone to break it. Before the laser beam is absorbed by the stone, the highly absorptive surrounding liq-

uid (water) is vaporised and a bubble is formed, connecting the fibre tip and the stone [18]. Formation of a vapour bubble is inevitable in this procedure, known as laser lithotripsy, and the bubble dynamics significantly affects the ablation outcome. Vaporisation is an essential part of laser-ablation of soft tissue across many medical disciplines, from neurosurgery to cardiovascular surgery [19]. For example, bubble formation significantly affects the mass removal rate in laser thrombolysis, a procedure to remove arterial blood clots [20]. In all medical procedures discussed above, understanding the bubble dynamics is crucial to predict the outcome, fully exploit the potentials, and prevent side effects. This thesis focuses on laser-generated cavitation on fibre optics which is an inevitable part of laser ablation.

1.2 Formation and dynamics of bubbles in laser ablation

1.2.1 Mechanism of laser-induced cavitation

The mechanism of bubble formation during laser ablation depends on the energy, wavelength, and duration of the laser pulse as well as the constituents of the ablation target. Absorption of the laser energy by the irradiated tissue depends on its composition. Proteins and DNA highly absorb ultraviolet radiation, melanin is the main absorber of visible radiation, and haemoglobin absorbs across the UV and visible spectrum [21]. Water, which constitutes

more than 70% of the weight of biological tissue, highly absorbs infrared radiation with a peak at $\lambda = 2.94 \mu\text{m}$ wavelength in the mid-infrared spectrum [22]. When scattering is negligible, the laser energy heats up the volume within the optical penetration depth and may lead to phase change from liquid to vapour. At extreme intensities, non-linear absorption and optical breakdown may lead to plasma formation, even if the irradiated material is transparent to the laser wavelength [23, 24]. Energy transport from this irradiated volume depends on the duration of energy deposition and the thermomechanical response time of the medium [17]. When the duration of energy deposition is shorter than the time needed for stress propagation, stress confinement occurs, leading to strong thermoelastic waves. When energy deposition is faster than thermal diffusion in the target material, thermal confinement occurs and traps intense heat within the irradiated volume. Both stress and thermal confinements can generate violent cavitation bubbles in the target material. The thermoelastic waves generated under stress confinement are compressive; however, due to conservation of linear momentum they carry subsequent tensile waves [17, 25] that can rupture the irradiated material and generate voids, i.e. cavitation bubbles in liquids. Intense heating under thermal confinement leads to superheating of the target material far above the boiling point and generates vapour bubbles after an explosive phase transition. Both of these mechanisms for bubble formation have been comprehensively investigated in the literature [17, 26–29].

Figure 1.1: Modification of the vapour transmission channel by combining two free running lasers. The erbium laser alone generates a thin elongated bubble (a). Adding a free running holmium laser with Δt delay significantly changes the bubble shape, (b) and (c). Reprinted with permission from Pratisto et al. [1]. Copyright the Optical Society.

1.2.2 Effects of cavitation on laser ablation

Cavitation plays two major roles during laser ablation in liquids. First, the bubble acts as a transmission channel that connects the fibre tip to the target. This means that there is usually an aqueous layer of fluid between the fibre tip and the ablation target that absorbs part of the laser energy and forms a bubble that connects the fibre tip to the target. The bubble acts as a transmission channel as it is usually transparent to the laser wavelength, allowing the energy to be transmitted from the fibre tip to the ablation target. The shape and dynamics of this transmission channels depend on the mechanism of bubble formation and the laser pulse duration. Q-switched lasers with nanosecond pulses commonly generate spherical bubbles that expand for a long time after the laser pulse. On the other hand, free running

Figure 1.2: Controlling the microjet generated after bubble collapse by modifying the probe shape. The microjet induced on the convex probe shape (a) is eliminated by using a concave probe (b). Adapted with permission from [2]. Copyright AIP Publishing.

Figure 1.3: Cavitation on a fibre optic tip in haemoglobin solution induced by a 115 ns excimer UV laser pulse (top row). Cavitation induced deformation of a blood vessel which causes mechanical trauma during laser angioplasty (bottom row). Reprinted with permission from [3]. Copyright Wolters Kluwer Health, Inc.

lasers with long pulse durations, usually hundreds of microseconds, vaporise the liquid throughout the pulse and generate elongated bubbles that expand as long as the laser is on. With equal energy, bubbles generated with Q-switched pulses expand to much larger extends compared to those created with free running lasers [30]. Modifying the shape of the vapour transmission channel can affect the efficiency and precision of laser ablation. For instance, Pratisto et al. [1, 31] combined two free running lasers into one setup, a Ho:YAG ($\lambda = 2.1 \mu\text{m}$) and an Er:YSGG laser ($2.79 \mu\text{m}$), to take advantage of two wavelengths: generation of a long-lasting bubble with the holmium laser and precise cutting with the erbium laser. Figure 1.1 depicts these modified vapour bubbles after adding a holmium laser with Δt delay after the erbium pulse begins.

The second role of cavitation during laser ablation is generation of mechanical effects and hydrodynamic forces that lead to material deformation and removal. A prominent example is the generation of rapid liquid microjets at the last stage of bubble collapse near boundaries [32], which may cause unwanted mechanical damage to healthy tissue during microsurgery [33]. Figure 1.2 shows an interesting way to eliminate the movement of such microjets away from the cavitation site towards healthy tissue by altering the shape of the probe used for bubble generation [2]. A dramatic example of adverse mechanical effects due to cavitation is in laser angioplasty, a procedure which aims to open coronary arteries blocked by fatty plaques. After years of clinical practice, it was revealed that the violent expansion and collapse of vapour bubbles in blood vessels under excimer laser angioplasty led to microsecond dialation and invagination of the vessels. These effects, depicted in figure 1.3, occurred far beyond the laser penetration depth and led to high rates of clinical complications in patients [3, 34]. While cavitation on the tip of fibre optics is an inevitable part of laser ablation, its effects should be understood and controlled to prevent trauma in the healthy tissue that surrounds the target.

This thesis explores on the formation and dynamics of cavitation bubbles, especially on the tip of fibre optics. We report on experiments, numerical simulations, and theories describing cavitation: from the mild expansion of bubbles with free-running lasers, to the violent dynamics of bubble generated by Q-switched lasers.

1.3 Guide to this thesis

We begin with the formation of a single bubble on a fibre optic due to heating with a free-running laser in chapter 2. Because of the long pulse duration, this bubble obtains an odd, elongated shape. A model is presented to link the bubble shape to the temporal pulse profile, and the effects of the bubble dynamics on fragmentation of urinary stones in laser lithotripsy are explained.

In chapter 3, numerical simulations are used to investigate the flow induced by a single bubble in a microfluidic confinement, focusing on unsteady boundary layers close to an oscillating bubble. We explain the fundamental flow features with the oscillating pressure gradient that drives the liquid motion. Biological cells of different size are exposed to this rapid cavitation-induced flow in an experiment where the shear stress deforms the cells.

Chapter 4 introduces the formation of double-bubbles on a fibre optic. The bubbles, generated by a high-repetition rate pulsed laser, induce a fast microjet along the fibre axis. This flow is experimentally investigated using particle image velocimetry and characterised as a synthetic jet. The mechanism of formation of double-bubbles is explained and a model is provided to link the bubble size with the laser pulse energy.

In chapter 5, generation of bubble clouds due to photoacoustic shock wave emission on the tips of fibre optics is studied. By modifying the shape of the fibre tip, the acoustic transients and the resulting cavitation cloud can be controlled. The pressure close to the fibre tip is probed and accurate simu-

lations are provided to optimize the fibre tip structure to achieve a desirable acoustic field.

Finally, the findings are summarised and concluded in chapter 6. The thesis ends with an outlook on future research directions and applications.

1.4 Summary

This thesis focuses on laser generated bubbles, especially on the tip of fibre optics. We explore a number of mechanisms to control these bubbles and their effects. For instance, the energy deposition rate of the laser controls the bubble shape and consequently the liquid motion (chapter 2). This liquid motion is also affected by nearby boundaries (chapter 3) and the repetition rate of the laser (chapter 4). Finally, shaping the fibre optic tip is an effective way to control the photoacoustic field and cavitation induced by shock waves (chapter 5). These control mechanism may improve microfluidic and clinical applications dealing with cavitation on fibre optics.

Chapter 2

Bubble dynamics in laser lithotripsy

2.1 Introduction

Intracorporeal laser lithotripsy has been in use for over two decades as a minimally invasive method for fragmenting urinary stones [35]. In this procedure, an endoscope is inserted into the urethra to locate calculi and a fibre optic is inserted through the working channel of the scope. A pulsed laser is coupled into the fibre to irradiate the stone, causing the stone to disintegrate into smaller fragments which are later washed out of the urinary tract. Using fibre optics for laser surgery is of course not limited to laser lithotripsy and a similar working principle is used for ablation of soft tissue, such as laser thrombolysis (fragmentation of blood clots), discectomy (intervertebral disc

surgery), and vitrectomy (surgery on the vitreous body of the eye) [19].

The ablation mechanism in these procedures depends on the laser wavelength, pulse energy and duration, and material properties of the irradiated medium [17]. The most commonly used laser lithotripter is the Holmium:YAG laser ($\lambda \approx 2.1 \mu\text{m}$) with a long pulse duration (e.g. $\tau = 300 \mu\text{s}$). In Ho:YAG laser lithotripsy, the fundamental mechanism of transfer of laser energy to the stone is through the formation of a vapour bubble that bridges the fibre optic with the urinary stone. Afterwards, the calculus absorbs the laser energy and is fragmented due to a photothermal mechanism [18].

To increase the ablation efficiency in laser lithotripsy, alternatives to the Ho:YAG laser have been explored. In the mid-infrared spectrum, Thulium ($\lambda \approx 1.96 \mu\text{m}$) and Erbium ($\lambda \approx 2.79 \mu\text{m}$) lasers have been able to increase the mass removal rate from the ablation target due to their stronger absorption in water [26, 36]. Thulium fibre lasers in particular can be coupled into fibre optics with smaller radii, reducing fibre failure and endoscope damage while allowing for improved visibility with greater irrigation rates [37].

A reported clinical challenge involved with laser lithotripsy is the retropulsive movement of the stone during ablation. Stone motion prolongs the operation time, leads to additional procedures for residual calculi larger than 2 mm, and causes clinical complications [38]. Mechanical backstops such as coiled wires can be inserted through catheters to prevent the motion of large calculi during the procedure [39]. To reduce stone retropulsion, the effect of several laser parameters has been experimentally investigated, including

optical pulse energy, fibre diameter, and the pulse duration [40–42].

In this chapter, we discuss the role of bubble formation in the ablation procedure and stone motion. Using high-speed photography, we present *in-vitro* experiments for different settings of a clinical Ho:YAG laser while recording the bubble dynamics, stone motion, and the temporal pulse profile. Finally, we provide a simple model to relate the laser pulse with the observed bubble dynamics.

2.2 Experiments: bubble dynamics and stone motion

2.2.1 Experimental setup

The experimental set-up is shown from the side and top view in figure 2.1. A transparent acrylic container (100 x 50 x 57 mm) is filled with water. An acrylic cube (6 mm) serves as a stone phantom. The 270 μm fibre from the clinical Ho:YAG (Dornier, Ho:YAG $\lambda = 2.1 \mu\text{m}$) is inserted from the right into the container. The dynamics of the bubble and stone movement is observed with a high-speed camera (Photron FASTCAM SA-X2). Illumination (200 W metal halide lamp) is provided from the back through a diffuser. The temporal profile of the laser pulse is measured with a reverse-biased InGaAs photodiode (Thorlabs, FGA20) by placing the detector in front of the fibre tip in air prior to the experiments. The experiments are conducted at Tan

Figure 2.1: Side and top view of the experimental set-up. The high-speed camera records the bubble dynamics and stone movement, and a photodiode measures the laser pulse profile.

Tock Seng Hospital, Singapore.

2.2.2 Sequence of events: bubble formation, ablation, and phantom motion

Figure 2.2: High-speed recordings of bubble dynamics (a), transmission of energy to the stone (b), and stone motion in Ho:YAG laser lithotripsy (c). Ablation region in the phantom is marked with a red circle. The laser is operating at setting 4. The inter-frame time step for each sequence is noted as Δt .

Figure 2.2 depicts the sequence of events from bubble formation to ablation and eventually phantom motion for a sample setting of the Ho:YAG

laser lithotripter. The first set, figure 2.2a, displays the growth and collapse of the elongated bubble during the laser pulse without the presence of a nearby object. The bubble grows from a pear shape to an elongated shape and collapses within 700 μs . The photodiode recordings of the laser pulse profile (see figure 2.3) reveal that the bubble grows as long as the laser is on, reaching the maximum size just before the laser pulse ends. Soon after, the bubble starts shrinking and collapses within a few hundred microseconds.

The thermodynamic process that leads to formation of elongated bubbles during mid-infrared laser ablation is the metastable vaporization of superheated liquid [17]. During lithotripsy with the commonly used Ho:YAG laser, the pulse duration, τ , is significantly longer than the stress relaxation time t_{st} , but shorter than thermal relaxation time t_{th} ; i.e.

$$t_{st} = \frac{\delta}{c} < \tau < t_{th} = \frac{\delta^2}{4k}. \quad (2.1)$$

Here c is the speed of sound, k is the thermal diffusivity of the liquid medium, and δ is the length scale of the irradiated volume, in this case the size of the fibre optic. When a Ho:YAG pulse of $\tau=300\mu\text{s}$ is delivered into water with a fibre diameter of 300 μm , since $t_{st}=100\text{ ns} < \tau$, stress confinement does not occur and no significant thermoelastic expansion waves are emitted. Shock waves upon bubble formation are only considerable when using Q-switched lasers lithotripters [30], such as Coumarin or Rhodamin dye lasers [23]. With the common clinical Ho:YAG laser lithotripters, however,

since $\tau < t_{th} = 39 \text{ ms}$, the liquid is heated under thermal confinement, and maximum possible temperature is obtained inside the irradiated volume. In this condition, the liquid temperature approaches the critical point before phase explosion leads to the formation of a vapour bubble. Optical temperature probing techniques show that water can be heated up to 280°C before phase transition occurs and the bubble is formed [43]. Therefore, in Ho:YAG lithotripsy, the laser pulse is transmitted through the transparent bubble and vaporizes the liquid layer by layer, as shown in figure 2.2a.

In the experiments with a phantom stone in front of the fibre optic, we observe that the formation of a vapour bubble is necessary for transmission of laser energy to the stone. Figure 2.2b shows that the water separating the phantom from the fibre tip is vaporized and the ablation process begins once the vapour bubble reaches the phantom. Due to transparency of the phantom, absorption of the laser light and the subsequent ablation becomes visible as a bright region within the phantom, see circle in figure 2.2b. During this process the stone does not move.

Figure 2.2c reveals that the stone moves after the bubble collapse, over a much longer time scale (100 ms) compared to the life time of the bubble (0.7 ms). We speculate that a ring vortex is generated during the bubble collapse, the origin of which could be the liquid jet moving towards the stone. The heat induced Schlieren in the water reveals the long lasting circulatory liquid motion.

Figure 2.3: Photodiode recordings of the Ho:YAG laser pulse. Energy of the laser increases from setting 1 to setting 4. Although the peak power does not differ much from the lowest to the highest pulse energy, the pulse duration varies considerably (marked with arrows).

2.2.3 Effective parameters on the bubble shape

Recording the laser pulse with a photodiode allows us to investigate the effect of the temporal pulse profile on the bubble shape. As shown in figure 2.3, although the peak power between the laser settings differs by less than 10%, their total pulse duration varies by 80%. Non-uniform energy deposition through this pulse affects the bubble shape, which is shown in figure 2.4. As the pulse energy increases from left to right, the bubble shape becomes more elongated. Earlier studies have shown that Q-switched lasers generate spherical bubble shapes with a faster expansion velocity, while longer pulses lead to elongated bubbles [30]. This is aligned with our observation based on

Figure 2.4: Evolution of the bubble shape for laser settings 1 (a), 2 (b), and 3 (c). The pulse energy increases respectively, according to the profiles shown in figure 2.3. The inter-frame time step is $\Delta t = 61.7 \mu\text{s}$.

Figure 2.5: The effect of separation distance between the fibre tip and the ablation target. Shortening the gap to the fibre tip from (a) to (b) results in larger mass removal, evident by a large annular ring of acrylic around the fibre tip that is marked with red. The laser is operating at setting 4 and the inter-frame time step is $\Delta t = 61.7 \mu\text{s}$.

Figure 2.6: The effect of separation distance on target movement. The stone movement is tracked after one pulse for two cases: a distance of 0.6 mm between the fibre tip and the target, and zero distance; i.e. contact mode. The laser is operating at setting 3.

the photodiode recordings: sudden energy deposition at the beginning of the pulse generates a spherical shape closer to the tip, followed by elongation of the bubble due to continuing energy deposition.

In addition to the laser pulse profile, the size of the fibre optic affects the bubble shape. A recent study of clinical Ho:YAG lasers reports that a smaller fibre diameter results in a more elongated bubble compared to a larger fibre [44]. Another factor that affects the bubble dynamics is elasticity of the ablation environment, where increasing the young modulus dampens bubble growth and the pressure transients [45]. The bubble shape in laser lithotripsy is closely linked to the ablation efficiency and stone motion. Recent studies have shown that for a given pulse energy, longer pulse durations and smaller fibres reduce stone retropulsion [40–42]. This means that elongated bubbles

compared to spherical ones mitigate stone motion in laser lithotripsy.

The separation distance between the fibre tip and the target also affects the bubble dynamics and the ablation process. Figure 2.5 shows that for the same laser pulse, placing the fibre tip closer to the target reduces the bubble size, leading to more energy absorption by the target in figure 2.5b compared to figure 2.5a. This results in greater mass removal, evident by the larger crater in the nearer target in figure 2.5b. Strong recoil momentum during the ablation procedure can push the debris toward the fibre tip, such as the ring of acrylic marked in figure 2.5b. Meanwhile, closing the gap between the fibre tip and the target increases the target motion due to the stronger recoil momentum. This is shown in figure 2.6, where ablation in contact mode has increased the retropulsion per pulse by 40%. Given that ablation in contact mode produces the greatest mass removal per pulse, statistical investigation over many pulses is required to find the optimum distance to the target. The bubble dynamics in contact mode is similar to figure 2.5b. The friction force between the phantom and the support also affects the retropulsion, yet this is not measured in our experiments.

2.3 Modelling the bubble shape

Despite the importance of bubble dynamics in laser lithotripsy, there is a lack of studies on modelling the bubble shape in the literature. This might be due to the complexity of simulating phase change while tracking the bubble in-

Figure 2.7: Simulation results for laser setting 1 compared with the experiments. Bubble length (Z) is along the axis of the fibre optic, while maximum bubble width ($R = R(z = 0, t)$) is normal to the fibre axis.

interface in a multiphase flow problem, which requires extensive computational effort. The collapse of spherical bubbles in similar surgical conditions is studied computationally [46, 47], but the non-spherical bubble shape produced by vaporization with long-pulsed mid-infrared lasers is not yet modelled. Here, we provide a simple, potential flow model to estimate the bubble shape based on the temporal profile of the laser pulse.

Similar to the modelling of elongated cavities for the classical water entry problem [48], we model the shape of the elongated bubble in laser lithotripsy by discrete layers, where each layer is governed by a two dimensional Rayleigh-Plesset equation, assuming radial potential flow. The 2D Rayleigh-Plesset equation for a non-confined bubble is

$$(R\ddot{R} + \dot{R}^2) \log\left(\frac{R_\infty}{R}\right) - \frac{1}{2}\dot{R}^2 = \frac{-p_\infty + p_b}{\rho_l} - \frac{2\nu\dot{R}}{R} - \frac{\sigma}{\rho_l R}, \quad (2.2)$$

Figure 2.8: Evolution of the bubble shape from the model (yellow) compared with high-speed recordings for laser setting 1. Using the laser pulse profile as an input allows the model to capture the pear-shaped bubble growth.

where overhead dots denote derivatives with respect to time, $\nu = 10^{-6} \text{ m}^2/\text{s}$ is the kinematic viscosity of the liquid, $\rho_l = 10^3 \text{ kg/m}^3$ is the liquid density, and $\sigma = 0.072 \text{ N/m}$ is the surface tension. The ambient pressure in the liquid is $p_\infty = 1 \text{ atm}$ and the pressure inside the bubble is p_b . Given the state of the initial vapour bubble, we model this internal pressure as an adiabatic process starting from the critical pressure of water; i.e. 218 atm. Finally, $R_\infty = 10 \text{ mm}$ is a characteristic length scale to bound the liquid momentum. We model the radius of the elongated cavity at any distance z away from the fibre tip ($z = 0$) and time t , i.e. $R(z, t)$.

In this model, two initial conditions are necessary to find the bubble shape at any time: initial radial velocity of each layer, and the time each layer starts expanding. These conditions are found through an energy balance argument based on hemispherical vaporization of the liquid at the fibre tip. At each layer, the bubble starts expanding with an initial radial velocity that is linearly correlated with the laser power $\dot{R}(z, t = t_z) = K_{exp} p_l(t = t_z)$. Here K_{exp} is the correlation constant which is found experimentally to be $5 \times 10^{-3} \text{ m/J}$, and $p_l(t)$ is the laser power as a function of time. The time t_z is the time when the laser has reached to depth Z , which is found from $t_z = E^{-1}(\frac{Z}{K_{exp}})$, where $E(t) = \int_0^t p_l(t) dt$ is the laser energy deposited in the liquid up to time t , and E^{-1} denotes the inverse of this function.

The photodiode measurements, see figure 2.3, are used as input for the bubble dynamics simulations for a sample setting of the Ho:YAG laser. Figure 2.7 compares the evolution of the bubble length $Z(t)$ and the maximum

bubble width $R = R(z = 0, t)$ between the model and the experimental results. The bubble length shows fair agreement with an average error of 20% with the simulation, while the bubble width and collapse time have a discrepancy of 10%. Figure 2.8 shows that the model is able to capture the growth of the pear-shaped bubble with fair agreement compared to the high-speed recordings, while the collapse is not captured accurately. The main culprit for the discrepancy during the bubble collapse is the radial flow assumption which neglects the liquid momentum along the fibre axis, resulting in a milder collapse prediction by the model. Despite the shortcomings of the model, it is a simple tool to estimate the shape and size of elongated bubbles in mid-infrared ablation.

2.4 Conclusion

While laser lithotripsy has been used as a standard method for fragmentation of urinary stones for over two decades, there are still unresolved problems regarding the physics of this procedure. One of the main challenges in laser lithotripsy, stone movement, happens over a long time scale after the bubble collapse. The stone moves due to the liquid motion that lasts for ~ 100 ms, while the bubble life time is only ~ 0.5 ms. By controlling the bubble dynamics, the liquid motion and consequently the stone movement could be mitigated.

The bubble shape depends on the laser pulse profile and the energy depo-

sition rate, allowing modifications by controlling the laser pulse profile. For instance, delivering an Er:YSGG pulse after a Ho:YAG laser significantly changes the bubble shape and results in high tissue ablation efficiency and cutting precision [31]. With a basic model for an elongated bubble, we show that it is possible to estimate the bubble shape from the laser pulse profile. The model can be useful in developing a laser setup with controllable temporal profile to modify the bubble shape and mitigate stone motion in laser lithotripsy. Such a laser system would act similar to dual-wavelength setups [31] but uses only one wavelength.

Chapter 3

Shearing flow from transient bubble oscillations in narrow gaps

3.1 Introduction

Strongly oscillating bubbles in microfluidic confinements are found in many applications of cavitation. These bubbles can be generated with focused laser pulses [4, 49, 50], acoustically excited capillary waves [51], or through spark discharges [52]. Applications of these transient pulsating flows span cell stretching [53, 54], liquid pumping [55], switching and sorting [56, 57], mixing [58], and droplet generation [59]. Many therapeutic applications, such as fragmentation of urinary stones in laser lithotripsy [44], ablation

of blood clots in laser thrombolysis [19], and gas embolotherapy for cancer treatment [60] involve bubble formation in confinements. While in some of these applications cavitation is generated on a fibre optic, in others, such as embolotherapy and sonoporation [12], cavitation begins with a spherical nucleus. Regardless of the bubble generation mechanism in these applications, it is essential to investigate the flow induced by expansion and collapse of a bubble in a confined geometry.

This cavitation-induced flow can be modelled to various degrees of sophistication. Over a sufficiently short time scale, the problem can be simplified as a planar inviscid flow leading to a Rayleigh type equation in cylindrical coordinates [61]. This potential flow description has been extended to non-spherical bubbles in narrow gaps [62]. However, a notable deficiency of these approximations is their inability to model boundary layers, which are important when dealing with suspended objects near the walls, e.g. flat red blood cells and thin elastic objects such as nanowires [63]. Our recent experiments on bubble-induced cell stretching, such as red blood cells [54], gave motivation to model the fluid flow in order to understand the underlying flow patterns that causes cell deformation. In this chapter, we focus on the structure of the liquid flow, i.e. the formation of boundary layers during the expansion and collapse cycle of a single transient bubble.

In general, this confined flow may be simplified to an axisymmetric radial flow forced by a time-dependent source at the origin. Axisymmetric radial flows in narrow gaps have been studied experimentally and analytically in

the past fifty years due to their relevance in industrial applications such as radial viscometers, radial diffusers, non-rotating air bearings, and disk type heat exchangers. For an oscillating source between two parallel plates, in which the source strength varies sinusoidally about a zero-mean value, Elkouh [64] obtained an analytical solution and reported reversed flow near the walls. Zitouni and Vatistas [65] provide an analytical power series solution to purely accelerating and decelerating flows between two flat disks, which has later been studied numerically by Ghaly and Vatistas [66] as well. Although the flow reversal is not captured with the solution provided by Zitouni and Vatistas [65], it can be deduced to occur once the derivative of the velocity in axial direction becomes zero at the wall. Von Kerczek [67] completed the work of Zitouni and Vatistas [65] by finding analytical solutions for the cases where flow reversal indeed happens, i.e. where the flow is neither purely accelerating nor decelerating.

Several groups have investigated bubble pulsations in a confinement. Cui et al. [68] studied analytically the response of an acoustically driven spherical bubble confined between two parallel plates. They reported that despite the small size of the bubble compared to the gap height, the channel walls affect the bubble oscillations. As the spherical bubble is confined between two plates, decreasing the channel height reduces the resonance frequency and the maximum response amplitude of the bubble. Large bubbles, in contrast, obtain a cylindrical shape bounded by the walls, which is referred to as a cylindrical bubble. Ilinskii et al. [69] obtained a solution for harmonic

cylindrical bubble pulsations in an infinite domain of compressible liquid. They compared their model with the Gilmore equation for cylindrical bubble oscillations and also with the commonly used 2-dimensional Rayleigh-Plesset equation, emphasizing on the role of liquid compressibility.

Considering the application of microbubbles as a pumping mechanism, Ory et al. [70] have simulated the viscous flow induced by growth and collapse of a bubble in a narrow tube. In a similar geometry, a detailed study of the liquid flow field induced by bubble activity has been reported by Ye and Bull [71]. They have conducted direct numerical simulation of microbubble expansion and shrinkage in a long tube, which represents the bubble activity in human vascular system during gas embolotherapy. An improved model with flexible walls is presented in [72]. In another biomedical application of microbubble expansion, the deformation of cells has been modelled with a boundary element method by Tandiono et al. [73], accounting for the membrane tension of the cell. They find that the deformation of a cell, modelled as a liquid droplet, is maximized when the resonance frequency of its surface mode matches the oscillation period of the bubble. This shape frequency is dependent on the density contrast between the liquids as well as the membrane tension. Their model provides a physical explanation for why the shape of an elastic object in a symmetric back-and-forth motion does not return to its original state, in contrast to a fluid particle in a homogeneous flow. Yet, the no-slip boundaries are ignored by Tandiono et al. [73].

In this chapter we focus on unsteady boundary layers generated in close

Figure 3.1: Sketch of the computational domain using axisymmetry and symmetry with respect to the centre-plane of the gap.

proximity of an oscillating bubble between two parallel disks using numerical simulations of the flow. The rapidly expanding bubble is assumed to be created by an intense, focused laser pulse as a method of impulsive energy deposition in the liquid. The simulation results are compared to an analytical expression for the planar flow induced by an oscillating pressure gradient. Finally, a simple experiment using deformable biological cells of different sizes is conducted, where the cells probe the shear stress through their deformation.

3.2 Numerical simulation

3.2.1 Computational domain

The problem modelled is the expansion and shrinkage of an initially spherical bubble, created by a focused, high-power laser pulse. The bubble is located at the centre and between two parallel discs separated by a gap of height $h = 20 \mu\text{m}$. Assuming axisymmetry, we only model the upper quarter of the gap and utilise the symmetry to reduce the computational cost. The computational domain is drawn to scale in figure 3.1, with the length being $\overline{AB} = 80 \mu\text{m}$ and the height $h/2 = \overline{DA} = 10 \mu\text{m}$.

The complex physical process of laser-matter interaction [24] leading to a rapidly expanding bubble is greatly simplified by starting the simulation with a bubble of finite size and filled with non-condensable gas, see [74] and [73]. The liquid motion around the bubble happens on a time scale of tens of microseconds, while the acoustic transients are significant for at most hundreds of nanoseconds [24, 75]. Therefore, for the purpose of resolving the pressure and velocity field in the liquid surrounding the bubble, liquid compressibility effects and acoustic transients can be safely neglected.

3.2.2 Numerical solver specifications

The multiphase flow problem, consisting of a compressible gas and an incompressible liquid, is modelled with the volume of fluid (VOF) method using ANSYS Fluent 14.0 [76], accounting for interfacial tension. The boundary

conditions as depicted in figure 3.1 are along $\overline{DA}$ axis of symmetry, between $\overline{AB}$ symmetry, at $\overline{BC}$ constant pressure p_0 , and no slip at $\overline{CD}$. In the VOF method a single set of momentum equations is solved for all phases, meaning that the pressure and velocity field are shared among all present phases, and the volume fraction of each phase is tracked throughout the domain. The momentum equation is

$$\frac{\partial}{\partial t}(\rho \vec{v}) + \nabla \cdot (\rho \vec{v} \vec{v}) = -\nabla p + \nabla \cdot (\mu (\nabla \vec{v} + \nabla \vec{v}^T)) + \vec{F} \quad , \quad (3.1)$$

where ρ is the density of the phases, $\vec{v}$ is the velocity, p the pressure, and $\vec{F}$ is the surface force. The surface force is modelled as a continuum surface force (CSF) as proposed in [77]. In ANSYS Fluent, the surface curvature is calculated from local gradients in the surface normal at the interface of phases. In solving the governing equations, the material properties, such as density, ρ , or viscosity, μ , are calculated as volume-fraction-averaged properties. Interface tracking is done by coupling the volume of fluid with the level-set method [78, 79]. This allows for accurate interface tracking as well as mass conservation, despite the large density difference between the bubble content and the liquid.

In the present simulation, the pressure-implicit with splitting of operators (PISO) scheme is used for pressure-velocity coupling. The pressure staggering option (PRESTO!) is chosen for the spatial discretisation of the pressure while second order upwind differencing is used in solving the governing

equations. The Geo-Reconstruct method is implemented for discretisation of volume fraction and interface reconstruction. An absolute convergence criteria of 10^{-6} is used for all governing equations and the time step in the simulation is 10^{-8} s. To assure the solution is mesh-independent, the simulations were conducted with two different grids, 5000 and 80000 elements, with an averaged element size of 0.4 and 0.1 μm , respectively. Both solutions are nearly identical, see figure 3.3, therefore grid independence is obtained. All results reported here are using 0.1 μm element size, while the mesh is refined in regions with high gradients, e.g. the initial bubble-liquid interface and near the channel walls. The simulation time on a 2.1 GHz Intel Core i7 personal computer with 8 GB of RAM is approximately 24 hours for a single cycle of bubble expansion and shrinkage.

The initial pressure in the gas bubble of $R(t = 0) = 5 \mu\text{m}$ is 100 bar. The liquid is initially at atmospheric pressure $p_0 = 1$ bar. The liquid is water with a density of $\rho_l = 998.2 \text{ kg/m}^3$ and a dynamic viscosity of $\mu_l = 1 \times 10^{-3} \text{ Pa s}$, while the gas viscosity is $\mu_b = 1.34 \times 10^{-5} \text{ Pa s}$, corresponding to water vapour. The ideal gas law is used for calculation of density of the compressible bubble content. This density calculation requires the solution to the energy equation [76], which is shared among both phases, similar to the momentum equation (3.1). The temperature field is initially assumed to be uniform in the computational domain and at 300 K. The interfacial tension coefficient is $\gamma = 7.2 \times 10^{-2} \text{ N/m}$.

Figure 3.2: Evolution of the bubble shape and the radial velocity profile at $r = 50 \mu\text{m}$ from the centre of the bubble. Please note the reversal of the flow direction near the boundaries occurring before the flow in the centre of the gap.

3.3 CFD results and analysis of the flow field

3.3.1 Bubble evolution and liquid velocity profile

Figure 3.2 depicts the temporal evolution of the bubble shape over a period of $10 \mu\text{s}$ in a gap of $h = 20 \mu\text{m}$. The spherical bubble quickly grows into a pancake shape, forming thin liquid films at the upper and lower solid walls. The maximum bubble radius is obtained after $4 \mu\text{s}$. During bubble shrinkage, the convex interface flattens and only increases curvature after $8 \mu\text{s}$; that is when the internal pressure increases and dampens the collapse of the bubble. The bubble collapses to its minimum volume at $10 \mu\text{s}$ and rebounds afterwards

Figure 3.3: Comparison of experimental cylindrical bubble dynamics [4] (dashed line with squares) with the VOF solution (filled line and dashed line for two computational grids). Additionally, the solution to the 2D Rayleigh-Plesset equation starting from the maximum bubble radius is shown with a dotted line.

(not shown here). Right to the bubble profile in figure 3.2, the radial velocity profile at a distance of $r = 50 \mu\text{m}$ from the bubble centre is shown. Initially, the liquid is at rest and rapidly develops into a plug flow with strong wall shear stress. Gradually, a more parabolic profile develops. At the later expansion stage, the velocity near the walls is reduced and its direction is reversed, while the liquid velocity in the centre of the gap is still outwards and positive. This flow reversal near the boundaries will be discussed in detail in section 3.3.2.

Figure 3.3 shows the projected bubble radius from figure 3.2 and compares

it with the experimentally determined radii [4] and a 2-dimensional Rayleigh-Plesset equation, e.g. see [61]. For the chosen initial conditions, i.e. radius $R(t = 0)$ and initial gas pressure, we find a good agreement between the VOF simulation and the experiment. In particular, the asymmetry of the bubble oscillation, having a faster expansion than the collapse, is captured in the VOF simulation. Interestingly, this asymmetry was attributed previously and in a different geometry to thermal effects [80]. We speculate that the faster expansion and slower collapses is caused by the asymmetry of the flow: during expansion a plug flow prevails which has a lower overall viscous pressure head. Later the boundary layers develop, retarding the flow and therefore lengthening the collapse.

The final stage of bubble collapse cannot be captured with the VOF model as it assumes that the liquid and bubble content are immiscible, while in experiments the laser generated bubble mainly consists of condensible vapour. Our simulation predicts a milder collapse with re-expansion of the bubble; however, the experiments find a much smaller minimum bubble radius followed by fragmentation of the bubble.

3.3.2 Flow reversal due to adverse pressure gradient

For clarity, we have divided the results into four stages and show the evolution of the liquid pressure, $p(r)$, and radial velocity $v_r(r = 50 \mu\text{m}, z, t)$ in figure 3.4. The four stages from left to right are early expansion, late expansion, shrinkage, and rebound. The upper frames in figure 3.4 show the unsteady

Figure 3.4: Evolution of pressure (top row) and radial velocity (bottom row) during early (a) and late expansion (b) and early (c) and late collapse (d). The times for the pressure and velocity profiles are indicated in the upper legend of each column.

pressure in the centre of the channel from $r = 0$ to the outlet, i.e. $p(r, z = 0, t)$. The axial pressure gradient is negligible in comparison to the significant pressure variation in the radial direction ($\partial p / \partial z \ll \partial p / \partial r$). Therefore, the pressure near the wall is nearly identical to the centre, see also figure 3.6. The bubble-liquid interface can be easily identified by the small pressure jump due to surface tension in the pressure profiles in figure 3.4. The lower frames of figure 3.4 show the velocity profile in the liquid at a fixed distance

of $r = 50 \mu\text{m}$ from the bubble centre.

During the first stage, figure 3.4a, $0 < t < 1.5 \mu\text{s}$, the initial gas pressure of 100 bar accelerates the liquid outwards from rest to almost 7 m/s. In consequence, the gas pressure drops within $1.5 \mu\text{s}$ below the outlet pressure. During this time a flat-top velocity profile develops.

The second stage, figure 3.4b, corresponds to the deceleration of the flow to the maximum bubble volume at $t = 4.3 \mu\text{s}$. As the pressure in the bubble drops, an adverse pressure gradient develops ($\partial p / \partial r > 0$). This leads to detachment of the boundary layer, where a reversed flow at the boundary sets in. This flow reversal is clearly visible at $t = 4 \mu\text{s}$ in the lower frame of figure 3.4b. At this stage, the flow profile possesses an inflection point; the liquid in the centre of the channel continues to flow towards $+r$ while at the boundaries the flow is directed towards the bubble.

In the third stage, figure 3.4c, the bubble shrinks, i.e. a net flow towards the bubble sets in and eventually a purely negative velocity profile builds up. In this stage the pressure gradient is stabilising the boundary layer and the pressure in the bubble steadily builds up.

In the last stage, the bubble reaches its minimum volume and rebounds, figure 3.4d, $8 < t < 10 \mu\text{s}$. Similar to stage one, the internal bubble pressure is higher than the liquid pressure, but here the liquid flow is towards the bubble. Therefore, the pressure gradient once again opposes the liquid flow and leads to flow reversal near the channel walls. The flow could be described similar to stage two, but with opposite signs ($\partial p / \partial r < 0$). The liquid flow

is eventually reversed at the walls in $t = 10 \mu\text{s}$, minimum bubble volume is reached, and the bubble begins to rebound.

3.3.3 Vorticity generation

Figure 3.5: The liquid streamlines during boundary layer separation, i.e. from $t = 3$ to $5.5 \mu\text{s}$. The radial velocity profile is plotted at $r = 50 \mu\text{m}$. Two counter rotating vortices form, move upwards, and merge.

Figure 3.5 depicts the instantaneous streamlines during the build up and decay of the adverse pressure gradient together with a radial velocity profile at $r = 50 \mu\text{m}$, corresponding to the late stage of bubble expansion in figure 3.4b. At time $t = 3.5 \mu\text{s}$ formation of vortices near the channel walls is observed. From $t = 3$ to $4.5 \mu\text{s}$ the adverse pressure gradient supports the

detachment of the boundary layer, leading to a recirculating flow. Due to symmetry two vortex rings are formed which are transported towards the centre of the channel. In figure 3.5 the horizontal distance between the vortex core and the bubble wall remains approximately constant at about $5\text{ }\mu\text{m}$. Additionally, the separation point of the boundary layer is relatively stable in space, here around $r = 44\text{ }\mu\text{m}$, and only moves toward the bubble in the last frame of figure 3.5, i.e. at $t = 5.5\text{ }\mu\text{s}$ when the bubble gains inward speed. The vortex rings merge as they migrate towards the centre of the channel, see figure 3.5, $t = 5.0$ and $5.5\text{ }\mu\text{s}$. From then on, the radial flow and the pressure gradient are aligned, stabilising the boundary layer.

3.3.4 Flow field contours

For clarity, we present a sequence of contour plots of the liquid flow during the bubble expansion and shrinkage in figure 3.6, in particular the pressure $p(r, z, t)$, the radial velocity $v_r(r, z, t)$, and the vorticity magnitude $|\nabla \times \vec{v}|$. The flow is separated into the same stages from section 3.3.2.

At the early expansion of the bubble, ($t = 0.5\text{ }\mu\text{s}$) in figure 3.6, the high pressure in the bubble accelerates the liquid outwards with a flat top radial velocity distribution with no vorticity in the central flow but shear and vorticity in the proximity of channel walls.

The late expansion and early shrinkage stage of the bubble ($t = 4.5\text{ }\mu\text{s}$) is accompanied by the presence of reversed flow and boundary layer separation due to adverse pressure gradient in the liquid. Close to the walls, the outward

Figure 3.6: Sequence of contour plots of the flow field during bubble expansion and shrinkage. Pressure field in both phases, radial velocity in the liquid, and vorticity in the liquid are shown at left, centre, and right, respectively. The bubble-liquid interface is shown by the black line, i.e. contour of 0.5 liquid volume fraction. The colour bar for each figure is given underneath. Each figure is plotted with its specific colour map to better distinguish the flow field. Results are shown for a sample time instant of each different stage of the flow.

flow is reversed and two counter rotating vortices are observed. The flow reversal and circulation in this stage cause significant shear stress near the walls.

As the inward flow sets in during the shrinkage ($t = 6.5 \mu\text{s}$), the two vortices migrate towards the channel centre until they eventually merge and circulation is not visible in the flow field any more. The liquid flow towards the bubble is not opposed but stabilised by the pressure gradient.

Finally, the shrinkage causes the pressure to build up in the bubble ($t = 9.5 \mu\text{s}$), which in turn acts as an adverse pressure gradient, opposing the inward liquid flow, and the bubble rebounds afterwards.

It is worthy to recall that the solution to the Navier-Stokes equations using the volume of fluid method is shared among all phases, i.e. the water vapour in the bubble and its surrounding liquid water. This means that all variables, including the pressure and velocity, are obtained for both phases. In this manuscript, however, we focus only on the liquid flow field induced by the shared pressure field. Therefore in figure 3.6, the radial velocity field and the vorticity field are shown only in the liquid.

3.3.5 Effects of initial pressure, channel height, and viscosity

We now examine the importance of the initial gas pressure, the gap height, and the viscosity of the liquid independently. In figure 3.7 the temporal evo-

Figure 3.7: The effect of initial internal bubble pressure p_i . Increasing the initial pressure results in a faster flow field and a larger bubble size, but important features such as flow reversal and vorticity generation are preserved. All simulations are done with $h = 20 \mu\text{m}$ and $\nu = 1 \text{ cSt}$.

Figure 3.8: Varying the channel height h from 20 to 30 μm . Flow reversal and vorticity generation near the walls are observed for a cylindrical bubble despite the gap height, but are more prominent in tighter confinements. Both simulations are done with $p_i = 100 \text{ bar}$ and $\nu = 1 \text{ cSt}$.

Figure 3.9: The effect of kinematic viscosity ν . Increasing the liquid viscosity, while keeping $p_i = 100$ bar and $h = 20$ μm constant, slows down the flow field considerably. The film thickness separating the bubble and channel walls is larger in a more viscous liquid.

lution of the bubble shape and the radial velocity at $r = 50$ μm are compared for increasing internal bubble pressure from $p_i = 25$ to 100 bar. The channel height $h = 20$ μm , the initial bubble size of 5 μm , and the liquid viscosity of 1 cSt are kept constant. By increasing the potential energy through internal pressure, the liquid is accelerated to a higher kinetic energy. The general features of the flow are observed across the tested pressure range, i.e. flow reversal and formation of vortices.

By increasing only the channel height, shown in figure 3.8, the bubble extends less in the radial direction. Meanwhile, it is observed that reversed flow and vorticity generation occur as long as the bubble covers the gap height. We can conclude that these flow features robustly occur for sufficiently large and rapidly expanding bubbles, regardless of the channel height.

As it is shown in figure 3.9, increasing the viscosity reduces the expansion velocity and consequently the radial velocity gradient outside the boundary layer. Assuming a stationary outward flow with velocity U at the top of the

boundary layer, we can relate the pressure gradient to the outward velocity as $\rho U dU/dr = -dp/dr$ based on the Bernoulli equation. Knowing that increasing the liquid viscosity reduces the outward flow velocity U and its radial gradient dU/dr , we can conclude that the pressure gradient also decreases at a higher viscosity. Since the pressure gradient is the driving force for separation, flow reversal is expected to be delayed or hindered at a higher viscosity. Figure 3.9 supports this idea: with increasing liquid viscosity the overall velocity is reduced. While flow reversal is still observed at the late expansion stage, it is less prominent in a more viscous liquid. Additionally, we observe that increasing the viscosity thickens the liquid film that separates the bubble from the channel walls.

3.4 Analytical approximation for liquid velocity profile: pulsating pressure-driven flow

The CFD solution reveals a complex flow pattern, yet we speculate that the main characteristics of the flow can be captured with fundamental solutions of unsteady flows. In the early stage of bubble expansion, the liquid in the channel is accelerated from rest to a velocity profile qualitatively similar to a Poiseuille flow. This acceleration of the liquid is accompanied by a rapid reduction in pressure inside the bubble.

Except for the short initial period described above, the pressure gradient acting on the liquid has a pulsating behaviour for the majority of the bubble expansion and shrinkage cycle. At the onset of bubble shrinkage, the outlet pressure is higher than the pressure inside the bubble. Therefore, the flow near the boundaries is opposed by an adverse pressure gradient and is eventually reversed. As the inward flow sets in, the bubble shrinkage results in an increase in density and pressure of the gas, which in turn leads to opposition against the inward flow. This situation resembles an oscillating pressure gradient acting on the liquid between two parallel plates. Here we compare the simulation results with an analytical solution for a 2-dimensional flow within a gap induced by an oscillating pressure gradient. Although the numerical solution is obtained for an axisymmetric geometry, we simplify the geometry to a planar flow in order to obtain an analytical solution.

Using a harmonic function for the pressure gradient, i.e. $\partial p / \partial x = \Delta p / \Delta x \sin \omega t$, we obtain the transient solution to the 2-dimensional Navier-Stokes equation for incompressible flow between two parallel plates. To be succinct, we do not repeat the steps here and refer the reader to section 7.8.3 of [81] for details of obtaining the analytical solution. The transient velocity profile $u(y, t)$ induced by this pulsating pressure-driven flow can be expressed as

$$u(y, t) = \frac{-1}{\rho \omega} \frac{\Delta p}{\Delta x} (f_c(y) \cos \omega t + f_s(y) \sin \omega t) \quad , \quad (3.2)$$

where $f(y)$ is given as a complex function with real and imaginary parts,

Figure 3.10: Comparison of (*left*) the VOF simulation of the radial velocity with (*right*) the analytical approximation of the planar velocity profile during bubble shrinkage (3.2).

$f_c(y)$ and $f_s(y)$, respectively. The complex function $f(y)$ is formulated as

$$f(y) = f_c(y) + if_s(y) = 1 - \frac{\cosh\left(\left(\frac{2y}{h} - 1\right)\frac{h}{2}\sqrt{\frac{-i\omega\rho}{\mu}}\right)}{\cosh\left(\frac{h}{2}\sqrt{\frac{-i\omega\rho}{\mu}}\right)} . \quad (3.3)$$

The pulsating velocity profile for the bubble shrinkage is approximated with the solution from (3.2) and is compared with the CFD solution in figure 3.10. A value of $\Delta p/\Delta x = 2.5 \times 10^4$ bar/m with a period of $2\pi/\omega = 10 \mu\text{s}$ is used for the harmonic pressure gradient. This simplified analytical description is able to capture the general features of the bubble-driven flow during bubble shrinkage. Development of a reversed flow near the channel walls is observed, which later on reverts completely towards the bubble. Later

on, the pulsating pressure gradient opposes and decelerates the inward flow, which is similar to the findings in the numerical solution.

3.5 Experiments: deformation of cells

The numerical simulation suggests that deformable objects located within the boundary layer and close to the bubble may be stretched considerably by the shearing flow. We test this hypothesis with deformable objects thin enough to be entrained in the boundary layer, and larger objects which cover most of the channel. As deformable objects we use biological cells in a suitable aqueous solution (saline buffer solution). The experiments were stimulated from previous work with red blood cells [53, 54]. Red blood cells are thin, bi-concave cells with a diameter of about $8\text{ }\mu\text{m}$ and a thickness of less than $2\text{ }\mu\text{m}$. After they are placed in the microfluidic gap, they sediment to the bottom because of their higher density. Figure 3.11 shows a typical configuration of red blood cells (RBCs) shortly before (top) and after the bubble oscillation (bottom), viewed from the top. RBCs near the bubble are largely stretched. Details of the experiment are available in [53, 54].

The simulations show that the boundary layer where stretching may occur is approximately $1\text{ }\mu\text{m}$ thick, which fits well with the thickness of red blood cells. Yet, one would expect that cells covering a large part of the gap would behave differently. This is tested in a second experiment where we insert colon cancer cells (RKO) which have a spherical shape with a diameter of 10

Figure 3.11: Deformation of red blood cells (RBC) due to the cavitation-induced flow in a $20\mu\text{m}$ thin gap. RBCs are shown just before a laser generates a single cavitation bubble and after the bubble collapses. RBCs are significantly stretched in this flow. Please see [54] for details of the experiment.

Figure 3.12: Deformation of colon cancer cells (RKO) due to the cavitation-induced flow in a 20 μm thin gap. RKO are much less deformed compared to RBCs while both are exposed to the same flow, see figure 3.11.

to $15\text{ }\mu\text{m}$, comparable with the $20\text{ }\mu\text{m}$ gap height. Figure 3.12 reveals that the RKO cells are transported with the flow but do not visibly stretch.

As the thin RBCs are comparable to the size of the vortex, they become exposed to a shearing force near the wall. From the simulation results, neglecting the presence of cells, we obtain a value of approximately $2 \times 10^6\text{ 1/s}$ for the magnitude of the shear strain rate, $1/2(\partial v_r/\partial z + \partial v_z/\partial r)$, near the walls, where the RBCs are located. This can be related to a stretching of a fluid particle assuming a characteristic height and duration. With a size of $2\text{ }\mu\text{m}$ and a duration of $2\text{ }\mu\text{s}$ we obtain a length increase of $8\text{ }\mu\text{m}$, a value which is of the order of the observed stretching in the experiments, e.g. [54]. Detailed analysis and quantitative measurements for stretching of RBCs with cavitation bubbles are presented in [54].

In contrast, the spherical RKO cells extend into the centre of the channel and are advected with the flow. The shear stress outside the boundary layer is considerably lower, leading to less deformation. We find about 10% strain at the centre of the channel over a thickness of $1\text{ }\mu\text{m}$. The resulting deformation of these cells may be well within the resolution limit of the imaging optics.

3.6 Discussion and Conclusion

A notable feature of the present numerical solution to the flow induced by a cavitation bubble in a narrow gap is the presence of boundary layer separation, reversed flow, and recirculation. In similar geometries, flow reversal

has been reported by other investigators. In the case of *steady*, radial flow between two flat disks, extension of Von Kerczek [67] on the power-series solution obtained in [65] shows the existence of boundary layer separation in radial diffusers. Prior to that, flow visualization done in [82] on radial flows with a steady influx showed the nucleation, growth, migration, and eventual decay of vortices in the outward flow. Mochizuki and Yang [82] also obtained finite-difference solutions to the unsteady vorticity transport equation which was in agreement with their experimental findings. In an analytical treatment of the unsteady axisymmetric flow between two flat disks, Elkouh [64] found that the radial velocity distribution induced by an oscillating source/sink exhibits flow reversal near the walls.

In a different confining geometry, an expanding bubble in a tube, direct numerical simulation of the flow reveals the presence of a recirculation region between the wall and the core flow at the end of bubble growth and beginning of shrinkage [71]. However, Ye and Bull [71] do not discuss this flow feature in greater depth, and instead focus on the wall pressure and shear stress which is aligned with the main application of their study, gas embolotherapy. Our findings about the wall shear stress due to bubble induced liquid flow in a microfluidic geometry agrees well with their report; thus, we refer the reader to [71] for a detailed discussion about wall shear stress. Interestingly, even inviscid boundary layer simulations may explain the deformation of dispersed droplets (being a simple model for a cell) if their interfacial tension is accounted for, see [73]. Tandiono et al. [73] simulated a

droplet at some distance from an oscillating bubble in an infinite liquid. The interfacial tension introduces a second timescale besides the period of bubble oscillation, i.e. it causes the droplet to oscillate in a surface mode.

Utilizing a viscous model in the present study, we find that during the late expansion of a cavitation bubble in a narrow axisymmetric gap, the adverse pressure gradient leads to boundary layer separation and flow reversal in the proximity of the walls. Although such complex flow patterns in axisymmetry require computational treatment of the problem, a simplified analytical expression for planar pulsating flow supports our understanding of the series of events we observe in the numerical simulations and experiments. A prominent result of the observed flow reversal and vorticity is generation of strong shear stress in the liquid close to the boundaries. The results indicate that some flows created by oscillating bubbles in narrow gaps may not be accurately described with inviscid potential flow in cylindrical symmetry. Particularly, in studying the mixing and emulsification of flows or the deformation of elastic objects, such as yeast cells [83], it is necessary to account for the vorticity generation and transport. For both applications, it would be interesting to extend the present simulation to two-way coupling.

Chapter 4

Synthetic jet generation by high-frequency cavitation

4.1 Introduction

Bubbles rarely collapse spherically; they often induce liquid jets at the last stage of their collapse. Rapid microjets are generated when bubbles collapse near rigid [32, 84], elastic [85], and free boundaries [86]. Jets generated during the collapse of cavitation bubbles have been studied extensively in the literature through experiments, numerical simulations, and modelling [87, 88]. These microjets, which could be as fast as 100 m/s, can penetrate into biological tissue [89, 90] or damage hydraulic machinery [91].

During stable bubble oscillations, vorticity generation within the boundary layer surrounding the bubble results in steady streaming in the liquid.

This streaming mechanism is comprehensively reported in the literature under the title of cavitation microstreaming [92–96]. Even gentle and linear bubble oscillations can induce streaming flows with enough shear force to manipulate and deform small objects [97]. By pairing up oscillating bubbles and micro-particles, the stream can be controlled to transport objects within a microfluidic chamber, eliminating the need for microchannel fabrication [98]. The stream can be collected for micropumping, for instance by placing a capillary tube above the oscillating bubble [99]. The interplay between the streaming-induced drag force and the Bjerknes forces enables cavitation microstreaming to be applied in selective particle trapping [100]. From switching [101] and micromixing [102] to a variety of other applications [103], lab-on-a-chip devices benefit from cavitation microstreaming.

While streaming from small-amplitude bubble oscillations is well known, it has received less attention with regard to strongly collapsing cavitation bubbles, i.e. inertial cavitation. These bubbles can be conveniently generated using pulsed lasers [4, 24], shock waves [6, 104], electrical discharge [105], or localized heating [52, 106]. Inertial cavitation is applied in various therapeutic and diagnostic medical procedures [16], for instance in laser surgeries and tissue ablation, such as ocular surgery [107], fragmentation of urinary stones [18], and removal of arterial blood clots [19]. In most applications involving laser-induced cavitation, the laser repetition period is much larger than the bubble lifetime. Yet in some applications, such as nanoparticle synthesis, kilohertz or even megahertz laser systems are used [108].

In this chapter, we study the flow induced by steadily generated cavitation bubbles at the cross over where the repetition period of the cavitation event is of the same order as the life time of the bubble. An experimental setup is presented to generate high-frequency cavitation and visualise the cavitation-induced flow using particle image velocimetry. We characterise the flow field as a synthetic jet and conclude with a model for the bubble formation process.

4.2 Experimental setup

To generate cavitation bubbles with high spatial and temporal repeatability, we use a pulsed thulium (Tm) laser (wavelength $\lambda=1.96\text{ }\mu\text{m}$) running at a high repetition rate [109]. The pulsed laser is delivered with a single-mode fibre optic of diameter $d = 250\text{ }\mu\text{m}$ into a transparent glass container ($75 \times 45 \times 20\text{ mm}$) filled with deionized water. The linear absorption of mid-infrared light by the liquid layer adjacent to the fibre tip leads to heating and eventually vaporisation, i.e. generation of a cavitation bubble. The dynamics of this bubble is recorded using a high-speed camera (Photron, FASTCAM SA-X2) with diffused back-illumination from an LED light source (REVOX, SLG-150V). The experimental setup is shown schematically in figure 4.1.

The flow field is visualised by seeding water with fluorescent microparticles which are $5\text{ }\mu\text{m}$ in diameter and 1050 kg/m^3 in density (Thermo Scientific, Fluoro-Max 36-2). Particle fluorescence is excited with a green light sheet from a continuous (CW) laser (Nd:YAG, $\lambda=532\text{ nm}$) focused with a cylin-

Figure 4.1: Experimental setup for high-speed photography and particle image velocimetry.

drical lens. The laser sheet illuminates the flow along the fibre axis in the axial and radial directions. Specular reflections are absent in the images due to a filter blocking the laser wavelength. The velocity field is extracted from the microparticle images through standard particle image velocimetry (PIV) analysis using the PIVLab toolbox [110].

Before each experiment, the laser pulse duration, τ , and its repetition rate f are measured with an InGaAs photodiode (Thorlabs, FGA20) connected to an oscilloscope (LeCroy, WaveRunner 64Xi-A). The average power of the laser, P , is measured with a power meter (Opir, 30A-BB-18), determining the energy of each pulse as $E = \frac{P}{f}$. The experiments are conducted with three different laser repetition rates, namely $f = 20, 30$, and 40 kHz. The laser power is varied between $P = 2.5$ and 20 W for each repetition rate and the pulse duration τ is between 800 ns and 1 μ s.

4.3 Jet generation by high-frequency cavitation

4.3.1 Bubble formation

Repetitive cavitation is realised with the pulsed laser running at a high repetition rate, as shown in figure 4.2. Upon delivery of the laser pulse, the liquid layer adjacent to the fibre tip absorbs the pulse energy and undergoes explosive vaporisation, forming an almost hemispherical bubble. This bub-

Figure 4.2: High-speed recordings of bubble formation with a high-repetition-rate pulsed laser delivered through a fibre optic. High-frequency cavitation occurs due to linear absorption of the laser energy by the liquid layer adjacent to the fibre tip, forming bubbles that expand and collapse before arrival of the subsequent laser pulse. The laser is operating at $f = 40$ kHz repetition rate and $P = 5$ W average power.

ble rapidly expands before collapsing under the ambient pressure. At this laser setting, the bubble has already collapsed before an identical bubble is generated with the next laser pulse.

The mechanism of laser induced cavitation depends on the pulse duration, the irradiation volume, and the absorption coefficient of the irradiated medium [17]. In our experiments, the pulse duration τ is longer than the stress relaxation time $\tau_{st} = \frac{d}{c}$, but shorter than the thermal relaxation time $\tau_{th} = \frac{d^2}{4k}$, i.e. $\tau_{st} < \tau < \tau_{th}$. Here c is the speed of sound and k is the thermal diffusivity in the liquid medium. This condition, known as thermal confinement, means that strong shock waves are not emitted from the irradiation site, yet the temperature in the irradiated volume can increase above the spinodal limit before phase explosion forms a vapour bubble [28].

A few seconds after switching on the laser, the bubble formation process becomes highly repeatable. Meanwhile, by increasing the laser power, the bubble may show a different periodicity from the laser pulse. We discuss the effects of laser power and repetition rate on bubble dynamics in section 4.4, but first we focus on the flow field induced by cavitation in the experiments.

4.3.2 Jet characteristics

Flow field visualisation reveals a stream of particles moving in the axial direction away from the fibre tip. Close to the fibre tip, the motion is complex, with particles following the oscillatory bubble expansion and collapse. Particles that are away from the fibre tip in the radial direction become gradually

Figure 4.3: Time-averaged velocity field induced by high-frequency cavitation visualised by PIV. The time-averaged flow field resembles a continuous free jet directed away from the cavitation site. Semi-transparent photos of bubble dynamics and the fibre optic are shown to scale. The velocity magnitude, vector field, and axial velocity profiles are plotted. The laser is operating at a repetition rate of $f = 40$ kHz with an average power of $P = 20$ W.

Figure 4.4: Transient and time-averaged axial and radial velocity, u and v , at point $(x = 2 \text{ mm}, r = 0)$ along the jet over a few periods of the cavitation event $T = f^{-1}$. The laser is operating at a repetition rate of $f = 40 \text{ kHz}$ with an average power of $P = 20 \text{ W}$.

attracted towards the fibre axis, accelerating as they approach the tip. Since the flow varies with time, we resort to analysing the time-averaged flow field to characterise the stream. Due to imaging limitations, we analyse the flow only at a distance of approximately three fibre diameters from the fibre tip.

Figure 4.3 shows the time-averaged flow field, while the time-variant velocity at a sample point along the stream is plotted in figure 4.4. On the left of figure 4.3 snapshots of the bubble and fibre tip are overlaid to scale. The time-averaged velocity field reveals a pronounced jet directed away from the fibre tip with an almost axisymmetric flow field, as indicated by the radial profiles of the axial velocity $u(x, r)$. The velocity profiles demonstrate that the jet expands as it travels away from the cavitation site. While the cavitation-induced jet is transient, the time-averaged flow field resembles a steady jet [111].

To study this flow field in more detail, we compare the cavitation-induced jet with a free axisymmetric jet without swirl. Sufficiently far from the origin, free jets are characterised by a self-similar velocity profile [112], i.e. their axial velocity is $\frac{u}{u_m} = (1 + \eta^2)^{-2}$, where $\eta = \sigma \frac{r}{x}$ is the similarity variable. Here σ is a constant and u_m is the axial velocity along the jet centreline, i.e. $u(x, r = 0)$. This similarity solution is plotted in figure 4.5 with a solid line, i.e. $\frac{u}{u_m} = f(\frac{r}{b})$. Here, b represents the jet half-width, where the velocity drops to half of the jet centreline velocity u_m , marked in figure 4.3. The jet half-width grows linearly along free jets, i.e. $b \propto x$ [111]. The time-averaged velocity profile of the cavitation-induced jet in the experiments matches well

Figure 4.5: Comparison between the self-similar velocity profile of the continuous free jet (solid line) and the time-averaged cavitation-induced jet (symbols). The laser is operating at $f = 40$ kHz and $P = 20$ W.

Figure 4.6: Decay of the jet centreline velocity $u_m = u(x, r = 0)$ for free jets (lines) and cavitation-induced jets (symbols). Origin of the free jets are fixed at the fibre tip and the cone angles are obtained from the experimental data. The laser is operating at $f = 40$ kHz and $P = 20$ W.

with the similarity solution for the axisymmetric free jet, particularly close to the jet centreline, $r = 0$. This is shown in figure 4.5, where the symbols represent various cross-sections along the flow marked in figure 4.3.

A second prominent property of the similarity solution for a free jet is that the centreline velocity drops as $u_m \propto \frac{1}{x}$ along the jet [111]. We put this property to test for jets generated at different laser powers. By placing the origin of the free jet at the centre of the fibre tip and estimating the cone angle of the jet from the experimental data [112], agreement is obtained for the time-averaged velocity drop $u_m(x)$ along cavitation-induced jets. The agreement closer to the tip improves as the laser power increases. Additionally, figure 4.6 reveals that with increasing laser power the cone angle of the continuous jet decreases, i.e. a stronger, more slender jet is produced.

4.3.3 Mechanism of synthetic jet generation

Despite the transient cavitation activity at the fibre tip, a time-averaged jet is produced which resembles an axisymmetric free jet. Looking in closer detail at the bubble dynamics, shown in figure 4.2, the cavitation bubble expands with its centre of mass moving away from the fibre tip, and during shrinkage moves towards it. During the collapse of a translating cavitation bubble a re-entrant microjet is generated in the direction of its motion [32, 113, 114], which is towards the fibre tip. Yet, we observe a large scale flow away from the fibre. The mechanism that drives the flow away from the boundary may be explained by the repelling motion of a stable vortex ring. The course

Figure 4.7: Schematic illustration of a collapsing bubble that directs the flow away from a rigid boundary. Important flow features are marked in red. (a) The collapsing bubble generates a rapid microjet towards the boundary. (b) After jet impingement, remnants of the microjet compete with a strong radial inflow, forming an annular stagnation ring and a rebounding splash. (c) The splash develops into a toroidal vortex. (d) The vortex moves away from the boundary and carries the bubble fragments along the axis of symmetry.

of events leading to the generation of this vortex ring after each cavitation event is schematically illustrated in figure 4.7. After jet impingement on a boundary due to the collapse of a sufficiently close bubble, the re-entrant jet competes with the radial inflow as the bubble is still shrinking. This leads to the formation of an annular stagnation ring. We speculate that the radial inflow is reflected away from the boundary as a splash [115, 116]. This flow may then close into a vortex ring which moves away from the fibre tip due to its rotation direction. An experimental visualization of a similar process is reported by Reuter et al. [117] in front of a large rigid boundary.

Streaming driven by trains of vortices has been comprehensively studied in the literature under the title of synthetic jets [118]. Synthetic jets are usually generated by alternating suction and ejection of fluid across an orifice [119]. In contrast to continuous jets, synthetic jets are produced without

injecting mass into the system. These jets work merely due to transfer of momentum, typically by oscillations of a membrane or diaphragm [120]. In the current setup, high-frequency cavitation provides the momentum for a synthetic jet, generating a stream with zero net mass flux. As the jet travels away from the cavitation site, it eventually spreads out, directing streamlines back to the cavitation site which complete a loop and feed the jet. In addition to pulsed lasers, spark generators and microheaters may be used to generate cavitation-based synthetic jets.

4.4 Bubble formation and dynamics

4.4.1 Secondary bubble formation

An increase in the energy input to the system not only strengthens the jet, as shown in figure 4.6, but also affects the bubble dynamics [121]. Figure 4.8 shows five series of images of bubble formation and collapse, where the laser power is increased from $P = 2.5$ to 20 W. The results are shown at a repetition rate of $f = 40$ kHz, but identical bubble dynamics are observed at other tested repetition rates. As it is shown in the two left rows of figure 4.8, at low laser power, a small bubble forms and collapses on the tip before the next laser pulse arrives. An increase in the laser power may drastically affect the bubble dynamics, as shown in the three right rows of figure 4.8. If the pulse energy is sufficiently high, the initial bubble does not collapse when the next pulse arrives. This allows the laser pulse to be transmitted through

Figure 4.8: The effect of increasing laser power on bubble formation with a high-repetition-rate pulsed laser operating at $f = 40$ kHz. By increasing the average laser power from left to right, the bubble size increases. Above a certain threshold, the formation of a secondary bubble at the tip of the initial hemispherical bubble is observed. This occurs when the lifetime of the first bubble is larger than the period of pulse repetition.

the shrinking bubble, leading to vaporisation of the liquid at its distal end and formation of a secondary bubble. While the first bubble shrinks and collapses on the fibre tip, the second one expands and moves away. This repelling motion, caused by out-of-phase oscillation of the two bubbles, may affect the jet generation mechanism [122–124]. Small vapour fragments of the secondary bubble are still visible when the next laser pulse is absorbed by the liquid layer adjacent to the fibre tip, restarting the bubble formation cycle. This double-bubble system, composed of the initial and secondary bubbles, follows a cycle with twice the period of pulse repetition.

4.4.2 Model for bubble growth

Here, we develop a model to predict the growth of the vapour bubble to its maximum size based on the laser pulse energy. Under thermal confinement, the laser pulse heats up the liquid over the fibre core to the critical point, where there is no phase boundary between the liquid and the vapour [125]. Thus, we start from an initial nucleus at the critical state where the pressure and density are $p_i = 218 \text{ atm}$ and $\rho_i = 322 \text{ kg/m}^3$. The growth dynamics is modelled as a hemispherical cavitation bubble undergoing explosive expansion without heat transfer. Therefore, the evolution of the bubble radius, $R(t)$, can be modelled with the Rayleigh-Plesset equation [126],

$$R\ddot{R} + \frac{3}{2}\dot{R}^2 = \frac{-p_\infty + p_g}{\rho_l} - \frac{4\nu_l\dot{R}}{R} - \frac{2\sigma}{\rho_l R} \quad , \quad (4.1)$$

Figure 4.9: Maximum bubble radius $R_{\max}$ as a function of the laser pulse energy E . The experimental results are compared with the model presented in section 4.4.2 for three different laser repetition rates.

Figure 4.10: Laser pulse energy E needed for secondary bubble formation as a function of the pulse repetition rate f . The grey colour marks the region and experimental data with only hemispherical bubble formation, and white denotes the formation of secondary bubbles.

where dots denote derivatives with respect to time. The material properties are the density of the liquid, $\rho_l = 10^3 \text{ kg/m}^3$, the kinematic viscosity of the liquid, $\nu_l = 10^{-6} \text{ m}^2/\text{s}$, and the surface tension $\sigma = 0.072 \text{ N/m}$. With negligible heat transfer and phase change over the bubble interface due to its rapid growth, the pressure in the bubble, p_g , evolves adiabatically, i.e. $p_g = p_i(\frac{R_i}{R})^{3\gamma}$, where $\gamma = \frac{4}{3}$ is the adiabatic exponent. The bubble expansion is countered by the ambient pressure, $p_\infty = 1 \text{ atm}$.

If we have the initial state of the vapour, the initial bubble radius R_i can be determined from the absorbed energy of the laser pulse E_{abs} , which is used to raise the state of the liquid to the critical point. The specific energy, i.e. energy per unit mass of the liquid, needed for this change of state is $h = 2.84 \times 10^3 \text{ kJ/kg}$. To calculate this, we use a thermodynamic path in which water is first heated up isobarically from ambient conditions to 373 K, phase change from liquid to vapour occurs at 373 K, the vapour is heated up under ambient pressure to the critical temperature of 647 K, and finally the steam at critical temperature is pressurized isothermally to the critical pressure of 218 atm [127]. Assuming that the initial vapour bubble at the fibre tip has a hemispherical shape, we can write the energy balance as

$$\frac{2}{3}\pi R_i^3 \rho_i h = E_{abs}. \quad (4.2)$$

In the current experiments, the liquid absorbs the laser energy through linear absorption, i.e. $E_{abs} = \alpha H_0 V_{irr}$, where α is the linear absorption

coefficient, H_0 is the laser fluence, and V_{irr} is the irradiated volume. For water, the absorption coefficient at $\lambda = 1.96 \mu\text{m}$ is $\alpha \approx 100 \text{ cm}^{-1}$. Using the fibre core radius $l_i = 25 \mu\text{m}$ as the irradiation length scale, the laser fluence H_0 is $\frac{E}{\pi l_i^2}$. Approximation of the irradiated volume as a hemisphere, $V_{irr} = \frac{2}{3}\pi l_i^3$, results in an absorbed energy of $E_{abs} = \frac{2}{3}\alpha l_i E$, which combined with equation 4.2 gives an estimation for the initial bubble radius,

$$R_i^3 = \frac{\alpha l_i}{\pi \rho_i h} E \quad . \quad (4.3)$$

Inserting the laser pulse energy from the experiments ($0.0625 \text{ mJ} < E < 1 \text{ mJ}$), we use the initial bubble radius R_i from equation 4.3 as the initial condition for the cavitation model in equation 4.1. After numerical integration, we obtain the maximum bubble radius R_{max} , which is compared with the experiments in figure 4.9. The maximum bubble radii from the experiments are estimated by averaging the bubble extensions in the radial and axial directions over multiple cycles. In general, the model agrees well with the experiments, despite some discrepancy at the lowest pulse energy. This model also allows us to determine the laser settings that lead to secondary bubble formation. As shown in figure 4.8, a secondary bubble forms when the initial bubble is large enough such that it has not collapsed when the next laser pulse arrives. This condition can be quantified as $2t_c > \frac{1}{f}$. Here, t_c is the Rayleigh collapse time of a bubble, which is known to increase linearly with the maximum bubble radius, i.e. $t_c = 0.915 \sqrt{\frac{\rho_l}{p_\infty}} R_{max}$ [126]. For a given laser

repetition rate f , the pulse energy E that is sufficiently large for secondary bubble formation is calculated from the Rayleigh-Plesset model and plotted as a function of the repetition rate f in figure 4.10. Overall, we find excellent agreement with the cavitation regime observed experimentally.

4.5 Discussion and conclusion

Synthetic jets are advantageous in various flow control applications as they eliminate the need to inject mass into the system. For instance, synthetic jet actuators can be mounted on an object to affect the hydrodynamic forces by interacting with the cross flow over the object [118]. Slow synthetic jets of millimetres per second velocity have been produced by oscillating sessile bubbles through the control of surface wetting [128]. In this work, we present a unique synthetic jet in which high-frequency cavitation provides the momentum for a jet up to 1.2 m/s in velocity, which speeds up with increasing laser power. The time-averaged velocity field for this synthetic jet agrees well with the self-similar velocity profile of the axisymmetric free jet, particularly along the axis and for stronger jets. The strength of a free jet is inversely proportional to its cone angle and is determined by its kinematic momentum K_a [112]. We speculate that for the cavitation-induced jet $K_a \propto U_v^2 R_{max}^2$, where U_v is the velocity of the collapse-induced toroidal vortex as it travels away from a rigid boundary. The motion of this vortex core depends on the bubble size and its distance from the boundary [117].

Synthetic jet generation from cavitation bubbles could be advantageous for flow actuation in microfluidics by eliminating the need for a mass source. By directing the flow, these jets can also reduce the need for microchannel fabrication [129]. Additionally, streaming from cavitation bubbles is relevant in medical applications, since these bubbles play an important role in laser surgeries by acting as transmission channels to ablation targets [17]. In particular, bubbles formed with mid-infrared lasers, such as the thulium laser used in the current study, are known to significantly affect the mass removal efficiency and resolution [1] and motion of the ablation target [42, 130]. We have found the strength of the synthetic jet to be proportional to the laser power, which in turn determines the bubble size. A model is presented to estimate the bubble size and determine the energy needed for secondary bubble formation, a cavitation regime that allows the generation of non-spherical bubbles with short energy impulses and leads to strong bubble-bubble interactions.

Chapter 5

Photoacoustic shock wave emission and cavitation on fibre optic tips

5.1 Introduction

The small size and flexibility of optical fibres allows their insertion into the urinary and vascular system to conduct laser-based microsurgeries, for instance fragmentation of urinary stones [131], angioplasty [34], and laser thrombolysis [46]. The mechanism of laser tissue ablation depends on the laser pulse energy, duration, and wavelength, as well as the material properties of the ablation region [17]. If the pulse duration τ is shorter than the acoustic and thermal relaxation time of the medium, strong photoacoustic

waves can be excited from the irradiated volume [132]. The stress and thermal relaxation times are t_{st} and t_{th} , where $\tau < t_{st} = \delta/c$ and $\tau < t_{th} = \delta^2/4k$. Here δ is the length scale of the irradiation volume equal to the fibre diameter, c is the speed of sound, and k is the thermal diffusivity of the absorber. Within the limit of linear absorption, the photoacoustic wave emitted can be modeled with a linear wave equation using a delta-type source term [133]. Above a threshold intensity the liquid is vapourised due to non-linear absorption which is accompanied by the rapid expansion of a vapour bubble and complex fluid flow [24]. Here we focus on pulses at sufficiently low-energy that result in linear absorption, see [25].

Using pressure measurements and numerical solutions of the wave equation, Paltauf et al. [27] and Frenz et al. [25] have demonstrated that laser irradiation of a liquid under stress confinement first compresses and then strains the liquid away from the fibre optic tip. The straining is caused by diffraction of the compressive wavefront at the circumference of the fibre tip, which generates a toroidal tensile wave that refocuses along the fibre axis. This tensile stress is important in laser ablation as it lowers the fracturing threshold of the target material. This tension also ruptures the liquid and forms a cloud of microscopic cavitation bubbles.

It is widely known that bubble-shock wave interaction can lead to intense bio-effects in medical applications [134]. A prominent example is extracorporeal shock wave lithotripsy (ESWL), an standard procedure for non-invasive fragmentation of renal stones where bubble-shock wave interactions are essen-

tial [6, 135]. Histotripsy is a novel therapeutic application of bubble-shock wave interaction which uses pulse trains of compressive and tensile waves generated by an extracorporeal acoustic source [136]. In this procedure, cavitation bubbles are nucleated by the tensile waves and are collapsed by the subsequent compressive shock fronts. These cycles generate intense shearing flows which liquefy the tissue to scales much smaller than the size of cells [7]. It would be advantageous to have such a shock wave source at the tip of an optical fibre to allow coupling into surgical catheters for highly targeted laser ablation.

In recent years remarkable progress has been made with laser generated ultrasound. Baac et al. [137, 138] focused laser generated shock waves down to 100 μm while achieving 57 MPa of positive pressure for biological applications. Based on a similar light absorbing film of carbon nanotubes, Colchester et al. [139] demonstrated laser generated ultrasound at the tip of optical fibres. Existing efforts to generate ultrasound via optical fibres have been focused on using chemical coatings, e.g. gold nanocomposites [140], or preparing micro-machined attachments for fibre tips [141]. However, the tension achieved via these acoustic emitters is far below the tensile stress needed to induce cavitation.

In this chapter, we use laser machining for shaping fibre tips, in order to emit complex waveforms at sufficient amplitude to induce cavitation. We report on direct visualization of the waves, numerical simulations, and localized pressure measurements. The results show that structured fibre tips

Figure 5.1: Schematic of the stroboscopic photography setup.

are capable of multiple shock wave emission from a single laser pulse, as well as steering and shaping of photoacoustic waves and cavitation clouds. This suggests that shaped fibre tips are able to generate suitable waveforms for cavitation-based microsurgery protocols.

5.2 Experimental setup

To obtain structured optical fibre tips, multi-mode silica fibres ($d = 600 \mu\text{m}$, FT600EMT, Thorlabs) are flatly cleaved at both ends, and laser micro-machining is done on one end with a picosecond infra-red laser (Duetto, Time-Bandwidth Products), for details see Wang et al. [142]. In the present

study, five tip structures have been tested: flat cleave, oblique cut, step, groove, and ridge, shown in figure 5.2 from right to left. Stress confinement is obtained by submerging the tip in an aqueous red dye solution (Cochineal Red, Bake King) and coupling a green nanosecond Nd:YAG laser pulse ($\lambda = 532 \text{ nm}$, $\tau = 6 \text{ ns}$, $E \leq 10 \text{ mJ}$, Orion, New Wave Research) into the cleaved fibre end. Using a spectrometer the absorption coefficient of the red dye was measured to be $\mu = 355 \text{ cm}^{-1}$ at 532 nm . The scene is illuminated with a red 6 ns pulse, generated through laser induced fluorescence emission from a Rhodamine dye. To achieve this, a second Nd:YAG laser is focused mildly into the fluorescent dye, coupled into a fibre, brought to the scene, expanded, and weakly diffused. By varying the delay between the two laser pulses, one inserted into the structured fibres and the other used for illumination, shadowgraph images of the shock waves and nucleation of cavitation can be recorded with a sensitive CCD camera (Sensicam QE, PCO). The setup is shown schematically in figure 5.1.

5.3 Shock wave emission and cavitation on structured fibre tips

Figure 5.2 shows the acoustic waves and cavitation in front of each fibre structure 300 ns after the laser pulse. The left column shows shadowgraph image from the experiments, while the right column presents the solution of the linear wave equation at the same time instant (discussed later). For the

Figure 5.2: Acoustic transients and cavitation induced by tensile stress in front of structured fibre optics after a single laser pulse ($\tau = 6$ ns, $H_0 < 1$ J/cm², and $\mu = 355$ cm⁻¹). Left column shows shadowgraph images and right column presents numerical solutions of the linear wave equation corresponding to the same instant of time.

flat and oblique cut, the acoustic field is composed of two main components: a planar compressive wave conforming to the shape of the fibre tip, and a toroidal diffraction wave caused by the edge of the tip [25, 27]. For the oblique cut the minimum pressure and the cavitation cloud is steered away from the fibre axis, i.e. perpendicular to the surface, but not parallel to the axis. The photoacoustic wave fields in front of the step, groove, and ridge structures are more complex: from a single laser pulse multiple shock waves are generated. A compressive wave is emitted from each liquid layer adjacent to an irradiated surface; i.e. two compressive waves are formed by the step structure, and three by the groove and the ridge. Tensile diffraction waves show an even more complex emission scenario for structured tips, which nevertheless can be easily explained with a linear wave equation in three dimensions.

The photoacoustic waves induced by the laser pulse are modeled by the linear wave equation, which is solved for the acoustic pressure p

$$\nabla^2 p - \frac{1}{c^2} \frac{\partial^2 p}{\partial t^2} = 0. \quad (5.1)$$

The shock wave emitted from the fibre tip is modelled as a high frequency, short pulse Dirichlet boundary condition on the tip surface; i.e. $p = p_a \sin(2\pi ft)$ for $t < \frac{1}{2f}$ and $f = 10$ MHz. The initial amplitude p_a does not affect the quantitative behaviour of the solution and it can be calibrated to the experimentally measured pressure, since the governing equation (Eq. 5.1) is linear.

The glass fibre boundary is modelled as a sound hard boundary condition; i.e. $\hat{n} \cdot \nabla p = 0$, where $\hat{n}$ is the normal vector at the surface. Planar symmetry boundary condition, i.e. $\hat{n} \cdot \nabla p = 0$, is utilised to reduce the computational cost. The outer limits of the liquid domain, modelled here as water, are implemented as plane wave radiation boundary condition

$$\hat{n} \cdot \nabla p + \frac{1}{c} \frac{\partial p}{\partial t} = 0. \quad (5.2)$$

We use a finite element solver (Mathematics Module, COMSOL Multiphysics) to solve equation 5.1 in 3D with the boundary conditions specified above. The structured fibre tips are modelled based on their measured dimensions from an optical microscope (VHX-2000, Keyence). For a sample tip structure, the computational domain and boundary conditions are shown in figure 5.3. Please note that we solve the wave equation only in the liquid domain and do not resolve elastic waves in the fibre to reduce the computational cost.

A time sequence of the shadowgraph images and numerical solutions of the acoustic wave equation are shown for the flat cleaved fibre and compared with one of the tip structures in figures 5.4 and 5.5. As described in previous studies [25, 27] on the flat cleaved tip, a shock wave is emitted from the irradiated liquid volume quickly after the laser pulse, see figure 5.4. While this shock wave is compressive, diffraction of this positive pressure wave from the edge of the optical fibre forms a toroidal acoustic wave with the opposite polarity; i.e. negative pressure. After tensile diffraction waves D1 and D2

Figure 5.3: Computational domain and boundary conditions.

overlap, the liquid along the fibre axis is ruptured and a cavitation cloud is formed, following the trail of the peak negative pressure. It is notable that this cavitation cloud is growing along the fibre axis, and extends deeper than the optical penetration depth. In addition to the planar shock wave and tensile diffraction wave that travel away from the fibre tip, an oblique pressure wave is observed in the liquid travelling in the opposite direction, marked in figure 5.4 with F1. This oblique pressure wave is due to radiation of a flexural wave that travels downward along the body of the fibre. Since the speed of sound in the silica glass fibre c_g is faster than in the liquid c_l , a Mach cone is formed in the liquid with the angle $\beta = \tan^{-1} \frac{c_l}{c_g}$.

Compared to the flat cleaved fibre, the acoustic transients in front of the

Figure 5.4: Shock wave emission and cavitation cloud formation from a flat cleaved fibre. The shadowgraph images from the left column are compared with the numerical solution of the linear wave equation in the middle. The right column shows evolution of the cavitation cloud based on the simulation data, generated by accumulation of pressure isosurfaces at the threshold $p = -0.5p_a$.

Figure 5.5: Shock wave emission and cavitation cloud formation from a stepped fibre tip. Similar to figure 5.4, experimental shadowgraphs are at the left, numerical solution in the middle, and prediction of cavitation clouds based on the simulation at the right column. The stepped tip emits multiple shock waves and forms two cavitation clouds.

structured fibre tip are more complex, shown in figure 5.5. Shortly after the single laser pulse, the liquid adjacent to the two distal fibre surfaces absorbs the pulse energy and two compressive pressure waves are emitted. At 100 ns, in addition to the positive pressure waves, three circles are observed which are tensile diffraction waves emitted from the tip edges. At 200 ns two of the diffraction waves, D1 and D2, overlap above the upper step surface, and a cavitation cloud begins to form perpendicular to the upper surface. Additionally, the three tensile waves D3, D4, and D5, which are emitted from the edges of the lower surface, overlap and overcome the yield strength of the dye solution near the vertical step surface, creating a prominent curved cavitation cloud. As shown at 300 and 400 ns, both the smaller and larger cavitation clouds follow the trails of peak negative pressure formed by the tensile waves. This is evident in the right row of figure 5.5, which is generated by the time-accumulated pressure isosurfaces at a cavitation threshold $p = -0.5p_a$. It should be noted that the pressure threshold for cavitation may be affected by the gas nuclei present in the water, yet this is not measured in our experiments.

5.4 Pressure measurements

The fibre optic hydrophone (HFO-690, Onda Corp.) allows measuring the pressure with a spatial resolution of 100 μm and a temporal resolution of 20 ns. In figure 5.6 we compare the numerical solutions with the measure-

Figure 5.6: Pressure measurements from a fibre optic hydrophone (blue, solid lines) compared with the numerical solution of the linear wave equation (red, dashed lines). The structured fibre tip generates multiple compressive and tensile waves on and off-axis from a single laser pulse.

ments from the flat and stepped fibre tip at two locations: probe 1 on the axis of cylindrical symmetry, and probe 2 away from the axis, marked in figures 5.4 and 5.5. Both probes are located on the symmetry plane of the fibre tips, at $x = 0$. Comparing figure 5.6(a) with figure 5.6(b) reveals two shock wave emissions by the structured fibre optic, while only one is generated by the flat cleaved fibre. For the flat tip in figure 5.6(c) the tensile stress is highly focused on-axis but rapidly drops off-axis, in agreement with earlier studies [25, 27]. In contrast, the structured fibre tip allows significant tension to be generated off-axis, as shown in Fig 5.6(d). Please note that we have

conducted the pressure measurements at 1 mJ laser energy, which is below the cavitation threshold, and only normalised pressure are shown in figure 5.6. The maximum measured tensile stress in figure 5.6 are 43.5, 31.8, 12.3, and 34.1 bar from left to right. Excellent agreement is achieved between the linear acoustic wave model and the experimentally obtained pressure-time traces.

5.5 Photoacoustic cavitation using fibre optics with coated tips

Shock wave emission due to the photoacoustic effect relies heavily on the absorption parameters of the irradiated medium. Paltauf et al. [27] report that by tuning the absorption coefficient of the irradiated liquid, the bubble cloud generated in front of a fibre optic tip changes drastically. The cavitation cloud is radially confined and extends along the fibre axis in a liquid with a high absorption coefficient, similar to the cloud shown in figure 5.4. In comparison, the bubbles are much closer to the fibre tip and cover nearly the whole tip area when the absorption coefficient is low, e.g. compare figures 4 and 5 in [27].

An effective way to achieve a consistent photoacoustic field in liquids with different absorption properties is coating the fibre tip with a strongly absorbing material. Colchester et al. [139] have demonstrated that a pulsed Nd:YAG laser ($\tau = 2 \text{ ns}$, $\lambda = 1064 \text{ nm}$) delivered through fibre tips coated

Figure 5.7: Photoacoustic shock wave emission and cavitation on the tip of a fibre optic coated with a CNT-PDMS composite.

with a composite of carbon nanotubes (CNTs) and polydimethylsiloxane (PDMS) can generate ~ 10 bar of pressure in water. Water is almost transparent to the Nd:YAG laser ($\alpha = 0.2 \text{ cm}^{-1}$ at $\lambda = 1064 \text{ nm}$); therefore, the shock wave is generated due to the absorption of the pulse energy by the CNT coating and thermoelastic expansion of the PDMS. The pressure amplitude from this coating is sufficiently high for diagnostic applications like high-resolution ultrasound imaging [143]. Here we investigate the capability of the CNT-PDMS coating in generating stronger effects, such as cavitation,

which is necessary for therapeutic applications.

We follow the procedure proposed by Colchester et al. [139] to prepare the CNT-PDMS coating for the fibre tips. Multiwall CNTs are functionalised according to a method described by Majeed et al. [144] to allow their dissolution in xylene. Following this method, for synthesis of oleylamine-functionalized pyrene, 3.5 mmol of 1-pyrenebutyric acid reacts with an excess of thionyl chloride by refluxing for 30 minutes at 100°C. After removing the unreacted thionylchloride by distillation, the remaining solid is dissolved in 5 ml of anhydrous tetrahydrofuran (THF) and reacts with 3.5 mmol of oleylamine and 7.1 mmol of triethylamine. After 12 hours at room temperature, diethyl ether is added to the mixture and washed with water. Sodium sulphate is added to remove water from the organic solvent, which is then evaporated under vacuum at 25°C. To functionalise the CNTs, oleylamine-functionalized pyrene is mixed with CNTs with 1:1 weight ratio in xylene and sonicated for 1 hour. Finally, to form the CNT-PDMS composite, a solution of CNTs (20 mg) and oleylamine functionalized pyrene (20 mg) in 0.2 ml of xylene is mixed with a solution containing 1.3 ml of xylene and 500 mg of PDMS (MED-1000, Nusil). After mixing with magnetic stirring, the solution is sonicated for 30 s, degassed for 1 minute, and immediately used to coat the fibre optics. Flatly cleaved fibre tips are dip coated in the CNT-PDMS composite and left at room temperature to cure overnight. Using a setup similar to the one shown in figure 5.1, a pulsed laser is coupled into the bare end of the fibre and stroboscopic photography is used to record the events

at the coated fibre tip. The major difference is that the pulse is delivered through the fibre into water instead of a dye. Water is almost transparent to the green Nd:YAG laser, which means that the coating absorbs the laser energy instead of the irradiated liquid.

Figure 5.7 shows similar acoustic transients in front of the coated fibre tip compared to the flat cleaved tip, see figure 5.4. Diffraction of the compressive shock front generates enough tensile stress along the fibre axis to create an elongated cavitation cloud. The cavitation cloud has a conical shape and is radially confined to the fibre axis. This shows that coated fibre tips can be potentially used to generate cavitation in the irradiated liquid, even if target material is transparent to the laser. Meanwhile, we should state that in many cases the CNT-PDMS composite did not adhere to the fibre tip strongly and the coating was peeled off after a few laser shots. As a result, we could not measure the absorption parameters of the CNT-PDMS coating. Improving the binding between the coating and the silica fibre ensures that the coating can be repeatedly used to generate a consistent acoustic field in front of the fibre tip.

5.6 Discussion and conclusion

While bubble-shock wave interactions are beneficial in various medical procedures, the focusing capability of commercial shock wave generators is limited. Even in high-intensity focused ultrasound (HIFU), the focal spot is in the mil-

limetre range (~ 2 mm) due to the low frequency generated by piezoelectric transducers (~ 1 MHz) [16]. Enhancing the spatial resolution and focusing to sub-millimetre scales could be advantageous for high-precision ablation, for instance in micro-vasculatures [137]. As an alternative to piezoelectric elements, optical ultrasound transducers readily generate frequencies of the order of ~ 10 MHz, thus allowing to focus with significantly higher resolution [137, 139, 143, 145].

In this work, we use structured fibre optics to generate intense shock waves, in the therapeutic pressure range, with high spatial and temporal resolution. The pressure measurements in front of the structured fibre optics, for instance the stepped tip in Fig 5.6, show that modifying the tip geometry leads to multiple shock wave emissions. This provides the possibility of generating bubble-shock wave interactions, which could be utilized for novel, fibre-based tissue ablation protocols. Additionally, structured fibre tips allow for altering the direction of the cavitation cloud, for example to steer the cloud from vascular lumen into the surrounding tissue. The fibre tip structure could be optimised for specific applications using the simple and computationally cheap linear wave model, while remaining within mechanical and manufacturing constraints.

Chapter 6

Conclusions and outlook

Motivated by the crucial role of cavitation bubbles in minimally invasive surgeries, we focused on laser-generated bubble dynamics, especially on the tip of fibre optics. Here, chapter by chapter, the findings are summarised and limitations of experiments and models are presented. We end the thesis with an outlook on future research directions and potential applications.

6.1 Conclusions

We began in chapter 2 with a single bubble generated by a free-running mid-infrared laser. Under thermal confinement, the laser vaporises the liquid continuously and results in an oddly-shaped elongated bubble. This bubble acts as a transmission channel during laser ablation by connecting the fibre tip to the ablation target, for instance in laser fragmentation of urinary

stones. The bubble also induces strong mechanical effects, such as motion of the ablation target. Using high-speed photography, we reveal that this motion happens over a much longer time scale compared to the bubble lifetime. Changing the bubble shape affects this motion [42, 44]. Simplifying the complex process of vaporisation while tracking the bubble interface, we provided a potential flow model to estimate the bubble shape based on the temporal laser profile. This simple model captures the growth of the pear-shaped bubble but falls short of predicting the bubble collapse, possibly due to neglecting the liquid motion along the fibre axis.

The rapid liquid motion induced by a cavitation bubble in a microfluidic confinement was studied numerically in chapter 3. The bubble acts as a source of oscillating pressure that drives the motion in the surrounding liquid. This induces flow reversal near the boundaries and generates vortical structures with strong shear stress. These features, here studied in an axisymmetric gap, are also important for bubble expansion in vessels, such as embolotherapy for cancer treatment [71]. The size of elastic objects exposed to this flow affects their deformation: smaller objects are strongly strained while larger ones are mainly transported. The response of biological cells to this flow can act as an important indicator for diseases, allowing potential applications in lab-on-a-chip devices.

In chapter 4, we studied cavitation on a fibre optic that delivers a pulsed laser at a high repetition rate. Particle image velocimetry reveals a stream driven away from the fibre tip which we characterise as a synthetic jet. This

jet is generated due to vortical structures induced repeatedly by collapsing bubbles. If the laser pulse is delivered through an existing bubble, a secondary bubble forms at the distal end of the first one. An energy balance argument allows us to estimate the bubble size based on the pulse energy and predict the laser settings that generate secondary bubbles. The localised microjets generated by these double-bubbles is similar to tandem bubbles, which have been successfully utilised to enhance drug uptake by biological cells [146, 147]. Compared to acoustic streaming, synthetic jets generated by high-frequency cavitation are significantly faster. These jets can be applied for flow control in microfluidics by eliminating the need for mass input to the system.

Finally, in chapter 5, we looked at shock wave emission and generation of cavitation clouds on a fibre tip at 100 million frames per second. Shaping the tip of the fibre optic alters the acoustic transients and the resulting bubble cloud. While flatly cleaved fibres, commonly used in minimally invasive surgery, generate tension and cavitation only along their axis, modified fibre tips generate complex acoustic fields and steer the bubble cloud. This allows for bubble-shock wave interactions on the tip of a fibre optic, which offers future opportunities for micro-scale histotripsy effects. We failed to repeatedly produce cavitation clouds on fibre tips coated with CNT-PDMS due to weak adhesion. Recipes that strengthen the adhesion of the coating to the fibre optic will be valuable in developing novel catheters for laser ablation. Shock wave emission is essential in laser ablation under stress confinement.

We find that the linear wave equation is a cost-effective model to accurately track the propagation and diffraction of shock waves on fibre optics.

6.2 Outlook

While the fundamentals of bubble formation in laser ablation are comprehensively studied, there are interesting control mechanisms worthy of future exploration. These include modifications on the laser system as well as the fibre optics used to deliver the laser pulse.

The potential flow model for elongated bubbles in chapter 2, as well as existing literature [42, 44], show that the temporal pulse profile of a laser affects the bubble dynamics (see figure 2.3), which is closely linked to the ablation safety and efficiency. However, commercially-available devices, such as clinical Ho:YAG lasers, do not allow modifications of the pulse profile. Developing a free running laser ($\tau \sim 100 \mu\text{s}$ to 1 ms) with a controllable temporal pulse profile provides a control mechanism for the vapour transmission channel during laser ablation which may result in less mechanical trauma to healthy tissue. Additionally, it presents interesting opportunities for photonics and fluids research. Simulating laser-induced phase change at a moving bubble-liquid interface, while accurately tracking the bubble shape, presents a challenging, yet valuable, problem in computational fluid dynamics. There is also a lack of study on how the confinement affects the bubble dynamics in laser lithotripsy: most experiments in the literature are conducted in free

field, but laser lithotripsy *in-vitro* takes place in a tube.

The mechanical effects induced by cavitation can be utilised for novel applications in biomedical research, such as cell stretching experiments presented in chapter 3. Our simulations and previous studies [71, 72] reveal interesting flow features induced by cavitation in confinements: flow reversal, vortical structures, and strong shearing. A more comprehensive picture can be provided by including biological cells or deformable objects in the vicinity of the bubble as a third phase in the simulation. This coupling can be used to simulate the cell response to the cavitation-induced flow and vice-versa. Recent open-source simulations that include shock wave emission [148] provide a valuable platform for such studies.

Another interesting research track is controlling the flow field induced by cavitation on a fibre optic tip. Specifically, we observe a jet away from the fibre tip in chapter 4 which is similar to the microjet induced by cavitation on endosurgical tools [2]. Recent particle image velocimetry experiments [117] have shown that toroidal vortices generated during bubble collapse can induce a flow away from a rigid boundary. It is important to extend such experiments or use simulations to reveal the cavitation-induced flow on fibre tips. The flow may be strengthened or eliminated by controlling the fibre tip shape, nearby objects, and bubble-bubble interactions. Delivering rapid streams via synthetic jetting may be relevant for medical or microfluidic applications, as it eliminates the need for flushing a lumen or channel with an external fluid. To use the synthetic jet in microfluidic applications, it is

necessary to conduct our free-field experiments in a confinement and consider the effect of nearby walls on the jet.

Finally, another clinically relevant problem is large deformation of blood vessels by bubbles, exemplified by the complications caused by excimer laser angioplasty [3]. During laser ablation with a fibre optic, a large bubble forms after vaporisation of the liquid layer adjacent to the fibre tip, e.g. see figure 4 in [27]. It is important to seek solutions that eliminate this large thermal bubble while keeping a precise ablation region within the fibre tip area. A potential solution can be coating the fibre tips with elastic absorbers, similar to the CNT-PDMS composite used in 5. It is worth exploring the abilities of elastic coatings that efficiently convert the laser energy into shock waves while adhering strongly to silica glass fibres. Shaping the fibre tip before applying the coating, as shown in chapter 5, can be a simple solution for controlling the acoustic field and the resulting cavitation. Linear wave simulations can optimise the shape of the fibre tip cheaply and accurately, presenting opportunities for development of novel microsurgical tools with precise control over the ablation region.

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