# Study of Temperature Changes Around Fibres in Super Pulse Thulium Fibre Laser in Vitro Lithotripsy

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**Purpose:** To investigate temperature changes around the fibres of the super pulse thulium fibre laser (SP-TFL) during in vitro lithotripsy.

**Materials and Methods:** Stones were placed in the in vitro model. The laser was continuously excited for 180 s; the probe was positioned 5 mm around the fibre; the laser power was set at 10, 15, 20, 25, and 30 W; and the irrigation rate was set at 0, 15, 25, 35 ml/min, using a domestic SP-TFL. The temperature variations around the fibre under different power settings and different irrigation rates were compared.

**Results:** Without irrigation, the temperature around the fibre rapidly reached the safety threshold of 43℃ at 24 s. At an irrigation rate 15 ml/min and laser power <15 W, the temperature around the fibre was < 43℃. Once the laser power increased to ≥ 20 W, the temperature around the fibre increased to >43℃ at its lowest plateau. At irrigation rate 25 ml/min and laser power ≤ 25 W, the temperature around the fibre was < 43℃. At irrigation rate 35 ml/min and laser power < 30 W, the fibre temperature was < 43℃. When laser power was ≥ 30 W, the fibre temperature was > 43℃.

**Conclusion:** In extracorporeal ureteroscope SP-TFL lithotripsy, when the laser power is ≤ 15 W, ≤ 25 W, and ≤ 30 W, the irrigation rate should be maintained at ≥ 15 ml/min, ≥ 25 ml/min, and ≥ 35 ml/min, respectively.

**Keywords:** Super-pulse thulium fibre laser; temperature; lithotripsy; safety threshold

## INTRODUCTION

Urological calculi are the most common disease in urology, and their incidence has increased worldwide over the past few decades.(1) With the development of endoscopic technology and the emergence of laser fibres,(2) ureteroscopic laser lithotripsy has become the most important treatment method for calculi because of its obvious advantages; it is minimally invasive, therapeutically effective, and considerable safety. Among the many lasers utilised, the holmium: yttrium-aluminium-garnet (Ho:YAG) laser has always been the "gold standard" for laser lithotripsy in the clinics.(3) Though at present, a new laser product, the super pulse thulium fibre laser (SP-TFL), has emerged in the market. Unlike the Ho:YAG laser, the SP-TFL consists of thulium-doped quartz fibres that are triggered during laser pumping and emit lasers with a wavelength of 1940 nm. In laboratory and clinical studies, it has shown an effect superior to Ho:YAG laser in stone ablation and stone retropulsion.(4-6) However, as the laser's mechanism of action can lead to an increase in ambient temperature (7) and cause thermal damage to surrounding tissue, the temperature safety limits of the laser has always been a predicament that urological surgeons pay close attention to and study extensively during surgery. There is currently no relevant research on temperature changes around the laser fibre in domestic SP-TFL lithotripsy procedures. From November to December 2021, this study investigated the temperature changes around the laser fibre during domestic SP-TFL lithotripsy procedures using an in vitro kidney model.

## MATERIALS AND METHODS

We used an in vitro kidney model designed by an independent research group (**Figure 1**). The in vitro device was comprised of two parts: the external section was a constant temperature water bath system, which used a combination of a thermoregulated water tank and a water pump to provide stable 36–37°C heated water, simulating human body temperature; the interior is an open cylinder at one end, with a length of 40 mm, radius of 8 mm, and a volume of 7.5 ml. The cylinder's open end is closed by a rubber stopper, with three openings for the ureteric access sheath (F12/14) and for two temperature probes to pass through. The stones were placed in the cylinder. The laser equipment selected was a continuous super-pulsed dual-mode output thulium-doped fibre laser surgical system with a peak of 500 W (hereinafter referred to as "SP-TFL") (LAKH Medical Instrument Co., Ltd., Beijing, China ), attached with a 272 µm core fibre and a pulse width of 7 ms. Using a multichannel real-time thermometer (Guangzhou Ruiman Instrument Technology Co., Ltd., Guangzhou, China), four temperature probes (recording temperature once per second) were used, with the first and second probes fixed at a position of 5 mm around the fibre to record the tem-

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Received June 2022 & Accepted December 2022

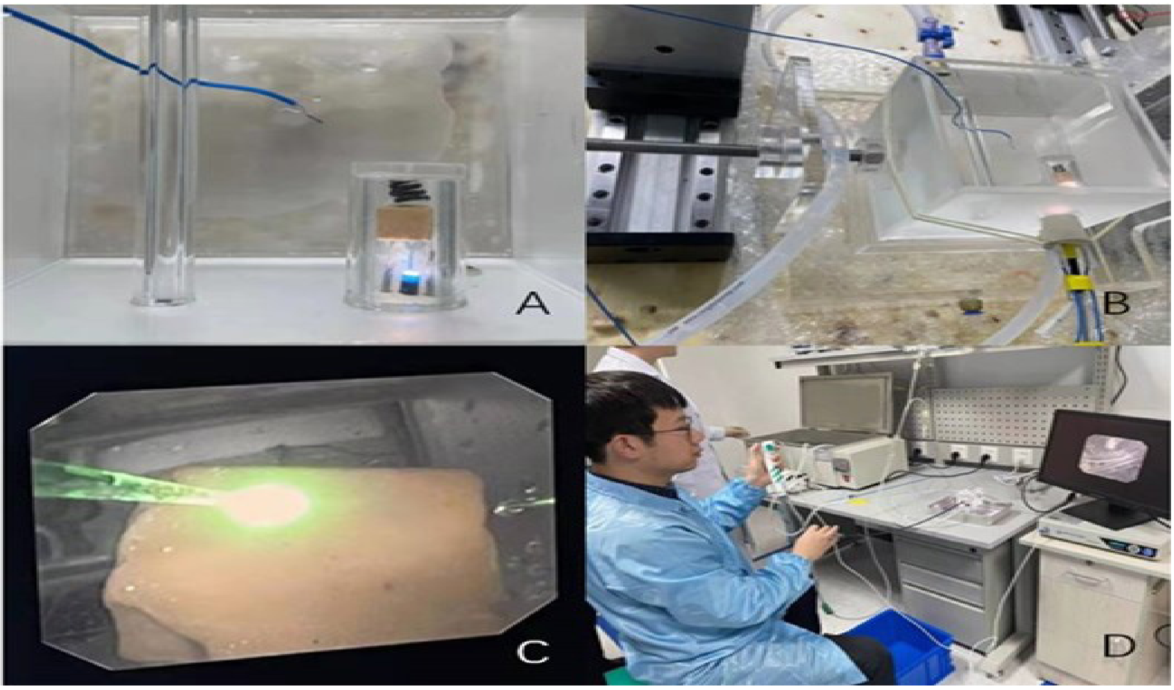


**Figure 1.** In vitro kidney model
**A:** In vitro kidney model (internal). **B:** In vitro kidney model (external). **C:** Lithotripsy laser ablating a stone. **D:** A senior doctor using soft ureteroscopy with laser for lithotripsy.

perature changes in the vicinity of the fibre. The third and fourth probes, recorded and monitored the temperature of the external water flow. Artificial stones (BEGO GmbH & Co. KG Germany) were used to simulate actual kidney stones.

Maintaining a room temperature of 22–26°C, the artificial stones were placed in an in vitro renal model (Figure 1). Using a flexible ureteroscope and the SP-TFL, a senior doctor who was blinded to the purpose of the study performed the lithotripsy procedure. Various laser power settings were used: 10 W (10 Hz*1 J), 15 W (10 Hz*1.5 J), 20 W (10 Hz*2 J), 25 W (10 Hz*2.5 J), and 30 W (10 Hz*3 J). Along with these, different irrigation rates were employed: 0, 15, 25, 35 ml/min (irrigation solution: normal saline, at 24°C). Before each experiment, the fibre was trimmed with a large core diameter fibre cutting knife and measured with a laser power meter to ensure that the laser power meets the national standard (100 ± 20%). Temperature readings from the two temperature probes, wrapped 5 mm around the fibre strand, were measured once every second, and the average reading of the two probes was recorded. The SP-TFL was set on continuous excitation for 180 s. After each experiment, the temperature

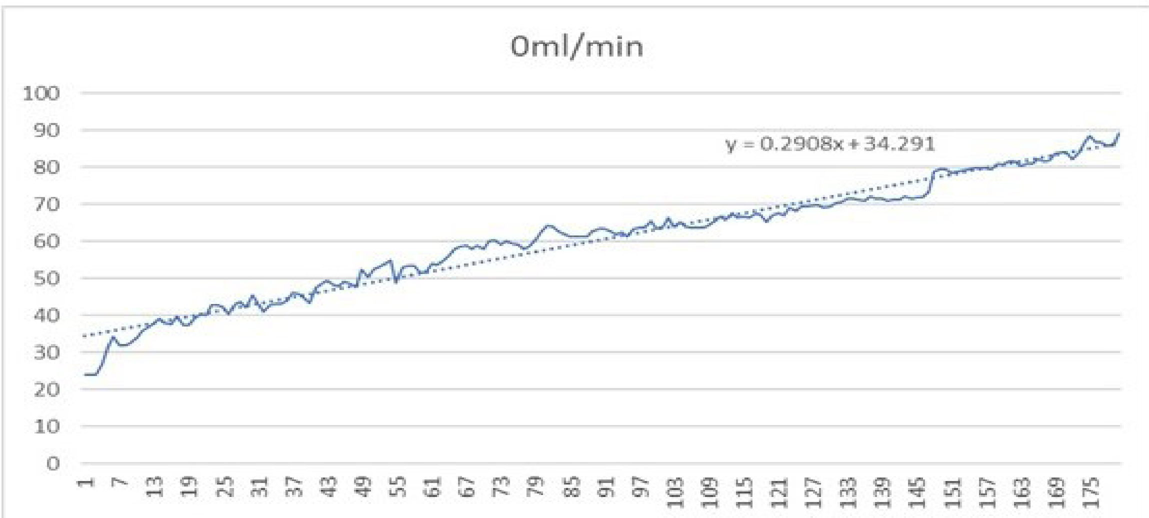


**Figure 2.** The relationship between time and temperature, in the absence of irrigation with a power setting of 10 W, in a SP-TFL in vitro setup.
The initial temperature in the in vitro renal pelvis is 24°C, and as time goes on, the water temperature incrementally increases. The temperature reaches the safety threshold of 43°C at 24 s and 89.1°C at the end of the simulation.

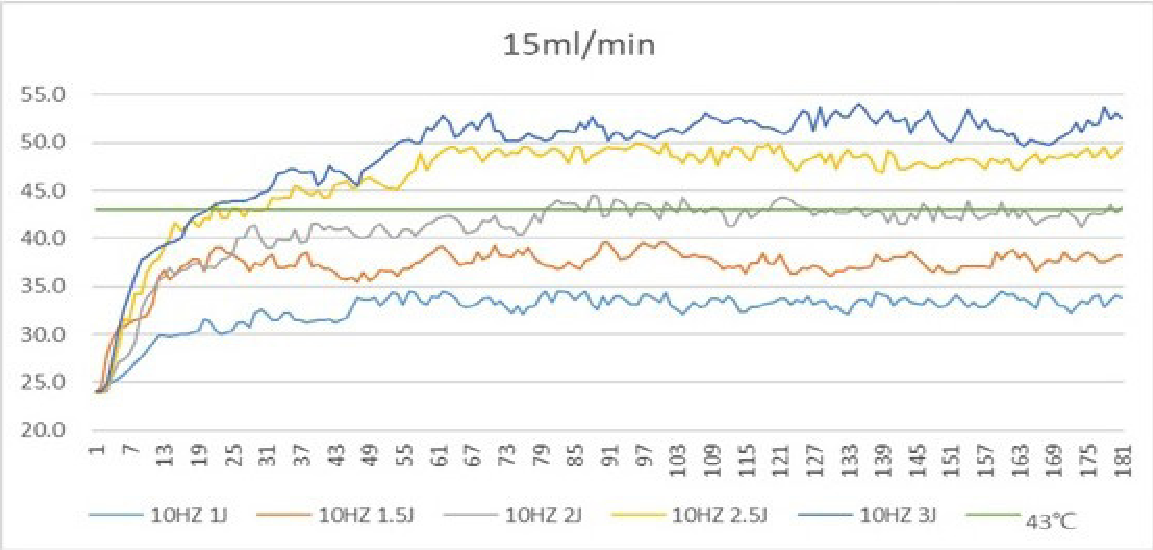


**Figure 3.** The relationship between time and temperature, at irrigation rate 15 ml/min, under different laser power settings.
At irrigation rate 15 ml/min with laser power ≤ 15 W, the plateau phase temperature is 35.5–39.6°C when the maximum temperature around the fiber is 35.5–39.6°C; the 43°C safety threshold is not exceeded.
With laser power ≥ 20 W, the plateau phase temperature around the fiber is 39.1–44.5°C, exceeding the safety threshold of 43°C.

in the simulated renal pelvis was reduced to 24°C before the next experimental set up was carried out. Each experiment was repeated three times under the same conditions to obtain the final average. A thermometer recorded the temperature in real time, while recording the temperature changes around the fibre under different power settings and irrigation flow rates.

## RESULTS

In the absence of irrigation (**Figure 2**), the ambient temperature increased by 0.3°C per minute, with the fibre reaching the safety threshold of 43°C from 24°C in 24 s. At irrigation rate 15 ml/min (**Figure 3**) and laser power ≤ 15 W, the plateau phase temperature was 35.5–39.6°C at the highest temperature peak around the fibre, which failed to exceed the temperature safety threshold (43°C). At laser power ≥ 20 W, the plateau phase temperature around the fibre was 39.1–44.5°C, exceeding the safety threshold of 43°C. At irrigation rate 25 ml/min (**Figure 4**) and laser power ≤ 25 W, the plateau phase temperature was 37.5–41.3°C when the maximum temperature around the fibre was 37.5–41.3°C, which did not exceed the safety threshold at 43°C. At laser power ≥ 30 W, the plateau phase temperature around the fibre was 40.3–43.5°C, which exceeded the safety threshold of 43°C. At irrigation rate 35 ml/min (**Figure 5**) and laser power ≤ 30 W, the plateau phase temperature during the highest peak temperature around the fibre was 35.7–37.2°C, which also did not exceed the safety threshold.

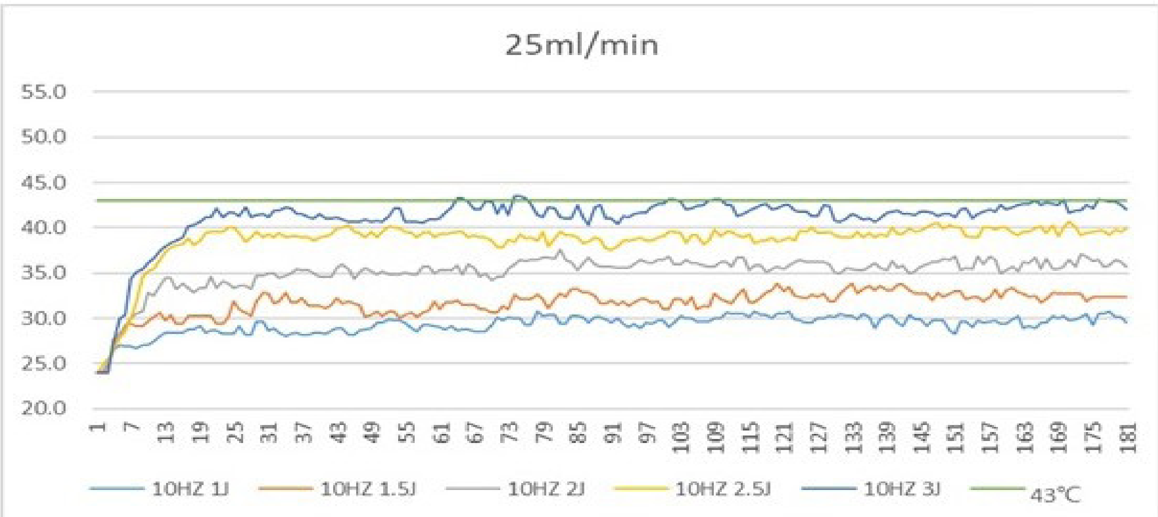


**Figure 4.** The relationship between time and temperature, at irrigation rate 25 ml/min, under different laser power settings.
At irrigation rate 25 ml/min with laser power ≤ 25 W, the plateau phase temperature is 37.5–41.3°C, and the maximum temperature around the fiber is 37.5–41.3°C; the 43°C safety threshold is not exceeded.
With laser power ≥ 30 W, the plateau phase temperature around the fiber is 40.3–43.5, exceeding the safety threshold of 43°C.

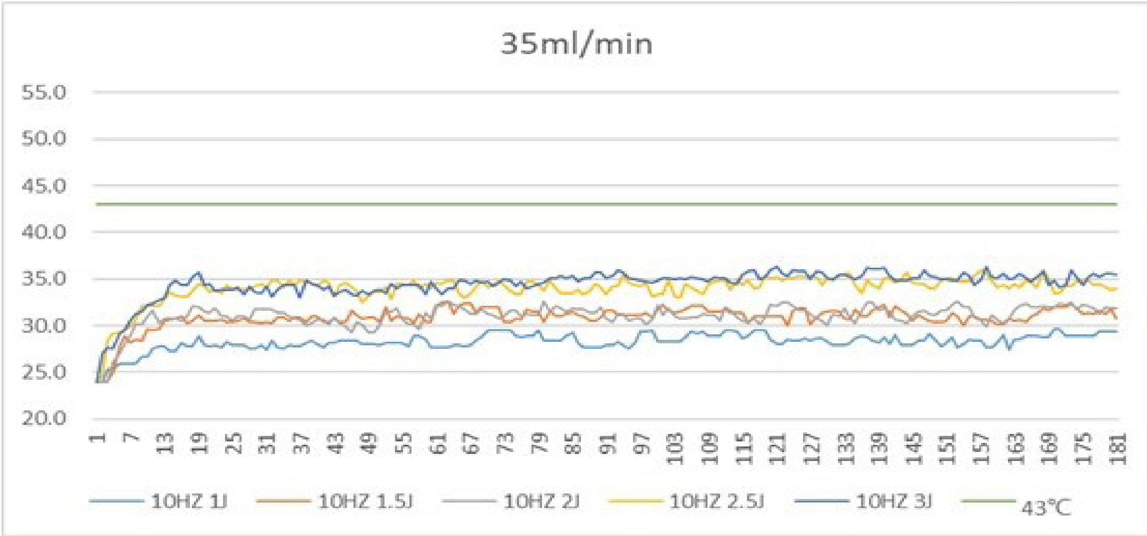


**Figure 5.** The relationship between time and temperature, at irrigation rate 35 ml/min, under different laser power settings.
At irrigation rate 35 ml/min with laser power ≤ 30 W, the plateau phase temperature at the peak ambient temperature of the fiber is 35.7–37.2°C, without exceeding the safety threshold.

## DISCUSSION

With the development of medical technology, laser lithotripsy has been utilised in more and more applications. The operational principle of laser lithotripsy is inherent in both its photothermal effect and photomechanical effect; its photothermal effect is caused by the accumulated heat emitted by the optical fibre which consequently ablates the stone; its photomechanical effect is caused by bubbles generated in the water through the excitation of the optical fibre, which leads to stone fragmentation.[7] The SP-TFL fibre laser operates using infrared light, with a wavelength of 1940 nm, and has a high absorption coefficient in water, leading to stone vaporisation after energy is absorbed in the stone cracks, thus ablating and dusting the stone.[8,9] As an effective and relatively safe surgical modality, ureteral laser lithotripsy has a low incidence of postoperative ureteral stenosis.[10] Although the incidence is low, the complications are more severe, and laser-induced tissue damage during lithotripsy is one of the causes. According to previous studies,[11-14] tissue damage begins at 43°C; accordingly, 43°C is determined as a safe temperature threshold for laser surgery. At temperatures exceeding 54°C, tissue essentially becomes completely necrotic.[15] Therefore, the damage to the surrounding tissue caused by the increase in ambient temperature during SP-TFL use is one of the most important indicators used to evaluate its safety.[16] Because of the high absorption coefficient of SP-TFL in water, the change of temperature around the fibre during the ablation process is more worthy of investigation.

In order to simulate an in vivo environment as closely as possible, reduce interference, and render the experiment feasible, the in vitro kidney model independently designed by a research group included an external thermoregulated water bath system, while the interior is a cylinder comparable to the actual volume of the kidney. According to Peng,[17] this renal model mimics the true volumetric environment in vivo as much as possible. One end of the cylinder is sealed by a rubber stopper equipped with three openings for the ureteric access sheath and two temperature probes to pass through, while the optical fibre and the temperature probe are fixed to the rubber plug, and the relative distance is fixed. To fragment the stone, it is placed in the cylindrical body; the soft ureteroscope is manipulated for stone ablation, and the irrigation solution is connected to the soft ureteroscope through a water pump. Subsequently, water flows out of the gap between the soft mirror and the dilated sheath. For 3D-printed kidney models, experienced soft ureteroscopic operators must operate the device, and during the operation, due to the displacement of stones or a blurred endoscopic field of view, the operator most likely stops laser excitation or adjusts the detection of temperature affected by the rate of irrigation fluid, which may result in increased experimental errors. The in vitro model we adapted for this study, is simple to operate; any first-time user can manoeuvre the endoscopic device in this experiment. Additionally, the model is made of transparent materials, allowing the stone to be observed simultaneously via the uteroscope or from the outside, without the impact of an obscured field of view. Furthermore, a circular spring placed in the cylinder body prevents stone displacement, so as not to affect the process of stone ablation, subsequently minimising the errors of intraoperative operation.

The results of this study suggest that in laser lithotripsy, the temperature around the fibre varies under different combinations of irrigation rate and laser power. Hein,[12] Taratkin,[18] and Peng,[17] by constructing an in vitro kidney model that excites the laser with different irrigation rates and power settings to detect changes in water temperature, concluded that laser power settings and irrigation rates play a key role in reducing water temperature, more specifically inferring that high and low irrigation rates may lead to potential tissue damage. In this study, the temperature curves caused by all laser power settings are similar to logarithmic curves, with the temperatures rising rapidly at the beginning, and then at a certain moment once the irrigation solution is balanced with heat, the curve exhibits a plateau phase. Once the irrigation rate is at 15 ml/min, the temperature increase caused by the laser power <15 W does not exceed the safety threshold, while the temperature increase caused by laser power ≥ 20 W exceeds the safety threshold. Maximum temperatures reached 52.1°C when the laser power was 30 W, which is similar to the results of Peng[17] et al., who stated that at uniform irrigation rates, 30 W of laser power will exceed the safety threshold due to the volume difference (20 ml and 7.5 ml) of the container used in the experiment; the larger volume leads to faster water flow mixing to decrease the temperature.[19] In our study, the volume selected was comparable to the volume of the renal pelvis and the results were more in line with the effect of a real in vivo experiment. Nonetheless, both experiments demonstrate that at an irrigation rate of 15 ml/min, a laser power <15 W is safe and reliable. In contrast, Taratkin[20] and Hardy[21], whose experimental models used a double-sided unobstructed tube with a mesh screen at the bottom, where the irrigation fluid flows from one end to the other, and the heat was greatly reduced by the liquid, came up with higher irrigation rates of 25 ml and 35 ml. While our in vitro model is closed on one side, the turbulence caused by the laser may affect the water temperature to some extent, but it can simulate the irrigation in the renal pelvis or the renal calyx, as the irrigation fluid flows out from the soft ureteroscope and the ureteral dilation sheath through the obstruction of the renal pelvis and the renal calyx, which explains the disparity between our findings and those of Taratkin[20] and Hardy[21]. Moreover, we also selected the more commonly used laser power setting of 10 W, which increased with time in the absence of irrigation and reached a safety threshold at 24 s. In in vivo laser lithotripsy, due to the effects of stone retropulsion or intrarenal hypertension, the operator deliberately reduces the intraoperative irrigation flow, and even suspends flushing to avoid stone displacement. This can cause insufficient local irrigation to accommodate heat in excess of the safe temperatures tolerated by the tissue, resulting in thermal damage. Thus attention should be paid to the suspension of the flushing time to avoid irreversible thermal damage to the tissue. In addition, this study observed that the temperature at the plateau phase attained using the same laser power settings decreased with increasing irrigation rates. However, caution must be practiced in increasing irrigation rates, as higher irrigation rates may lead to an increase in intrarenal pressure, which can then lead to postoperative infection complications and even the occurrence of urinary sepsis. Tokas[22] has shown that intrarenal pressure in percutaneous nephrolithotomy is closely related to postoperative infection and postoperative rehabilitation. Even if the intraoperative visual field is clear during soft ureteral lithotripsy, cases of intrapelvic hypertension still occur. Alsyouf[23] used an irrigation pump to maintain the average irrigation flow and concluded that intraoperative intrarenal pressure changes are related to irrigation pressure and irrigation time. Therefore, we did not select a high irrigation rate and restricted our irrigation fluid use to below 50 ml, which should meet the clinical requirement to avoid the occurrence of intrarenal hypertension.

As this study is a preliminary in vitro experiment using domestic SP-TFL, it has certain limitations. First, the experimental device does not accurately simulate the true anatomy of the human renal system nor the flexibility of the tissues involved; however, it can still prove the relationship between the laser power settings during the lithotripsy process and the temperature changes around the optical fibre. Further studies using animal models may help solve this limitation. Second, intraoperative influencing factors were not considered, such as the intrapelvic pressure and intraoperative bleeding during the procedure. The secretions of the renal collection system and the renal circulatory system also play important roles in preventing damage to the surrounding tissue.[12] Finally, this study only used a single energy combination, not multiple combinations, of ablation parameters. In subsequent studies, we will set up a combination of multiple ablation parameters. Nonetheless, our preliminary findings provide a valuable reference for in vivo experiments and clinical applications.

## CONCLUSIONS

In summary, our preliminary findings show that in in vitro flexible ureteroscope SP-TFL lithotripsy, the safe irrigation rate is correlated with laser power. The irrigation rates should be maintained at >15 ml/min, >25 ml/min, and > 35 ml/min for laser power <15 W, <25 W, and < 30 W, respectively.

## REFERENCES


1. Zeng GA-O, Mai Z, Xia S, et al. Prevalence of kidney stones in China: an ultrasonography based cross-sectional study. (1464-410X (Electronic)).
2. Coptcoat MJ, Ison KT, Watson G, et al. Lasertripsy for ureteric stones in 120 cases: lessons learned. Br J Urol. 1988;61:487-9.
3. Fried NM, Irby PB. Advances in laser technology and fibre-optic delivery systems in lithotripsy. Nat Rev Urol. 2018;15:563-73.
4. Traxer OA-O, Keller EA-O. Thulium fiber laser: the new player for kidney stone treatment? A comparison with Holmium:YAG laser. (1433-8726 (Electronic)).
5. Kronenberg P, Traxer O. The laser of the future: reality and expectations about the new thulium fiber laser-a systematic review. (2223-4691 (Print)).
6. Rapoport LM, Gazimiev MA, Korolev DO, et al. [Flexible ureteroscopy for lower pole renal stones: novel superpulse thulium (TM) fiber laser lithotripsy]. Urologiia. 2020:89-92.
7. Teichman JMH, Qiu J, Kang W, et al. Laser Lithotripsy Physics.
8. Wieliczka DM, Weng S, Querry MR. Wedge shaped cell for highly absorbent liquids: infrared optical constants of water. Appl Opt. 1989;28:1714-9.
9. Ventimiglia E, Doizi S, Kovalenko A, et al. Effect of temporal pulse shape on urinary stone phantom retropulsion rate and ablation efficiency using holmium:YAG and super-pulse thulium fibre lasers. BJU Int. 2020;126:159-67.
10. Dong H, Peng Y, Li L, et al. Prevention strategies for ureteral stricture following ureteroscopic lithotripsy. Asian J Urol. 2018;5:94-100.
11. Thomsen S, Pearce JA. Thermal Damage and Rate Processes in Biologic Tissues.

12. Hein S, Petzold R, Schoenthaler M, et al. Thermal effects of Ho: YAG laser lithotripsy: real-time evaluation in an in vitro model. World J Urol. 2018;36:1469-75.
13. Aldoukhi AH, Ghani KR, Hall TL, et al. Thermal Response to High-Power Holmium Laser Lithotripsy. J Endourol. 2017;31:1308-12.
14. Sapareto SA, Dewey WC. Thermal dose determination in cancer therapy. Int J Radiat Oncol Biol Phys. 1984;10:787-800.
15. He X, McGee S, Coad JE, et al. Investigation of the thermal and tissue injury behaviour in microwave thermal therapy using a porcine kidney model. Int J Hyperthermia. 2004;20:567-93.
16. Cinman NM, Andonian S, Smith AD. Lasers in percutaneous renal procedures. World J Urol. 2010;28:135-42.
17. Peng Y, Liu M, Ming S, et al. Safety of a novel Thulium fiber laser for lithotripsy: an in vitro study on the thermal effect and its impact factor. J Endourol. 2020;34:88-92.
18. Taratkin M, Laukhtina E, Singla N, et al. Temperature changes during laser lithotripsy with Ho:YAG laser and novel Tm-fiber laser: a comparative in-vitro study. World Journal of Urology. 2020;38:3261-6.
19. Andreeva V, Vinarov A, Yaroslavsky IA-O, et al. Preclinical comparison of superpulse thulium fiber laser and a holmium:YAG laser for lithotripsy. (1433-8726 (Electronic)).
20. Taratkin MA-OX, Laukhtina E, Singla N, et al. Temperature changes during laser lithotripsy with Ho:YAG laser and novel Tm-fiber laser: a comparative in-vitro study. (1433-8726 (Electronic)).
21. Hardy LA, Wilson CR, Irby PB, et al. Thulium fiber laser lithotripsy in an in vitro ureter model. (1560-2281 (Electronic)).
22. Tokas TA-O, Herrmann TRW, Skolarikos A, et al. Pressure matters: intrarenal pressures during normal and pathological conditions, and impact of increased values to renal physiology. (1433-8726 (Electronic)).
23. Alsyouf M Fau - Abourbih S, Abourbih S Fau - West B, West B Fau - Hodgson H, et al. Elevated renal pelvic pressures during percutaneous nephrolithotomy risk higher postoperative pain and longer hospital stay. (1527-3792 (Electronic)).