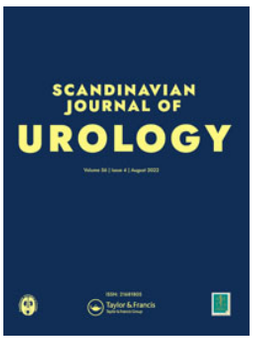

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ARTICLE

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Temperature profiles during ureteroscopy with thulium fiber laser and holmium:YAG laser: Findings from a pre-clinical study

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ABSTRACT

Objective: The aim of this study was to investigate temperature profiles in both the renal pelvis and parenchyma during Thulium Fiber Laser (TFL) and Holmium:yttrium-aluminium-garnet (Ho:YAG) laser activation in an ex-vivo porcine model.

Methods: Three porcine kidneys with intact renal pelvis and proximal ureters were used in the study. A temperature sensor was inserted through a nephrostomy tube into the renal pelvis and a second sensor was inserted directly into the renal parenchyma. Temperatures were recorded during continuous laser activation for 180 s, and for an additional 60 s after deactivation. TFL (150 μm and 200 μm) and Ho:YAG (270 μm) laser delivered power at settings of 2.4 W, 8 W, 20 W and 30 W.

Results: Intrapelvic temperatures correlated directly to power settings. Higher power produced higher temperatures. For example, using a 150 μm fiber at 2.4 W resulted in a 2.6 °C rise from baseline ($p = 0.008$), whereas using the same fiber at 20 W produced a rise in temperature of 19.9 °C ($p = 0.02$). Larger laser fibers caused significantly higher temperatures compared to smaller fibers using equivalent power settings, e.g. mean temperature at 20 W using 150 μm was 39.6 °C compared to 44.9 °C using 200 μm, $p < 0.001$. There was a significant increase in parenchymal temperatures when applying 20 W and 30 W of laser power with the two larger fibers.

Conclusion: In this ex-vivo study, renal temperatures correlated directly to power settings. Higher power produced higher temperatures. Furthermore, larger laser fibers caused higher temperatures. These findings could help guide selection of safe power settings for ureteroscopic lithotripsy, but future clinical studies are needed for confirmation.

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Ureteroscopy; thulium fiber laser; Ho:YAG; temperature profiles

Introduction

The prevalence of kidney stone disease has increased in the modern era and is currently estimated at 10% in industrialized countries [1]. Moreover, it is recognized as a chronic disease with a recurrence rate close to 50% [2]. As renal stones frequently require surgical intervention, the number of procedures performed, particularly ureteroscopic lithotripsy, has increased accordingly [3].

The introduction of laser technology has been pivotal to the advancement of ureteroscopy (URS) as a minimally invasive treatment of urolithiasis [4]. The Holmium: Yttrium–Aluminum–Garnet (Ho:YAG) laser has served as the leading choice for endoscopic laser lithotripsy for more than three decades and is recognized as the gold standard [5]. This is owed to its versatility and favorable properties, including the ability to break all stone types and delivery in the form of thin, flexible fibers [6]. Development of higher power laser technology has allowed the limits of what can be achieved with URS to be set even higher [6,7]. In comparison to stone fragmentation, techniques such as dusting and popcorning often implement higher power settings (≥20 W), and the efficacy associated with different strategies has been the focus of many studies [8–10]. In recent times, the intraoperative safety of URS has gained increasing attention, particularly with regard to high intra-pelvic pressure (IPP) and raised temperature levels [11–14]. The results of such studies include new recommendations for procedural techniques aimed at reducing the risk of complications. One such example is the avoidance of pressurized irrigation pumps [13].

With the advent of the Thulium fiber laser (TFL), discussion surrounding high temperatures during lithotripsy and resultant thermal damage have been fueled further [15]. TFL offers pulse frequencies up to 2,400 Hz and reduced retropulsion [16]. These properties incentivize dusting, which generally translates to high power settings and more concentrated active laser time when compared to stone fragmentation and extraction [17].

Research investigating the implications of using novel techniques in combination with new technologies is of great importance for clinical practice and patient safety. A recent clinical randomized trial demonstrated excellent results after URS lithotripsy with TFL even at very low settings of 2.4 W

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[18]. To date, laser temperature experiments have mainly focused on high power settings and a study demonstrating limited temperature elevation using low power (<10 W) could contribute to optimal recommendations for laser settings.

The aim of this study was to investigate the temperature profiles in both the renal pelvis and parenchyma during TFL and Ho:YAG laser activation in an ex-vivo porcine model as well as investigate the impact of fiber size(s) and different laser settings that are commonly used in a clinical setting. Furthermore, we wanted to study if our method of temperature measurement was feasible, so that it can be adapted for future clinical studies.

Materials and methods

Outcomes of interest

Primary outcomes were the observed differences in temperature profiles in both the renal pelvis and parenchyma using different laser settings and fiber sizes. The secondary outcome was success of temperature measurements using a temperature sensor inserted through a regular 10-fr nephrostomy tube.

Experimental set-up

Three fresh porcine kidneys with intact renal pelvis and proximal ureters were obtained for this experiment. The setup is presented in Figure 1. A digital flexible ureteroscope (URF-V, Olympus Corporation, Japan) was inserted into the renal pelvis through the ureter. Room temperature (23 °C) 0.9% saline was used for irrigation *via* the working channel (3.6 Fr) with gravitational pressure at 60 cmH_2O. One millimeter diameter K-type thermocouple sensor (RS Pro, United Kingdom) was employed for temperature measurements. Recording temperatures at the tip of the ureteroscope has been reported to provide less accurate results of intra-renal temperatures [19]. It was therefore inserted via a 10-Fr nephrostomy tube, which was placed in a suitable upper or lower calix and clamped to prevent outflow. A second thermocouple was placed directly into the renal parenchyma at a central position. These sensors were connected to two separate data loggers (Vernier Go Direct® Thermocouple, USA) to record real-time temperature measurements. Temperature range of the data loggers was −200 °C to 1,400 °C with an accuracy of ±2.2 °C. Connection was established with two different computers via Bluetooth 4.2. Temperature data was registered in Vernier Graphical Analysis™ v5.8.0–387 (Vernier Software & Technology, USA) and the data acquisition rate was set at twice per second. The temperature sensors were calibrated before starting the experiment.

Three different laser fibers were used; Thulium 150 µm, Thulium 200 µm and Ho:YAG 270 µm. The laser fiber was inserted through the working channel of the ureteroscope and positioned in a calix adjacent to the nephrostomy tube. Temperature data was recorded during and after laser firing. Graphs were plotted in real-time for measurements taken

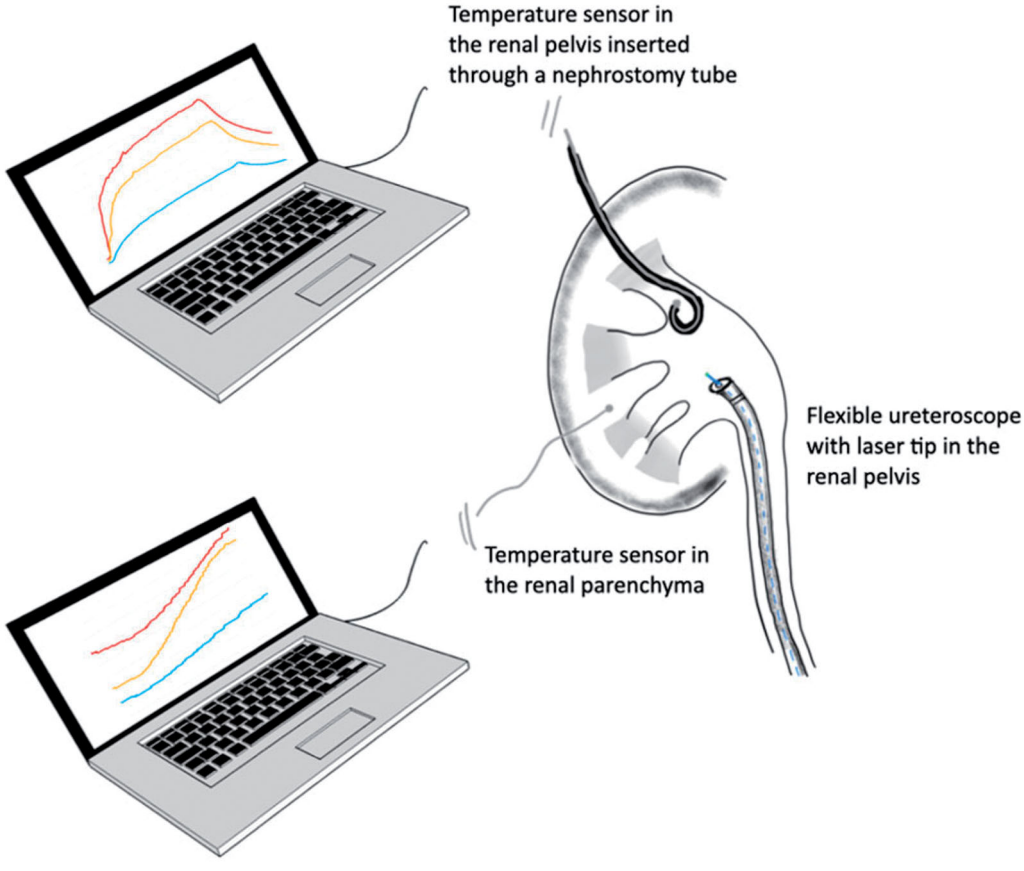

Figure 1. Experimental set-up with porcine kidney, temperature sensors and flexible ureteroscope with laser fiber.

from both the renal pelvis and parenchyma. Continuous irrigation was used throughout the experiment. Irrigation rates were measured to 15 ml/min, 12 ml/min and 9 ml/min when the 150 μm, 200 μm and 270 μm laser fibers were inserted through the ureteroscope, respectively.

TFL (Olympus Soltive™ Premium 60 W, USA) and Ho:YAG laser (Dornier Medilas H Solvo 30 W, Germany) were used to deliver laser energy. The first series of the experiment was performed with a 150 μm fiber starting at laser settings of 0.4 J/6 Hz (2.4 W). Continuous laser firing was maintained for 180 s before deactivation. Temperature recording was continued for a further 60 s thereafter, which we refer to in this study as the idle laser time. After completing the measurements, the irrigation was increased for 4 minutes to achieve a steady baseline temperature. The procedure was then repeated with the same laser fiber but different settings of 0.8 J/10 Hz (8 W) and 0.2 J/100 Hz (20 W). The experiment was then repeated using a 200 μm fiber and, finally, a 270 μm fiber. For the two latter fibers we also included a laser setting of 30 W (1 J/30 Hz TFL and 3 J/10 Hz Ho:YAG). Given the Ho:YAG laser cannot deliver 20 W at 0.2 J/100 Hz, a setting of 2 J/10 Hz was used to achieve equivalent power. After completing all measurements in the first kidney, the experiment was repeated for the remaining two kidneys. This enabled us to gain results from a total of three kidneys and average readings were calculated.

Ethics and statistics

When planning the study, ethical approval was cleared with the Norwegian Food Safety Authority.

Peak vs. baseline temperatures were compared using paired-samples *t*-test for different laser settings and laser fibers. Wilcoxon test for related samples was used comparing the continuous temperature profiles with different laser settings or laser fiber sizes.

IBM SPSS Statistics 25 (IBM, Armonk, NY) was used for statistical analysis. Statistical significance was defined as $p < 0.05$ for all tests.

Results

Using a novel method, temperatures in the renal pelvis and parenchyma could be registered during laser activation at different settings and fiber sizes. Baseline temperatures, T_0, in the renal pelvis varied between 23.6 °C and 24.6 °C at the beginning of the series in all three kidneys. The corresponding baseline temperatures in the renal parenchyma varied between 23.3 °C and 23.7 °C. Due to laser activation, the kidneys were gradually warmed up, which caused a slight rise in baseline temperatures throughout the experiment.

Renal pelvis

In regards to the primary outcome, Figure 2 illustrates temperature profiles in the renal pelvis comparing different laser settings using the same fiber across all the sizes. Comparing temperature profiles revealed that significantly higher temperatures are generated over time when increasing laser power using the same fiber size.

A comparison of temperature profiles using different fiber sizes at the same laser settings is shown in Figure 3. Increasing fiber size while maintaining the same laser settings caused a significant rise in temperature profiles, e.g. mean temperature at 20 W using 150 μm was 39.6 °C compared to 44.9 °C using 200 μm, $p < 0.001$.

The threshold for thermal cell injury (43 °C) was never reached at laser settings of 2.4 W and 8 W, irrespective of fiber size. For 20 W settings, this threshold was surpassed after a laser activation period of 44 s and 40 s with the 200 μm and 270 μm fiber, respectively. For 30 W settings, it was exceeded after an average of 24 and 18 s with the 200 μm fiber and 270 μm fiber, respectively.

During the 60 s of idle laser time, there was a marked fall in temperature recorded across all three fibers, as illustrated by the temperature profiles. The endpoint temperature was higher for the largest laser fiber. For 20 W settings, endpoint temperature was almost 5 °C higher for the 270 μm fiber (34.3 °C) compared to the 150 μm fiber (29.6 °C), $p = 0.032$. Temperatures did not return to baseline during the 60 s of idle laser time in any of the series, irrespective of fiber size.

Table 1 shows baseline and peak temperatures (T_{max}) in the renal pelvis and parenchyma after 180 s of continuous laser firing using the different laser fibers and settings. The highest peak temperatures and most pronounced temperature changes from baseline were observed at the higher power settings (20 W and 30 W).

Renal parenchyma

Figure 4 illustrates the temperatures in the renal pelvis and parenchyma for different laser settings using the 150 μm fiber. Compared to intrapelvic temperatures, parenchymal temperatures lagged behind and continued to rise during the 60 s of idle laser time. Table 1 shows that there was no significant rise in parenchymal temperatures from baseline for any of the laser settings when the 150 μm fiber was used (Figure 4). In contrast, there was a significant increase in the parenchymal temperatures recorded with the 200 μm fiber with 30 W settings as well as the 270 μm fiber with both 20 W and 30 W settings.

Discussion

In this study we have investigated the temperature profiles in the renal pelvis and parenchyma during URS laser activation in an ex-vivo porcine model. Regardless of power settings and size(s) of laser fiber, there was an increase in intrapelvic temperature from baseline after activating the laser. Intrapelvic temperatures correlated directly to power settings. As expected, higher power levels produced a more pronounced rise in temperatures (Figure 2).

Thermal cell injury is shown to occur at 43 °C [20,21]. Aldoukhi et al. [22] reported findings from an in-vivo porcine model study and recorded temperatures up to 50.1 °C after only 10 s when 40 W was applied with medium irrigation

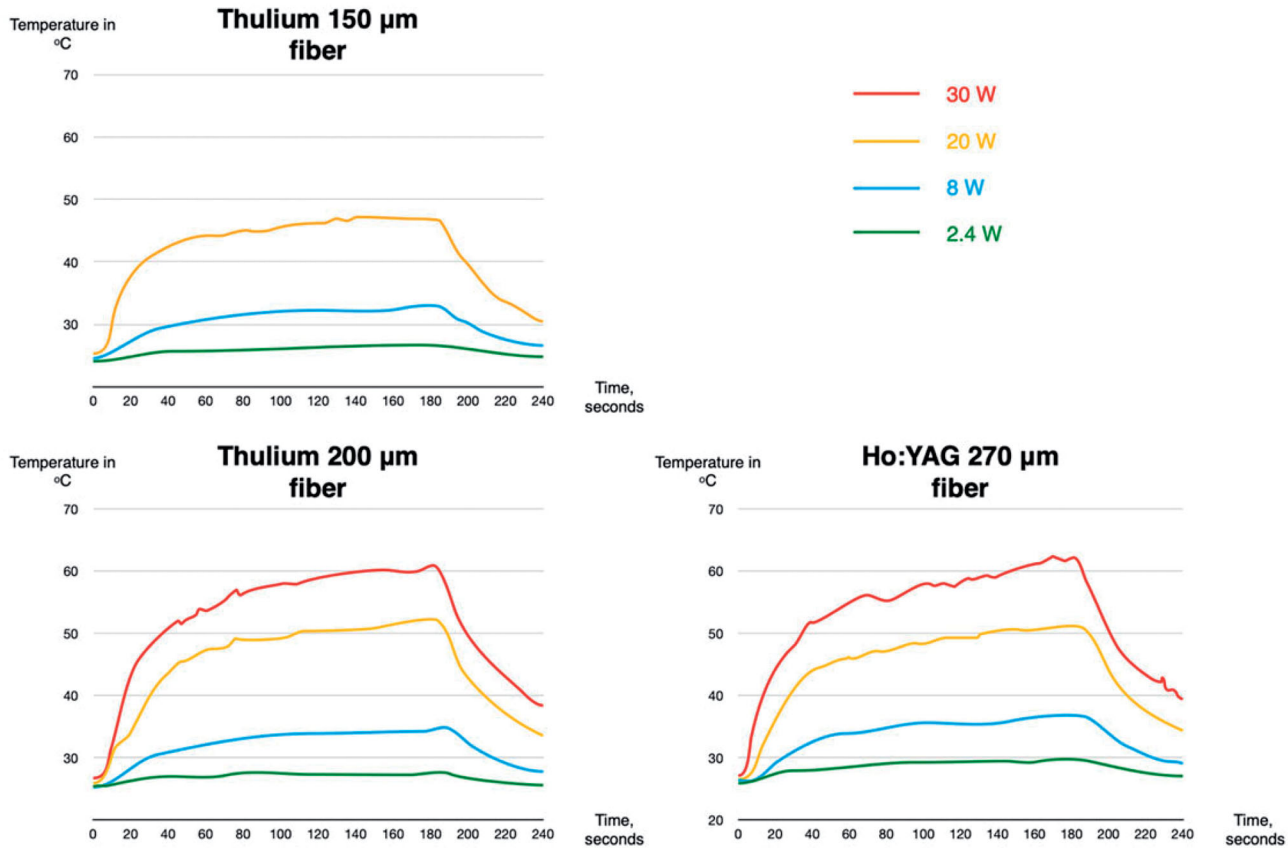

Figure 2. Temperature profiles in the renal pelvis for different laser settings using 150 µm, 200 µm and 270 µm fibers. For every laser fiber, 20 W produced significantly higher temperatures than 8 W and 8 W significantly higher temperatures than 2.4 W, $p < 0.001$. For 200 µm and 270 µm fibers, 30 W produced significantly higher temperatures compared to 20 W, $p < 0.001$ (Wilcoxon).

(15 mL/minute). After 3 × 60 s trials with laser activation and a variety of irrigation settings, inspection of the kidneys revealed gross pathological tissue coagulation and injury. Another study by the same authors examined patterns of laser activation and found the duration of pedal activation to be as high as 182 s, although the average time was much less [23]. While the threshold for thermal cell injury was exceeded with high power levels, laser settings of both 2.4 W and 8 W failed to reach this threshold in the present study. Although caution should be taken when drawing conclusions based on ex-vivo studies, this adds to the already favorable properties of low power levels for URS lithotripsy [16,18]. Additionally, it further supports the recommendation that laser settings should be low powered. This is especially true for TFL, which has demonstrated excellent stone free rates using very low power levels [18].

Larger laser fibers caused significantly higher temperatures when compared to smaller fibers using equivalent power settings (Figure 3). This can be explained by the lower irrigation rate observed with the use of larger laser fibers. The latter occupy more of the available working channel diameter and therefore reduce irrigation flow, which decreases the cooling effect in the renal pelvis accordingly.

There was a significant increase in parenchymal temperatures when applying 20 W and 30 W with the two larger fibers. This increase in parenchymal temperatures was delayed compared to intrapelvic temperatures and the former continued to rise even after deactivating the laser. To our knowledge, this is the second study to demonstrate a rise in parenchymal temperatures due to laser activation in the renal pelvis [19]. This indicates a possible risk of thermal injury occurring deep in the renal parenchyma during laser lithotripsy. Furthermore, it emphasizes the importance of caution regarding selection of laser fiber and power settings.

Temperatures did not return to baseline during the 60 s of idle laser time in any of the series, regardless of laser fiber size. However, the 150 µm fiber demonstrated a significantly lower endpoint temperature when compared to the two larger fibers. In addition to this, when applying 20 W/30W with the 200 µm/270 µm fiber, it took more than 20 s for the temperatures to fall below 43 °C (Figure 2). Our results indicate that very short breaks in active laser time, as can be common in clinical practice, may only have a negligible effect on renal temperatures. Furthermore, this implies that extended breaks in active laser time are necessary to allow the temperature to fall when high laser power settings are applied.

Most previous studies have measured temperatures with an active laser time of 60 s or less [24,25]. In contrast to this, we chose to activate the laser continuously for 180 s to both increase our understanding of temperature profiles beyond 60 s as well as mirror the ranges performed in clinical practice. While the dusting technique is becoming the preferred

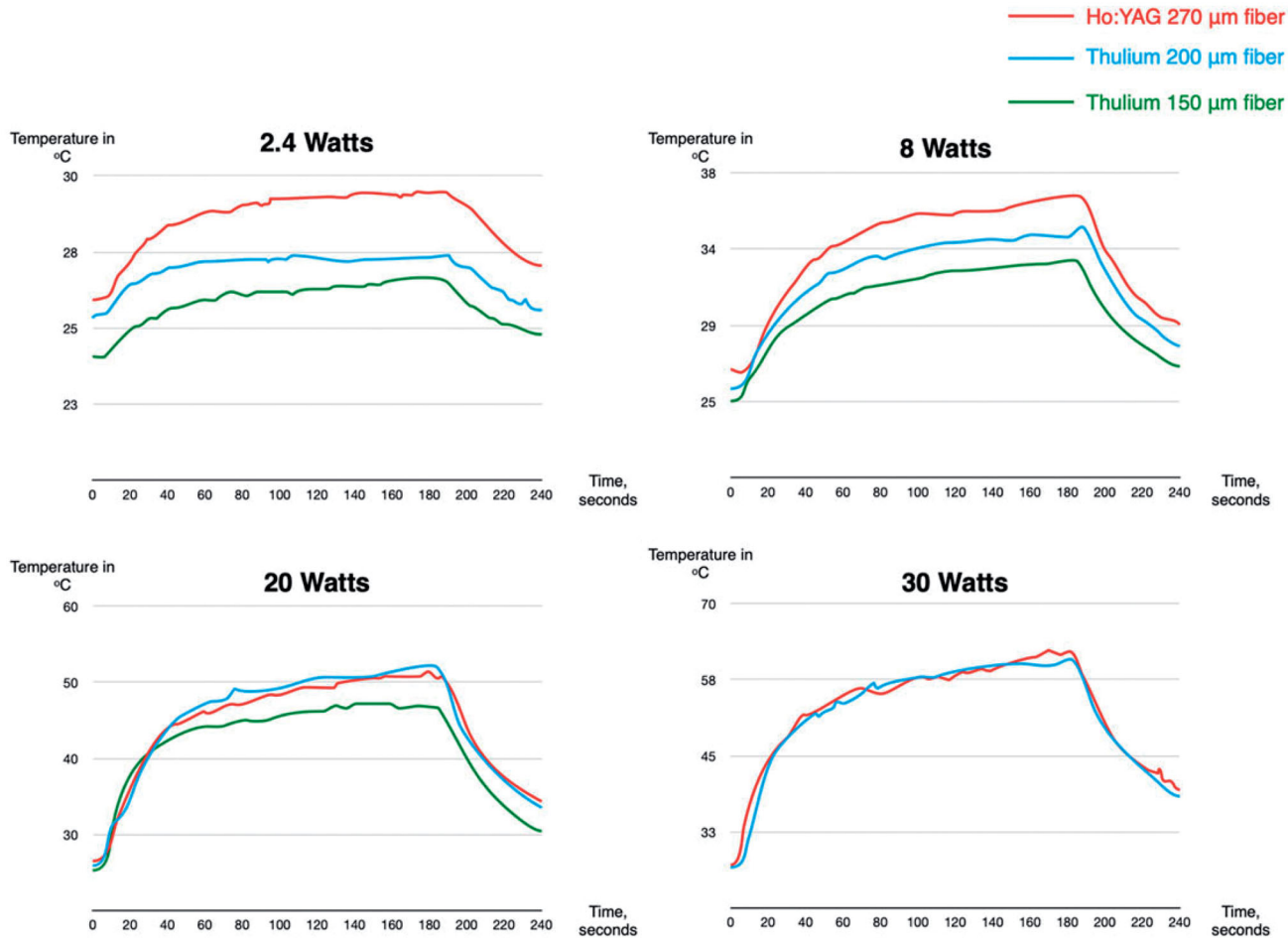

Figure 3. Temperature profiles in the renal pelvis comparing 150 µm, 200 µm and 270 µm fibers at different laser settings. Besides the 20 W settings for the two largest fibers, the 270 µm fiber produced significantly higher temperatures than the 200 µm fiber and the 200 µm fiber significantly higher temperatures than the 150 µm fiber for any laser setting, $p < 0.001$ (Wilcoxon).

Table 1. Comparing baseline and peak temperatures in the renal pelvis and the parenchyma.

		Temperature (°C), renal pelvis			Temperature (°C), parenchyma		
Fiber	Laser settings	Baseline, T_0 (range)	Peak, T_{max} (range)	*p*-value	Baseline, T_0 (range)	Peak, T_{max} (range)	*p*-value
Thulium 150 µm	0.4 J/6 Hz (2.4 W)	24.1 (23.6–24.6)	26.7 (25.8–27.2)	0.008	23.5 (23.3–23.7)	24.1 (23.7–24.3)	0.095
	0.8 J/10 Hz (8 W)	24.5 (24.2–24.8)	32.8 (31.6–34.6)	0.011	24.1 (23.9–24.1)	25.9 (24.5–26.7)	0.098
	0.2 J/100 Hz (20 W)	25.3 (25.2–25.4)	45.2 (42.3–50.9)	0.020	25.6 (24.9–26.5)	29.5 (26.3–31.7)	0.051
Thulium	0.4 J/6 Hz (2.4 W)	25.3 (25.2–25.4)	27.3 (27.0–27.8)	0.013	26.7 (26.5–26.9)	*	NA
200 µm	0.8 J/10 Hz (8 W)	25.3 (24.8–25.6)	34.8 (34.2–36.1)	0.003	25.8 (25.5–25.9)	26.9 (25.7–27.1)	0.051
	0.2 J/100 Hz (20 W)	25.9 (25.6–26.2)	52.1 (47.5–55.8)	0.009	26.8 (26.1–27.3)	30.8 (26.9–32.3)	0.085
	1 J/30 Hz (30 W)	26.7 (26.4–27.4)	60.9 (56.7–67.5)	0.010	29.3 (27.9–31.7)	31.6 (28.9–32.7)	0.006
Ho:YAG	0.4 J/6 Hz (2.4 W)	25.9 (25.6–26.4)	29.4 (29.0–30.0)	0.015	29.2 (28.3–29.7)	*	NA
270 µm	0.8 J/10 Hz (8 W)	26.4 (26.0–26.8)	36.7 (34.4–40.7)	0.044	27.8 (27.1–28.1)	*	NA
	2 J/10 Hz (20 W)	26.5 (25.6–27.4)	51.3 (48.5–56.4)	0.011	27.5 (27.1–27.9)	29.0 (27.7–28.9)	0.013
	3 J/10 Hz (30 W)	27.1 (26.8–27.4)	61.8 (57.8–68.3)	0.008	28.8 (27.7–29.9)	30.4 (28.5–30.5)	0.020

Values = mean for all three porcine kidneys (range).
*In these series, the baseline parenchymal temperatures were falsely elevated following activation of 20 W and 30 W laser power in previous measurements. Due to this, data registration was initiated before a true baseline temperature was achieved. This resulted in a paradoxical fall in parenchymal temperatures while applying 2.4 W with the 200 µm and 2.4 W and 8 W with the 270 µm fibers.

strategy for lithotripsy, extended active laser time may be a more accurate representation of current clinical laser use. Using low laser settings aid maintaining a clear endoscopic view, which in turn allows for extended laser activation [18]. This also emphasizes the relevance of monitoring the temperatures for a long period of time as done in the present study. In addition, new technology makes it feasible to treat larger stones, which also facilitates extended periods of continuous laser activity. By employing a longer active laser time than has been used in previous studies, this allowed us to observe a delayed rise in parenchymal temperatures. This may also be true in a clinical setting.

Higher irrigation rates have previously been shown to reduce temperatures caused by laser activation [22,26]. In

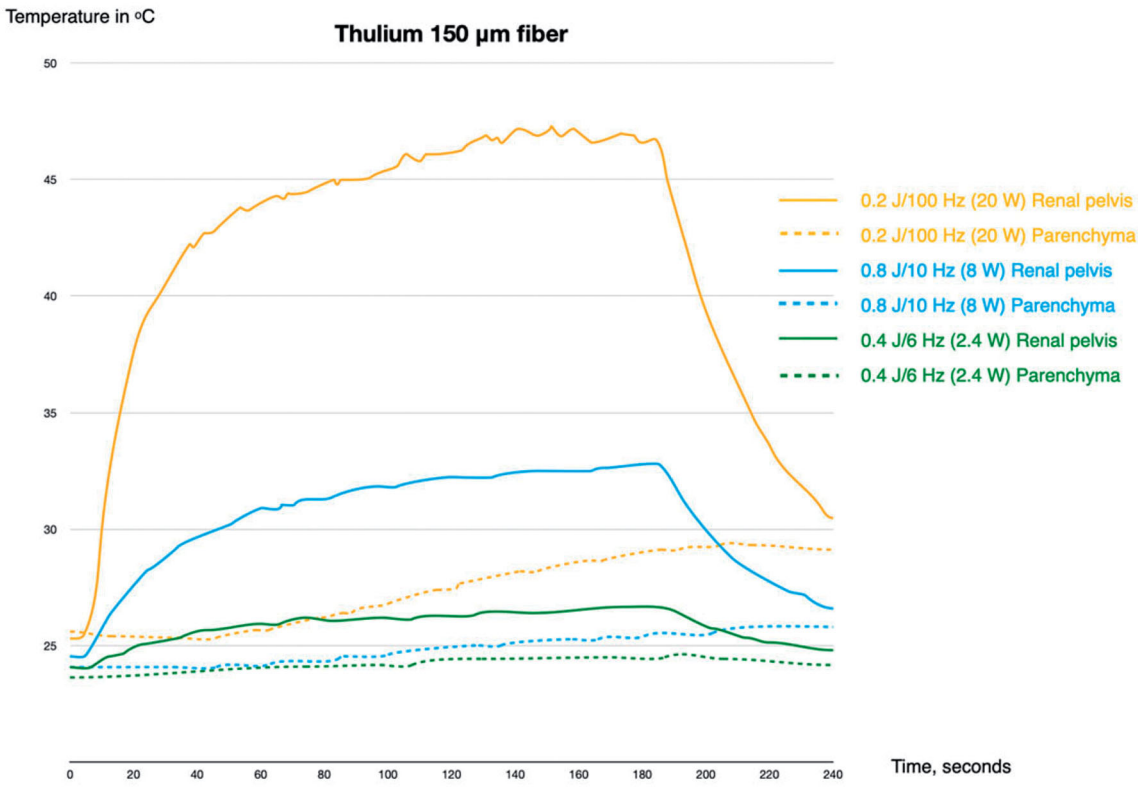

Figure 4. Temperature profiles in the renal pelvis and parenchyma for different laser settings using 150 µm fiber. The laser was activated for 180 seconds before being idled for 60 seconds.

our experiment, we utilized room temperature irrigation fluid at 60 cmH_20 gravitational pressure. Increasing irrigation pressures can cause high IPP levels and an increased risk of complications, such as sepsis [11]. A study examining renal pelvic pressures in patients undergoing URS without ureteral access sheath (UAS) for renal stones, demonstrated mean IPPs of 63 cmH_20 with a continuous irrigation pressure of 80 cmH_20 [27]. Considering that it is advised to maintain IPPs under 30 cmH_20 to avoid renal backflow, one should be cautious to counterbalance high temperatures by increasing irrigation pressures.

Utilizing UAS can allow for higher irrigation while maintaining lower temperature and IPP levels as a result of continuous outflow [5,12,28]. However, despite the recognized benefits associated with use of UAS, concerns over ureteral damage persist and long-term data is lacking [5,29]. Meier et al. [30] reported outcomes after URS in a prospective study of 22 centers. Not only did use of UAS vary hugely (1.9–96%) but the authors found their use did not increase the likelihood of achieving a stone-free status. UASs were also associated with an increased risk of postoperative emergency department visits and hospitalization. An alternative to increasing the irrigation rate in order to maintain safe temperatures in the renal pelvis is to avoid prewarmed irrigation fluids. Instead, one should consider keeping irrigation fluid at room temperature, or even cooled, when laser lithotripsy is performed [31].

We used an ex-vivo porcine model for our experiment. A porcine kidney is recognized as the most accurate comparative model for human renal anatomy. It has a multipapillate system with associated major and minor calyces. Furthermore, both the kidney, renal pelvis, and ureters are of similar size to that of human renal anatomy [32]. However, the ex-vivo porcine model used in our study differs from a clinical scenario in several aspects and caution should be taken when drawing clinical recommendations from benchtop studies. First, the URS procedure itself was different from the actual surgery performed in clinical practice. As we used ex-vivo kidneys there was no muscle tone in the renal pelvis or ureter, which could affect outflow and irrigation rates. Also, the porcine kidneys had a baseline room temperature and there was no blood perfusion. However, it is noteworthy that in previous in-vivo porcine studies, baseline intrapelvic temperatures were 23–25 °C as continuous room tempered irrigation lowers the intrapelvic temperature to that of the irrigation fluid when the laser is not activated [12]. Nevertheless, the absolute temperatures found in our ex-vivo model are not necessarily accurate for a clinical setting and future clinical studies are therefore needed. Another limitation was that the kidneys used in this study were gradually warmed up during the experiments due to accumulation of laser energy. This caused a slight rise in the baseline temperature throughout the experiment. As a result of this, comparing changes in temperature rather than absolute temperature was determined to be more appropriate.

While the 150 µm fiber is only available for TFL, we did not perform head to head comparisons using a 200 µm Thulium fiber and Ho:YAG fiber of the same size. Our fiber

selection mirrored three types we typically use in our day-to-day practice.

Conclusion

In our ex-vivo porcine model, both intrapelvic and parenchymal temperature rose with increasing laser power and fiber size. The rise in temperature was negligible when using very low power. Furthermore, the rise in parenchymal temperature was delayed and continued to rise even after laser deactivation. Our findings could help guide selection of safe power settings for ureteroscopic lithotripsy, but future clinical studies are needed for confirmation. Until then, careful attention should be paid to avoid thermal injuries when using lasers with high power ($\geq$20 W) and fiber sizes $\geq$200 μm.

Disclosure statement

Øyvind Ulvik is a consultant for Olympus but they were not involved in the design, analyses, interpretation or writing of the manuscript. The remaining authors have nothing to disclose.

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