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Transmyocardial laser revascularization: Are new approaches with new lasers possible?

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The purpose of this article is to briefly describe the main physical mechanisms of laser ablation of biological tissues and to analyze the prospects of new laser systems for transmyocardial laser revascularization (TMLR).

From a physicist point of view, the TMLR-procedure is generally based on the ability to concentrate laser light in a thin beam of very high intensity, which effectively ablates soft biological tissues. For a surgeon, TMLR is a non-trivial and elegant way to help patients with coronary artery disease. Throughout the world, many investigators utilize and further develop this procedure (see e.g. publications [1-5]) as it could, hopefully, complement or in some cases even substitute traditional angioplasty and complicated heart-bypass surgery. However, further research and clinical data are needed to explain the mechanisms involved. The discussion particularly focusses on how long the created transmyocardial channels could stay open; how these channels affect the electrical conductivity of the perforated region; how extensive the thermal damage, shock waves and mechanical disruptions caused in the myocardium is/are. Today, a *prior* optimization of laser parameters and even the selection of the best laser type for TMLR is difficult. Instead, one could try different operation modes and different lasers, which meet several basic requirements:

- the laser must be able to “drill” the transmyocardial channel with a diameter $d \approx 1$ - 1.5 mm,
- with an acceptable thermal damage of surrounding tissue and
- it should be suitable for the application in an operation room, thus reliable, simple in manipulation, relatively compact, with no electrical interference etc.

Using laser light, three general approaches to heart perforation are possible:

1) creation of the transmyocardial channel with one single powerful laser shot,
2) utilization of a burst of quickly following short laser pulses,
3) use of multiple weak laser pulses, not synchronized with heart rhythm.

The parameters of the laser systems having been evaluated with respect to TMLR have been summarized in Table 1, together with the results published so far. If possible, the important ablation parameters such as peak intensity I (W/cm^2), pulse energy density Φ (J/cm^2) and specific ablation energy w_{abl} (J/mm^3) have been recalculated on the basis of those publications. The following gives a brief account of the essential properties of each laser type and discusses the results obtained.

CO_2 lasers for TMLR.

- **CO_2 laser** emits non visible infrared (IR) light at a wavelength λ = 10.6 µm. The laser is widely used in surgery because of high tissue absorption at this wavelength, good beam focusability, high power, high efficiency and reliability of the system. The light is generated in a mixture of the inert gases CO_2/N_2/He usually excited by means of radio-frequency (RF) or direct current (DC) electrical discharge. In so-called “super-pulsed” or “ultra-pulsed” systems the RF discharge can be pulsed with a periodicity of up to several kHz. Here, instead of continuous oscillation (cw), pulses of several hundred µs or several ms in duration are emitted with a peak power exceeding the cw level by several times. Much shorter (100 ns – 1 µs) and more powerful pulses are generated by transversely excited atmospheric gas pressure (TEA) CO_2 lasers. The TEA systems are, however, less reliable and more cumbersome due to very powerful electrical discharge circuits. The cw power level of medical CO_2 lasers normally varies between 10 and 200 W. The overall electrical efficiency η (“from plug”) is 3-6 %, thereby being considerably higher than the one of most other lasers. An essential disadvantage of the system is not associated with the laser himself, but arises from the absence of thin and flexible light-guides for middle IR.

Compared to other medical applications, a much higher laser power is required in order to create a transmyocardial channel with one single shot. Until now, 800 W quasi-cw (long pulse) CO_2 lasers, produced by PLC Systems Inc., USA, have been employed for this purpose. The laser light is delivered to the beating heart through an intercostal incision 15-25 cm long using an articulated mirror arm and a long focusing optic. New specialized thoracoscopic applicators also allow for surgery through several small (2 cm) incisions [11].

The CO_2 laser pulse duration and thus the total energy applied is typically adjusted to between 20 and 90 msec, depending on the thickness of the myocardium and fat layer. The ECG-triggered pulse is applied at the end of a diastole and does not disturb the heart operation. The entire ventricle perforation procedure requires minimal time (~2 min for 30-50 holes). The observed thermally altered zone in the channel is typically about 200-300 μm thick. The measured damage thickness depends, however, on various experimental conditions and the evaluation method applied. Sensitive methods like birefringence microscopy reveal an extended thermally damaged zone: for a 10 msec pulse duration, publication [5] reports a damage zone of 100 μm in the direction along myofibrils and of 550 μm across them, whilst 300 μm and 650 μm respectively are reported for a pulse duration of 50 msec.

Also an *in vitro* attempt has been made using a burst of msec-pulses generated by an "ultra-pulsed" CO_2 laser [9]. The laser system used ("UltraPulse 5000", Coherent) emitted 0.5 J pulses of 1.4 msec duration at a repetition rate f = 200 Hz. To compensate for smaller average power and to deliver the ablation energy needed (~5 J/mm^3), the beam must be more tightly focused (d = 0.5 mm) and the burst duration has to be increased to up to 0.2 sec. However, it remains to be proven whether this procedure is suitable for TMLR on a beating heart.

XeCl, Nd:YAG and Ho:YAG lasers.

In the next approach, low (1-10 W) average power pulsed laser systems were used for TMLR. Experiments with Ho:YAG (2.1 μm) [4, 6, 7, 9, 10], Nd:YAG (1.44 μm) [6, 7] and XeCl (308 nm) [6, 7, 9] lasers have been reported (Table 1). The small single pulse energies of these systems are compensated for by a short pulse duration, so that the peak light intensities are higher or even much higher than those of the quasi-cw CO_2 laser. This results in strong overpressure of the ablation products and the formation of irregular channels with wall ruptures at the sides. The effect is particularly pronounced for XeCl laser (τ = 120-150 ns, I = 30-80 MW/cm^2, Δp ~ 500 bar). Several tens or even hundreds of laser pulses are necessary to "drill" the channel using low pulse energy. That also adds to the irregularity of the channel geometry.

The long "drilling" duration found with the low average power laser systems forces surgery to be effected on a non-beating heart. Otherwise, a light-guide fiber and/or a special contact applicator has to be used for fixing the "drilling" direction.

Complicated TMLR operation at a non-beating heart with open chest and use of a lung-heart machine is only reasonable when combined with bypass surgery. On the contrary, the use of a highly flexible and thin (0.3-1 mm) quartz light-guide fiber or a bundle of 50-100 μm fibers with a total diameter of up to 1.5 mm provides for the advantages of minimally invasive surgery. The fiber could be directed to the myocardium through a very small incision or even percutaneously [4] through the femoral artery like in balloon angioplasty. TMLR from inner endocardial side would likely provide for some additional advantages, e.g. no outward bleeding if the TMLR channel end is close to the outer epicardial heart layer. The direction of the TMLR channels can also be easily regulated using a fiber.

The availability of thin and flexible low-loss silica (quartz) fibers for the spectral region of 0.2-2.3 μm was one of the main reason to try Ho:YAG, Nd:YAG and XeCl laser. The specific ablation energy values given for each of these systems (see Table 1) are only estimated, because a precise geometry of the ablation channels (ablated volume) was not reported in the referred publications. The specific energy found typically varies between 2 and 5 J/mm^3. The value of 2 J/mm^3 just represents the energy needed to heat up and evaporate the tissue water.

- **XeCl lasers** belong to the family of excimer lasers. Constructionswise, they are very similar to TEA CO_2 lasers, but use a gas mixture of $Xe/Cl_2/He(Ne)$ as active medium. The excited atoms in the discharge form an XeCl dimer (excimer), which dissociates very fast, emitting short (10 – 200 ns) ultraviolet (UV) light pulses at λ = 308 nm. The typical "surgical" XeCl laser is relatively compact and emits several tens of mJ with f < 100 Hz in a rather non-uniform beam profile, which may contain undesirable "hot spots" (local intensity peaks). Its overall "plug" efficiency is η < 1 %. In addition, the corrosive and toxic laser gas mixture must be exchanged after several million shots and also other technical services become necessary every few months.

An extremely small specific ablation energy, w_{abl} = 0.04 J/mm^3 at Φ = 2.7 J/cm^2, and a short "drilling" time (< 1 sec) was reported for XeCl lasers in [9]. From other publications [6, 7] a “normal” w_{abl} value of 2-4 J/mm^3 at Φ = 5-10 J/cm^2 could be recalculated. The difference most probably relates to an artifact. In [9], a 950 μm fiber was pushed into the myocardium with a static force of 0.8 N (80 g). So, we suppose, it was rather “laser-assisted” mechanical perforation than ablation in that case. If such mechanical effect is predominant, the final channel may be smaller than the fiber’s (applicator’s) diameter because of the myocardium’s elasticity. In this situation, the thermal damage would be minimal. In any case, according to [6, 7] the excimer lasers provide for the smallest thermal damage i.e. less than 50 μm. It has already been reported [6] that operations on a beating heart using a XeCl laser at f = 20 – 40 Hz and no ECG-triggering are well tolerable.

- **<u>Nd:YAG lasers</u>** are solid-state lasers (1-2 % of Nd-ions in the Yttrium Aluminum Garnet crystal matrix) with a main emission wavelength of 1.064 μm in the near-IR region. The crystal rod is excited using flash-lamps (the lamps must be exchanged after ~10^6 shots; poor laser beam quality, η < 3 %) or laser diodes (nearly perfect Gaussian output beam, η is up to 10 %). Nd-laser light causes energetic blood and tissue coagulation and is less effective with respect to the ablation when compared to the other laser systems described.

The Nd:YAG laser used in [6, 7] was especially adjusted to λ = 1.44 μm (at the expense of laser efficiency), where muscle tissue absorbs better than at the normal wavelength of 1.06 μm. However, the absorption at 1.44 μm is still lower than at longer IR wavelengths or in the UV region, so that the specific ablation energy amounts to 5 J/mm^3 causing larger thermal damage than the other systems. To our knowledge, there are no reports on any *in vivo* TMLR attempts with this laser.

- **<u>Ho:YAG lasers</u>** use - like the Nd-laser - a YAG crystal matrix doped with Ho and additionally with Cr and Tm ions. The pulse duration usually varies between 100 and 600-800 μs; the pulse energy amounts to few J; repetition rate f < 40 Hz; efficiency η < 1 %.

The absorption of soft biotissues increases in the IR region. The Ho:YAG laser (2.1 μm) emits the longest wavelength, which could still be guided by silica fiber. Commercial Ho:YAG laser systems are now marketed being equipped with special TMLR fibers (e.g. from CardioGenesis Corp., USA). Interesting results of *in vivo* studies on canine subjects have been published, indicating that both epicardial [10] and percutaneous endocardial TMLR [4] are well bearable when using fiber-equipped Ho:YAG lasers and do not lead to any ventricular dysfunction.

The above revue of the literature data and different laser systems does not yet allow to decide on the best alternative to high-power CO_2 TMLR lasers. To evaluate the prospects of new systems an analysis of the main physical mechanisms of laser-tissue interaction is required.

Basic physical mechanisms of laser tissue ablation.

Laser light can be absorbed, scattered and partially reflected by tissue. The reflection at muscles and blood is small. However, scattering is a significant factor, particularly in the near-IR ($0.76 < \lambda < 2.5$ μm), visible (400-760 nm) and near-UV region, where it even dominates absorption for the most tissues (see Table 2). In that region, the wavelength is comparable to the size of numerous intercellular elements. Thus, the so-called Mie scattering is very effective, preferentially in forward direction (anisotropy factor $g > 0.8$). In the UV spectral range ($\lambda < 400$ nm) Rayleigh scattering (scattering coefficient $\mu_s \sim 1/\lambda^4$) is considerable large as well.

Absorption in the infrared region is associated with vibrational excitation of molecules. Because of the high water content (70-85 %), the absorption coefficient μ_a for soft biotissues is very close to the μ_a value for H_2O at $\lambda > 1.5$ μm. An especially strong H_2O absorption band at 3 μm practically coincides with the wavelength of the Er:YAG laser. A high absorption is also observed in the UV region due to electronic absorption bands of proteins. In the visible and near-IR region hundreds of chromophores (e.g. hemoglobin) contribute to the absorption, but it is much weaker than in the IR or UV region.

The absorbed energy can induce ablation in different ways. Two mechanisms are most important for soft biotissues: thermal ablation (vaporization) and direct photoablation. If high-energy UV photons are applied, molecules can be excited in highly lying repulsive electronic states. The excitation is followed by “immediate” molecular dissociation and ejection of fragments. Such a direct photoablation is characterized by very clean ablation craters. Only a minor part of the laser pulse energy remains in the tissue in form of heat, so that the inherent thermal damage is very small. The efficient direct photoablation is observed with an ArF laser

(193 nm), whose photon energy (6.4 eV, Table 2) is just large enough to break the strong double C=C chemical bond. The 193 nm laser light can not however be satisfactory transmitted through light-guide fibers. Other excimer system, a KrF laser (248 nm, 5 eV), is suspected to cause mutagenic effects in the cells and is therefore not applied in medicine. The photon energy of the XeCl laser (4 eV) is high enough to break only single chemical bonds in organic molecules, so that the thermal mechanism competes with the photoablation at this wavelength.

With the thermal ablation a very fast evaporation of the tissue water occurs. The energy, initially deposited into the molecular internal motion, is transformed into heat at a picosecond time scale. The following explosion-like vapor expansion causes a precise removal of the affected tissue. The vapor does not only carry away the cell debris but also most of the absorbed energy. The specific ablation energy w^{o}_{abl} (J/mm^3) can be roughly estimated as the energy needed for heating up the tissue from 37 up to 100°C and evaporating it:

$$w^{o}_{\mathrm{abl}} = \rho\,(c \cdot \Delta T + L\,) \approx 2\ \mathrm{J/mm^3}\,, \tag{1}$$

with the myocardium density ρ = 1.06-1.08 g/cm^3, the heat capacity c = 3.55-3.72 J/g/K, the latent evaporation heat $L \approx$ 1800 J/g and ΔT = 63K. Any additional losses due to heat transfer into the surrounding tissue, also partial light absorption by the ablation debris etc. have not been accounted for.

The calculated "internal" specific ablation energy w^{o}_{abl} does not depend on laser parameters. Contrarily, a light energy density Φ^{o} (J/cm^2), which should be delivered to the tissue for the ablation, depends on the absorption (μ_{a}) and scattering (μ_{s}), and thus on the laser wavelength. The laser light intensity in tissue quickly (~ exponentially) drops with the depth. About 2/3 of the energy is lost after an effective absorption length,

$$\begin{array}{ll} l_a = 1/\mu_a\,, & \text{at } \mu_{\mathrm{a}} >> \mu_{\mathrm{s}}\,, \text{ or} \\ l_a = 1\Big/\sqrt{3\mu_a\left(\mu_a + \mu_s(1-g)\right)}, & \text{in case of a strong scattering.} \end{array} \tag{2}$$

To vaporize the layer of the depth l_{a} one needs obviously at least an energy density

$$\Phi^{o} = w^{o}_{\mathrm{abl}} \cdot l_{\mathrm{a}}\,. \tag{3}$$

The calculation of the l_{a} and Φ^{o} values for different laser types is given in Table 2. The smallest Φ^{o} values are achieved with Er:YAG (Φ^{o} = 0.2 J/cm^2), followed closely by CO_2 laser.

For XeCl laser reference [7] reports a value l_a of only 6 μm, which leads to Φ^o = 1.2 J/cm^2. Other sources ([12-15]) published considerably higher values for biological tissues (e.g. $l_a \approx$ 100 μm for aorta wall). Ho:YAG and Nd:YAG lasers demonstrate much poorer ablation characteristics because of weak absorption and strong scattering at their wavelengths.

An efficient ablation is always accompanied by a minimal thermal damage of the affected tissue. The following rules can be set up:

a) The absorbed laser energy has to be concentrated in a minimal volume (easy achievable for tissue with high μ_a and low μ_s). The residual light, which penetrates deeper into the tissue (beyond the ablation volume) does not generate enough heat for evaporation, but induces thermal damage.

b) At a fixed pulse energy a shorter, thus more intensive laser pulse induces faster evaporation than a longer low-intensity pulse. At a sufficiently high intensity I, the evaporation front moves into the tissue much faster than heat could do. Then, the energy loss to the surrounding tissue is minimal and the thermal damage zone at the bottom of the ablation channel is approximately l_a . As a criteria, one could compare the characteristic ablation time $\tau_{abl} = \Phi^o/I$ to the thermal relaxation time $\tau_T = l_a^2/4\delta$ (thermal conductivity δ = 0.0012-0.0015 cm^2/sec for myocardium). The thermal relaxation time shows how fast the heat deposited in a thin absorption layer l_a, is transferred to the deeper tissue. The value of τ_T is only 2 μs for the Er:YAG laser but 3.5 sec for the Nd:YAG laser. Nevertheless, at their typical pulse duration and intensity (see Table 1) all lasers included in Table 2 meet the criteria $\tau_{abl} \ll \tau_T$. We like to emphasize, that the often used comparison of the thermal relaxation time τ_T with laser pulse duration is actually of no meaning since the thermal relaxation occurs independent of the laser pulse width. A short ($\tau \ll \tau_T$) but not intensive enough ($\tau_{abl} > \tau_T$) pulse produces a massive thermal damage.

c) Edges and walls of the ablation channel reveal an extended thermal damage due to interaction with the low-intensity wings of the laser beam profile (heating only) and because they are affected by multiple laser pulses. This damage could be reduced applying a “hat” intensity profile of the laser beam.

d) Short sub-μsec laser pulses cause an additional photo-mechanical ablation effect due to a very high overpressure of the evaporated liquid. Induced microexplosion is able to mechanically obliterate some extra tissue volume. Such an ablation is characterized by less thermal damage and by a certain degree of mechanical ruptures. Experiments with 100-ns-pulses of TEA CO_2 lasers confirm this suggestion [16-18]. The reported threshold of 0.7 J/cm^2 in [16] is more than 4 times smaller than the Φ^{o} value for 10.6 μm.

New lasers for TMLR?

The above consideration indicates that the Er:YAG laser is an excellent ablation instrument. That is why Er:YAG systems have been so intensively developed over the last years (see e.g. [19]).

- **Er:YAG laser** (2.94 μm) is similar to Nd:YAG or Ho:YAG laser. A modern Er:YAG system delivers up to several hundreds of mJ with $f \approx 100$ Hz, or even several J at smaller repetition rates. The pulse duration is $\tau = 100$ - 800 μs; the overall efficiency is $\eta \approx 1$ %.

The achieved Er laser pulse energy is not high enough to "drill" a transmyocardial channel with one shot. A light-guiding fiber application is impossible due to the lack of appropriate mid-IR fibers. The most advanced Zirconium-Fluoride fiber is brittle and hygroscopic. Other existing mid-IR light-guides manifest high losses or some other drawbacks. However, a further rapid progress in the development of mid-IR light-guides is to be expected. Here, we refer to e.g. the interesting reports about hollow glass light-guides with inner metallic/dielectric layers [20,21] and the liquid-core light-guides [22] recently developed by our group.

In our opinion, the utilization of short sub-μsec pulses generated by a CO_2 laser could be very promising for TMLR applications. Such pulses could provide for the already mentioned advantages of a smaller ablation threshold, higher efficiency and less thermal damage. However, the use of CO_2 lasers prevents the application of flexible light-fibers and the light must be delivered to the myocardium in a single burst of many pulses with an optimal pulse energy and peak intensity. The total burst duration can be very short due to the high pulse repetition rate.

We recommend the use of a "Q-switched" high-repetition-rate CO_2 laser (see Figure) as a new source of sub-μsec IR pulses.

- **Q-switching** allows time controlled laser oscillation and is widely used in laser technology. Blocking the laser action within the cavity by some means, the usual population inversion in the laser active medium can largely be increased. Then, with an appropriately fast switching method the quality Q of the cavity can be brought back to a large value resulting in a very short and intensive emission of laser light.

Our laser system has been originally developed for photochemical laser enrichment of stable isotopes [23]. It is based on a fast mechanical Q-switch and provides for a convenient way of generating short and powerful pulses at a high repetition rate. Commercially available cw or super(ultra)-pulsed CO_2 laser systems could be upgraded with this Q-switch technology. For this, one has to install an additional optical telescope and a fast rotating chopper disk which interrupts the beam in the telescope focus plane into the laser cavity. As a result, very short, 100-500 ns, laser pulses are generated at a peak power being several hundred or even several thousand times higher than the cw laser power level. The pulse repetition rate could be adjusted to between 100 Hz and 100 kHz. At a high repetition rate (tens of kHz) the average output power of the optimized Q-switched laser is not far from the cw power level.

The parameters of our research Q-switch CO_2 laser system are summarized in Table 1. The duration of a single pulse is 250-350 ns. The pulse energy amounts to up to 50 mJ at 200 Hz and 20 mJ at 10 kHz resulting in an average power of up to 400 watts. Focused down to a diameter of 1 mm, the beam yields a peak intensity $I \geq 10$ MW/cm^2. Our preliminary experiments have shown, that at such intensity the ablation was distinctly promoted by additional photo-mechanical effects whilst at the same time undesired plasma formation could be avoided. A relatively strong acoustic effect accompanies tissue incision. It can be easily monitored with a simple microphone and used for process control. A detailed investigation of the physical ablation parameters and thermal effects is currently under progress.

Conclusion.

The high-power quasi-cw CO_2 laser has been tried for TMLR operations on beating heart for several years. The main requirement for the development of this "heart" CO_2 laser was the need to "drill" the transmyocardial channel with only one shot. The new results show, however, that TMLR can also be successfully done with a thin light-guide "slowly" penetrating into the myocardium and transmitting multiple weak laser pulses to the tissue. Several disadvantages of the high-power CO_2 laser became evident. The system is cumbersome and light delivery relies on an articulated arm mirror system, which strictly limits the operation technique. Excess laser pulse energy is required in order to "drill" the channel with certainty. That leads to evaporation of blood at the epicardial side and formation of gas bubbles in the ventricle. To control the procedure, a transesophageal echocardiographical probe must be inserted into the gullet. The extent of the thermally damaged zone around the channel is at least 0.2 – 0.3 mm, and can be considerably larger according to some reports.

Smaller thermal damage and more efficient ablation could probably be attained with a burst of short sub-μsec pulses of CO_2 laser. We have already demonstrated [24], that 300 ns pulses of a high-repetition-rate Q-switch CO_2 laser "cleanly" ablate a bone tissue. Current experiments with soft biological tissues are intended to clarify, how much the ablation efficiency might be improved and the thermal damage reduced, as compared to quasi-cw or super-pulsed CO_2 laser system.

It is known, that superficial ablation characteristics can be realized in the IR with an Er:YAG laser at $\lambda = 2.94$ μm. Unfortunately, the lack of suitable fiber light-guides for $\lambda > 2.3$ μm still prevents the application of this system, which is otherwise very favorable for medical applications. Promoted by several companies, other IR medical systems like the Ho:YAG laser (2.1 μm) work well with silica-glass fibers and are convenient to use. However, their ablation characteristics are much poorer than those of Er:YAG, CO_2 or excimer lasers.

Of all systems under evaluation, the XeCl laser so far caused the smallest thermal damage. The system's efficiency in terms of ablation is ensured by extremely high photon energy and peak intensity of the laser pulses, as well as by the high tissue absorption in UV. The possibility to use a thin and flexible light-guide fiber largely facilitates surgery and permits the application of the percutaneous technique. The low average power excimer laser is also much smaller and not so expensive as the "heart" CO_2 laser. These features make the XeCl laser a very interesting alternative to the high-power CO_2 laser at TMLR.

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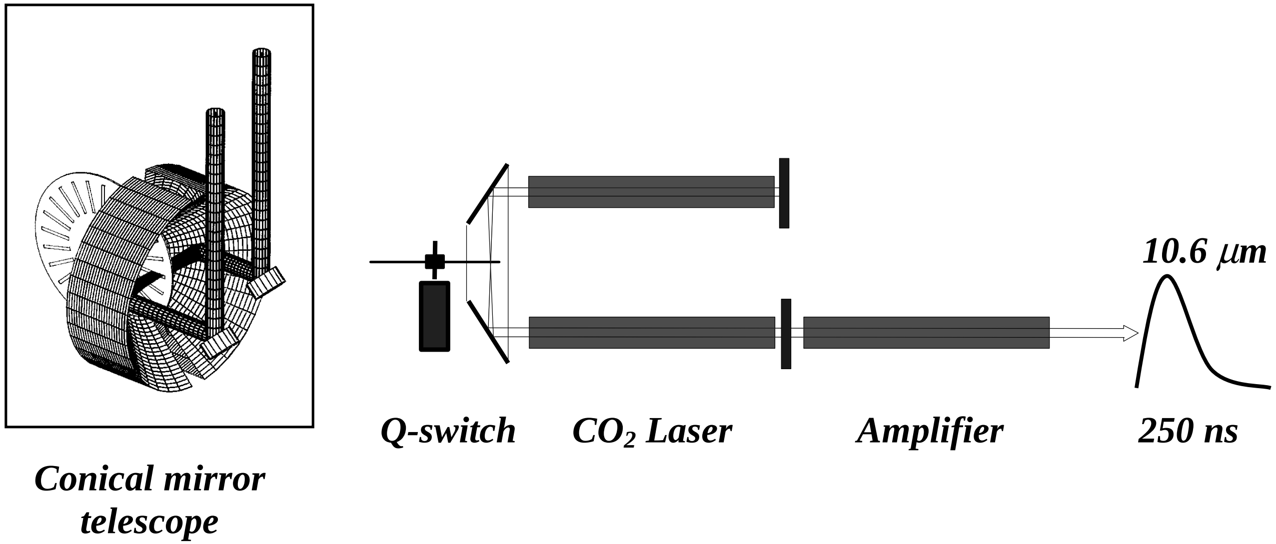
Table 1. Laser systems for TMLR.

Laser:	**CO_2** quasi-cw	**CO_2** ultra-pulsed	**CO_2** Q-switch	**Ho:YAG**			**Nd:YAG**	**XeCl**		
wavelength, λ	10.6 μm			2.1 μm			1.44 μm	308 nm		
pulse energy, E	15 - 40 J	0.5 J	0.02-0.05 J	1 - 2 J	0.75 J	2 J	1 - 2 J	20 - 40 mJ	20 mJ	21 mJ
pulse duration, τ	20 - 50 ms	1.4 ms	250-350 ns	100 - 500 μs		~ 350 μs	600 μs	150 ns	120 ns	
repetition rate, f	single shot	200 Hz	200-20000 Hz	5 - 20 Hz	5 Hz	3 pulse per diastole	5 - 20 Hz	40 Hz	40 Hz	40 Hz
average power, P	(800 W cw)	100 W	10-400 W	5 - 10 W	3.75 W		5 – 10 W	0.8 - 1.6 W	0.8 W	0.82 W
beam delivering	articulated mirror arm			silica (quartz) fiber of 0.3-1 mm core diameter or fiber bundle						
channel diameter, d	1 mm	0.5 mm	1 mm	1 mm	1 mm	1.75 mm	1 mm	0.5 - 1 mm	0.5 mm	(≤ ?) 1 mm
peak intensity, I	0.1 MW/cm^2	0.2 MW/cm^2	~10 MW/cm^2	1.2-0.5 MW/cm^2		0.24 MW/cm^2	0.2-0.4 MW/cm^2	70-30 MW/cm^2	~ 80 MW/cm^2	
time per channel	20 - 50 ms	200 ms		4 - 8 s	4 s	~ 5 s	4 - 8 s	3 – 6 s	10 s	< 1 s
pulses per channel	1	40		40-80	20	12/ 8 mm	40-80	120-240	400	30
pulse energy density, Φ	2000-5000	250 J/cm^2		130-250 J/cm^2	100 J/cm^2	85 J/cm^2	130-250 J/cm^2	10-5 J/cm^2	10 J/cm^2	2.7 J/cm^2
specific ablation energy, w_{abl}	**2 - 5 J/mm^3**	**~ 5 J/mm^3**	reduced ?	**~ 5 J/mm^3**	**~ 1 J/mm^3**	**1.25 J/mm^3**	**~ 5 J/mm^3**	**~ 2 J/mm^3**	**~ 4 J/mm^3**	**≥ 0.04 J/mm^3**
thermal damage	> 200 μm		reduced ?	~ 300 μm		~ 75 μm	~ 350 μm	10 - 50 μm	30 μm	
shock wave pressure	~ 3 bar			~ 10 bar			~ 10 bar	~ 500 bar		
geometry	smooth channels			irregular channels with ruptures to the side						
specimen (m. = myocardium)	**beating heart**	*in vitro* bovine m. 2 cm thick		*in vitro*	*in vitro* bovine m. 2 cm thick	**beating dog heart**	*in vitro*		*in vitro*	*in vitro* bovine m. 2 cm thick
additional conditions	ECG triggering		proposal, HHU Düsseldorf		35 g pushing force	ECG triggering				80 g pushing force
ref.	[6-8]	[9]		[6,7]	[9]	[10]	[6,7]	[6]	[7]	[9]

Table 2. Optical and ablation characteristics of some human tissues at different wavelengths.

Laser:		**CO_2**	**Er:YAG**	**Ho:YAG**	**Nd:YAG**	**Nd:YAG**	**XeCl**	**KrF**	**ArF**
wavelength, λ		10.6 μm	2.94 μm	2.1 μm	1.44 μm	1.064 μm	308 nm	248 nm	193 nm
photon energy, eV		0.1	0.4	0.6	0.9	1.2	4.0	5.0	6.4
coefficients of ***absorption, μ_a/ scattering, μ_s***, cm^{-1} (ref. in [13-15])	*blood:*	$\approx \mu_a(H_2O)$	$\approx \mu_a(H_2O)$	27.91 / 203.5	34.35 / 359.4	9.77 / 508.6	250-260		
	muscle:	$\approx \mu_a(H_2O)$	$\approx \mu_a(H_2O)$			2 / 215			
	myocardium:	$\approx \mu_a(H_2O)$	$\approx \mu_a(H_2O)$			0.3-1.4 / 180-320			
H_2O μ_a, cm^{-1} [12]		860	12000	36	11	0.61	0.0058	0.018	0.1
absorption length, l_a		15 μm	1 μm	~ 200 μm	~ 300 μm	≥ 1300 μm	6 μm ?		
threshold ablation energy density, Φ^o		3 J/cm^2	0.2 J/cm^2	~ 40 J/cm^2	~ 60 J/cm^2	≥ 250 J/cm^2	> 1 J/cm^2		



Mechanically Q-switched high repetition rate CO_2 laser-amplifier system with conical mirror telescope.