

Investigation on Laser Dental Implant Decontamination

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During the last decades, the use of dental implants for tooth replacement became state of the art in dental prosthetic therapy. The high survival rate of osseointegrated dental implants is well documented, but it is becoming increasingly evident that even successfully integrated implants are susceptible to diseases that finally may lead to implant loss. One of the most frequent reasons for implant failure nowadays is periimplantitis caused by pathological inflammatory changes in the tissue being adjacent to a load bearing implant. Clinically it is associated with pocket formation, suppuration, bleeding and implant mobility.

Numerous methods for implant surface decontamination have been suggested as a part of the surgical treatment of periimplantitis. Actually, decontamination of infected implant surfaces can be achieved very effectively by application of laser radiation. Hence, the purpose of our study was to evaluate suitable parameters for laser decontamination of titanium plasma-sprayed implant surfaces by employing laser pulses in the microsecond range generated by two laser systems (Diode and frequency-doubled Nd:YAG). In this paper the results for temperature development and surface morphology of different laser parameters and their clinical relevance and limitations are reported.

Keywords: dental implants, implant surface, periimplantitis, laser decontamination, temperature development

1. Background and objective

During the last decades, the use of dental implants for replacement of missing or extracted teeth became state of the art in dental prosthetic therapy. Actually, current goals of dental therapy include the development of prosthetic fixtures for this purpose being easy to fabricate and implant. The high survival rate of osseointegrated dental implants is well documented, but it is becoming increasingly clear that successfully integrated implants are susceptible to disease conditions that may finally lead to their loss. As with any treatment modality, however, failure can occur with implants. One of the most probable reasons for implant failure nowadays is periimplantitis.

Generally, periimplant disease refers to the pathological inflammatory changes that take place in the tissue surrounding a load bearing implant. This type of inflammation is associated with the typical clinical symptoms of periimplantitis like increased probing depth (i.e. pocket formation), bleeding, mobility and suppuration for the implant representing. Implant failure may be the final result of one or several pathogenic factors such as infection, implant contamination, trauma during surgery, impaired healing and/or premature loading during the healing process.

During periimplantitis therapy, contaminants such as bacteria and their products, calculus, and soft tissue cells should be removed without modifying the implant surface. However, it is still unknown to which extent these contaminants have to be removed to achieve a successful treatment outcome [1].

Various treatment possibilities for plaque-induced inflammatory lesions in the periimplant tissues have been reported in literature [e.g. 2-5]. Thus, mechanical debridement, antiseptics, antibiotics, surgical procedures and explantation were suggested depending on the severity of the clinical and radiographic manifestations of the lesions. Surgical treatment of periimplantitis lesions can be performed in cases with considerable pocket formation (larger than 5 mm) and bone loss, after the acute infection has been resolved and proper oral hygiene has been established [6]. It has been suggested that the provision of an implant surface conducive to bone formation is a prerequisite for successful regenerative treatment of periimplantitis [7]. The present study puts its focus on the issue of applying lasers for decontamination of periimplantitis lesions around commercially available titanium implants. The results should provide a solid base for recommendation of laser parameters for safe therapy.

The development of laser technology is in a fast progress particularly concerning its adaptation for medical applications. Nevertheless, it still requires profound knowledge to select the right laser device and the parameters fitting for the planned therapy.

The clinical target of periimplantitis therapy is mostly the elimination of the inflammatory lesion caused by bacterial infection around the implant, the inhibition of further progression of the disease and the maintenance of the implant function in situ with sound re-attached bone and soft tissues [12-14].

In several studies the bactericidal effect of high-power microsecond-pulse laser irradiation on contaminated dental implant surfaces already has been demonstrated. Usually, the laser energy is strongly absorbed in water, thus causing preferentially the evaporation of water-rich tissues. In bacterial cytoplasm this effect causes cellolysis. Publications have shown the effectiveness of lethal photosensitization in decreasing the viable count of periodontal pathogens in periimplantitis lesions without damage of the dental implant surface [8, 15].

The diode laser as a typical medical laser device with a broad range of possible applications has already been investigated in literature for temperature development during simulated laser periimplantitis treatment. Temperature elevations during irradiation were registered for a period of 120s. An estimated critical threshold of 47°C was exceeded after 9.0s at 2.5W, 12.5 s at 2.0W, 18.0s at 1.5W and 30.5s at 1.0W. Specific implant surface characteristics were not reported to have a significant effect on the temperature elevations. The study recommended that with respect to the applied energy, implant surface decontamination with an 809nm Diode laser must be limited in time to allow the implant and bone to cool down [9].

The estimated threshold for temperature increase in biological tissue is based on the finding that temperatures beyond 47 to 50°C (i.e. a temperature increase of 10–13°C above body temperature) can induce tissue damage in the bone [10, 11]. It must therefore be ensured that this threshold is not exceeded during laser application.

Another problem is that many authors just reported the parameter settings they have chosen on the control panel of their devices but never checked for the real output. Especially for Diode lasers the real output parameters can differ significantly from the programmed settings, as older devices often do not calculate for the specialties of chopped mode operation in their displayed values.

However, the effects of heat application to already osseointegrated implants are still not sufficiently known and therefore further research to clarify the safety of such laser techniques is necessary.

According to this, the primary goal of the present study was the evaluation of the commonly suggested laser parameters regarding the temperature development during laser implant decontamination. Hence, the experiments were performed following established protocols for periimplantitis therapy under clinical conditions, as particularly implant surface alterations possibly inhibiting osseointegration and temperature rise inducing bone necrosis above a certain level were considered to lead to implant failure.

2. Material and methods

For the investigation of the implant temperature development during laser decontamination, two laser systems with different wavelength were applied: a diode-pumped frequency-doubled Nd:YAG laser (mostly referred to as “KTP” in medical applications, $\lambda=532\text{nm}$) with pulse durations of 20 and 50ms and a PRR of 10Hz and a Diode laser ($\lambda=810\text{nm}$) with pulse durations in the μs range and a very high pulse repetition rate (PRR) in the 20kHz regime. Both lasers were using 300 μm fiber as a delivery system. Their average power settings were chosen according to the values suggested by the manufacturer for this purpose, i.e. 1.05,

1.5 and 1.95W for the Diode laser, and 1.0, 1.5, 2.0 and 2.5W for the KTP laser. It is important to mention that in medical applications, mostly just the displayed output powers are considered by the user with almost no attention paid on rechecking the differences between the displayed output power and the real output power. This fact can directly influence the temperature results according to our practical observations and recorded data.

The actual output powers for both lasers were recorded by a thermal measurement head (Ophir Optonics 10A-FS) to be compared with the displayed power.

For the investigation of possible surface alterations of the implants during irradiation, different output powers were applied on new titanium plasma sprayed implants of 4.1mm diameter and 12mm length (Straumann, Switzerland). The implants were fixed in a mechanical holder and the laser radiation was applied under an irradiation angle of 30° between implant surface and fiber. All irradiated implants were investigated under ESEM for potentially induced morphological damages.

For the temperature evolution measurements, a block of polyoxymethylene (POM) having a comparable heat conductivity to bone tissue was prepared with a geometry being in the order of magnitude of a human jaw bone. An implant bed with artificial cavity, i.e. a deep pocket lesion, was processed in the block to simulate the typical shape of a periimplantitis lesion. An implant of 4.1mm diameter and 12mm length was mounted in this bed to simulate the in-vivo situation of the implant with adjacent bone defect. Before and during the experiments, the block was placed in a water bath having a constant temperature of 37°C for simulation of the intra-oral in vivo conditions (see Figure 1).

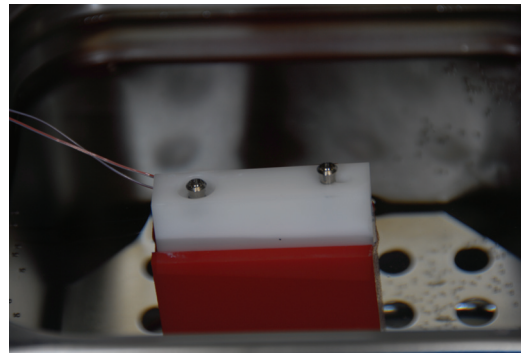


Fig. 1 For evaluation of the temperature development during laser irradiation, the implants were embedded in a POM block model placed in a water bath at 37°C. Two T-type thermocouples were connected to the implant surfaces in the apical (lower third) and coronal (upper third) region through two small holes in the block.

For the temperature recordings, two T-type thermocouples were connected to the implant surfaces in the apical (lower third) and coronal (upper third) region through two small holes in the block. The measured temperature values were directly transferred to a personal computer connected to the electronic measurement device (see Figure 2). Twelve temperature measurements were recorded for each power setting and the average, maximum and minimum temperature course was calculated.



Fig. 2 Experimental setup for temperature recordings. The thermocouples attached to the implant surfaces were connected to the electronic measurement device and the temperature was directly read out to a personal computer.

The implants were irradiated in cycles of 4x5s. After each 5s of laser irradiation under constant, slow movement of the laser beam over the exposed implant surface from the bottom of the lesion to the neck of the implant, the implant surface was allowed to cool down for a certain intermission time. To investigate the influence of this cooling time, experiments with 1s and 5s intermission time between the single irradiation periods were performed. This irradiation procedure was according to usual clinical procedure for dental pocket and periimplantitis decontamination. The temperature was recorded continuously from the start of the irradiation and subsequently till the implant was cooled down again to the initial temperature. After laser irradiation the cover of the water bath was closed during cooling of the implant to simulate the oral cavity closed by the patient after treatment.

Opposite to the cited literature, a lower temperature limit of 4°C above body temperature (i.e. about 41°C absolute) was chosen, as this is a level where still no significant inhibition of enzymatic activities being important for tissue survival takes place. Higher temperatures already result in inhibition of enzymatic processes by destabilization of proteins, thus representing a first induction of processes for permanent cell damage [16].

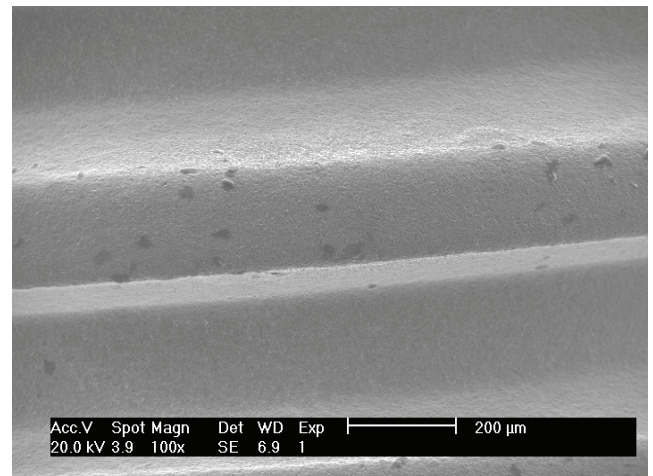
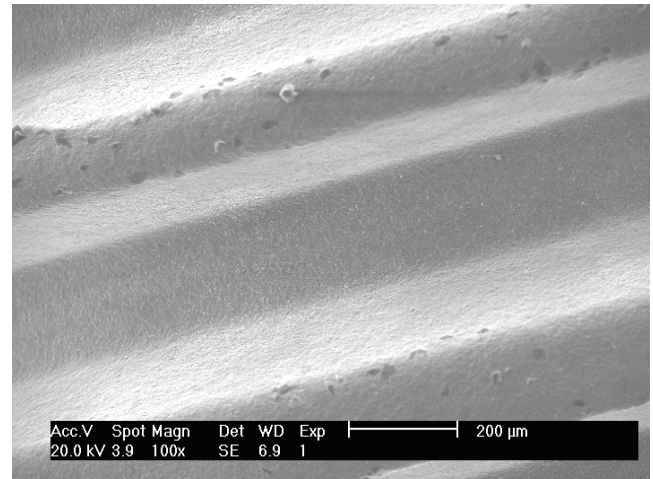


Fig.3 Screw-shaped Titanium plasma-sprayed (TBS) implant surfaces investigated under ESEM (100x magnification). The grooves representing the threads of the implant showed no structural modifications or damaged regions in the irradiated sites. Above: Implant irradiated with Diode, $\lambda=810\text{nm}$, 20Hz, 1.05W. Below: Implant irradiated with frequency-doubled Nd:YAG, $\lambda=532\text{nm}$, 10Hz, 50ms, 1.0W. The small particles on the surface are artifacts placed during examination.

Output power	Maximum temperature rise											
	coronal						apical					
	Diode		KTP 20ms		KTP 50ms		Diode		KTP 20ms		KTP 50ms	
	$t_i=1\text{s}$	$t_i=5\text{s}$	$t_i=1\text{s}$	$t_i=5\text{s}$	$t_i=1\text{s}$	$t_i=5\text{s}$	$t_i=1\text{s}$	$t_i=5\text{s}$	$t_i=1\text{s}$	$t_i=5\text{s}$	$t_i=1\text{s}$	$t_i=5\text{s}$
0.2W			0.7°C	0.7°C					0.4°C	0.5°C		
0.3 W			1.5°C	0.8°C					0.8°C	0.4°C		
0.5 W			2.3°C	1.1°C	1.9°C	0.9°C			1.2°C	0.5°C	1.3°C	0.6°C
0.8 W					3.4°C	1.6°C					1.5°C	0.9°C
1.1W	10.6°C	2.4°C			5.6°C	2.2°C	5.8°C	2.3°C			2.2°C	1.1°C
1.3W					6.9°C	2.9°C					3.8°C	1.7°C
1.4W	11.6°C	2.3°C					7.1°C	2.2°C				
1.7W	13°C	2.5°C					7.5°C	2.3°C				

Table 1: Maximum values of the temperatures at the apical and coronal regions of the implants recorded during irradiation with the listed output powers of the applied laser systems. Significant differences in the temperature rise were recorded between the two selected intermission times for cooling, $t_i = 1\text{s}$ or 5s , during the applied treatment cycles. Of course the diode laser yields a higher temperature increase due to the generally higher output power. Nevertheless, comparing the range of identical output powers, the frequency-doubled Nd:YAG shows significantly less temperature rise for 1s intermission time. For 5s intermission time the differences are reduced and both lasers show acceptable results.

3. Results and discussion

3.1 Surface morphology

The observation for the lased implants under ESEM showed a sound and homogeneous surface structure after treatment for all applied laser wavelengths and power settings with no molten or altered areas (see Figure 3). Hence, the suggested prerequisites for supporting a good re-attachment of the bone structures [7] are perfectly met.

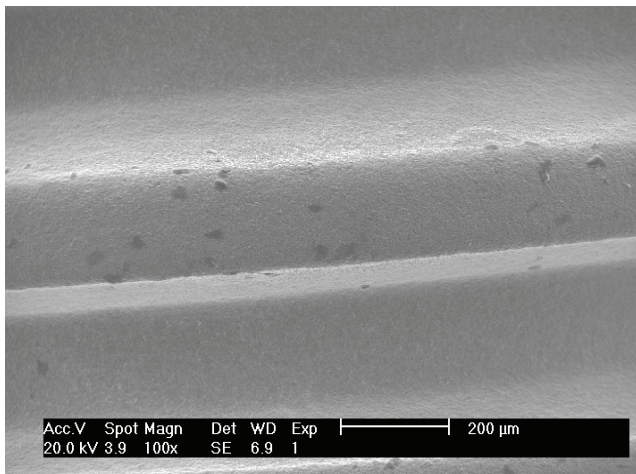
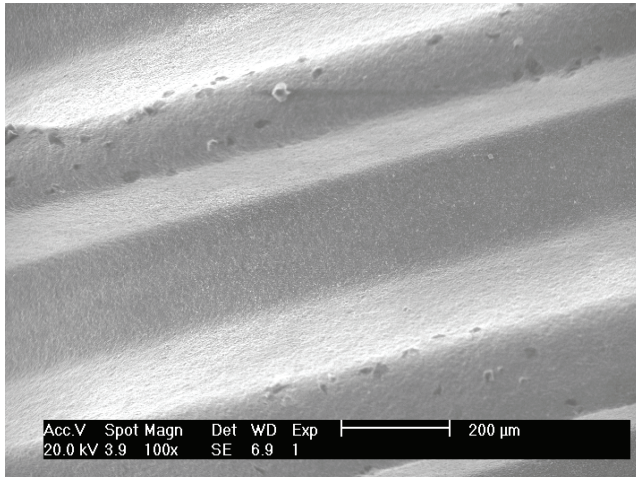


Fig.3 Screw-shaped Titanium plasma-sprayed (TBS) implant surfaces investigated under ESEM (100x magnification). The grooves representing the threads of the implant showed no structural modifications or damaged regions in the irradiated sites. Above: Implant irradiated with Diode, $\lambda=810\text{nm}$, 20Hz, 1.05W. Below: Implant irradiated with frequency-doubled Nd:YAG, $\lambda=532\text{nm}$, 10Hz, 50ms, 1.0W. The small particles on the surface are artifacts placed during examination.

3.2 Recorded output power

For the Diode laser, the recorded output power was in good correlation to the displayed power settings (compare Table 2). Opposite to this, the measured output power of the KTP showed remarkable aberrations. This can be explained by the chopped operation mode and the internal software not calculating for these conditions and thus not correcting the displayed values. Chopped operation mode means that the laser is set to a certain value representing the maximum

output power (or, in other terms, the peak power) of the emitted laser pulses. With this maximum power, the laser is turned on and off according to the chosen pulse duration and frequency. As just the maximum power is displayed on the control panel, the real average power depends on the duty cycle of the chopped operation, i.e. in other terms the ratio of pulse period (in our case 100ms) and the pulse duration (20 and 50ms, respectively). Hence, depending on the chosen pulse length, the real average output power will just be 20% or 50% of the displayed value. Considering this specialty, the measured output power is in good accordance to the theoretically possible values.

3.3. Temperature measurements

Table 1 shows the recorded temperature rise according to the different output powers for the investigated lasers. The temperature measurements clearly indicate that lower irradiation powers cause a lower temperature rise in the implant in an almost linear correlation. Beside this linear correlation, the Diode laser shows a higher temperature rise compared to the frequency-doubled Nd:YAG laser (KTP) at the implant surface. This effect is more significant for 1s than for 5s cooling per cycle, probably because of the lower cooling time. It will have to be investigated if this effect occurs due to a difference in the absorption of the applied wavelengths at the implant surface or any other effect.

The results of Table 1 are also graphically displayed in Figure 4. In this graph it easily can be seen which power settings exceed the allowed limit of 4°C and also how strongly the inter cycle cooling time influences this result. Hence, treatments according to this protocol should be safe for clinical application.

Pointing out e.g. the Diode laser at 1.1W, the coronal temperature rise reaches 10.6°C for $t_c=1\text{s}$, whereas for $t_c=5\text{s}$ it just goes up to 2.4°C. Moreover, the difference between apical and coronal temperature decreases significantly for longer cooling times.

Summing up it can be stated that all results for $t_c=5\text{s}$ are acceptable and cause no damage to the implant as well as to the POM jaw bone model.

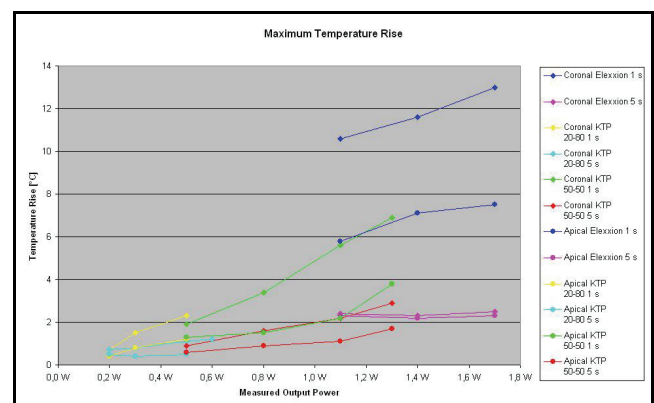


Fig. 4 Maximum temperature elevations after laser periimplantitis treatment in a POM jaw bone model recorded in the apical and coronal regions of the implant surface. An intermission cooling time of $t_c=5\text{s}$ is required to keep the temperature rise within the accepted clinical limits.

The reason for this effect is that, of course, for longer cooling times more energy can be transferred to the surrounding air by convection, but mostly conduction via the block where the implant is embedded in will take place. Hence, also the heat from the coronal region standing free in the air can be removed more effectively by transportation through the implant into the block. For shorter intermission (cooling) times this heat conduction can not be performed so well, thus the heat stays more concentrated in the coronal region, consequently yielding a higher temperature rise in this part. It is likely that similar results will be achieved for the implants being in situ in the real bone during periimplantitis treatment, as bone and POM have comparable heat conductivities.

Display output power [W]			Measured output power [W]		
Diode laser	KTP 20ms	KTP 50ms	Diode laser	KTP 20 ms	KTP 50 ms
1.05	1.0	1.0	1.15	0.23	0.53
1.5	1.5	1.5	1.43	0.33	0.8
1.95	2.0	2.0	1.7	0.46	1.1
	2.5	2.5		0.55	1.3

Table 2 Overview of the differences between displayed power settings and measured output powers. The Diode laser showed rather comparable values for both measured and displayed output, while the KTP device showed about 50% to 80% reduction from the display output power, according to its operation mode and non-consideration of the specialties of chopped mode operation.

4. Conclusion

Based on the procedure applied in this study to simulate the clinical decontamination with lasers, several aspects have to be taken into consideration like e.g. the intercycle cooling time. By using two different time durations as a break between each 5s of laser irradiation, variable values for the temperature rise were recorded resulting in the expectable finding that short intermission times for cooling yield higher temperatures rise.

During the measurements, also changes of the hand-guided movements and the position of the fiber during lasing caused variations in the resulting temperature. This indicates that it will be necessary to standardize the treatment procedure and that the operator very strictly obeys the suggested protocol.

The bactericidal effect of lasers in periimplantitis therapy has already been proved in many studies. As a result of this work, the Diode laser showed a higher temperature rise compared to the frequency-doubled Nd:YAG (KTP) laser at the implant surface at 1 and 5 s cooling cycle.

Hence, according to our results, the range of safe application is situated between 0.2-1.2W real output power for the frequency-doubled Nd:YAG (KTP) laser and below 1.0W for the diode laser, when combined with an intercycle intermission time of 5s. At output powers higher than 1.0W, both, coronal and apical region of the implant, showed a temperature rise exceeding the 4°C limit.

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References

1. A. Mombelli: *Periodontol* 2000, 28, (2002) 177–189
2. A. Mombelli, N.P. Lang: *Periodontol* 2000, 17, (1998) 63–76
3. A. Mombelli: “Proceedings of the 3rd European Workshop on Periodontology. Implant Dentistry” ed. by N.P. Lang, T. Karring, J. Lindhe (Quintessenz, Berlin, 1999) p.281–303
4. M. Baron, R. Haas, O. Dörtbudak, G. Watzek: *Int J Oral Maxillofac Implants*, 15 (2000) 533–544
5. B. Klinge, A. Gustafsson, T. Berglundh: *J Clin Periodontol*, 29, (2002) 213–225
6. S. Schou, T. Berglundh, N.P. Lang: *Int J Oral Maxillofac Implants*, 19, (2004) 140–149
7. Z. Schwartz, K. Kieswetter, D.D. Dean, B.D. Boyan: *J Periodontal Res*, 32, (1997) 166–171
8. J.A. Shibli, M.C. Martins, L.H. Theodoro, R.F.M. Lotufo, V.G. Garcia, E. Marcantonio Jr: *Journal of Oral Science*, 45(1), (2003) 17-23
9. M. Kreisler, H. Al Haj, H. Götz, H. Dushner, B. d’Hoedt: *Laser Surg. Med.*, 30, (2002) 233-239
10. A.R. Eriksson, T. Albrektsson, T.: *Journal of Prosthetic Dentistry*, 50, (1983) 101–107
11. G. Romanos: *Journal of the Canadian Dental Association*, 71(2), (2005) 122
12. H. Deppe, H.H. Horch, J. Henke, K. Donath: *Int J Oral Maxillofac Implants*, 16(5), (2001) 659–67.
13. C.M. Block, J.A. Mayo, G.H. Evans: *Int J Oral Maxillofac Implants*, 7(4), (1992) 441–9.
14. G.A. Catone: “Laser application in oral and maxillofacial surgery” ed. by G.A. Catone, C.C. Aling (Saunders, Philadelphia, 1997) p. 181–96.
15. D.W. Coffelt, C.M. Cobb, S. MacNeill, J.W. Rapley, W.J. Killoy: *J Clin Periodontology*, 24, (1997) 1–7.
16. P. Kassak, L. Šikurova, P. Kvasncka, M. Bryszewska: *Physiol. Res.* 55(2006): 189-194.

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