CORNEAL PHOTOABLATION BY MID-INFRARED LASER INSTRUMENTATION AND RESULTS

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ABSTRACT:

Infrared-laser radiation is capable for corneal as well as lens photoablation. The Raman-shifted Yag-Laser (wave length = 1.5 or 2.8 μm) offers the advantage of minimal thermal side effects and the possibility of fibre optics delivery. Histological evaluation of laser keratectomies demonstrates the thermal loaded zone to be as small as 3 μm surrounding the excision.

Laser keratectomy with the Raman-shifted Yag-Laser

The interaction of laser radiation with human tissue depends highly on the amount of its absorption. The absorption of infrared light in the cornea is predominantly caused by water which represents about 80% of the corneal tissue. In the case of high absorption, the penetration depth is low and the radiation energy is absorbed in a very thin layer of the tissue.

Picture 1 shows that absorption of water is highly dependent on the choice of wave length. The highest absorption in the whole area of infrared radiation takes place around the wave length of 3 μm whilst relative maxima are seen at 2 and 1.5 μm. For laser surgery also wave lengths beyond 3 μm could be used but there are no advantages compared with the CO_2-Laser with its well-known difficulties and problems.

There are various interaction mechanism of laser radiation with tissue. The best-known interaction is **photo-coagulation** which is widly used in ophthalmology and which is the basis for the retinal application of lasers. Photo-coagulation is performed by heating the tissue below its melting point.

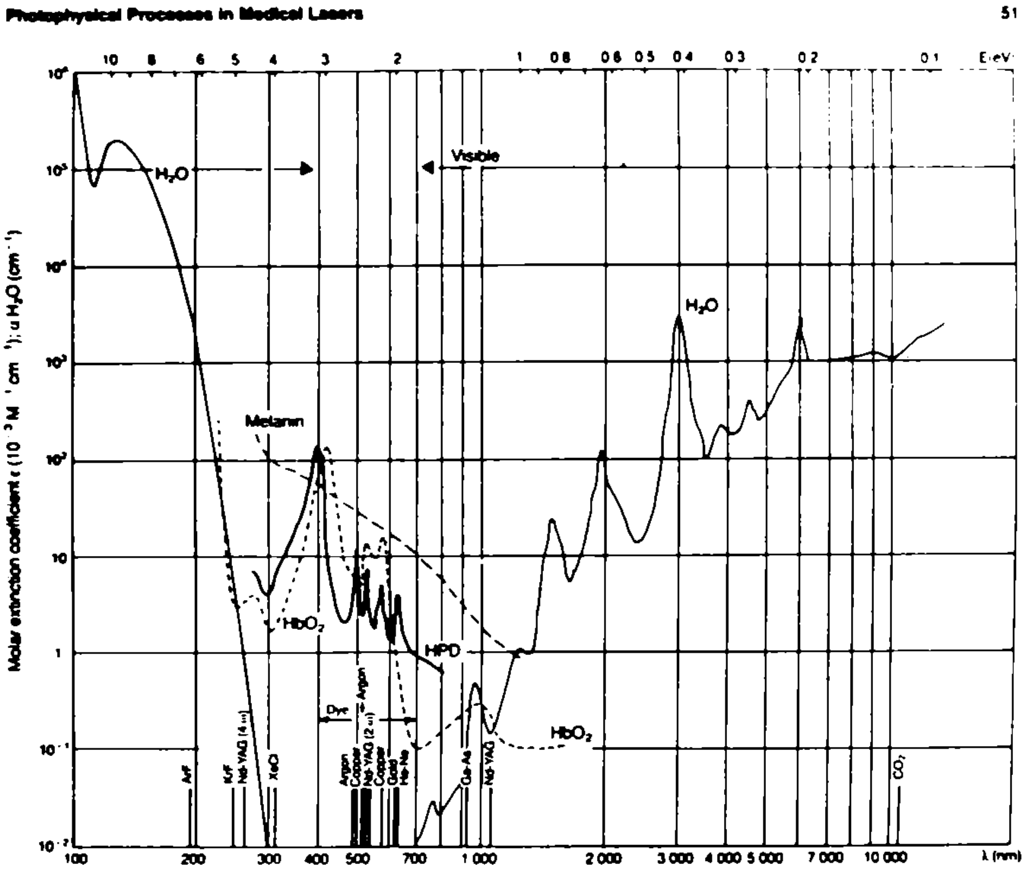
If the thermal interaction is increased i. e. by temperatures which are regularly higher than the melting point, we speak of **photo-vaporization.** Here, the tissue is not only coagulated but mostly vaporized.

In the case of **photo-ablation,** a non-linear optical effect occurs, with the result that the closed molecular compound of the tissue is fragmented into small molecular parts which are ejected as a gas. The process of **photo-ablation,** takes place in less than 10 ns, which means that laser radiation which interacts longer than 10 ns with the tissue does not contribute to **photo-ablation,** but only heats the tissue up and causes necrosis.

In the case of **photo-disruption,** however, the tissue is fragmented into atoms. Here, high electrical fields cause an optical breakdown and, as a consequence, more or less severe shockwaves are produced which in most cases cause considerable damage to the tissue.

If it is intended to cut tissue without or with a minimum of damage, the thermal stress on the tissue has to be minimized which means that photo-coagulation and photo-vaporization have to be avoided. On the other hand, destruction of the adjacent tissue by shockwaves must also be prevented which means that also photo-disruption has to be eliminated (See picture 1).

Therefore, when having in mind to use the laser as a light scalpel, only the interaction mechanism of photo-ablation is left. This can be done by the ultraviolet Excimer-Laser or by several infrared-lasers. The main advantage of the laser surgery compared with the dissection by knives is the non-contact application procedure. Knives always produce a mechanical pressure on the tissue which leads to uncontrolled depths of incision, to mechanical stress of the adjacent area or to pressure trauma. In the case of laser-excision, a high grade of accuracy can be achieved which makes this method superior to any kind of mechanical scalpel. With each laser pulse, only a small but well defined layer of tissue is ablated, so by applying a predefined quantitity of pulses the depth of the excision can precisely be predetermined. In the case of the 1.5 μm laser, for instance, we find depending from the pulse energy, ablation depths between 5



PICTURE 1

and 25 μ m per pulse, so that the depth of a laser keratectomy can theoretically be predetermined better than +/ —5%. Seiler and co-workers in Berlin have recently shown that clinical keratectomies with Excimer-Lasers can be performed with an accuracy of +/ —4.5%. This is by a factor of 2 to 3 less than the deviation from the precalculated value which can be achieved by using diamond knives.

In the mid-infrared wave length range, three types of lasers are actually available for medical application: the Erbium-Yag-Laser, the HF-Laser and the Raman-shifted Yag-Laser. (See picture 2).

The **Erbium-Yag-Laser** is a solid state laser which operates at 2.94 μ m, that means in the absorption-maximum of water and which from this point of view is perfectly suitable for surgery at the cornea. Its main disadvantage, however, is the fact that it produces only very long pulses in the range of about 200 μ s which cause considerable coagulation zones around the incision. A really technically perfect Q-switching of this laser is not yet available.

Mid-Infrared Lasers

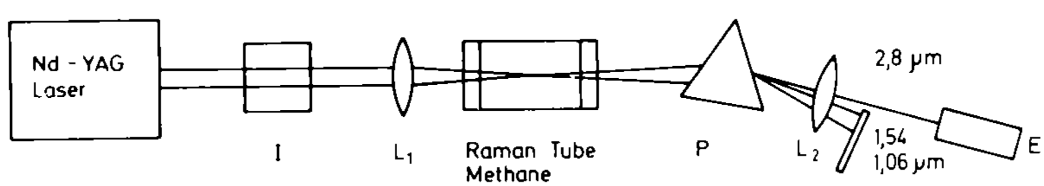
Er:YAG	3 µm	200 µsec	500 mJ
HF	2.7 - 2.9 µm	50 nsec	100 mJ
Raman-shifted	2.94 µm	5 nsec	10 mJ
YAG	1.48 µm	5 nsec	20 mJ

PICTURE 2

The **HF-Laser** is a chemical laser pumped by the chemical reaction of hydrogen and fluorine with considerably shorter pulses (around 50 ns) but with rather poor beam quality (high beam divergence and impossibility of focussing to a small spot). Here, we also find thermally affected or damaged zones of about 20 μ m around the incision. Furthermore, the HF-Laser operates with toxic gases and produces during its operation the highly aggressive silicic acid. These are problems which prevent this type of laser from a clinical application.

The **Raman-shifted** laser is basically an ordinary solid state laser, actually a Q-switched Nd: Yag-Laser as it is widely used in ophthalmic departments. The resulting radiation of 1064 nm and of 10 ns pulse duration is focused into a so called Raman-cell containing a high pressure gas or liquid, in our case methane. (See picture 3).

Stimulated Raman Emission



Spectral Distribution of Infrared Emission

I — Isolator (Faraday Rotator)

L_1 — Focusing Lens

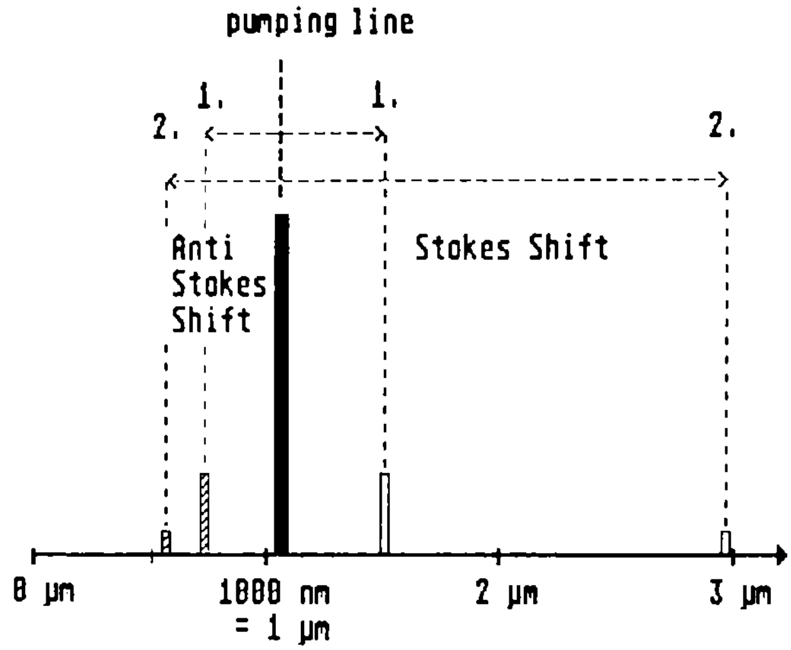
P — Quartz Prism

L_2 — CaF_2 Lens

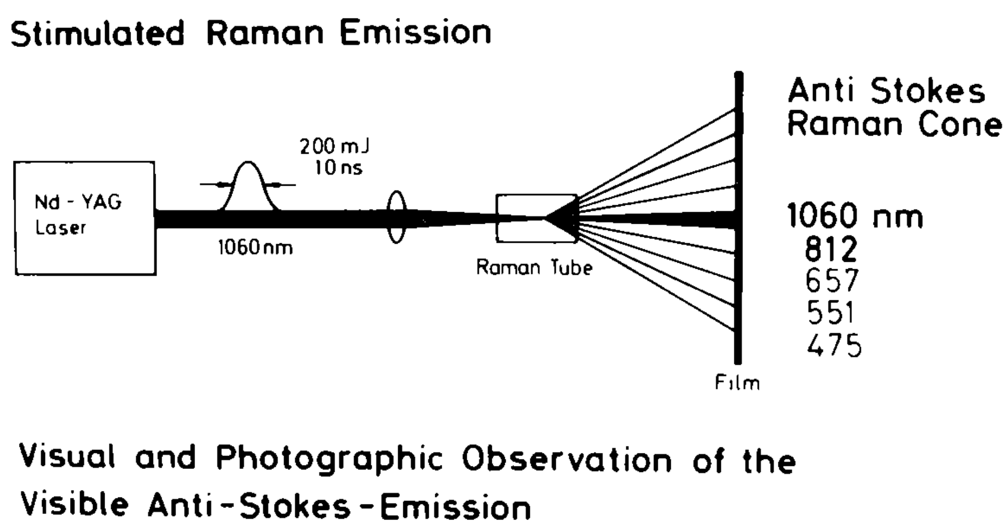
E — Energy - Detector

PICTURE 3

The wave length shift is provoked by the interaction of the laser light with the molecular vibrational states of the Raman medium. This is a non-linear optical effect. The shift to longer wave lengths than the original 1.06 μm Yag is called Stokes effect and produces radiation of 1.5 or 2.8 μm. Since a similar effect occurs in the opposite direction, namely into visible red and green emission, both the Stokes and the Anti-Stokes effects are performed simultaneously. (See pictures 4 and 5).

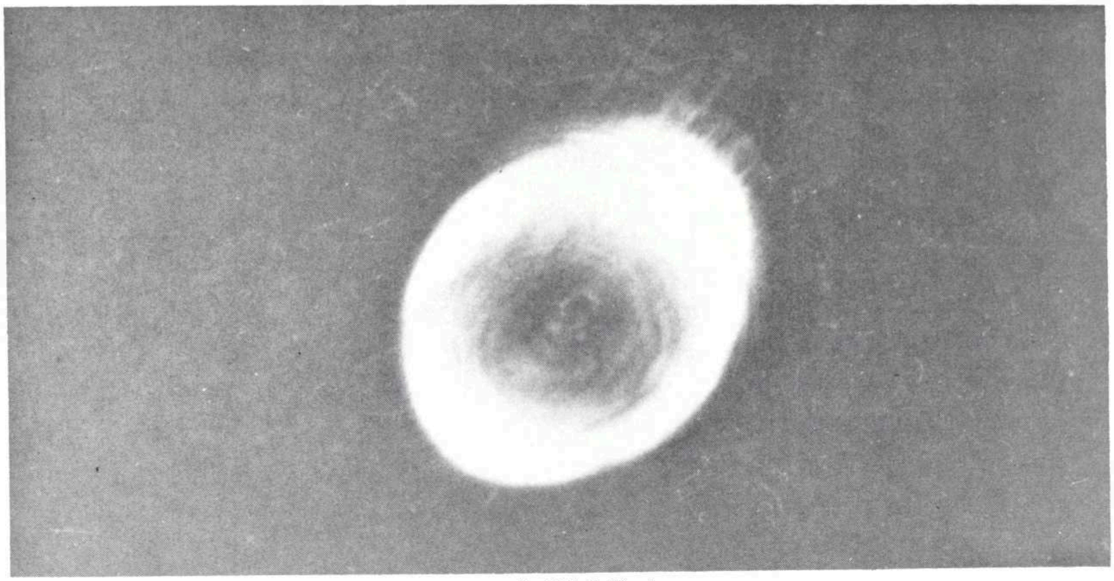


PICTURE 4



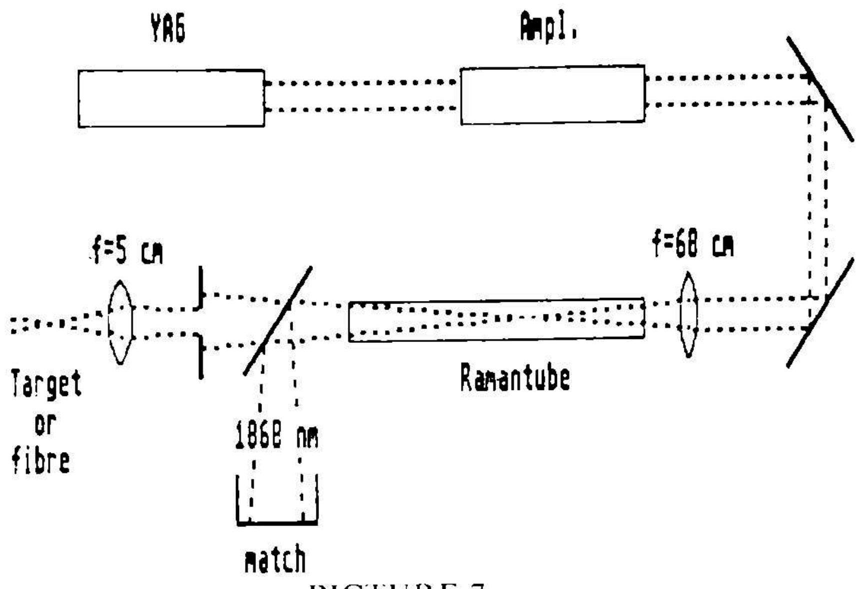
PICTURE 5

In picture 6, the visible portion of the Raman-shifted radiation is shown: the red and the green light which combine in the center to yellow. Also in the center of this spot, we find the non-visible infrared radiation which is used for laser keratectomies. Since a radiation of 1.5 μm wave length can easily be guided by

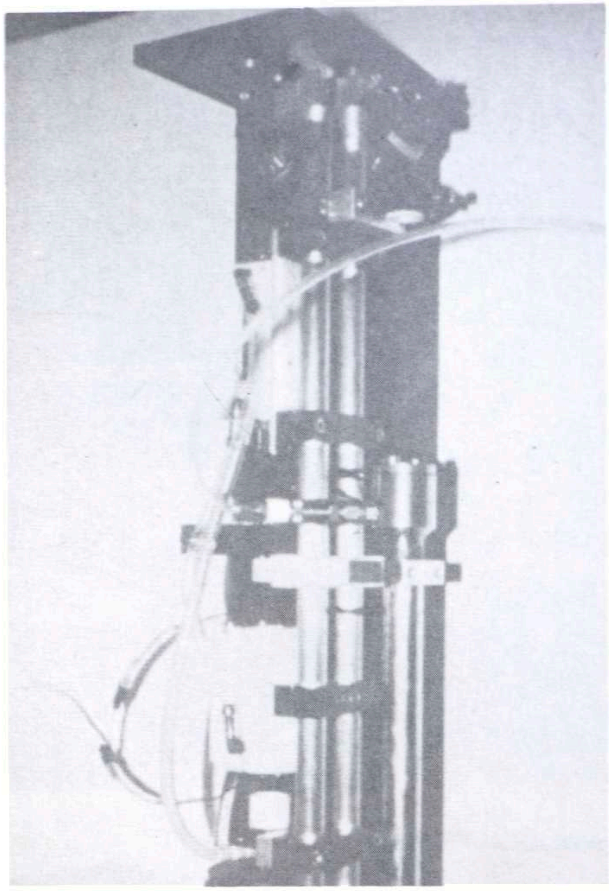


PICTURE 6

commercially available fibre-optic-cables, this choise of wave length also offers the advantage of an intra-ocular application. In pictures 7 and 8, a prototype of such a Raman-shifted Yag-Laser is presented. It produces a laser-spot of 300 μ diameter and allows photo-ablation at energies below 5 mJ corresponding to a fluence of less than 6 J/cm² With these parameters, each pulse ablates a tissue layer of 10 μm thickness.



PICTURE 7



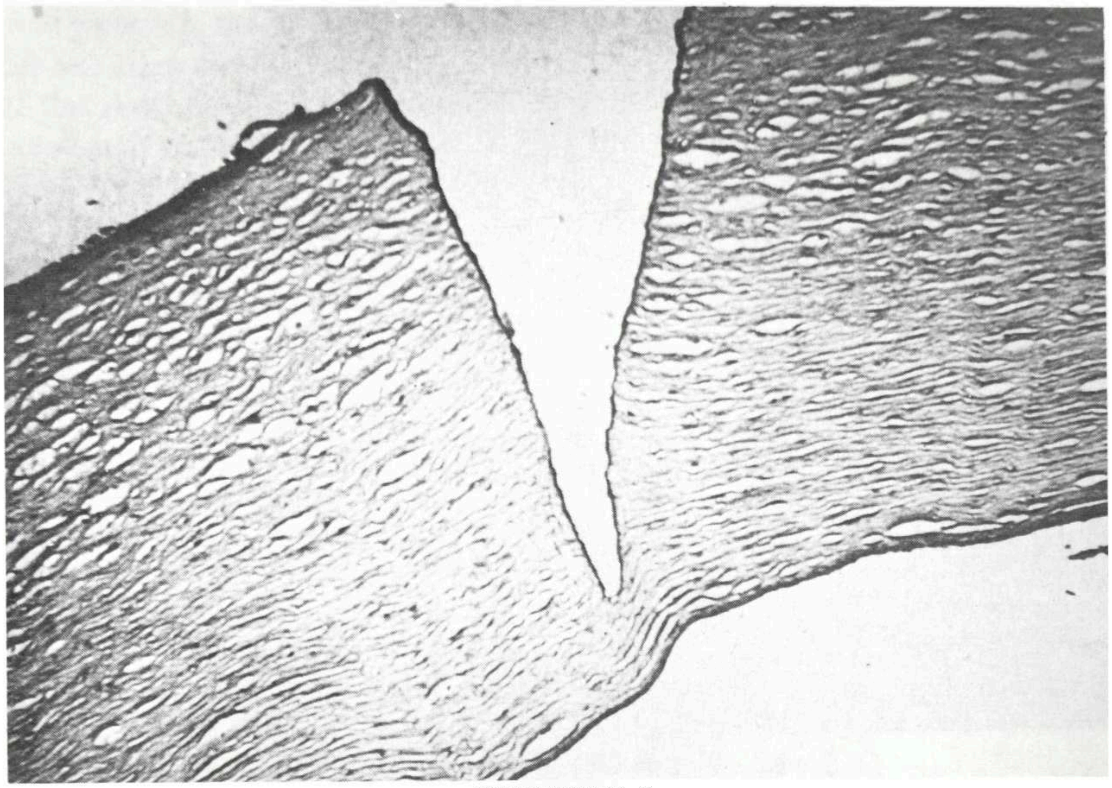
PICTURE 8

Such a photo-ablation is shown in pictures 9 and 10. In picture 9 ablation is made outside of the focus-plane, which means that the energy density is slightly below the ablation treshhold. The result is a strong thermal heating of the tissue and the effects are very similar to those of CO_2-Lasers. In picture 10, ablation is made in the focal plane. We achieve a fine keratectomy with clear boundaries, and the thermally affected area is smaller than 3.5 μm, a result which inspite of our optimism was really surprising.

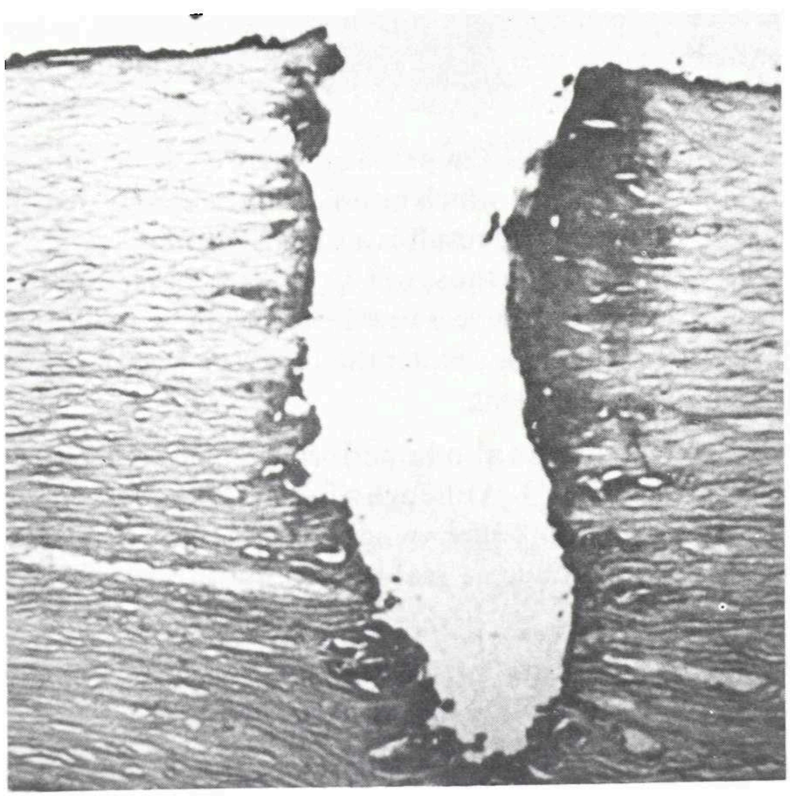
Photo-ablation in the lens can also be performed by using a fibre optic cable. The result is shown in picture 11. Although a relatively large thermally damaged area around the photo-ablation crater in the hard lens core can be seen, there was obviously no problem to produce a real photo-emulsification of the lens.

In these days, at the University Eye Clinic of West-Berlin several studies are made comparing the healing procedures of laser keratectomies with infrared-lasers and Excimer-Lasers in rabbit corneas.

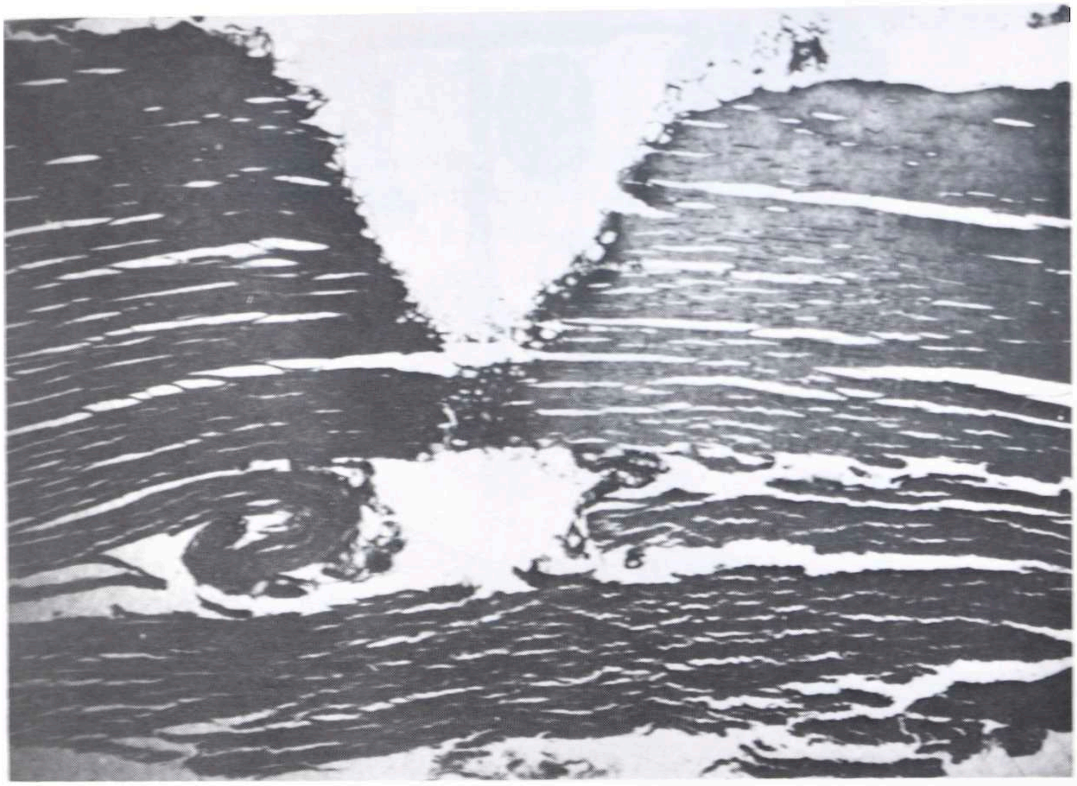
Besides these clinical investigations, technical experiments and studies are made in the laboratories of the Rodenstock Company in Munich and of the



PICTURE 9



PICTURE 10



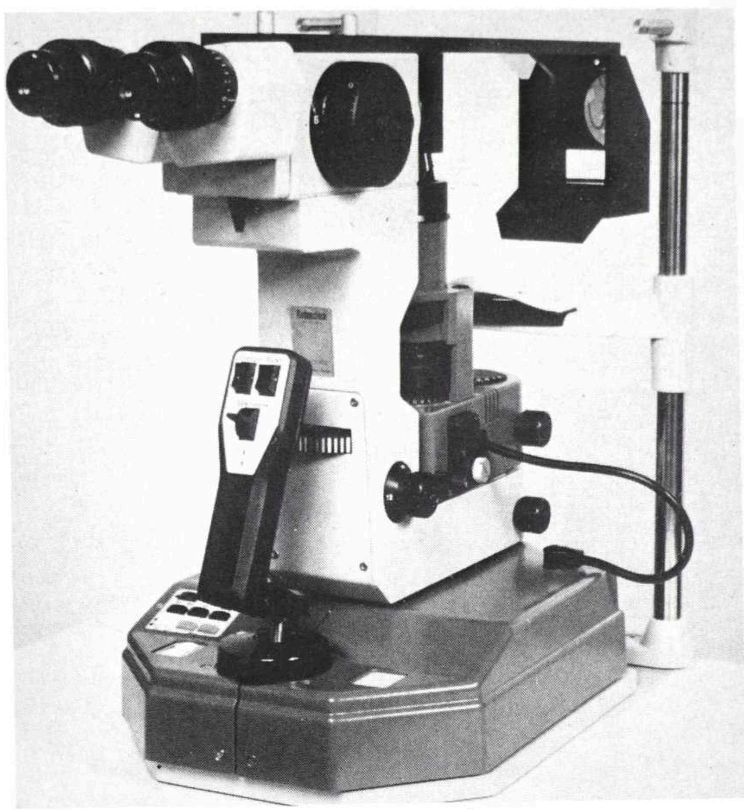
PICTURE 11

Technical University of Berlin with the objective to optimize the Raman-shifted lasersystem. These efforts concentrate on two major fields which are:

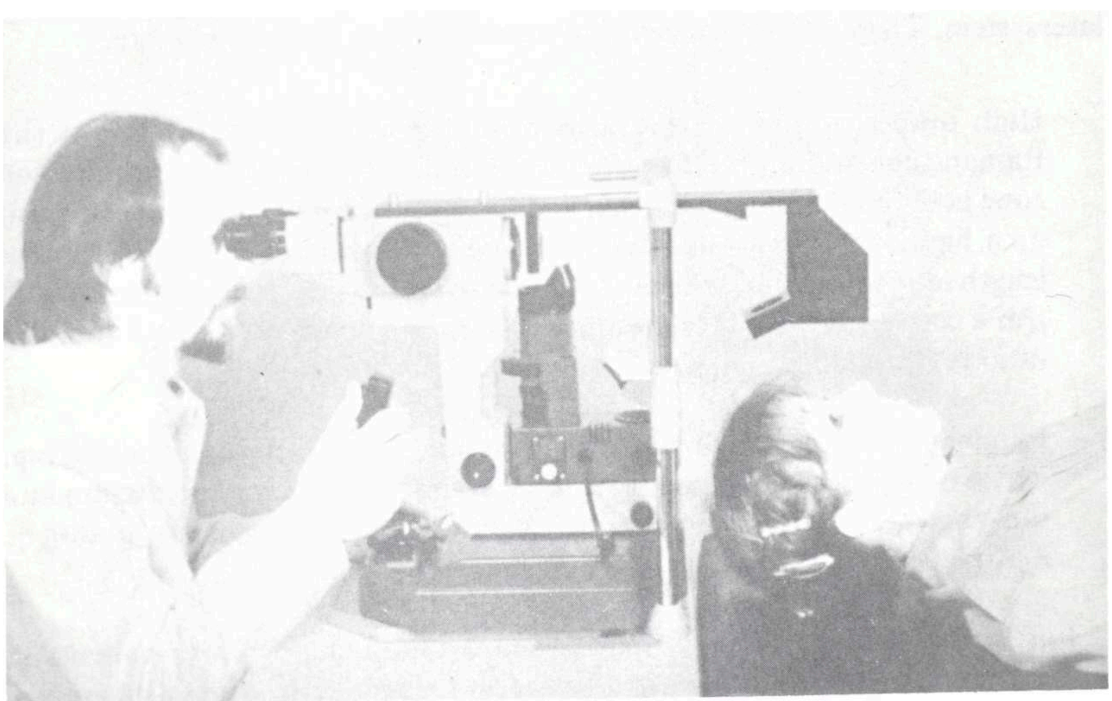
- High output/imput energy ratio which depends on the length of the Raman-tube and on the pressure of the Raman-medium (long interaction zone between pumping radiation and Raman-medium) on the one side and on a high conversion rate which is roughly a square function of the pulse length (due to non-linear conversion) on the other side. It seems, that for 2.8 μm a conversion rate of more than 10% can be achieved. For 1.5 μm, this ratio is considerably higher.

  Elimination of unwanted side effects, as for instance Brillouin scattering, which causes highly damaging back reflection of the Raman-emission into the Yag-resonator and which can be prevented by a Faraday rotator or $\lambda/4$-plates.

Furthermore, we have developed a delivery system for lying patients, a prototype of which is shown in pictures 12 and 13. This device consists mainly of a conventional laser-slit-lamp, with its optical components specially coated for



PICTURE 12



PICTURE 13

infrared light, and a deflection device transmitting slit-illumination and laser-radiation via prisms to the eye.

The infrared beam can be coupled to this slit-lamp by a fibreoptic cable. In case of 1.5 μm wave-length, an ordinary quartz fibre can be used which there has an absortion minimum. At 2.8 μm wave-length, however, a fibre-cable consisting of Zirkonium fluoride (ZrF_4) has to be used which is rather awkward to manipulate because it is very brittle and furthermore it is hygroscopic with the result that the endings are immediatly contaminated.

As already mentioned, the fibre can also be used separately from the slip-lamp as a laser scalpel. We guess, that this application is limited to quartz-fibres.

We also try to develop a deflecting system which guides the laser-spot electronically controlled over the cornea in order to produce incisions or ablations of any predetermined size and configuration. We have successfully done a first step in this direction by an electronically actuated wobbling device for our regular Argon-Laser-Systems. This device is under clinical evaluation right now, but there is still a long way to go until an infrared laser-system can be fully electronically and safely controlled.

CONCLUSIONS:

As a conclusion, we can state that the Raman-shifted Yag-Laser represents an infrared laser system which is not only suitable for any kind of corneal surgery and phaco-photo-ablation but also delivers precise and predictable clinical results with a minimum of tissue damage. The system can be further optimized with regard to energy, pulse length and even wave length while its optical beam quality is already perfect compared with all other mid-infrared lasers. The quality of the photo-ablative incisions is comparable to those of Excimer lasers but the well-known disadvantages, of the Excimer laser such as toxic gases, rigid optical elements and the non-availability of fibre-optic cables are fully excluded.