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**A SCANNING ELECTRON MICROSCOPIC
STUDY OF LASER-ETCHING OF HUMAN
DENTAL ENAMEL**

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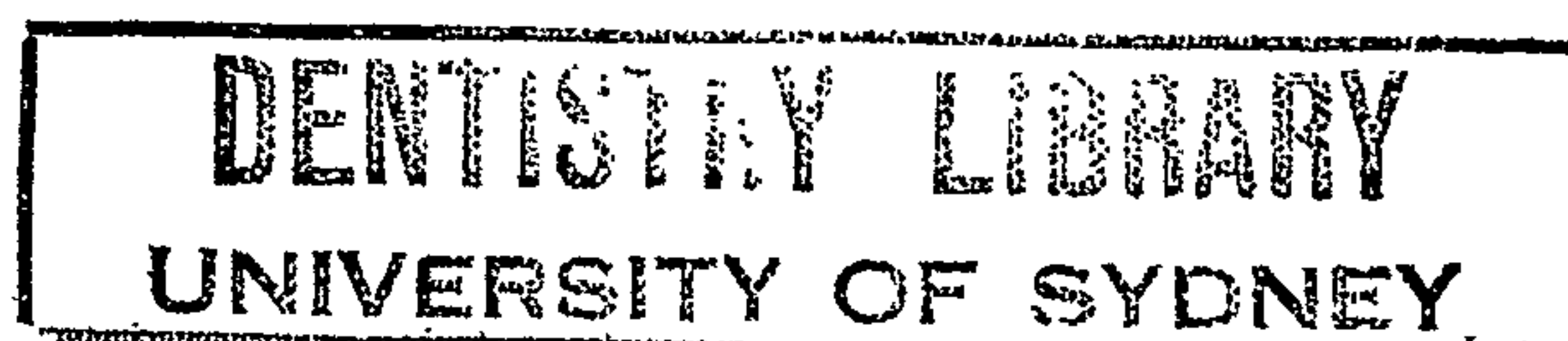
**A thesis submitted in partial fulfilment
of the requirements for the degree of**

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ABSTRACT

This experiment used scanning electron microscopy to investigate the similarities and differences in the morphological changes to enamel created by the two etching techniques of phosphoric acid and Nd:YAG laser. Also, the effect of varying the laser parameters of energy density, pulse rate and initiator use was studied to determine the ideal setting for etching.

Four different changes to the enamel were observed - cracking, flaking, pitting and cratering. The last two appeared to be dependent on the energy density used and provided the only means of increasing the surface area sufficiently to enhance bonding via mechanical interlocking. Both were directly related to impact disks which were visible clinically. These areas corresponded to absorption of the laser beam by the initiator which transferred the heat to the enamel surface. Without the initiator there were no changes observed. Increasing the pulse rate did not affect the changes but did increase the proportion of enamel covered by the impact sites. Cracking was present in all samples except one, and resulted in flaking in more susceptible areas.

The only instance in which the lased surface resembled that etched by acid was the roughened enamel exposed after surface enamel flaking. This may have led to confusion in some previous studies that described the surface changes created by the two techniques as being alike. These results are very similar to those found using either CO₂ or Nd:YAG lasers. This would imply that the associated changes are not characteristic of a particular wavelength as is seen with soft tissue. Rather, they seem

to be only dependent on the energy density imparted to the surface of the tooth.

The manufacturers recommendation of an energy density of 62.2 J/cm^2 is insufficient to provide enough cratering for significant mechanical interlocking for bonding. An energy density between 124.3 J/cm^2 and 159.2 J/cm^2 gives the best energy to cratering ratio. Overall, the Nd:YAG laser could be used as an alternative to acid for etching of enamel prior to orthodontic bracket placement as it does provide a means of increasing mechanical retention. However, a number of concerns still need to be addressed with regard to its long term effects, particularly at debonding. At this stage it does not provide a viable alternative to acid-etching as used in orthodontics.

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Finally, thank you to my wife, Kipling and my family for taking the journey with me on the long road to completing this thesis.

DEDICATION

This work is dedicated to my uncle, Dr Geoffrey Scarlett,
who told me which fork to take in the path of my career.

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CHAPTER 1

FUNDAMENTALS OF LASERS

1.1 Laser Physics

The essential physics necessary for the development of the laser was formulated in 1917 by Albert Einstein in his treatise "The Quantum Theory of Radiation". In this he proposed the concept of *stimulated emission*.

According to the quantum theory an atom can be "excited" to a higher energy state by the absorption of thermal, electrical or optical energy. This energy may only be added in well defined amounts, in other words, it is quantised. Excited states are, however, unstable and the atom will revert back to the ground state by emitting the excess energy in the form of a photon of electromagnetic energy. This is called *spontaneous emission*. The time spent in an excited state is variable but is typically a few thousandths of a second (Taylor *et al*, 1965). What Einstein theorised was, that an atom, which is already in an excited state, may be stimulated to emit a photon in response to an incident photon corresponding exactly to the wavelength of the absorption energy. This is called *stimulated emission* (Figure 1-1).

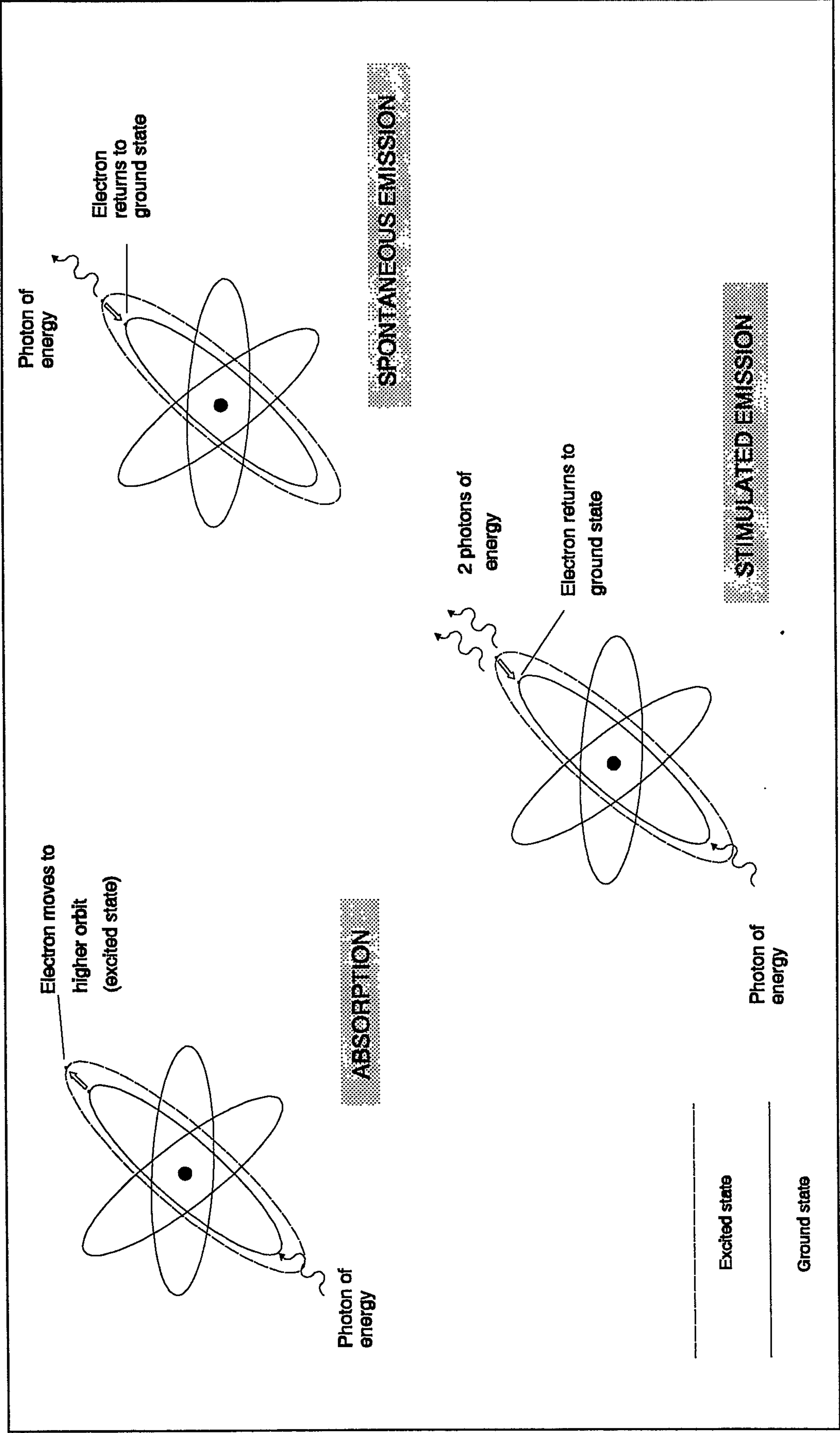


Figure 1-1 Energy diagram illustrating the process of absorption, spontaneous emission and stimulated emission. From Apfelberg (1987).

When energy is supplied to a collection of atoms in certain materials a population inversion may occur. That is, there are more atoms in the excited state than the ground state rather than the other way around. Due to the instantaneous nature of the photon emission a population inversion is necessary to make enough excited atoms available (Welch *et al*, 1990). Thus if one atom undergoes spontaneous emission producing a photon, that photon can cause the stimulated emission of another photon giving two photons. These two photons can generate the emission of two more photons giving four photons and so on. One photon can therefore stimulate the emission of many others with the same energy and wavelength and travelling in the same direction (Robertson and Clark, 1988).

This is now Light Amplification by the Stimulated Emission of Radiation or **LASER**.

1.2 Basic Components

The basic requirements to produce a laser beam ensure that stimulated emission is much more probable than spontaneous emission (Jeffrey *et al*, 1990a) (Figure 1-2):

1. Active medium

The selected material must have an appropriate energy state which will emit a photon of suitable energy and be sufficiently long lived for a population inversion to occur. There are currently many materials capable of undergoing

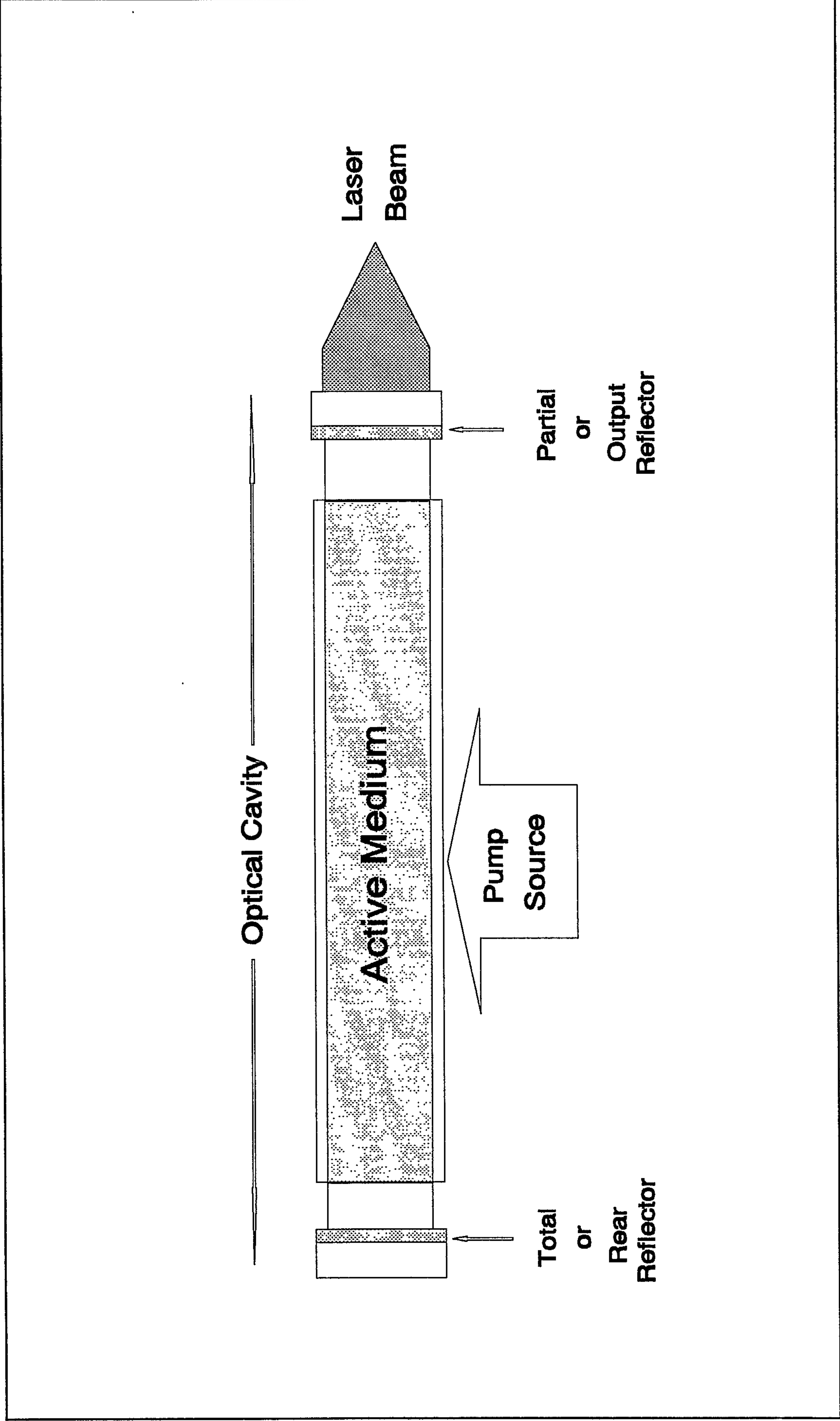


Figure 1-2 Basic components of the laser. From Fuller (1987).

stimulated emission. Typically, they are divided into 3 main classes based on the phase of the laser medium; gas, liquid or solid. These classes are further subdivided according to the nature of the laser gain medium (Robertson and Clark, 1988) (Table 1-1). The materials used mostly in medical applications, either clinical or research, are carbon dioxide (CO₂), argon (Ar), neodymium doped yttrium-aluminium-garnet (Nd:YAG) and helium neon (HeNe).

2. Excitation mechanism

In order for amplification to proceed the majority of atoms must remain in an excited energy state rather than in a lower energy state (population inversion). This is achieved by a device that directs energy into the active medium. This stimulation of the electrons in the medium to raise them to an excited state is called *pumping*. If the medium is a gas the input energy may be an electrical discharge (Robertson and Clark, 1988). Liquids and solids generally use a flash lamp or another laser. This may be a continuous arc light if continuous wave (CW) output is desired or an extremely brilliant flash lamp for pulsed lasers (Johnson, 1990).

3. Optical feedback system

An optical cavity or resonator encloses the active medium in a long thin tube with parallel mirrors at either end. The rear mirror is totally reflective for 100 per cent of the impingement energy. The front of the mirror reflects only a portion of the light therefore transmitting some fraction of the total energy. Stimulated emission would not continue long if the photons were allowed to

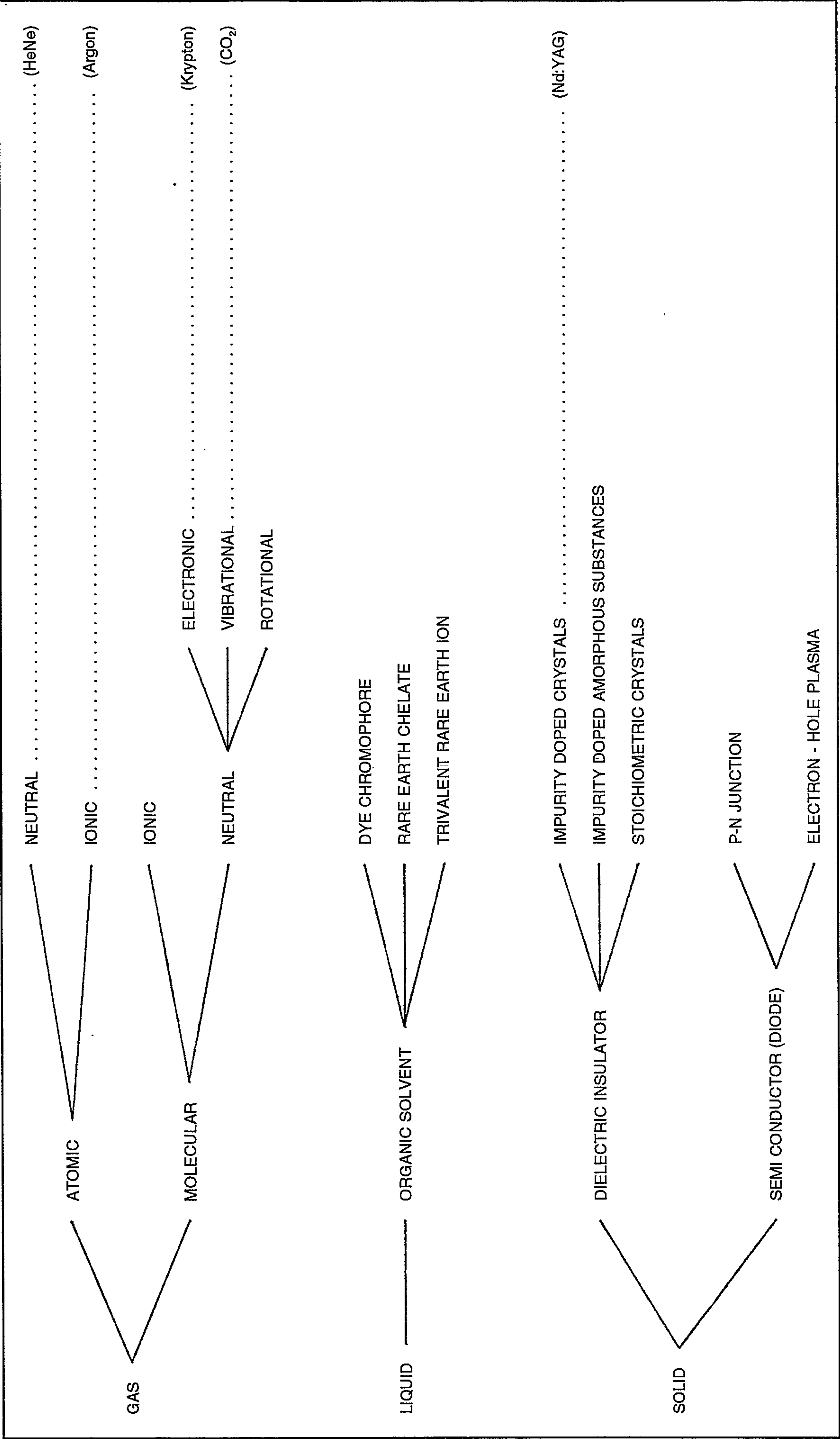


Table 1-1 Laser classes according to the active medium

completely exit the active medium. Therefore, the beam of light oscillates between the mirrors which increases the activity and intensity exponentially (Myers and Myers, 1989; Robertson and Clark, 1988).

As the excited electron steps down to a lower, more stable level, the energy absorbed earlier is released. The step taken is characteristic of that particular type of atom. The size of the energy step is directly related to the wavelength of the emitted photon (Welch *et al*, 1990).

$$\text{Wavelength} = \frac{hc}{(E_E - E_R)}$$

h = Planck's constant (6.3×10^{-34} joule seconds)

c = velocity of light in a vacuum (3×10^8 m/sec)

E_E = energy at excited state

E_R = energy at resting state

Thus, the larger the step, the shorter the wavelength.

This process of energy absorption and release continues as long as the pumping energy is applied to the active medium. Photons of light created from the spontaneous emission will radiate in all directions but they will not come across excited nuclei towards the side of the tube and will eventually be dissipated as heat (Fuller, 1987). A small percentage will travel along the axis of the laser cavity. These photons will

collide with atoms in an excited state, thereby stimulating the emission of radiation. At the partial reflector, a portion of the energy is permitted to escape as an intense, parallel, coherent beam of light (Figure 1-3).

1.3 Properties Of The Beam

After emerging, the photon beam has four major properties:

1. High intensity

Due to amplification the beam of light is far more intense than any other source. For example, a laser operating at 80 watts with a 5 mm diameter beam has an intensity of approximately 4 million watts / square metre compared to 6 watts / square metre for an 80 watt light globe.

2. Coherent

All the photons are "in-step" or "in-phase" both spatially and temporally. That is, the electric field vibrations of the emitted radiation and the incident radiation are perfectly synchronised (Jeffrey *et al*, 1990a).

3. Monochromatic

There is only an extremely small range of energies in a laser beam which present as a very narrow spectral colour or wavelength. A laser can thus be

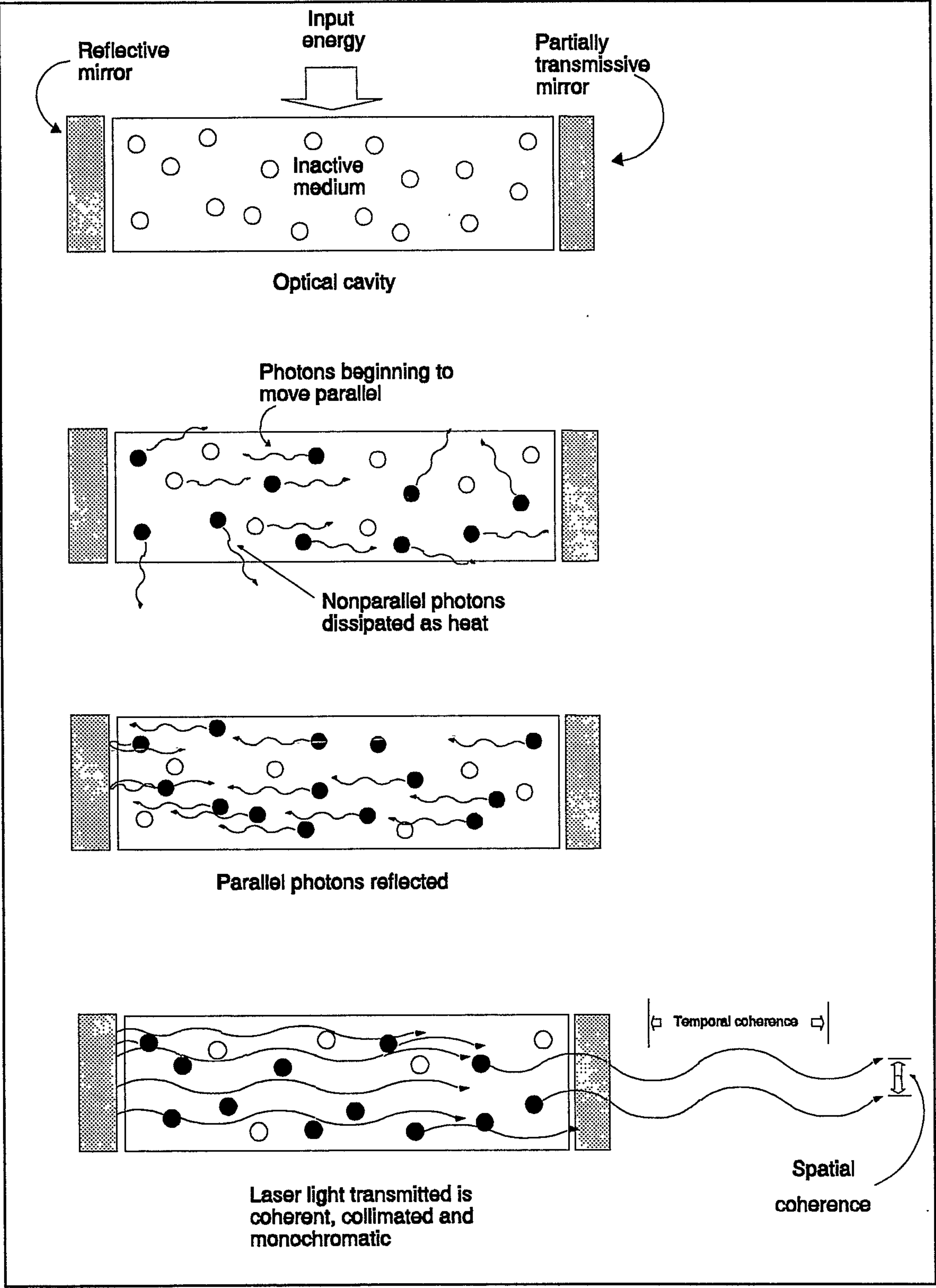


Figure 1-3 Generation of a laser beam. From Robertson and Clark (1988).

selected for its absorption characteristics which permits very specific targeting of laser light to tissue. The whole of the beam can be absorbed rather than some reflected or transmitted.

4. Collimated

The photon beam has minimal divergence so that an increase in distance away from the source means virtually no increase in the target spot size. Medically, the major advantage of this is the high degree of accuracy in focusing and cutting.

1.3.1 Beam output

The laser beam may be generated for different periods of time. There are 5 different modes available (Johnson, 1990):

1. Continuous wave (CW)

This denotes continuous excitation of the laser medium and subsequent continuous output of the beam. A typical output graph is shown in Figure 1-4. Most continuous wave devices for medical applications provide a power range up to 100 watts. However, in many applications it is desirable to pulse the output energy.

2. Gated

This permits the relatively low power CW energy to be delivered for a short duration by opening and closing the shutter of the CW laser. This gives a burst of laser energy that lasts approximately 0.1 to 0.9 seconds with an intensity less than that of a true pulsed laser. This results in a smaller "heat affected zone" along the edge of the impact site (Johnson, 1990). An output graph is shown in Figure 1-5.

3. Pulsed

Energy is stored in a capacitor within the electrical power supply and discharged in very high power, short duration pulses to give intermittent excitation. This results in instantaneous peak powers of several hundred watts for a few thousandths of a second. Pulse widths generally vary from 10^{-8} to 10^{-6} seconds (Weber, 1982). An output graph of a normal pulsed laser is shown in Figure 1-6.

4. Q switched

Both CW and pulsed lasers can be forced into a mode of operation known as Q-switched. Usually, in pulsed and CW systems, the cavity reflectivity of "Q" is held at a constant moderate value. By holding the reflectivity at a very low level there is a build up of particles at higher energy levels. The reflectivity is then rapidly raised to a moderate level to give a pulse of energy of very short duration (10^{-8} - 10^{-6} seconds) with an intensity several magnitudes higher than without Q-switching (Robertson and Clark, 1988) (Figure 1-7).

Figure 1-4 Graph of the power output of a continuous-wave laser.

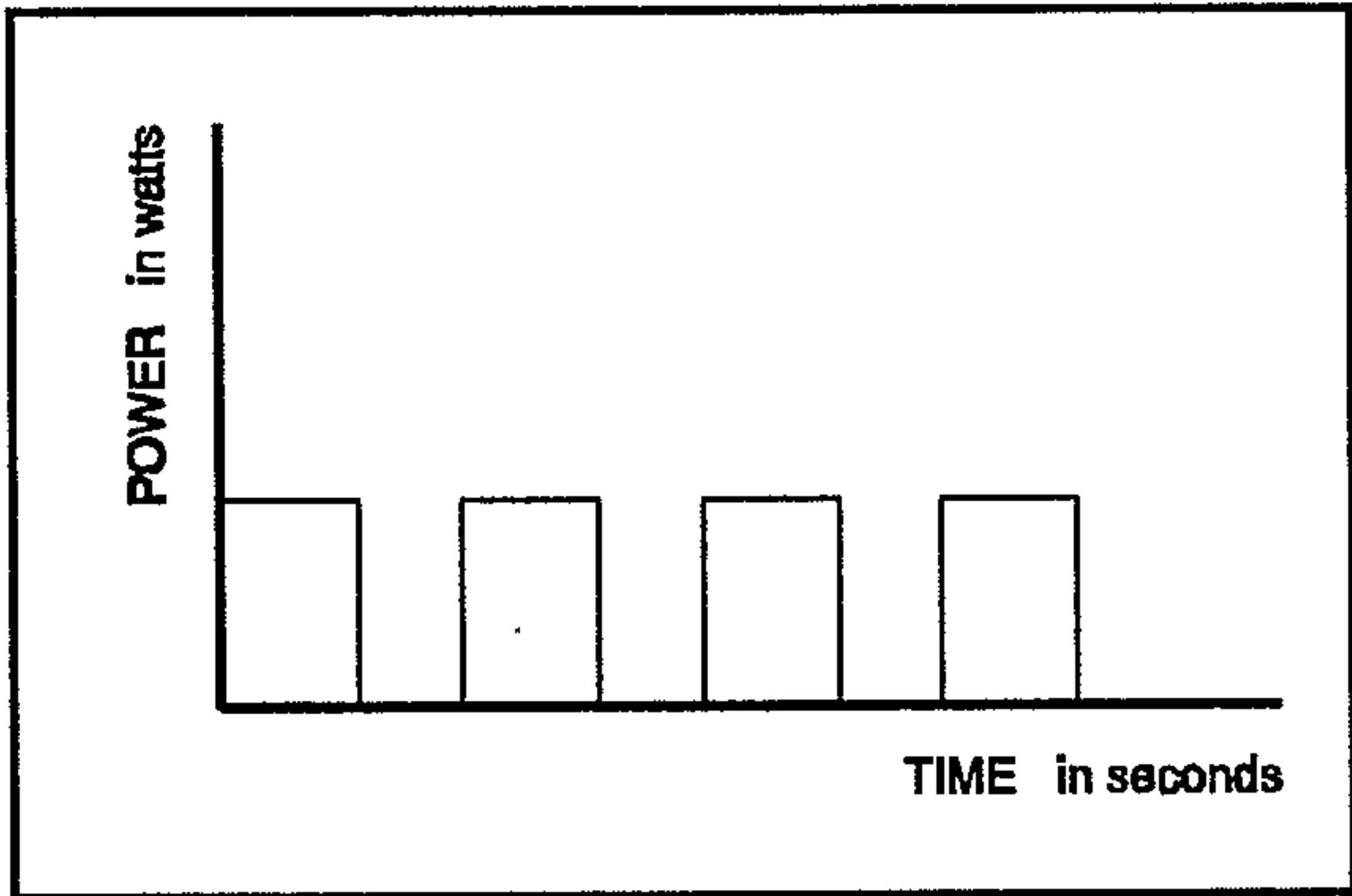
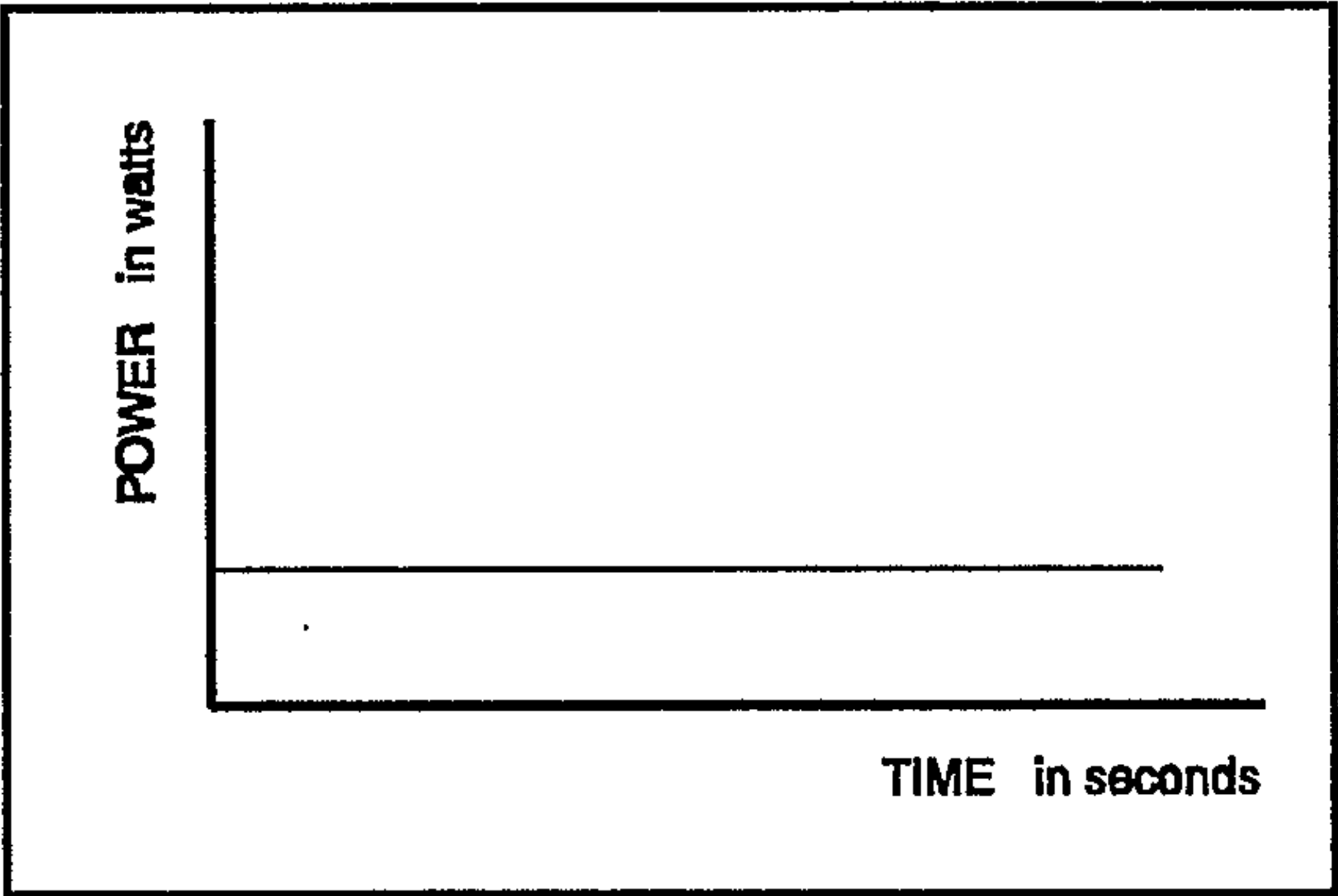


Figure 1-5 Graph of the power output of a gated laser.

Figure 1-6 Graph of the power output of a normal pulsed laser.

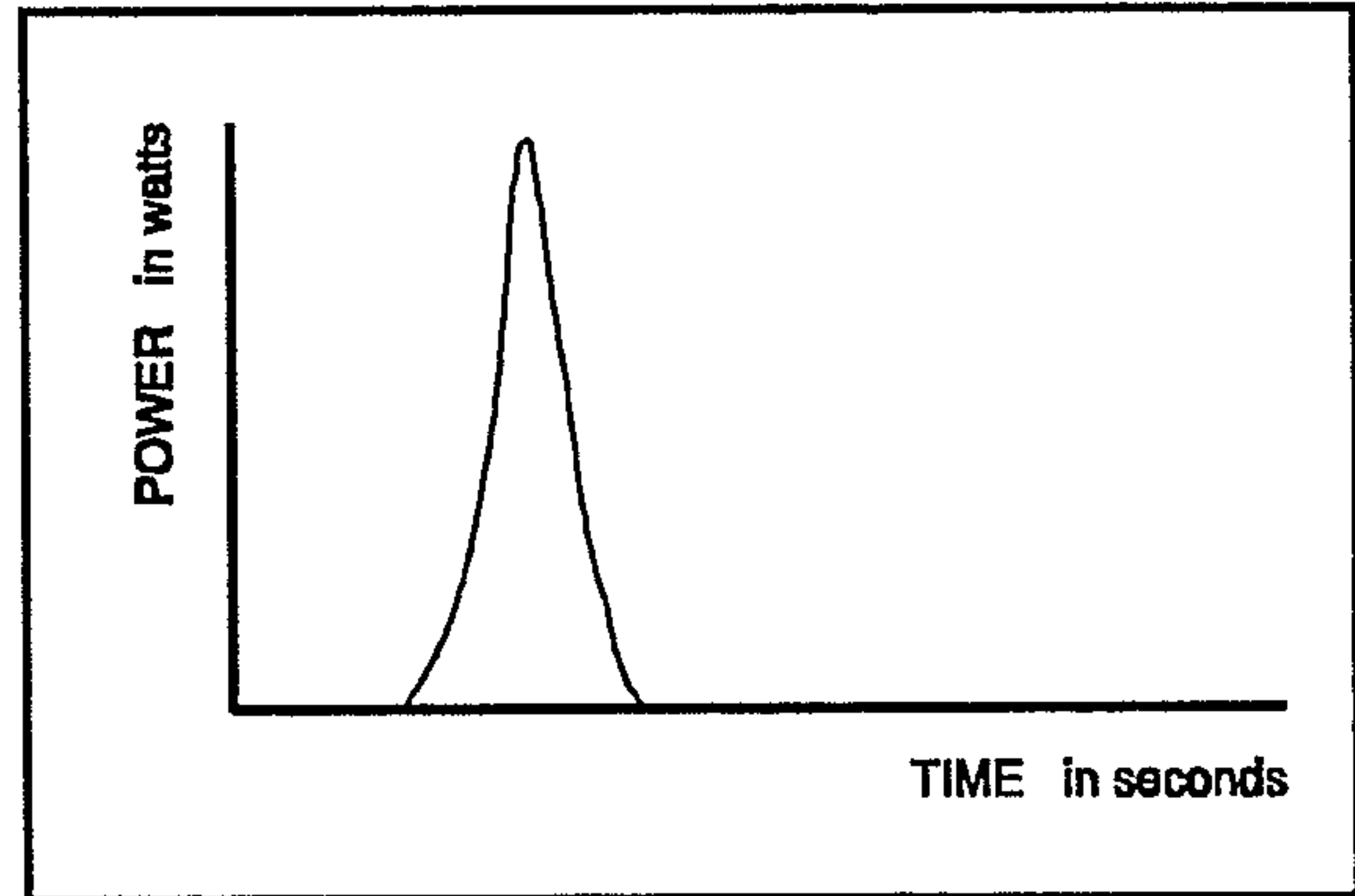
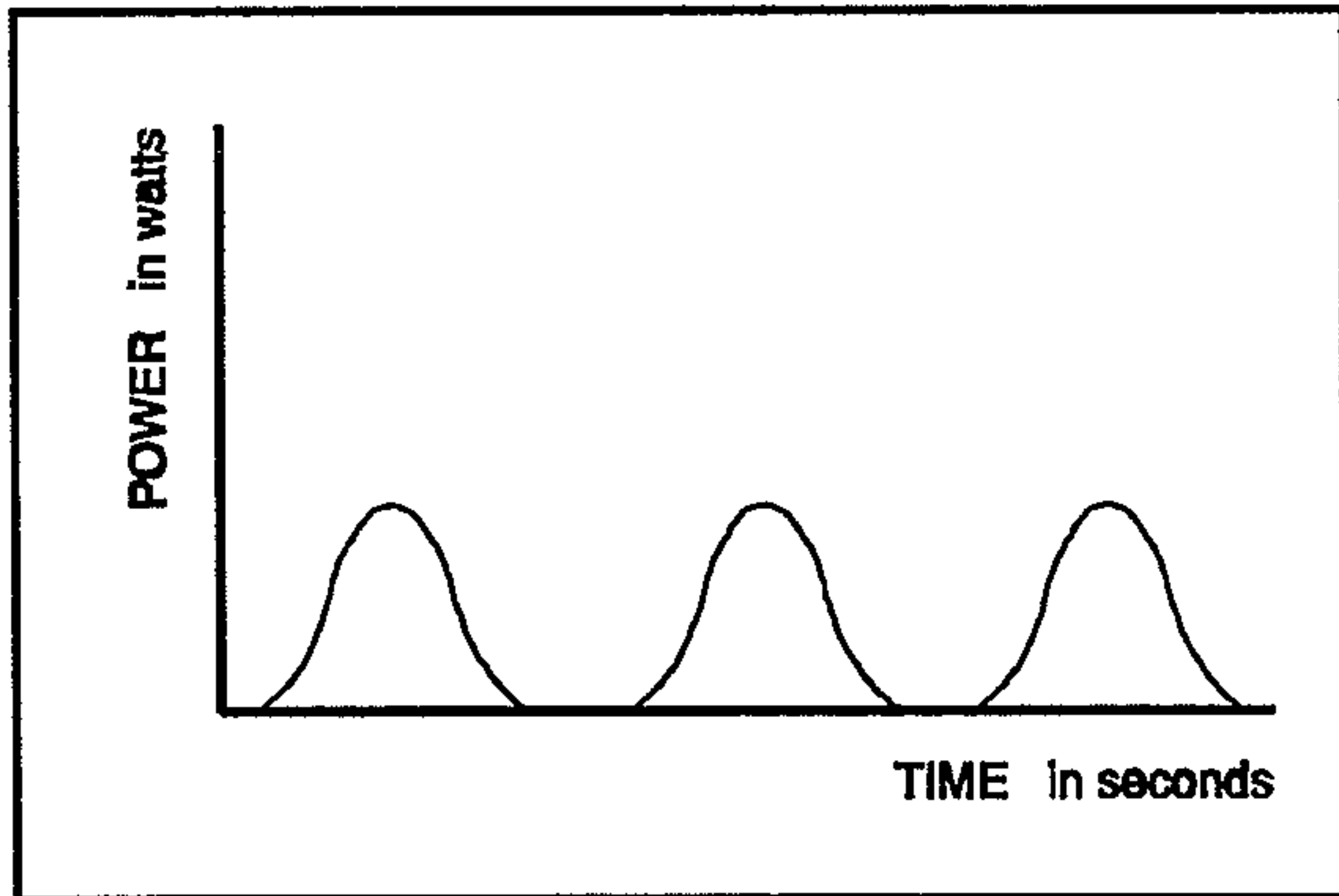


Figure 1-7 Comparative graph of the power output of a Q-switched laser.

5. Mode Locked

This is a technique using pulsed or continuous pumps to create pulses lasting only a few millionths of a second.

1.4 The Electromagnetic Spectrum

As mentioned previously the wavelength of the emitted light depends directly upon the magnitude of the energy transition. Several wavelengths have been found to be of medical use due to their biologically interactive properties. Laser beams can be separated into categories depending on their position on the continuum of electromagnetic radiation (Figure 1-8). The argon laser, which lies within the visible light range, does not require a separate aiming beam. In contrast Nd:YAG and CO₂ are both within the infrared region and are therefore invisible (Apfelberg, 1987).

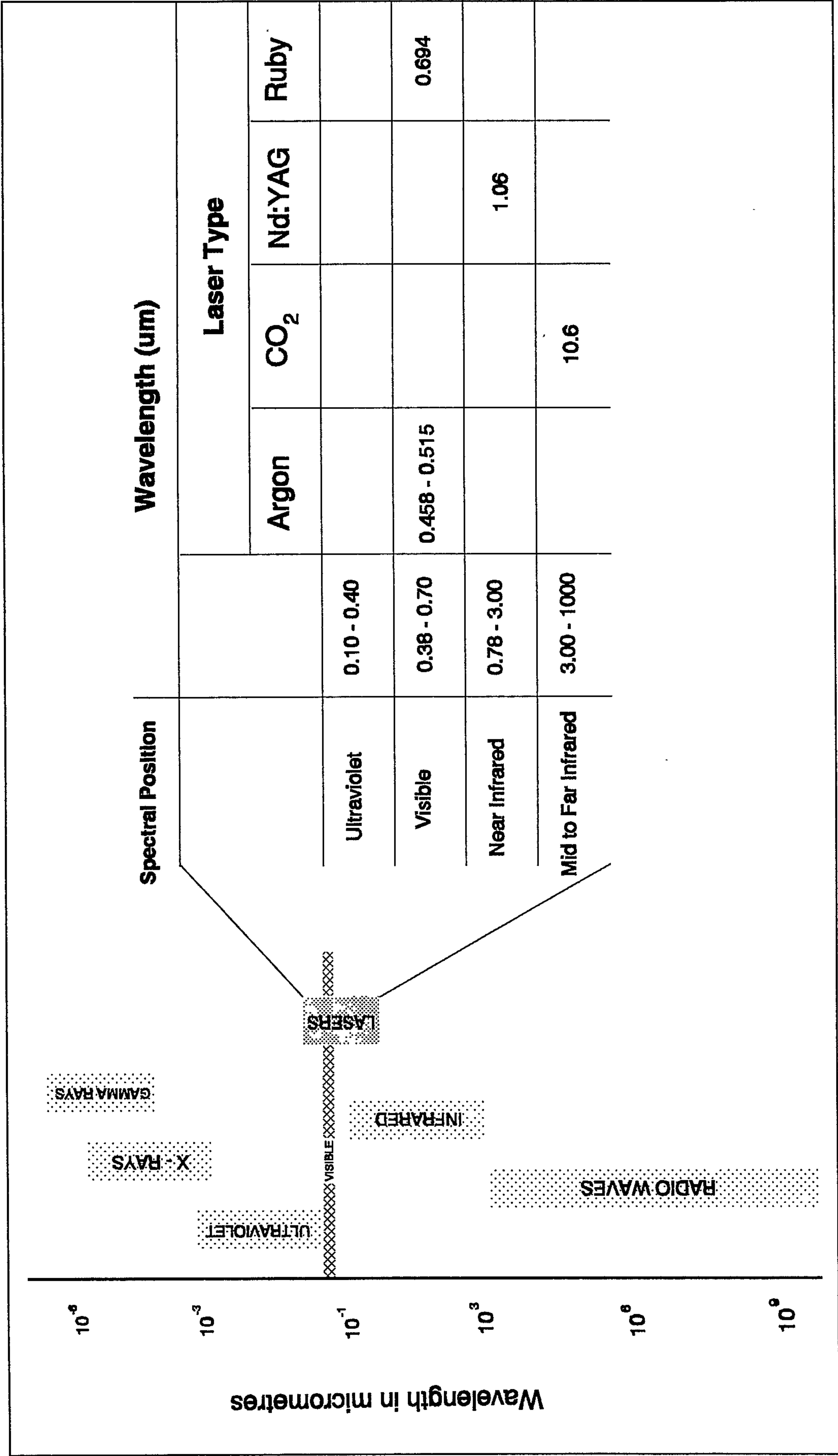


Figure 1-8 Laser type according to wavelength. Adapted from Apfelberg (1987).

CHAPTER 2

DEVELOPMENT OF THE LASER

2.1 History

In 1955, Gordon outlined the technology for a device called a *maser* which was the microwave forerunner of the laser. Schawlow and Townes published a paper in 1958 extending the maser principle to the infrared and visible regions of the spectrum. The theory was applied by Maiman in 1960 to produce stimulated emission at a visible wavelength - deep red 0.694 μm . This first laser used an artificially created clear sapphire doped with chromium (which gives it the red colour) that was excited with a pulsed flash lamp (Taylor *et al*, 1965).

Other materials were quickly tried. Neodymium ions were successfully used in a number of different host crystals from 1961 with the best performance given by yttrium-aluminium-garnet (Nd:YAG) developed by Johnson. The first report of laser oscillation in Nd:YAG was by Geusic *et al* in 1964. However, it was not applied medically until a continuous wave version was developed in 1975 (Enderby, 1987). The first gas laser was also developed in 1961 by Javan *et al* using a mixture of

helium and neon. It did not produce a beam powerful enough for thermal reactions but did provide an excellent method of aiming non visible lasers.

The argon laser, emitting energy in the blue-green portion of the spectrum, was developed by Bennett and colleagues in 1962. In 1964 Patel, at Bell laboratories, developed the CO₂ laser. The first commercially available CO₂ medical laser was introduced by American Optical Co. in 1967. It operated continuously rather than in the pulsed mode with an output beam in the far infrared. This became one of the most widely used medical lasers over the next 20 years due to two great attributes. Firstly, it is an inherently powerful device, delivering tens of watts rather than the thousandths of a watt which typify other gas lasers. Secondly, its energy is highly absorbed by water, which means that the depth of the incision in tissue can be precisely controlled (Carruth and M^cKenzie, 1986).

Fibreoptics were investigated as a transmission mode as early as 1964 (Goldman *et al*) but were unable to transmit high energy laser beams without severe destructive changes. By the 1970's the output beam of argon (wavelength 0.458 μm - 0.515 μm) and continuous wave Nd:YAG (wavelength 1.06 μm) lasers had been successfully directed through an anti-reflective coated lens into a fibre some 400 μm in diameter. A helium-neon laser beam was also introduced into the fibre to serve as an aiming beam for the invisible Nd:YAG beam on the target site (Enderby, 1987). CO₂ laser emission (10.6 μm wavelength) is absorbed by all presently available fibre optic materials that may be used for transmission. Recently, a flexible hollow tube called

a "wave guide" has been developed with a transportable CO₂ laser, although simultaneous transmission of an aiming beam is not possible.

In 1983 Myers pioneered research into developing possible applications of the laser for use in general dentistry. The advantages of Nd:YAG over gas are that the concentration of active Nd ions may be as much as 10^{26} per cubic metre compared to only 10^{23} per cubic metre for gases. The YAG crystal is hard and stable and optically isotropic, so that it is transportable without the risk of damage to the laser systems (Enderby, 1987). YAG crystals can also be grown with high optical perfection and exhibit minimal loss for the laser radiation (Thornton *et al*, 1969). Unfortunately, Nd:YAG energy is reflected by enamel and must be coloured with agents such as methylene blue or Indian ink to obtain action.

By 1987 a prototype Nd:YAG laser for study on patients was approved by the FDA with American Dental Laser becoming commercially available in 1989. A typical Nd:YAG laser is pumped with a 9 kW krypton arc lamp, cooled by deionised water flowing around the rod in a quartz jacket and delivering 100 watts (Carruth and M^cKenzie, 1986). Working efficiency is about 1% (Robertson and Clark, 1988). The rod itself is 100 mm long and 6 mm in diameter. One mirror is 99% reflective, the other is 80%. The power is variable between 0.3 watts and 3 watts and is passed up a fibre-optic system to a choice of handpieces similar in size to conventional dental instruments. This provides excellent access which in turn leads to precise directional control in the oral cavity. There is a contact handpiece which is used for procedures

of less than one watt, for example etching, caries removal and dentine desensitisation. The contact handpiece allows use of the full range of energy, targeting in on the area it touches with a spot diameter of 0.2 mm to 0.3 mm. The energy is pulsed at approximately 150 microseconds.

2.2 Medical Lasers

There are 3 popular lasers used in medicine. These are argon, CO₂ and Nd:YAG. Each emits in a different part of the spectrum and therefore has different properties and affects tissues differently.

2.2.1 Carbon dioxide

This laser has emission in the middle of the infrared portion of the spectrum at 10.6 μm with an efficiency of approximately 13% (Weber, 1982). This wavelength is heavily absorbed by water. Animal tissues are composed of 70% - 90% water (Fuller, 1987) and thus there is very little scattering or reflection regardless of pigmentation (Robertson and Clark, 1987). Risk to underlying structures is minimised due to the lack of penetration beyond the surface (approximately 0.17 mm) (Fuller, 1987). All the energy is essentially used to dehydrate cells, with local tissue temperatures rising rapidly as vaporisation occurs (Johnson, 1990).

For human dental hard tissues the absorption peak for enamel is in the infrared region from 9 μm - 11 μm (Nelson *et al*, 1986a) which is near to that emitted by a CO_2 laser at 10.6 μm (Jeffrey *et al*, 1990a; Stern *et al*, 1972). This would appear to make it the ideal choice for use on enamel. However, it is also absorbed by all presently available fibre-optic materials that may be used for transmission (Jeffrey *et al*, 1990a).

The CO_2 laser's major disadvantage is that the beam is usually delivered to the point of use coupled to an operating microscope or with the aid of articulated mirrors. This means a cumbersome arrangement that is difficult to adapt for intraoral use. Very recently a commercially available laser more suitable for dental use has become available. This machine can be used in both continuous and pulsed mode depending upon the application, with a power range from 2 watts to 20 watts. The beam is delivered to the mouth by a "wave guide" which is a flexible hollow tube and a variety of straight and contra-angled tips. An aiming laser is not incorporated as it was not possible to transmit both beams down the same flexible tube.

2.2.2 Argon

This laser is capable of emitting different wavelengths in the blue (0.458 μm) to green (0.515 μm) portion of the visible spectrum simultaneously. Efficiency is about 0.1% with power outputs up to 20 watts (Edwards *et al*, 1983). The emission is

primarily absorbed in tissue by melanin and haemoglobin (Fuller, 1987). It is ideal for thermally destroying tissue without vaporisation (photocoagulation) (Robertson and Clark, 1988). Delivery to the tissue is via fibreoptics. Principal applications have been found in dermatology, ophthalmology and plastic surgery rather than dentistry. About 40% - 60% of the energy is scattered leading to tissue penetrations of approximately 1 mm (Robertson and Clark, 1988).

2.2.3 Nd:YAG

The lasing material for this device is a single crystal of yttrium-aluminium-garnet (YAG) doped with 1% to 3% of neodymium (Nd) ions. The neodymium is pumped with energy from a xenon or krypton arc lamp. This results in emission in the near infrared portion of the spectrum (1.06 μm). It is less absorbed by blood (absorption coefficient $\alpha \cong 4 \text{ cm}^{-1}$) than either argon ($\alpha \cong 34 \text{ cm}^{-1}$) or CO_2 ($\alpha \cong 230 \text{ cm}^{-1}$) and water absorption is minimal (Fuller, 1987). It's penetration into tissue is very deep and therefore can be used to coagulate blood vessels (Fuller, 1987). It is not absorbed effectively by human dental enamel (Spitzer and ten Bosch, 1975; Morioka *et al*, 1984) therefore the surface must be coloured with an agent to obtain any action. The laser energy is delivered by flexible fibreoptics with the distal end typically coupled to a hand probe for focussing. The laser is first absorbed in the coating material, then the heat produced at the surface of the tooth is transmitted into the adjacent enamel (Tagomori and Morioka, 1989).

CHAPTER 3

TISSUE EFFECTS

3.1 Biological Interaction

Surgical lasers are basically destructive devices with this destruction taking one of three forms - thermal, hydrodynamic or photochemical (Robertson and Clark, 1988).

The majority of tissue effects seen are due to heat, which occur only as a result of the absorption of the laser photon by biological tissue (Robertson and Clark, 1988).

This is determined by the following parameters (Fuller, 1987):

1. Absorption and scattering coefficients at a particular wavelength.
2. Power density of the optical radiation on tissue.
3. Duration of exposure.
4. Size of the radiated area.
5. Cooling component of the tissue, that is, blood flow.

Depending on the wavelength of the laser and the optical absorption characteristics of the tissue being irradiated, the laser beam may be absorbed, reflected, transmitted

or scattered within the tissue (Figure 3-1). Only absorbed light exerts an effect on tissue. Heat is produced in the absorbing material, with thermal penetration depending on the ratio of absorption versus scattering (Apfelberg, 1987).

The three primary absorbers in the body are water, melanin and haemoglobin (Johnson, 1990). Thus, for a given procedure that requires heating, the primary absorber is determined first and then the laser wavelength is chosen for maximum absorption. The CO₂ laser has strong tissue absorption with very little scattering, being absorbed at the surface with a small depth of penetration. This is in contrast with the Nd:YAG laser which has more optical scattering and deep tissue penetration. The argon laser is in between these two (Apfelberg, 1987). For example, CO₂ is best for surgery, as there is high absorption in cellular water with rapid heating and vaporisation at the impact site (Johnson, 1990). Argon, on the other hand, is absorbed better by haemoglobin and is therefore good for differentiating port wine stains from normal tissue where the haemoglobin concentration is lower (Figure 3-2).

Whether the tissue is warmed, dehydrated or vaporised depends on the energy per unit area or energy density. Energy input and output are conventionally expressed in joules (1 joule = 1 watt of power for 1 second).

$$\text{Power (watts)} \times \text{Exposure time} = \text{Energy (joules)}$$

$$\text{Power density (W/cm}^2\text{)} \times \text{Exposure time} = \text{Energy density (J/cm}^2\text{)}$$

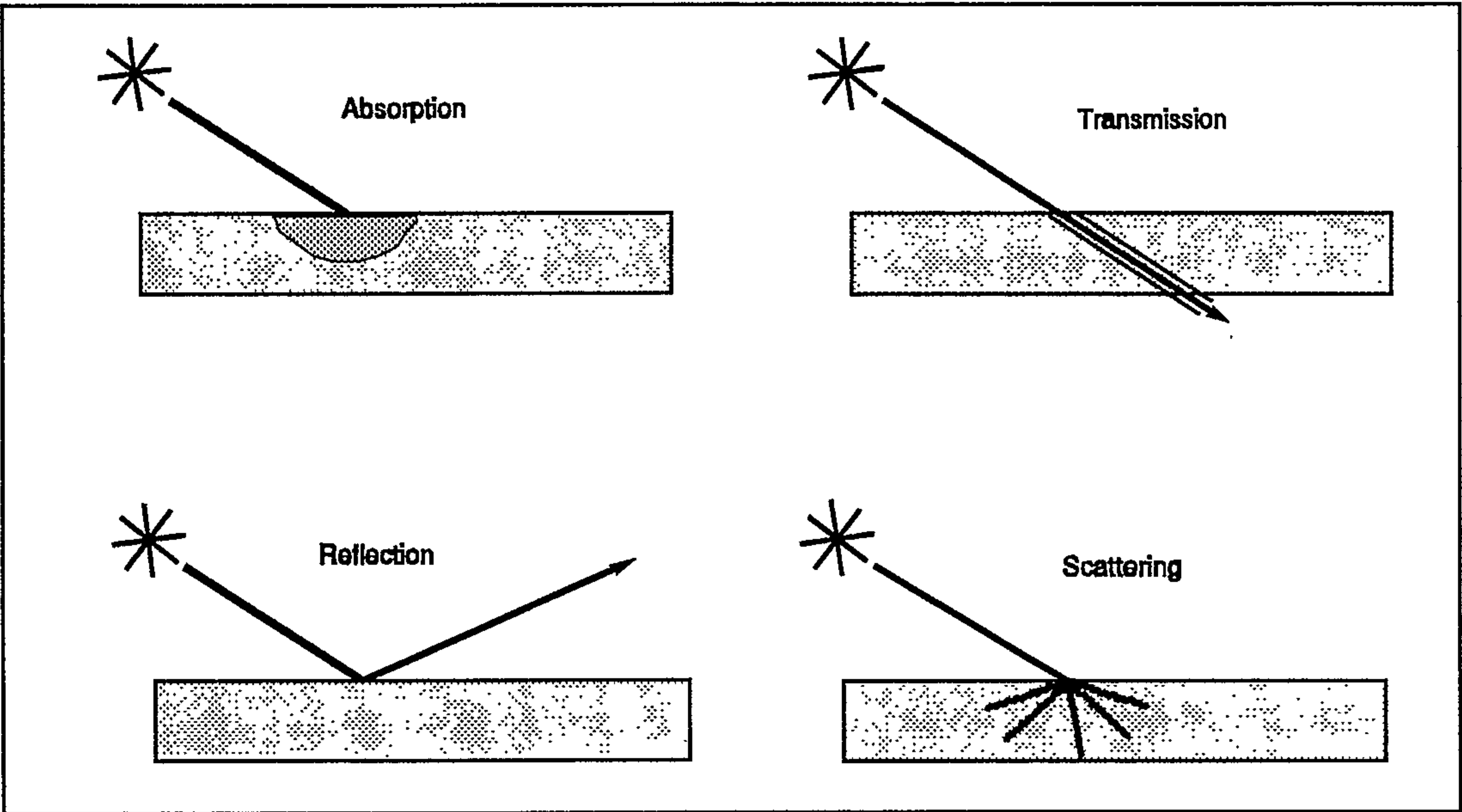


Figure 3-1 Interactions with biological tissue. From Robertson and Clark (1988).

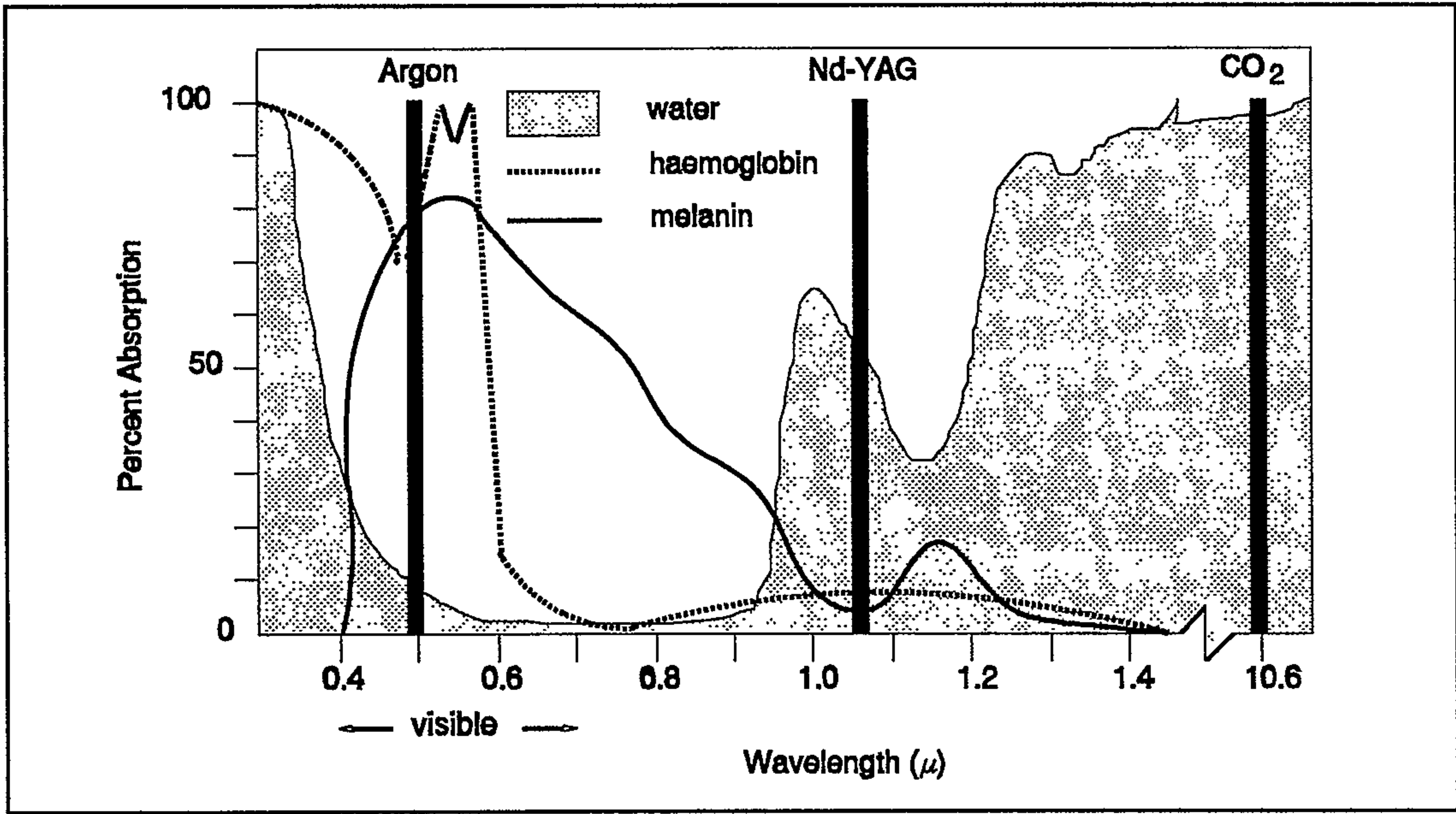


Figure 3-2 Comparison of wavelength-dependent absorption curves. From Johnson (1990).

Power and energy are dependent upon the laser type and design. Power density (irradiance) depends on the concentration of the absorbed light and is very closely related to the temperature increase at the absorbing site (Johnson, 1990). This can therefore be controlled directly by altering the power of the laser or by changing the area of the beam. When the beam width is greatly reduced tremendous power can be pulsed on a small area with concentrations up to $1 \times 10^6 \text{ W/cm}^2$ (Kinersly *et al*, 1965).

Duration of exposure of the energy is an important variable in controlling both desirable and undesirable effects on tissues. Exposure is dependent upon both the laser design and movement of the beam across the material. When tissue is heated above 60°C for a brief time protein denaturation and coagulation will take place resulting in necrosis (Johnson, 1990). As the temperature approaches 100°C , water begins to evaporate, causing shrinkage and tissue loss. Vaporisation of tissues occurs when it is heated above 100°C which is the basis for cutting and ablating (Johnson, 1990). If the energy density is increased, more energy is absorbed by the tissue and vaporisation occurs at a deeper level. There is heat transfer from the impact area to the adjacent tissues for as long as the laser beam is on (Fig 3-3). To minimise thermal damage to surrounding tissue, the beam should be on for the shortest time possible.

The disadvantage for ultra-short exposures is that the heating at the impact site is so intense, rapid and confined that an elastic or thermoacoustic transient pressure wave is produced (Johnson, 1990). Continuous wave lasers in medicine only generate minor acoustic waves with no deleterious effects. High power pulse lasers (particularly Q-

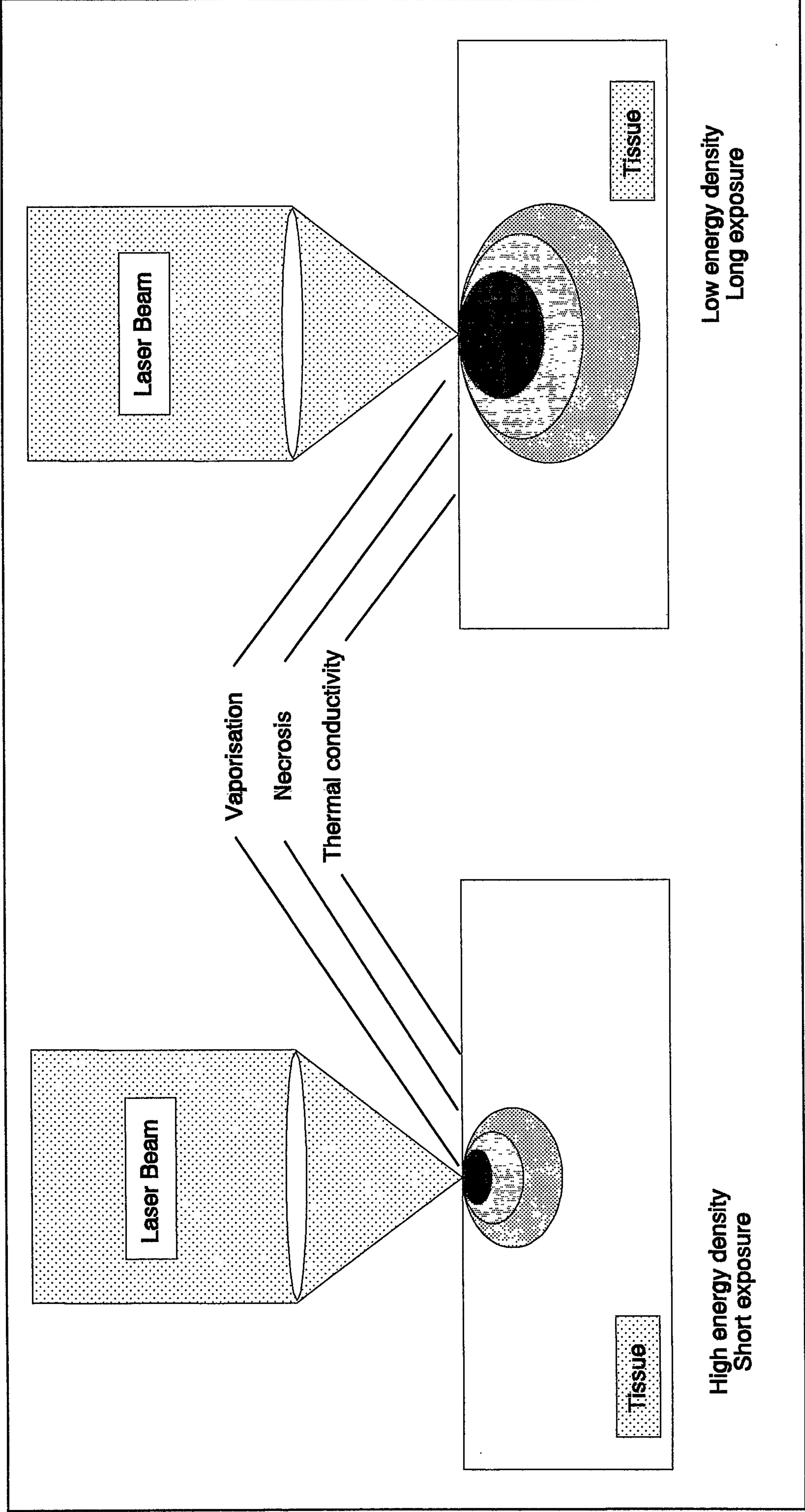


Figure 3-3 Heat transfer according to energy density and exposure time. Adapted from Johnson (1990).

switched) are capable of producing intense pressure waves that can tear tissue (Fuller, 1987; Johnson, 1990).

The majority of therapeutic procedures are performed at a minimal beam diameter where the power is concentrated into a small spot. Power density (PD) varies directly with the power wattage and inversely with the area of the focal spot ($PD = \text{watts per unit area}$) (Apfelberg, 1987). Therefore the larger the surface area the lower the concentration or power density. Two times the diameter equals a reduction in the PD by a factor of four. Also, as the distance to the tissue decreases the spot size more closely approximates the fibre diameter.

3.2 Dental Tissues

Applications of lasers to dentistry were reported only 3 to 4 years after Maiman's successful trial with the ruby laser. The limiting factor in all in vivo laser applications to dental hard tissues is that the temperatures and changes induced by the irradiant conditions do not damage oral tissues. Primary consideration is for (Fowler and Kuroda, 1986):

1. Pulpal integrity.
2. Enamel integrity.
 - a. resistance to caries
 - b. morphological changes.

3.2.1 Pulpal integrity

For lasers to be acceptable as a means of treating calcified structures in the oral cavity the effect of irradiation on the underlying pulp and soft tissue must be minimised (Stern and Sognnaes, 1964) since a pulpal temperature rise of 5.5°C or higher causes irreversible damage (Zach and Cohen, 1965). Early proposals of lasers in dentistry centred on altering calcified tooth structure using continuous wave ruby lasers with high energy densities (1000 J/cm²) coupled with long interaction times (up to 2 seconds) (Stern and Sognnaes, 1964; Goldman *et al*, 1965; Kinnersly *et al*, 1965; Stern *et al*, 1966). It was suggested, based purely on "theoretical calculation", that limited pulpal temperature changes were induced when the surface enamel was exposed to the laser beam. However, at the power required to effect morphologic changes such as cavitation in the surface enamel, there were deleterious effects to the pulp (Taylor *et al*, 1965). Animal experimentation showed that these effects appeared to be consistent with trauma due to excess heat with evidence of pulpal degeneration away from the point of irradiation (Taylor *et al*, 1965).

Haemorrhage was the most common finding, then coagulation necrosis of odontoblasts, oedema and inflammatory cell infiltration (Adrian *et al*, 1971). It appeared that to produce even a minimal recordable defect in the enamel one must use energy levels that would produce pulpal necrosis and therefore render lasers unsuitable for general use (Adrian, 1977). Less severe pulpal degeneration was noted in teeth subjected to less power or teeth at some distance from the focal point (Taylor

et al, 1965). Stern *et al* (1969) experimented further and noted minimal pulpal changes at energies sufficient to produce enamel changes that could prevent the formation of in vitro incipient carious lesions without altering the enamel macroscopically.

In addition to this possible thermal injury is the impact pressure which may also be deleterious to the pulp as it is essentially a closed liquid system (Stern *et al*, 1969). The impact pressure waves are dependent on the energy density. Therefore studies are needed to determine both the temperature effects and the pressure wave effects (Stern *et al*, 1972). They concluded that lasing of dental enamel at such high energy densities was not practical (especially with the ruby laser) if there were concomitant pulpal alterations. Their analysis of the ruby laser found that at its wavelength of 0.694 μm the enamel behaves like a diathermanous solid (laser energy divided in two with a large component reflected and a small component absorbed). This ratio in the ruby laser is 80 : 20. Therefore with an energy density of 500 J/cm² approximately 400 J is reflected (Stern *et al*, 1972). The portion entering the tooth causes heating and damage to the pulp.

However, the CO₂ laser is absorbed more efficiently at the enamel surface (behaving like an opaque solid) so that heat penetration that may otherwise injure the pulp is dissipated within a few microns of the surface (Stern *et al*, 1972). It would thus be advantageous to have a laser which allows a reduction in the energy density and an increase in the surface absorption (Stern *et al*, 1972). For example, a CO₂ laser with

a wavelength of 10.6 μm and an energy density of 13 J/cm^2 - 50 J/cm^2 produces enamel effects which could only be achieved with the ruby laser at 200 J/cm^2 - 700 J/cm^2 .

Other studies using a CO_2 laser on animal teeth (Melcer *et al.*, 1985; Melcer *et al.*, 1987; Jeffrey *et al.*, 1990a; Friedman *et al.*, 1991) and human teeth (Franquin and Solomon, 1986; De Raad *et al.*, 1988) agreed that pulpal reactions were mainly dependent on the laser intensity and focal spot size as well as the thickness of the residual dentine. The inflammatory changes appeared to be reversible and followed by new calcified dentine formation. Caution concerning therapeutic indications was advised, and thermal damage to the pulp was still regarded as a definite possibility (Jeffrey *et al.*, 1990b). As previously mentioned, pulsing a laser provides a way to enhance the peak power density whilst keeping the energy density constant.

Temperatures of more than 1000°C are required to fuse the surface of the enamel in order to weld cracks in the enamel (Renneboog-Squilbin *et al.*, 1989). Continuous wave argon laser at 4000 W/cm^2 for periods of 2 or 4 seconds generated temperatures in the pulp chamber which were less than those caused by contact with water at 54°C to 55°C for 1 to 2 seconds.

Initial reports on the Nd:YAG laser showed that cavitation in the enamel produced a pulpal response which was sharply demarcated from the surrounding normal pulp (Adrian, 1977). There was only mild haemorrhage and focal necrosis compared to

that produced by the ruby laser. It was suggested that the pulp was more resistant to injury by Nd:YAG without any real evidence to substantiate the claim. McCarthy (1990) used an Nd:YAG laser for a variety of treatments ranging from etching to removal of decay to desensitising dentine. The results suggested little or no change in tooth vitality. Launay *et al* (1987) compared the thermal effects of CO₂, Nd:YAG and argon medical lasers using surface infrared thermography and internal thermocouples. Contrary to McCarthy (1990), it was found that Nd:YAG irradiation diffused deeply through enamel and dentine. This led to pulpal overheating in proportion to incident energy, with the temperature in the pulp cavity exceeding healthy limits. They concluded that the Nd:YAG laser is the "wrong tool for dental treatment of hard tissues."

Arcoria *et al* (1991) related surface roughness produced by the laser to pulpal response, and found a 2 second pulse from a continuous wave coaxial CO₂ / Nd:YAG laser produced varying degrees of pulpal necrosis. The greatest increase in surface roughness was produced by the highest power laser treatment, but this also produced the greatest deleterious effects on the pulp compared to low power.

3.2.2 Enamel integrity

The ruby laser was most commonly used in early experiments on the effects of the laser on dental hard tissues as it was the first to be developed and was capable of reliable high energy output (Stern and Sognnaes, 1964; Kinersly *et al*, 1965; Goldman

et al, 1965; Taylor *et al*, 1965; Stern *et al*, 1966; Peck and Peck, 1967). The effects of energy densities for the 3 most popular medical lasers; ruby, Nd:YAG and CO₂ on oral tissues have been studied. However, the variety of wavelengths (0.694 μm , 1.06 μm and 10.6 μm), the differing energy densities and variations in the laser interaction times used make direct comparison between the studies not really possible (Jeffrey *et al*, 1990a; Nelson *et al*, 1986b). Visible light lasers are unsuitable due to the amount of radiation reflected, the large energy densities required and the less efficient absorption by enamel which increases the probability of damage to soft tissues and the pulp (Nelson *et al*, 1986b). There are only two contemporary lasers available primarily for dental applications; Nd:YAG and CO₂.

3.2.2.1 Resistance to caries

Early studies on the effect of lasing dental enamel indicated that it may play more of a preventive role rather than an alternative method of cutting enamel. The laser may impart protection that decreases acid dissolution and partially inhibits caries formation (Stern *et al*, 1966). The energy density required seemed to be relatively low (10 J/cm² - 15 J/cm²) but high enough to partially fuse the natural surface irregularities (Stern *et al*, 1972). This led to the presumption that the resistance was due to a *physical* alteration in the enamel surface giving a decrease in *permeability* to demineralising agents. A series of experiments using a Q-switched Nd:YAG laser tended to confirm this assertion (Yamamoto and Ooya, 1974; Yamamoto and Sato, 1980a; Yamamoto and Sato, 1980b).

Borggreven *et al* in 1980, on the other hand, speculated that the resistance may actually be due to *chemical* changes within the enamel itself causing a decrease in *solubility*. This view was supported by Kantola *et al* (1973) and Kuroda and Fowler (1984) using X-ray diffraction studies. They verified compositional changes due to rapid heating of the enamel by showing recrystallisation and growth in hydroxyapatite crystal size as well as the presence of the high temperature form of the apatite phase, α tri-calcium phosphate. Nelson *et al* (1986a) believed that the laser altered the permeability and hydration of the enamel which in turn alters the enamel solubility during artificial lesion formation. Goodman and Kaufman (1977) also reported a change in enamel solubility but put this down to the presence of fluoride at the time of lasing.

Ferreira *et al* (1989) took the opposite point of view, believing that the laser-induced effects of crazing and cratering of enamel did not reflect an overall improvement in the enamel structure. In fact, they believed that it increased the permeability of enamel leading to an increased loss of mineral rather than an improvement in the resistance to demineralisation.

In 1989 Tagomori and Morioka used an Nd:YAG laser with a varying energy density of up to 10 J/cm^2 at a pulse rate of 20 pps for 0.5 seconds (10 pulses). Both the structural and compositional changes seen were similar to previous studies. Acid resistance was also increased along with increased permeability of the enamel given the depth of penetration of fluoride into lased enamel. Oho and Morioka (1990) using

a continuous wave argon laser also found an increased permeability. They suggested that there was the formation of micropores in the enamel, a decrease in lattice strain and a decrease in the contents of water, carbonate and organic substances. This led to the hypothesis that ions released by demineralisation would be trapped in lased enamel whereas they would diffuse to the surrounding solution in unlased enamel.

The nature of the changes in dental enamel causing the reduction in demineralisation still remain unclear. It is feasible to attribute the improved acid-resistance to changes in both permeability and solubility (Tagomori and Morioka, 1989).

3.2.2.2 Morphological changes

For enamel, two surface processes are important (Nelson *et al*, 1986a);

1. Drilling and cutting - require long duration and high energy densities.
2. Rapid melting and recrystallisation - requires short duration and low energy densities, as in a pulsed laser.

Early experimentation concentrated on the former process in an effort to determine if the laser could replace the dental drill. Similar destructive changes were noted in several studies (Stern and Sognnaes, 1964; Goldman *et al*, 1965 ; Kinnersly *et al*, 1965; Taylor *et al*, 1965; Peck and Peck, 1967; Adrian, 1977). Grossly, the impact area consisted of a small shallow depression or crater with a white charring and

drying of the surrounding enamel (Stern and Sognnaes, 1964; Goldman *et al*, 1965). The surface in this zone was slightly roughened (Taylor *et al*, 1965). The white region was thought to be due to desiccation by the extreme heat (Adrian, 1977). The surface of the craters were coarsely textured and non-reflective with a noticeable demarcation between the lased areas and the bordering smooth enamel surface (Peck and Peck, 1967). Microscopically, the craters displayed a glaze-like lustre with the enamel at the periphery appearing to be fused (Abed *et al*, 1990; Cooper *et al*, 1988).

The severity of the enamel changes induced by the ruby laser appeared to be directly related to the intensity of the laser beam. Energy densities of considerable magnitude were required to produce even minimal defects in the enamel (Goldman *et al*, 1965; Taylor *et al*, 1965). These were also most likely to produce pulpal necrosis (Adrian, 1971). Kinersley *et al* (1965) made the observation that carious areas which were stained, dull and rough were more readily affected than the pale, shiny and substantially smooth virgin enamel, and therefore was not an ideal substance for maximal absorption of the laser. It was soon realised that the laser was not a practical alternative to the drill in cavity preparation.

Attempts at fusing filling materials in cavities and carious regions failed (Goldman *et al*, 1965) but Peck and Peck in 1967 were able to fuse powdered porcelain to itself as well as with its plate glass matrix. Later, welding of cracks on extracted human teeth using an argon laser was shown microscopically to be effective for half the enamel thickness (Nammour *et al*, 1987).

The 1970's and early 80's were dominated by research with the CO₂ laser. The approximation of the CO₂ wavelength to the enamel absorption peak increased the surface energy absorption, decreased energy loss by reflection and transmission and was thus significantly more efficient (Stern and Sognnaes, 1972). This permitted the use of much lower energy densities - approximately one tenth of that used with earlier ruby lasers.

Stern and Sognnaes (1964) used a pulsed CO₂ laser to determine to what extent macroscopic changes can be induced in the enamel and how the appearance was affected by differing energy densities. Pulses of 13 J/cm², 25 J/cm² and 50 J/cm² were 50 milliseconds long and 100 milliseconds apart. At 13 J/cm² there was superficial surface fusion. In some regions, flaking away of enamel exposed normal looking structures. At 25 J/cm² surface alterations were more pronounced. Enamel fusion made surface characteristics less recognisable. The enamel appeared similar to a fine crust which at high magnification was cracked and irregular. Some lased portions showed surface flaking and others showed cracks and fissures. The exposed subsurface enamel appeared to have melted crystals that subsequently resolidified. At higher magnifications the small fissures were connected by micropores. At 50 J/cm² gross enamel changes were apparent immediately after laser exposure. The surface was crazed and mottled with white opacities and light brown discolouration. They also predicted that the thermal diffusion of CO₂ is such that one would expect a temperature rise just below the surface (10 µm) of 5% of the incident energy.

Other SEM views of enamel lased by CO₂ and pulsed Nd:YAG lasers showed

morphological changes consistent with that of Stern *et al* in 1972. At low magnification there was a distinct surface melt with extensive surface roughening near the centre but smooth and glaze-like near the edges (Nelson *et al*, 1986a, 1986b, 1987). At higher magnification there appeared to be two different changes in the enamel - crazing alone, and crazing and cratering (Ferreira *et al*, 1989). In regions where the enamel prisms alternated in parallel and perpendicular planes (Hunter-Schreger-bands) the appearance of the lased enamel depended on this pattern. Prisms oriented in a parallel plane typically showed rough exposed enamel whilst prisms that were perpendicular to the surface often showed crazed and cratered and/or crazed enamel (Ferreira *et al*, 1989). The fused glaze-like surface commonly "blew off", revealing rough exposed enamel (Ferreira *et al*, 1989; Nelson *et al*, 1986a) that appeared to be a partially fused mass of crystallites (Nelson *et al*, 1987). In cross-section, the depth of melting, fusion and recrystallisation of enamel ranged from 2 μm to 11 μm (Ferreira *et al*, 1989; Nelson *et al*, 1986a, 1986b, 1987). The surface regions was at times separated from the underlying intact enamel (Nelson *et al*, 1987).

Microscopic shallow cracks were produced that occurred at 10 μm - 20 μm intervals parallel to the enamel surface (Tagomori and Morioka, 1989). Ferreira *et al*, (1989) felt that the cracking was due to a combination of expansion and contraction, with heating and shock waves due to beam interaction with the tooth. Cracks were less than 0.3 μm and extended to 60 μm - 70 μm in depth. Underneath the surface zone was a denatured zone of 10 μm - 20 μm . The temperature rise in this region was

large enough to affect the organic component of enamel (Nelson *et al*, 1987). Heat treatment of dental enamel in the 60°C - 300°C region results in dehydration and denaturation of the organic matrix (Nelson *et al*, 1986b).

CHAPTER 4

ENAMEL ETCHING

4.1 Enamel Ultrastructure

Enamel is almost entirely inorganic, consisting of 95% crystalline calcium phosphate (hydroxyapatite) by weight (Maas, 1991). Various ions may also be incorporated into the apatite crystals if present during initial crystal formation (Boyde, 1989). The other 5% is organic material and water which appears as a "fine, lacy network" fully occupying the inter-crystal spaces (Ten Cate, 1985).

According to Maas, there are two major structural categories of enamel:

1. **Non Prismatic.**

Enamel crystallites are perpendicular to the tooth surface with their long axes predominantly parallel to one another, giving a relatively homogenous structure. This type of enamel usually occurs in the gingival third of secondary teeth (Ten Cate, 1985).

2. Prismatic.

Bundles of similarly oriented crystals form prisms which are separated from other prisms, with differently oriented crystallites, by interprismatic material. This gives a heterogenous structure which may present in a number of different packing patterns (Boyd and Martin, 1987).

Every enamel rod has an undulating course with rows of rods arranged circumferentially around the long axis of the tooth and perpendicular to the surface of the tooth. The prisms or rods have an average width of approximately 5 μm but this varies throughout the thickness of the enamel (Ten Cate, 1985). The striae of Retzius are incremental growth lines reflecting successive enamel-forming fronts and are seen as concentric rings (Ten Cate, 1985). At the surface, the enamel topography in young permanent teeth is characteristically a series of concentric, alternating ridges and grooves that encircle the coronal aspects of the teeth (Hoffman, 1972). The grooved perikymata represent the site of surface enamel immediately overlying these striae and appear as shallow furrows. Imbrication lines are a representation of the surface terminal extensions of these striae and appear as convex ridges (Hoffman, 1972).

Acid-etching of the enamel surface causes selective dissolution of crystals with the pattern of etch depending on the crystal orientation in the prisms or rods. Crystals lying perpendicular to the surface are most vulnerable (Ten Cate, 1985). The effects of laser etching, as mentioned previously, are also dictated by the crystal pattern.

Cratering and crazing was found by Ferreira and colleagues (1989) to also be preferentially associated with crystals perpendicular to the surface.

4.2 Acid-Etching

Enamel surfaces can be chemically treated to improve the adhesion between tooth structure and restorative materials. The currently accepted method is acid-etching which was introduced by Buonocore in 1955. Consideration was given to the possibilities by which acid-conditioning enhances bonding;

1. An increase in surface roughness and therefore an increase in the bonding area.
2. Exposure of organic material of the enamel which may serve as a network in, about, and to which the resins can adhere.
3. Creation of spaces in depth along interprismatic areas into which the resins can penetrate.
4. Removal of old enamel surfaces exposing a fresh reactive surface more favourable for adhesion.
5. Increased wettability of the surface with an adsorbed layer of highly polar phosphate groups derived from the acid.

Microscopic examination of the bonded surface showed the presence of "prism-like"

tags without which poor bonding resulted (Buonocore *et al*, 1968). Initial scanning electron microscopy studies indicated that the mechanism of acid attack was primarily a penetration to depths of 50 μm to 60 μm through the interface between the rod core and the rod sheath (Hamilton *et al*, 1972). No tag-like extensions of the adhesive were observed in tooth enamel that was not acid-etched before bonding (Myers *et al*, 1974). Penetration into enamel is a major key to the enhanced resin bond strength (Gwinnett and Buonocore, 1965) which relies on mechanical interlocking (Retief, 1978). Silverstone *et al* (1975) later described and classified three types of etching patterns found in the coronal and middle thirds of the buccal surface;

1. Prism cores preferentially removed.
2. Peripheral regions removed leaving the cores intact.
3. Areas corresponding to both 1. and 2.

Galil and Wright (1979b) added two further types which were mainly found in the cervical third of the tooth;

4. Porous.
5. Smooth.

Since then, bonding, rather than banding of orthodontic attachments has become a routine clinical practice.

The advantages of this are (Retief *et al*, 1970):

1. Reduced gingival irritation.
2. Improved aesthetics.
3. Ability to maintain better oral hygiene.
4. Elimination of band occupying interdental space.
5. Decreased risk of decalcification due to leakage beneath bands.

Crucial to these bonding procedures is the *quality* of the enamel etch (Galil and Wright, 1979b). The effects on etching patterns with different acids of different concentrations for different times has been widely studied (Galil and Wright, 1979a and 1979b; Oliver, 1987; Bryant *et al*, 1987; Kinch *et al*, 1988; Soetopo *et al*, 1978; Mardaga and Shannon, 1982). It was concluded that 30% unbuffered phosphoric acid provided the most suitable conditions for bonding (Dogon and Silverstone, 1975). With respect to orthodontic brackets, a time of 15 - 30 seconds fulfils the criteria of minimal strength to resist occlusal forces (Newman, 1965; Reynolds, 1975). However, it is difficult to confine the phosphoric acid solution solely to the area that will be covered by the bracket. Concern has thus been expressed regarding the etching of enamel adjacent to the brackets since removal of fluoride-rich surface enamel may predispose these surfaces to decalcification during orthodontic treatment (Ceon and Gwinnett, 1981).

4.3 Laser-Etching

It has been proposed that the laser may be used as an alternative to acid-etching prior to bonding of orthodontic brackets. Initial studies that specifically compared the morphological appearance created by acid-etching and laser-etching of human dental enamel were not favourable. Winkler *et al* (1976) examined 5 extracted bicuspid under reflected light. The buccal surface had been acid-etched with 30% phosphoric acid and the lingual surface lased with a 2 millisecond pulse at 12.1 J/cm^2 from a neodymium laser. In *section*, they observed "no significant differences between the two techniques". The bond strength of composite to the acid-etched surface was, however, greater. They concluded that there was no significant advantage offered by laser etching over the currently used acid-etching for retention of resin restorations.

In 1977 Goodman and Gwinnett conducted a similar study using 50% phosphoric acid for 60 seconds and a continuous wave argon laser with an energy density of 300 J/cm^2 for half a second. The extent of resin penetration into both etched surfaces was also examined. Ten molar and incisor teeth were used. Their results highlighted 3 important differences in the morphological appearance of the enamel surface between acid-etching and laser-etching.

1. The enamel rods of the lased surface had a globular, coalesced appearance consisting of cracks, pits, fractures, blisters and craters

contrasting markedly with the discrete, uniform, spicular surface obtained with acid.

2. The heat involved with lasing melted some of the exposed crystals producing a non-porous surface compared to the increased porosity found with acid-etching.
3. Resin penetration in the form of tags was extensive when related to the acid-etched surfaces, but absent in the lased surfaces.

At this stage, laser-etching, particularly with an argon laser, was not regarded as a practical, desirable or economically sound alternative to acid-etching.

A more promising result was obtained in 1984 by Liberman *et al* when they compared bond strengths of composite materials to enamel etched with acid and a CO₂ laser. They did not look at the surface created, but presumed that the hard tissue changes previously reported of fissures, cracks and micropores increase the surface area which may then afford the additional retention. They concluded that the bonding strength following laser pretreatment was comparable to that of acid-etching. Other studies have also indicated that etching with the CO₂ laser may have the potential to produce a rougher surface than standard acid-etching (Macdonald *et al*, 1990; Zakariasen *et al*, 1991).

The Nd: YAG laser has also been used successfully in the pretreatment of enamel surfaces prior to bonding of orthodontic brackets (Abed *et al*, 1990; White *et al*, 1991a). Forces required to debond the brackets were the same or higher than with the use of an acid gel, but below that obtained with liquid-etched enamel (Abed *et al*, 1990). Even bond strength of composite to dentine has been shown to be increased between 100% (White *et al*, 1991b) and 300% (Cooper *et al*, 1988).

A number of studies (Myers, 1990; Myers and Hess, 1990; Myers and Riddle, 1990) support the view that both high and low power SEM photos of human dental enamel lased with the Nd:YAG show comparable surface morphology to that obtained with phosphoric acid. Using a pulsed Nd:YAG laser, human enamel and dentine samples were exposed in vitro and then examined both histologically and under scanning electron microscopy. The average energy density was 0.8 J/cm^2 delivered as 0.8 W at 10 pps. Morphologically, the lased surface was described as being covered by multiple overlapping circular patterns which took on a "soda cracker" appearance at higher magnification. The depth of enamel cratering was measured at between 15 μm and 40 μm , although no mention was made of how this was calculated.

Myers stated that the morphological changes in the enamel were very similar between the two techniques (Myers and Riddle, 1990). Of greater significance was the finding that both microscopically and histologically there was a total absence of cracking or fissuring in the adjacent unlased tissue. Sadly, no mention is made of the presence or absence of cracking in the *lased* area, even though it was admitted that it "is

indicative of thermal damage resulting in weakening of the strength of the enamel."

In a similar study, Hess (1990) irradiated extracted human teeth also with an Nd:YAG laser. Eight teeth were coated with laser initiator. Four were irradiated at 95.5 J/cm^2 and four at 238.8 J/cm^2 until the initiator was completely removed. The specimens were then examined under SEM. Grossly, a well defined opaque surface limited to the area coated with initiator was viewed with adjacent enamel appearing normal. Microscopically, the surface showed a variable appearance depending on the energy used. At the lower energy density he reported many small bubble-like cavities making up a pockmarked surface with indistinct craters. At 238.8 J/cm^2 these craters presented a similar appearance as that described by Myers (1990) - more distinct, overlapped and approximately $230 \mu\text{m}$ - $250 \mu\text{m}$ in diameter. The overall impression was of bubbling, molten enamel that had solidified leaving circular cavities and pores.

Interestingly, microcracks appeared in all the specimens within the laser-etched surfaces but not in the control areas. A later experiment by Hess (1991) found the cracks ranged from depths of $10 \mu\text{m}$ - $15 \mu\text{m}$ and resulted, not from an artefact due to preparation of the sample, but from expansion of gases and other materials from the surface. He was unsure to what extent they would contribute to retention of adhesive materials.

Combining laser-etching with laser-induced resistance to decalcification of enamel would result in the bonding of orthodontic brackets with a decreased likelihood of

"white spot" formation. However, the effects are accomplished using different exposure parameters and it is doubtful whether etching would simultaneously provide preventive benefits. Zakariasen *et al* (1991) thought that exposures used for etching would only roughen a thin layer of the enamel surface. The underlying layer would be unaffected by the laser and therefore would not have increased resistance to decalcification. Their opinion was that laser-etching, providing the enamel roughness, would be completed first and then the surrounding enamel laser treated separately to reduce the probability of decalcification.

Most investigators agreed that more scientific investigation needs to be accomplished before procedures such as laser-etching can be reasonably advocated for clinical practice. In particular, the disruption to the enamel surface needs to be examined to determine if some of these alterations, such as cracking, are more deleterious than the overall benefit gained clinically. As yet, there is no consensus in the literature as to the most suitable energy density for etching although some information is known about the effect of altering this parameter. The physical means by which resin interlocking is obtained could then be compared directly to bonding strengths, so the point of maximum benefit with minimal disadvantages could be determined.

CHAPTER 5

EXPERIMENT

5.1 Aims

1. To match the clinical process of laser-etching as closely as possible in preparation for the bonding of orthodontic brackets, and to examine the subsequent changes induced in human dental enamel both clinically and with scanning electron microscopy.
2. To compare the qualitative effects of varying the energy density, pulse rate, and initiator use, on the surface appearance of the lased human dental enamel.
3. To compare the topographical and cross-sectional appearance of laser-etched human dental enamel with both normal and acid-etched human dental enamel.

5.2 Materials and Methods

5.2.1 Laser

The laser used was the American Dental Laser (Birmingham, Michigan).

Active medium	Nd:YAG
Aiming beam	HeNe (wavelength = 0.63 μm)
Beam type	Pulsed
pulse width	150 μsec
pulse rate	10 pps - 30 pps
Delivery system	Fibre optic
fibre diameters	320 μm and 200 μm
fibre type	Contact
Power output	0.3 W - 3.0 W
Energy at fibre end	30 mJ - 180 mJ
Energy density	Pulse energy divided by area. Calculated according to $4\text{PE} / \pi D^2$ where PE = pulse energy and D = fibre diameter. Measured in J/cm^2 .

5.2.2 Tooth preparation

Human premolar teeth extracted for orthodontic purposes were obtained from the

Department of Oral and Maxillofacial Surgery at Westmead Hospital. The teeth were kept in 70% alcohol until required. They were then debrided of soft tissue, cleaned and examined under a light microscope for defects in the labial enamel. The sound teeth were then divided into 2 major groups, those to be laser-etched and those to be acid-etched. The roots were retained for ease of handling.

5.2.2.1 Laser-etching

Group 1:

The teeth were dried with a triplex syringe. Laser initiator¹ was applied to an area on the labial surface approximately 4 mm in diameter with a dycal applicator. The amount of ink applied was kept as uniform as possible between the samples. The parameters for the power / pulse rate setting on the laser and the diameter of the fibreoptic tip were varied to give a range of energy densities (Table 5-1). Three teeth were lased per subgroup.

The fibreoptic tip was held approximately 1 mm - 2 mm from the coated enamel and passed slowly across the surface until no further "sparking", denoting the end of laser activity, was observed. Groups 1A and 1J represent, respectively, the minimum and maximum energy densities obtainable at the fibre end. Group 1B is the power and pulse rate setting recommended by the manufacturers for etching of enamel. Groups

¹ Black Sumi Ink KF2, Yasutomo and Co., San Francisco, California.

GROUP	FIBRE DIAMETER (μm)	PULSE RATE (per second)	POWER (W)	ENERGY PER PULSE (mJ)	ENERGY DENSITY (J/cm^2)
1A	320	10	0.30	30	37.3
1B	320	15	0.75	50	62.2
1C	320	30	1.50	50	62.2
1D	200	10	0.30	30	95.5
1E	320	15	1.50	100	124.3
1F	200	15	0.75	50	159.2
1G	200	30	0.75	50	159.2
1H	320	20	3.00	180	223.8
1I	200	15	1.50	100	318.3
1J	200	20	3.00	180	572.9

Table 5-1 Range of laser settings used.

1C and 1H received the same energy density as the standard setting with both the 320 μm (Group 1B) and 200 μm fibres (Group 1F) but at twice the pulse rate giving two times the power in the same time. Group 1E also imparts twice the power as Group 1B in the same period by keeping the pulse rate constant but doubling the energy per pulse. By altering the fibre diameter the power, pulse rate and energy per pulse can remain the same but the actual energy density is increased.

Group 2:

The teeth in this group were irradiated at the recommended parameters of 50 mJ per pulse at 15 pps with the 320 μm fibre (giving a displayed power of 0.75 watts and an energy density of 62.2 J/cm^2) but the amount of initiator applied was varied. Three teeth were lased per subgroup.

Group 2A	No initiator
Group 2B	0.1 μL initiator
Group 2C	0.5 μL initiator
Group 2D	50:50 water : initiator

Group 3:

To compare the appearance of the lased enamel with that of acid-etched enamel, teeth were prepared individually, or on either side of the tooth crown as set out for Group 1 laser-etching or acid-etching. Three teeth were acid-etched alone. Four teeth were divided along the long axis on the labial surface. Three were designated for half

laser-etched / half acid-etched and one for half laser-etched / half normal. The lasing parameters used were 50 mJ per pulse at 15 pps with both 200 μm and 320 μm diameter fibres giving energy densities of 159.2 J/cm² and 62.2 J/cm² respectively. One of each of the teeth that was half lased / half etched and half lased / half normal were also designated for coronal sectioning.

5.2.2.2 Acid-etching

37% phosphoric acid gel was used. The teeth were pumiced and washed. The gel was applied for 30 seconds washed for 30 seconds and dried with a triplex syringe.

5.2.3 Scanning electron microscopy

All samples to be viewed topographically were attached to aluminium stubs and gold coated. They were examined in the Jeol JSM-840A scanning electron microscope (operating at 15 kV) housed in the education clinic of the dental school. The teeth to be used for the cross-sectional view were then removed from the stub and the root cut off. The crown was then coated with cold cure acrylic resin to prevent enamel fracture during preparation. These teeth were then sectioned coronally using a steel circular blade. Both occlusal and cervical sections were then mounted on stubs and carbon coated for use with back scattered electrons.

To avoid operator bias in selecting areas that may not be truly representative of the alterations induced in the enamel surface, a method of random selection was used for each of the teeth in groups 1 and 2. The tooth, positioned with the long axis vertically, was magnified until the etched area, delineated by the initiator coverage, filled the entire viewing screen of the SEM. This was then accepted as the central point of the lased area. Photographs were taken at 3700x, 1000x and 200x in this position, as well as a low magnification picture of the area as a whole. Using the calibrated X and Y axis markings on the microscope, the specimen was moved alternately 1 mm to the right and 1 mm to the left and the photographs repeated. A total of ten photographs per tooth were taken.

CHAPTER 6

RESULTS

6.1 Clinical Appearance

All teeth had a well demarcated area of "etching" that was easily distinguished from non-lased surfaces, and corresponded to application of the laser initiator. The appearance of this area varied according to the energy density used. At low densities, the surface was flecked as a result of initiator remaining between the individual impact sites of the laser beam. Overall, the enamel appeared to be only slightly less reflective than the adjacent unlased enamel (Fig 6-1 and Fig 6-2). With increasing energy density, the area became progressively lighter in colour with less residual initiator present (Fig 6-3 and 6-4). The impact areas became more opaque and ultimately less reflective, so that at the highest energy density the total area was not unlike acid-etching, albeit not as uniform. The point at which the impact sites were distinguished due to their own opacity, rather than lack of initiator, occurred between 95.5 J/cm^2 and 124.3 J/cm^2 .

Clinically, an accurate assessment of the type or extent of surface alterations could

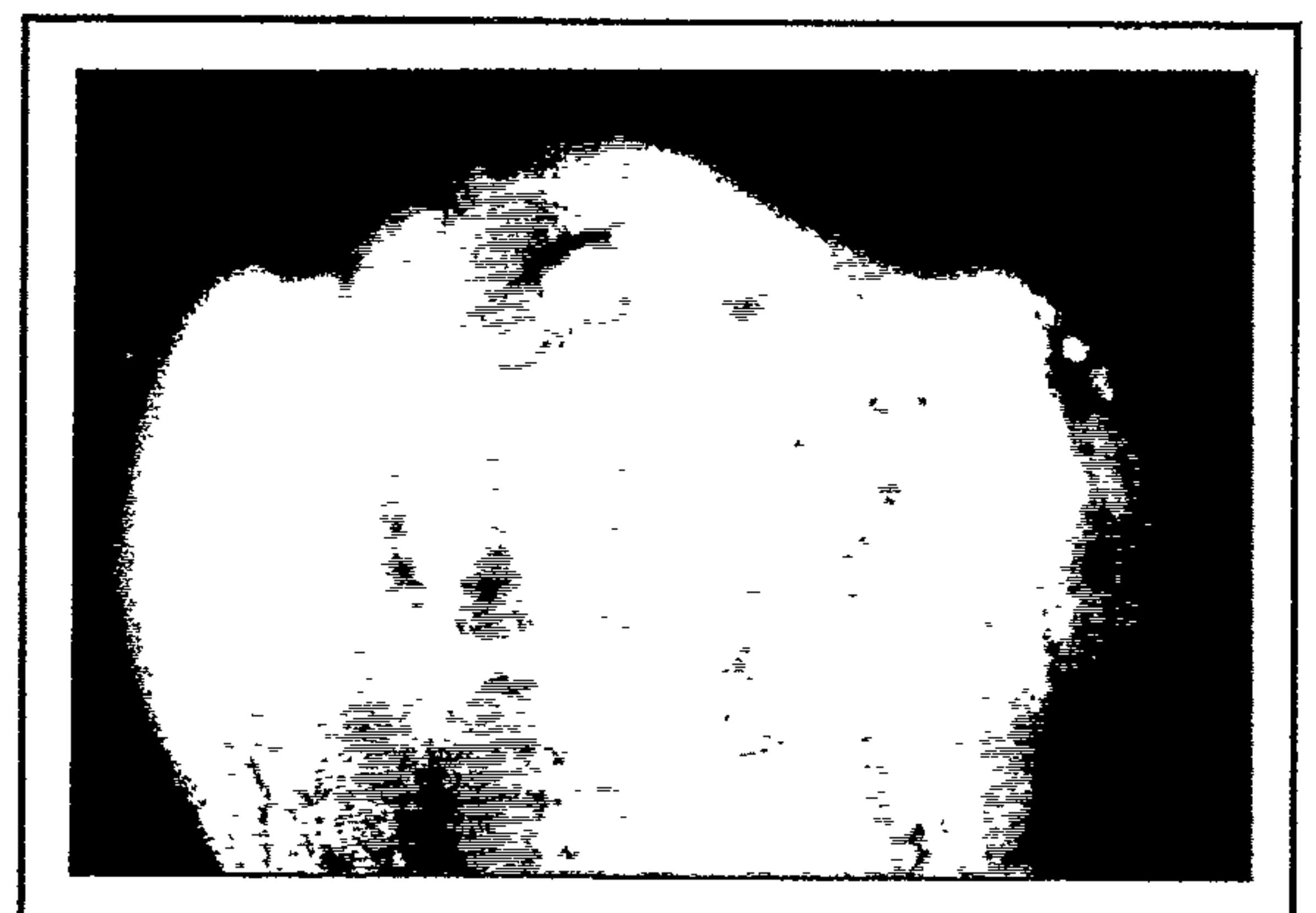
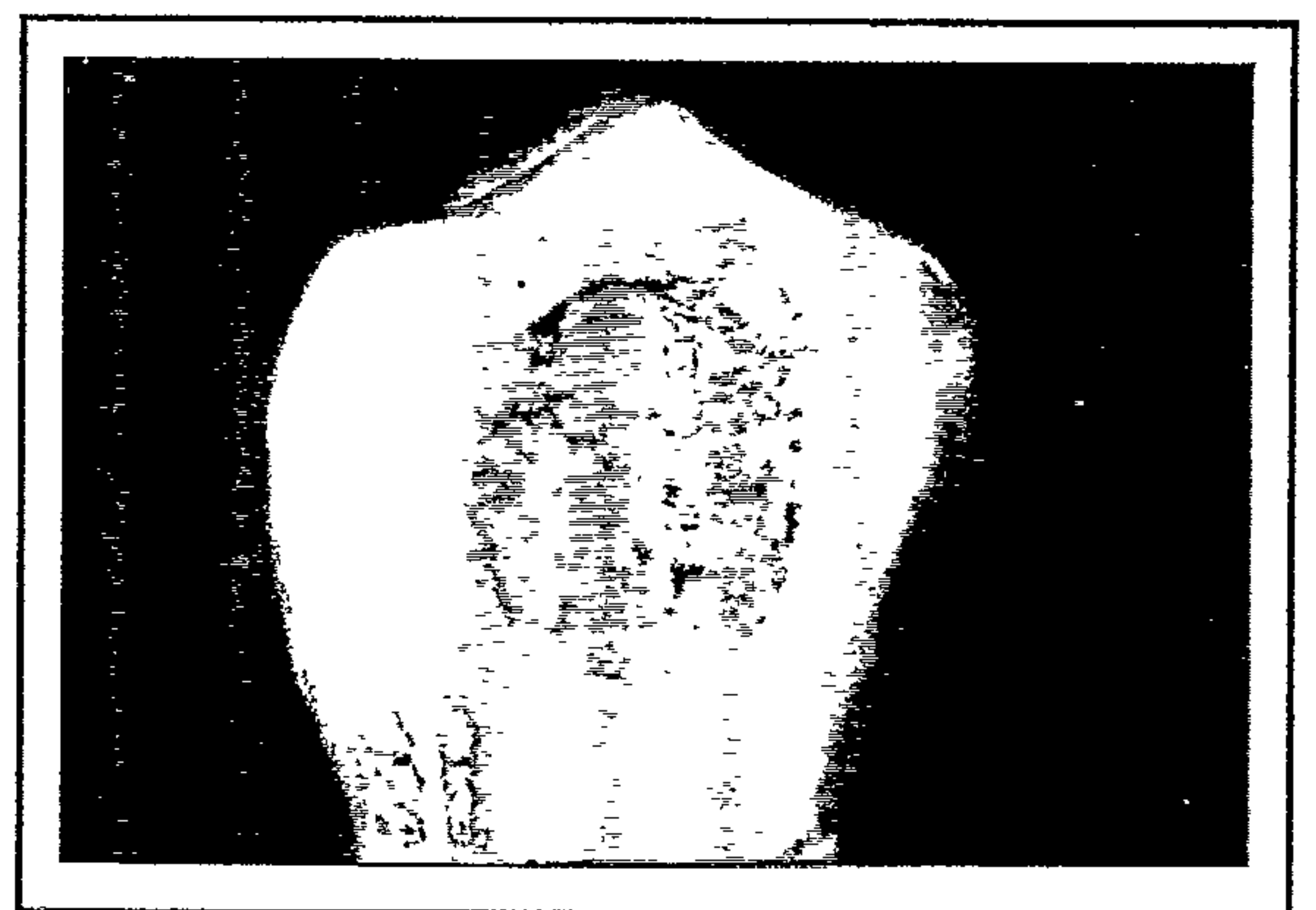
Figure 6-1 Clinical view of laser-etching at an energy density of 37.3 J/cm².

Figure 6-2 Clinical view of laser-etching at an energy density of 95.5 J/cm².

Figure 6-3 Clinical view of laser-etching at an energy density of 223.8 J/cm².

Figure 6-4 Clinical view of laser-etching at an energy density of 572.9 J/cm².

Results



not be made by visual inspection of the lased area. However, it would provide an indication of the energy density used. At this stage, it could not be said if the tooth was "well etched" or not.

6.2 Surface Alterations

The surface of each of the teeth was examined using scanning electron microscopy (SEM). Magnification of 30x - 400x showed the overall pattern of surface changes and 1000x - 4000x was sufficient for detail of these enamel alterations. Variation was found with changes in the energy density, pulse rate, as well as the amount of initiator used (Table 6-1).

6.2.1 Energy density

Microscopically, differences in the enamel surface are apparent even at the lowest energy level. The area delineated by the initiator is "peppered" by discrete disks with diameters corresponding closely to that of the laser fibreoptic. This would imply that these disks are the sites of direct absorption of energy by the initiator overlying that point of the tooth surface. They overlap in some places and are spaced in others giving two general surfaces - *disk area* and *interdisk area*. The microscopic distinction between these areas was not always readily apparent. At low energy

		MICROSCOPIC APPEARANCE	
LASER VARIABLE	ENERGY DENSITY	37.3 - 95.5 J/cm ²	Disk and interdisk areas not well delineated. Enamel flaking occurred preferentially over imbrication lines. Micro-cracking present in all samples except one. Areas of both sparsely distributed and concentrated pitting. Minimal areas of random cratering.
		124.3 - 223.8 J/cm ²	More obvious demarcation between disk and inter-disk areas. Less pitting and more cratering - definite pattern emerging with peripheral distribution in disk areas. Cracking and flaking present in the same dimensions and quantity.
		318.3 - 572.9 J/cm ²	No improvement in the quality of disk cratering for mechanical interlocking. Some impact areas showed more "destructive" effects. Cracking and flaking present.
	PULSE RATE	10 - 30 pps	Increasing the pulse rate did not change the quality of the surface alterations but it did increase the proportion of disk area to inter-disk area.
	INITIATOR	None	No surface alterations.
		0.1 - 0.5 µL	Reduction in the quality of enamel alterations. Increased amount of residual debris present.
		50:50 H ₂ O:Initiator	No change in the surface alterations in 2 specimens. One specimen showed areas of pitting without cratering.

Table 6-1 Summary of microscopic enamel alterations with respect to laser variables.

Figure 6-5 Low magnification SEM view of laser-etched area showing featureless impact disks.

Energy density = 37.3 J/cm^2

Bar = 1 mm

Figure 6-6 Low magnification SEM showing impact disks with obvious changes in the enamel surface.

Energy density = 223.8 J/cm^2

Bar = 1 mm

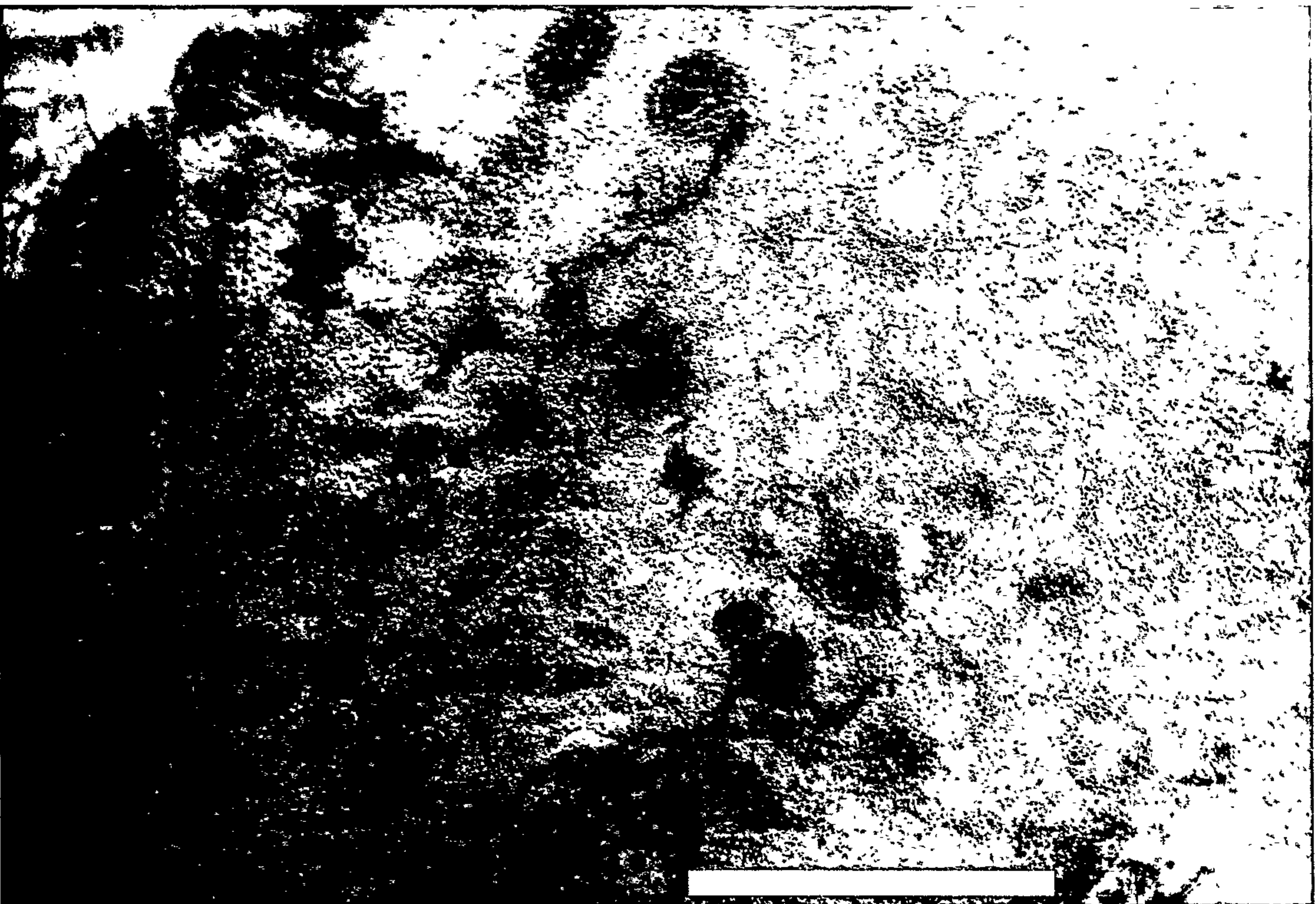
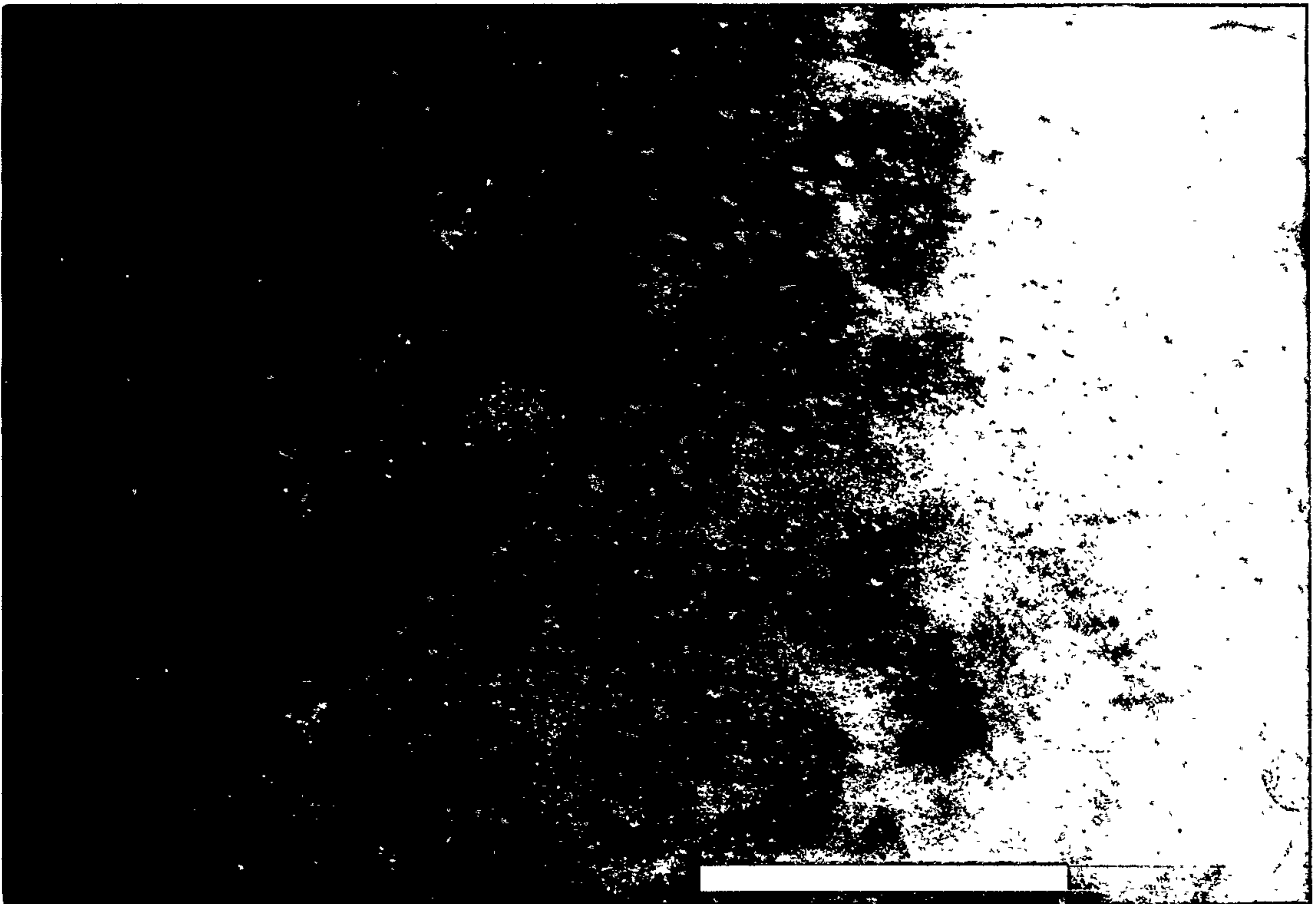


Figure 6-7 Surface cracking in both the disk and interdisk areas of enamel.

Energy density = 124.3 J/cm²

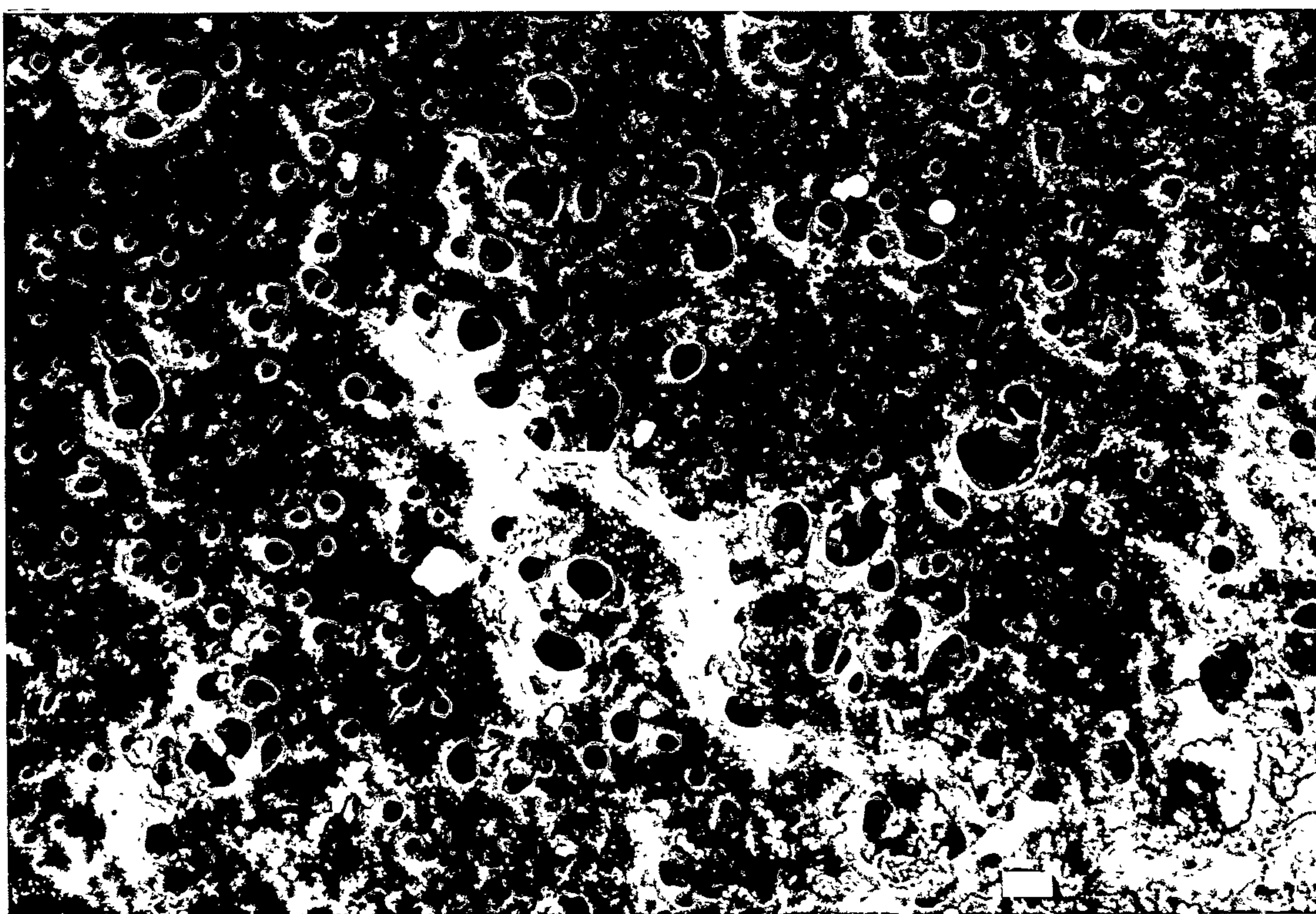
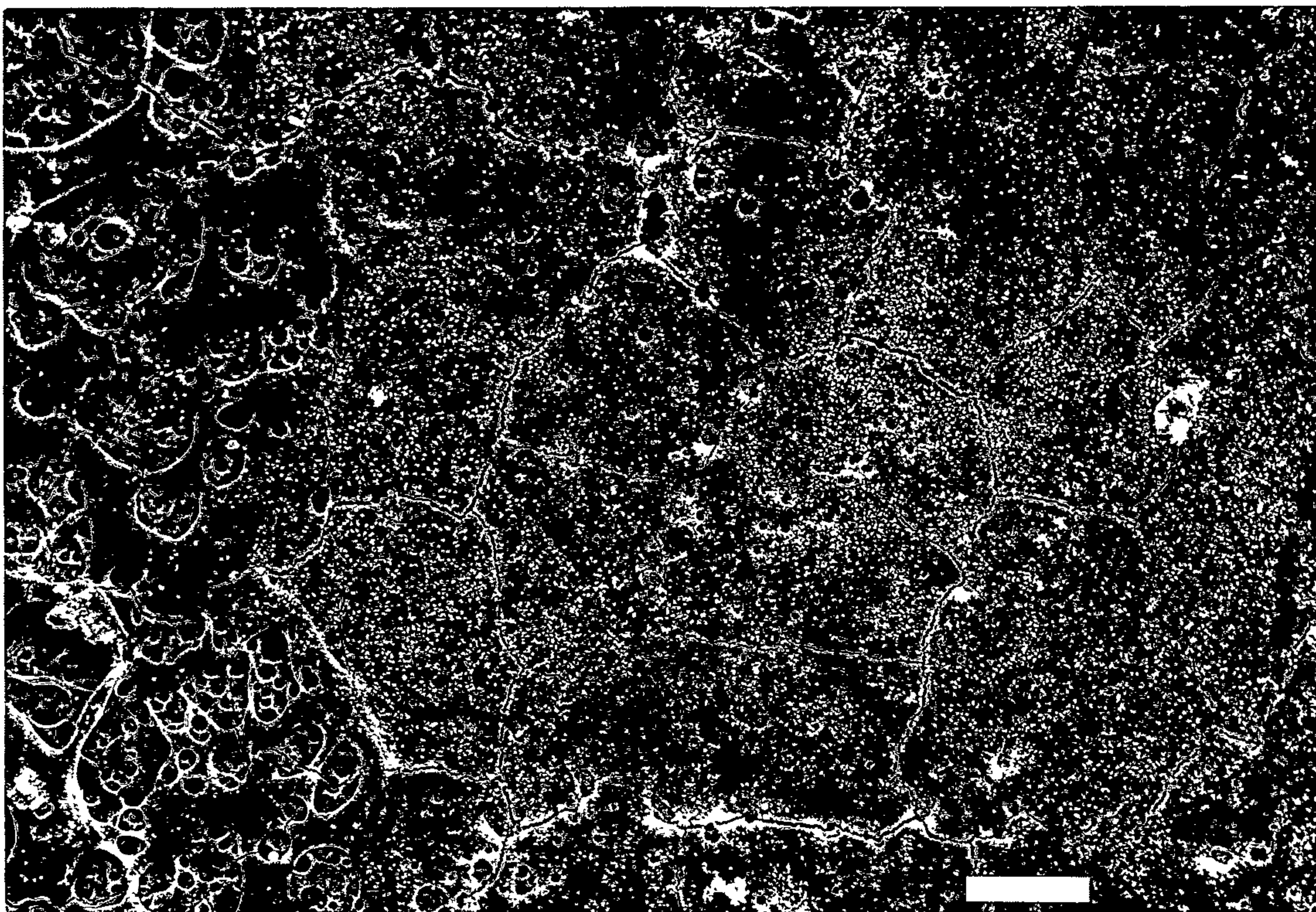
Bar = 10 μm

Figure 6-8 Absence of surface cracking associated with small pitting of the surface.

Energy density = 62.2 J/cm²

50 : 50 water : initiator

Bar = 1 μm



densities the disks appear "softer" with no absolute changes visible in the enamel (Fig 6-5). With increasing energy, the impact disks are more defined and subsequent surface changes are more prominent (Fig 6-6). Consequently, a number of alterations was observed at varying energy densities which comprised four basic changes.

1. Cracking

Almost all specimens showed fine cracking of approximately 0.1 μm to 0.5 μm in width throughout the entire lased area. The cracking did not extend into the adjacent unlased enamel. It was found in both disk and interdisk areas (Fig 6-7). There was no increase in either the amount of surface cracking or the width of the cracks with increasing energy. However, one sample used with a combination of 50 : 50 water and initiator had areas of pitting without any cracking at all (fig 6-8).

2. Flaking

Some areas of enamel of varying size had actually "flaked off" exposing a roughened undersurface (Fig 6-9). At times, the associated cracking extended below the surface layer into the exposed roughened enamel. Figure 6-10 demonstrates imminent ejection of a piece of enamel lateral to a larger area of exposed enamel. The enamel flaking tended to occur preferentially over the convex ridges, or imbrication lines (Fig 6-11 and 6-12).

Figure 6-9 Enamel flaking.

Areas of enamel of various sizes lifted off from the surface with roughened enamel exposed.

Energy density = 159.2 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

Figure 6-10 Enamel flaking.

Central area of enamel prior to ejection.

Energy density = 37.3 J/cm^2

Bar = $10 \text{ }\mu\text{m}$

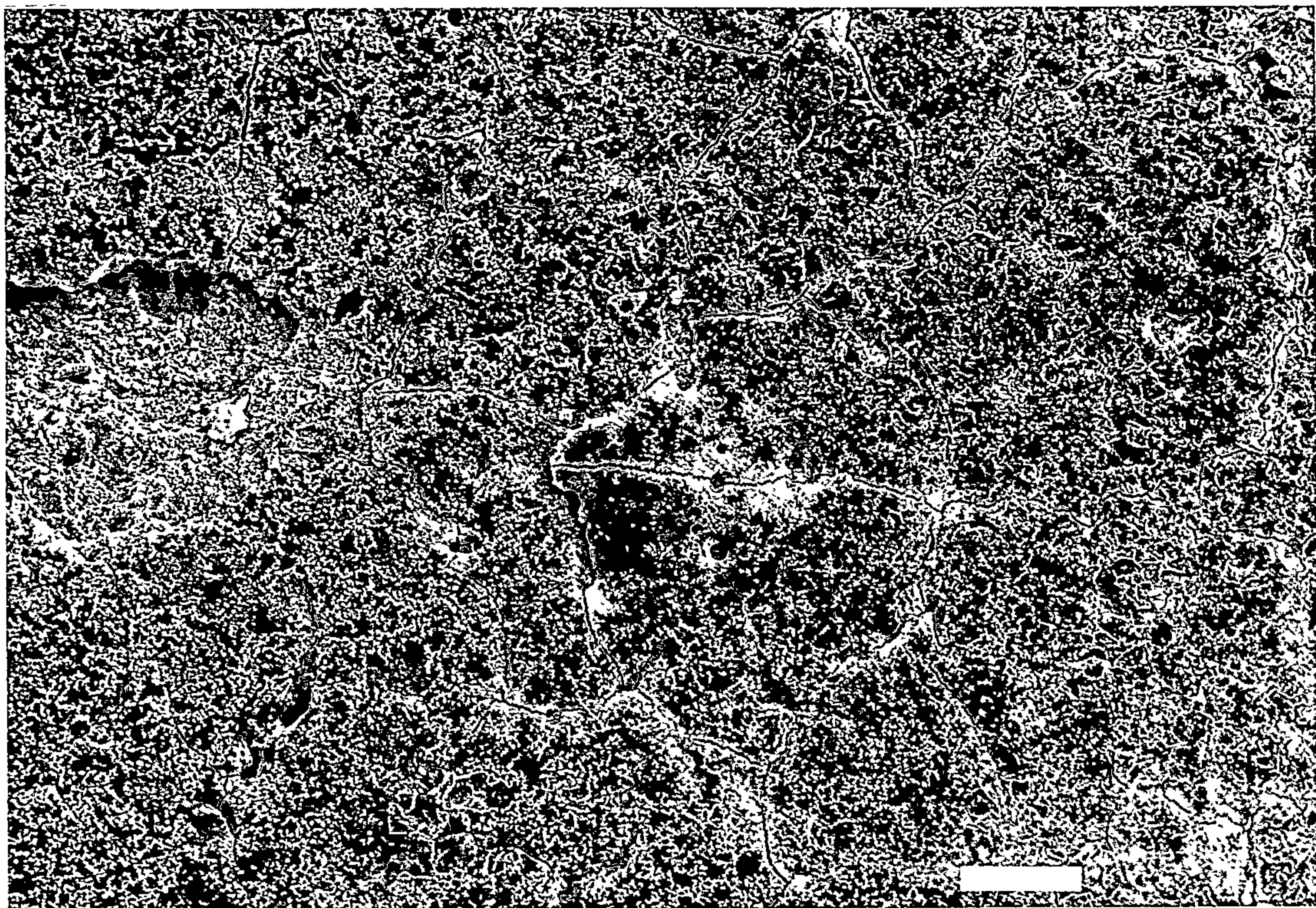
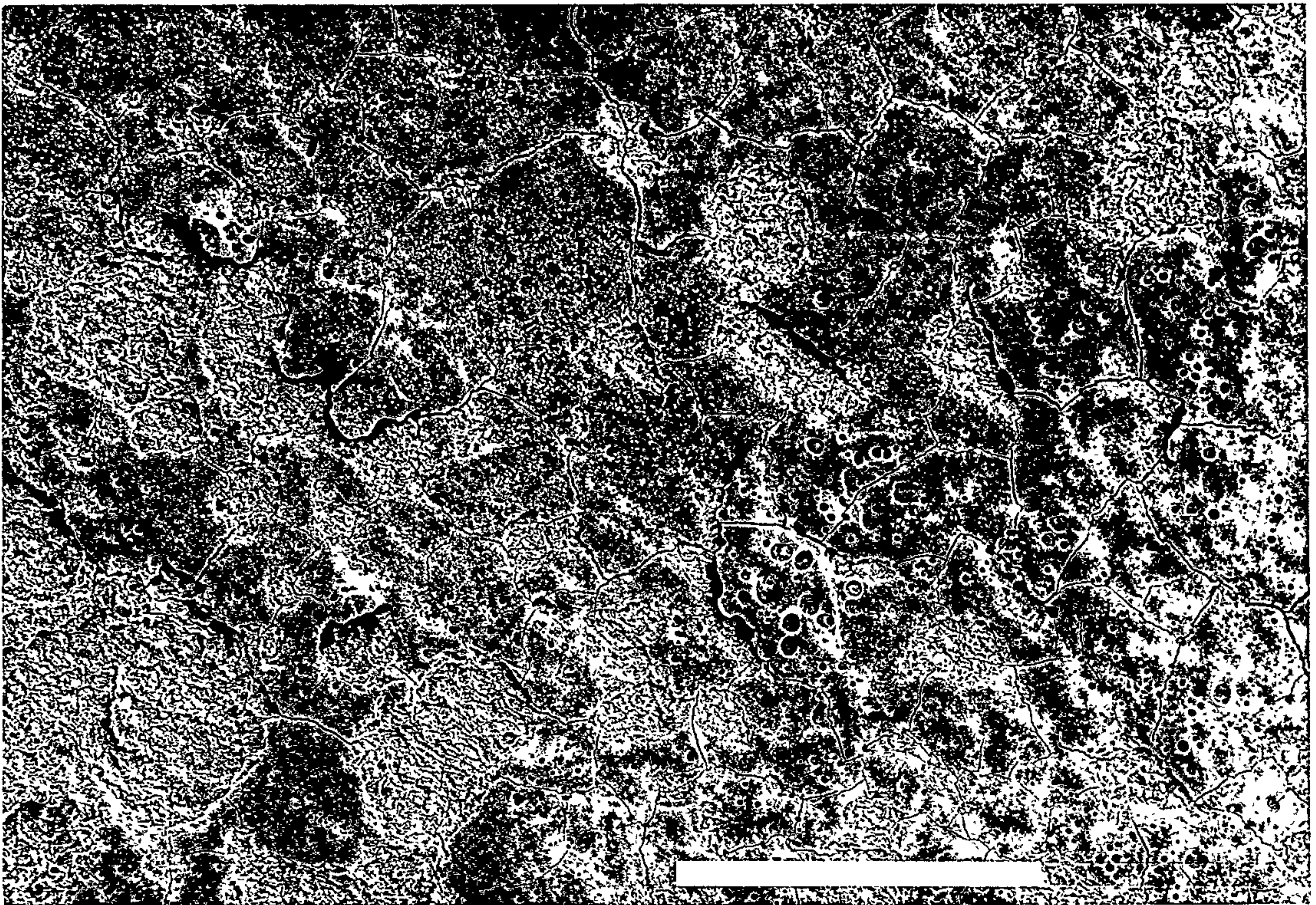


Figure 6-11 Areas of enamel flaking related to imbrication lines. Cratering of the enamel does not seem to follow the same trend.

Energy density = 159.2 J/cm²

Bar = 100 μm

Figure 6-12 Areas of enamel flaking as well as pitting related to imbrication lines.

Energy density = 95.5 J/cm²

Bar = 100 μm

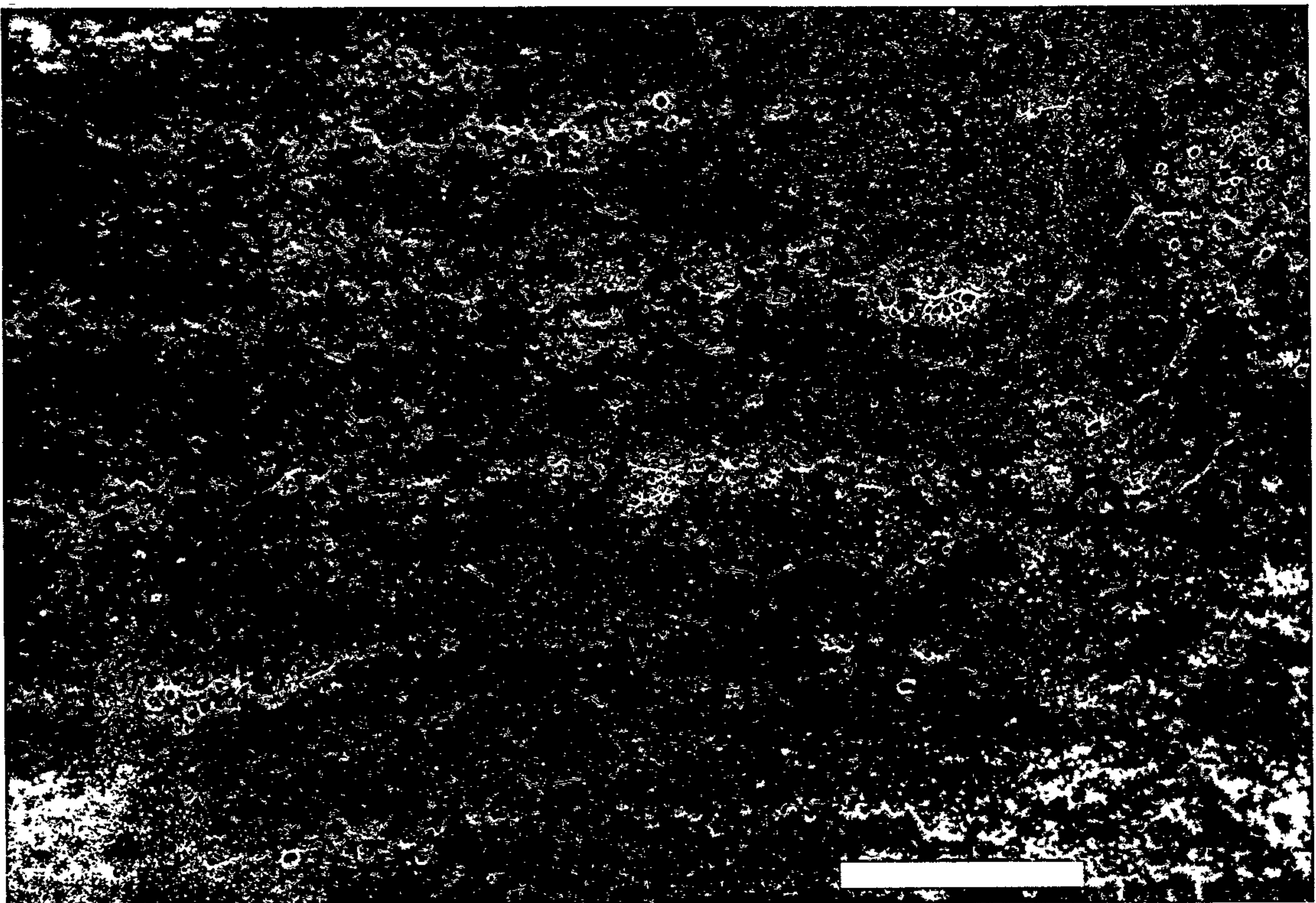
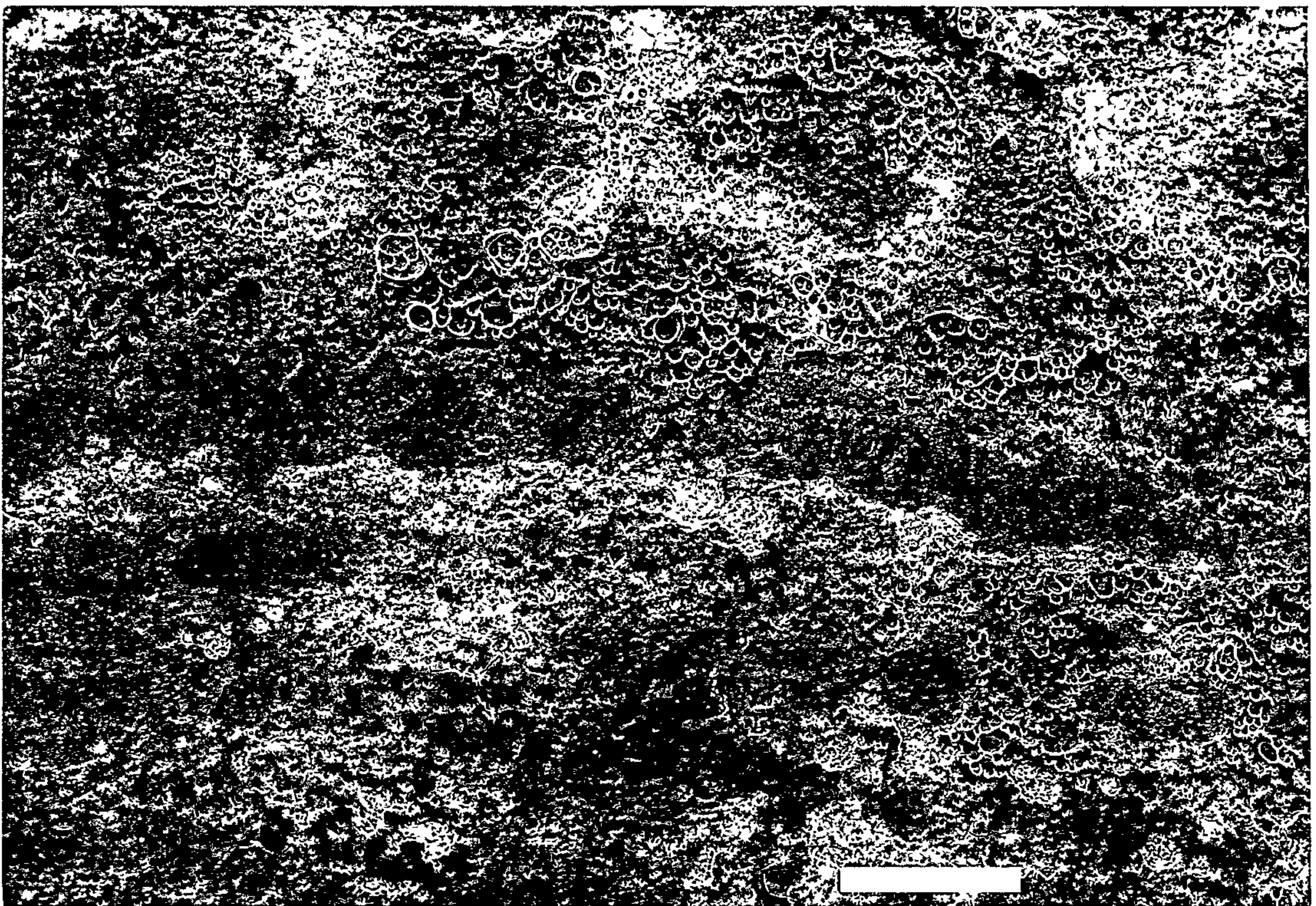


Figure 6-13 Pitting.

Sparsely distributed and relatively small in size.

Energy density = 62.2 J/cm²

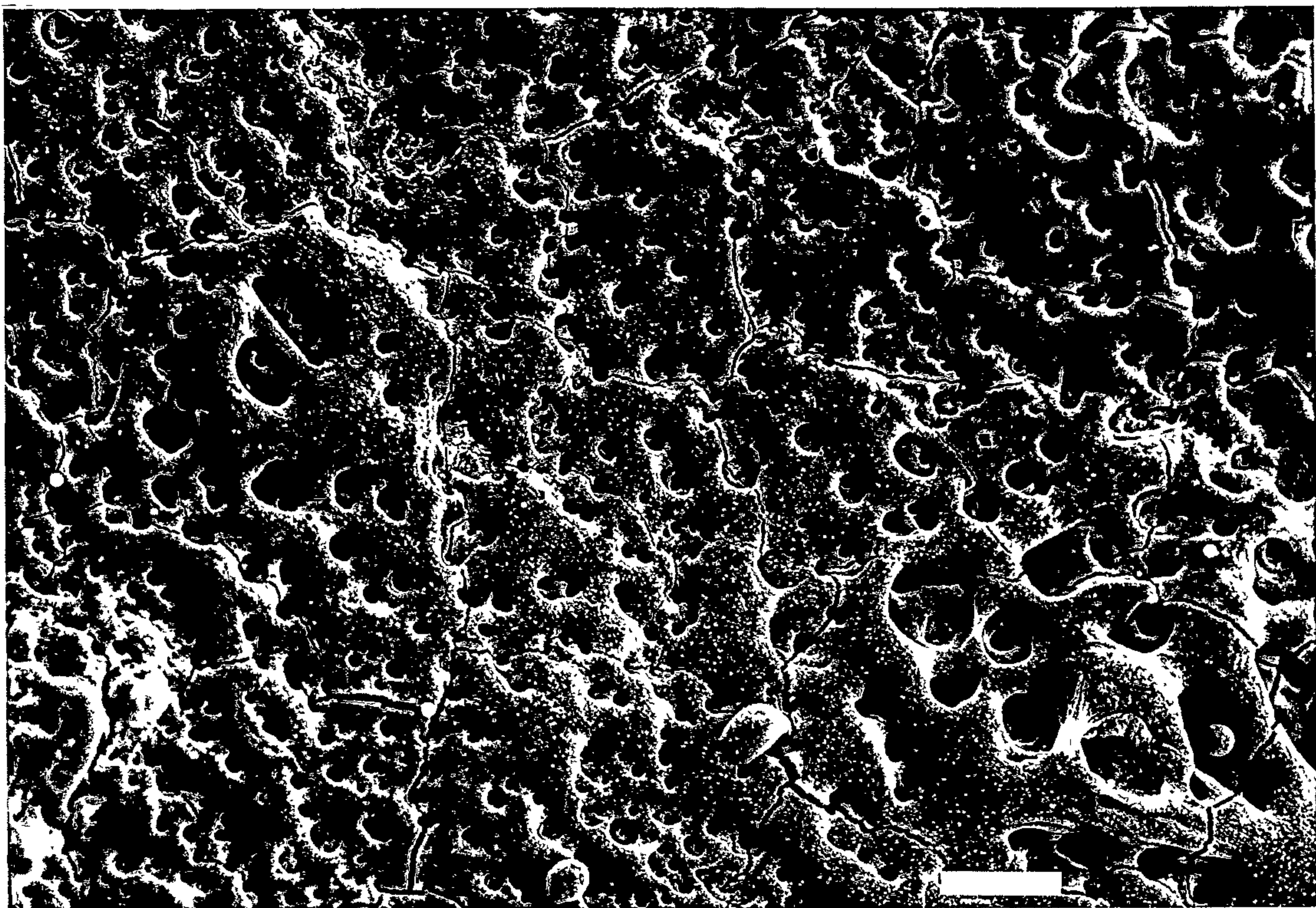
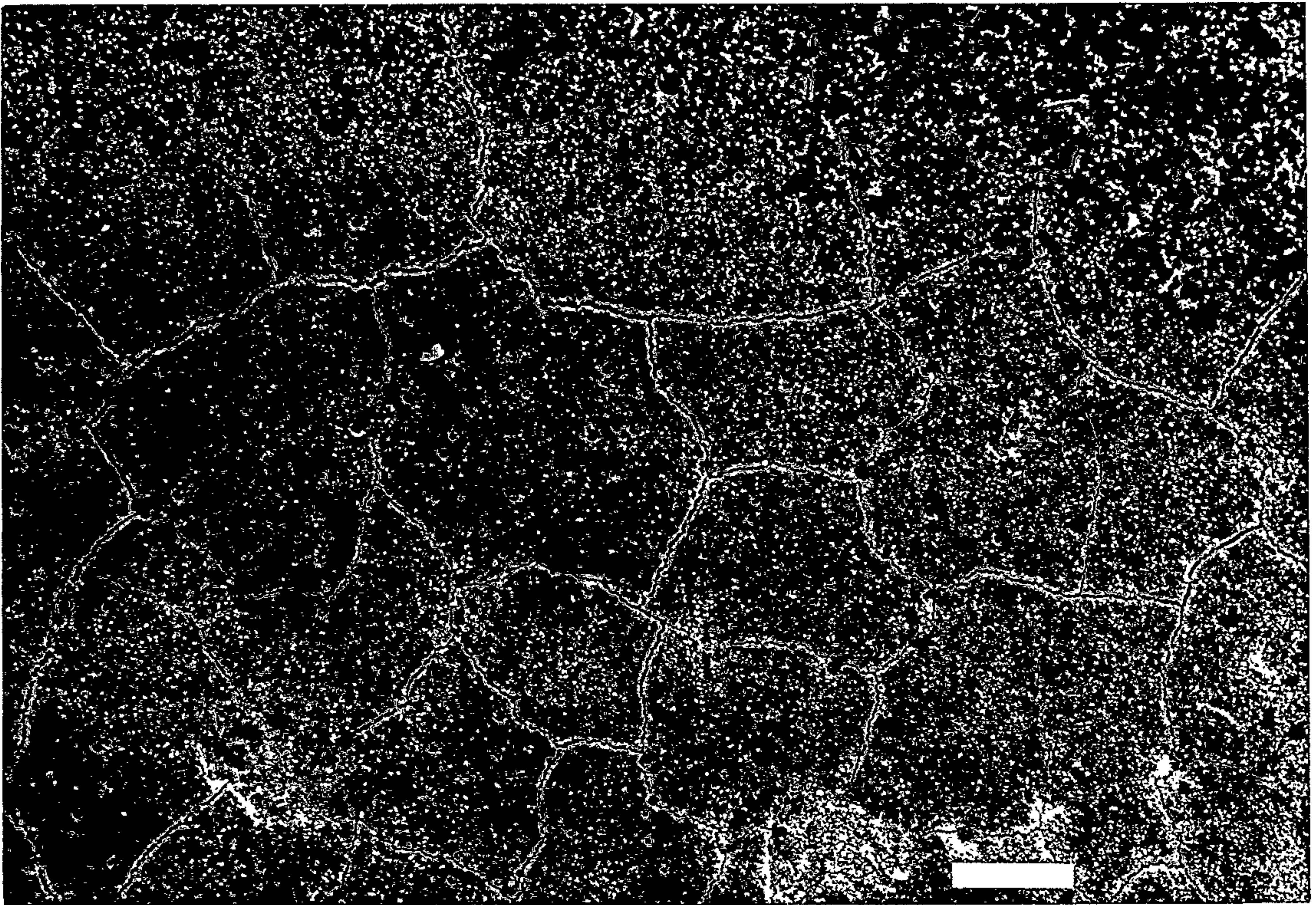
Bar = 10 μm

Figure 6-14 Pitting.

Densely distributed and of varying size. Some coalescing to form cratering is visible in the bottom right.

Energy density = 159.2 J/cm²

Bar = 10 μm



3. Pitting

Small pores of varying widths were found sparsely distributed over a large area at the lower energy densities (Fig 6-13). With increasing energy settings, they tended to become more concentrated in the areas corresponding with the impact craters (Fig 6-14).

4. Cratering

This was the most obvious change and seemed to be energy dependent, although some variation was found between the teeth. Not all teeth responded in exactly the same way, but trends did emerge. At 37.3 J/cm^2 cratering was not present. Between 62.2 J/cm^2 and 95.5 J/cm^2 , some of the pitted areas appeared to be coalescing to form larger loculated regions without a definite pattern or order (Fig 6-15). At 124.3 J/cm^2 and 223.8 J/cm^2 the cratering was less random and more or less localised to the periphery of the impact areas (Fig 6-16). This was even more pronounced at 318.3 J/cm^2 and 572.9 J/cm^2 (Fig 6-17). At the highest energy density, the impacted enamel occasionally took on a more "splattered" look with a much depressed central area (Fig 6-18). Solidification of the once molten enamel in this pattern suggests significant power.

Individually, each disk presented with roughly the same characteristic features (Fig 6-19). The periphery is raised and has the appearance of molten enamel that has "bubbled" and then cooled to form rounded,

Figure 6-15 Random cratering of enamel related to impact sites.

Energy density = 124.3 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

Figure 6-16 Cratering of enamel with a definite peripheral distribution related to the impact craters.

Energy density = 223.8 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

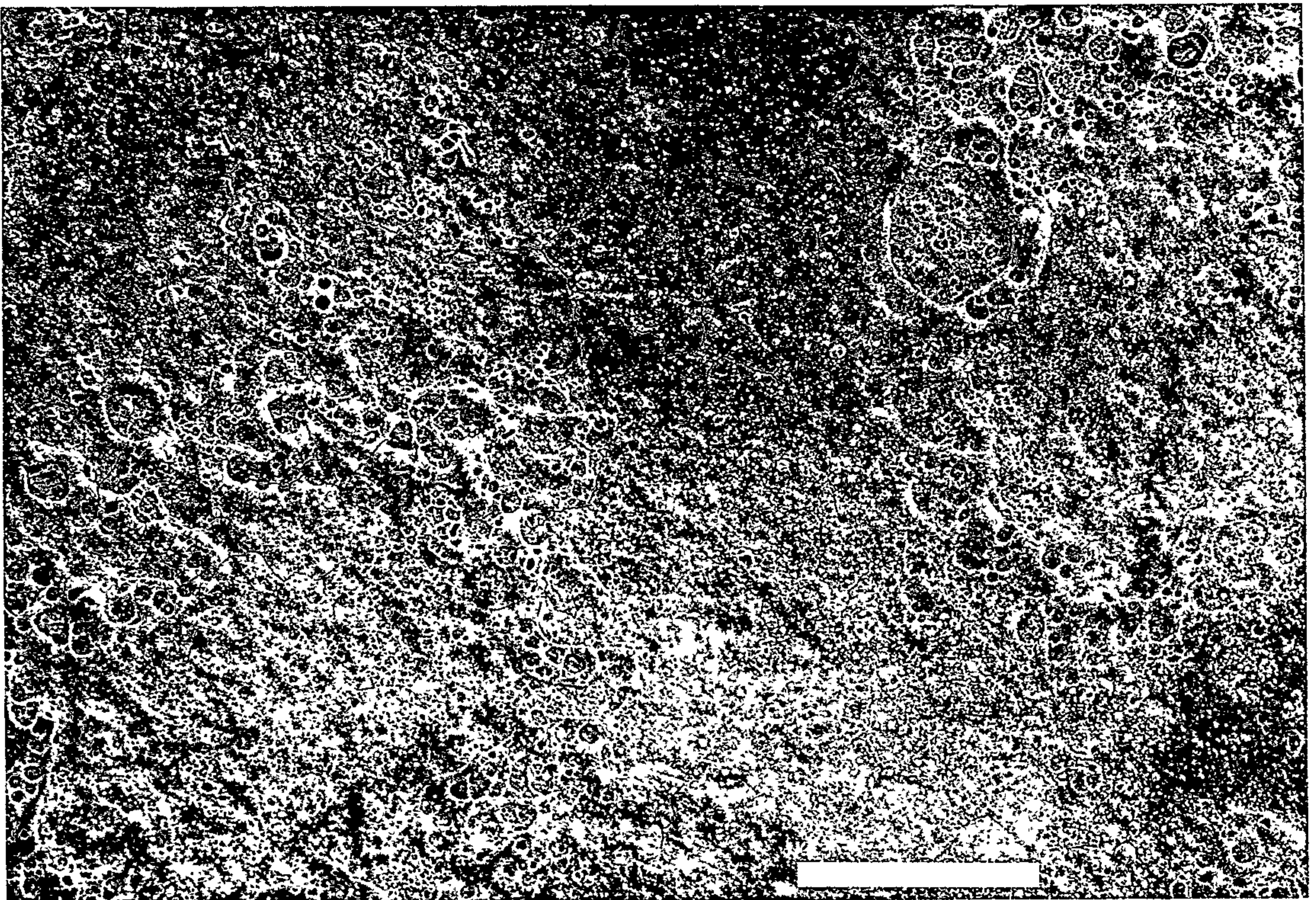
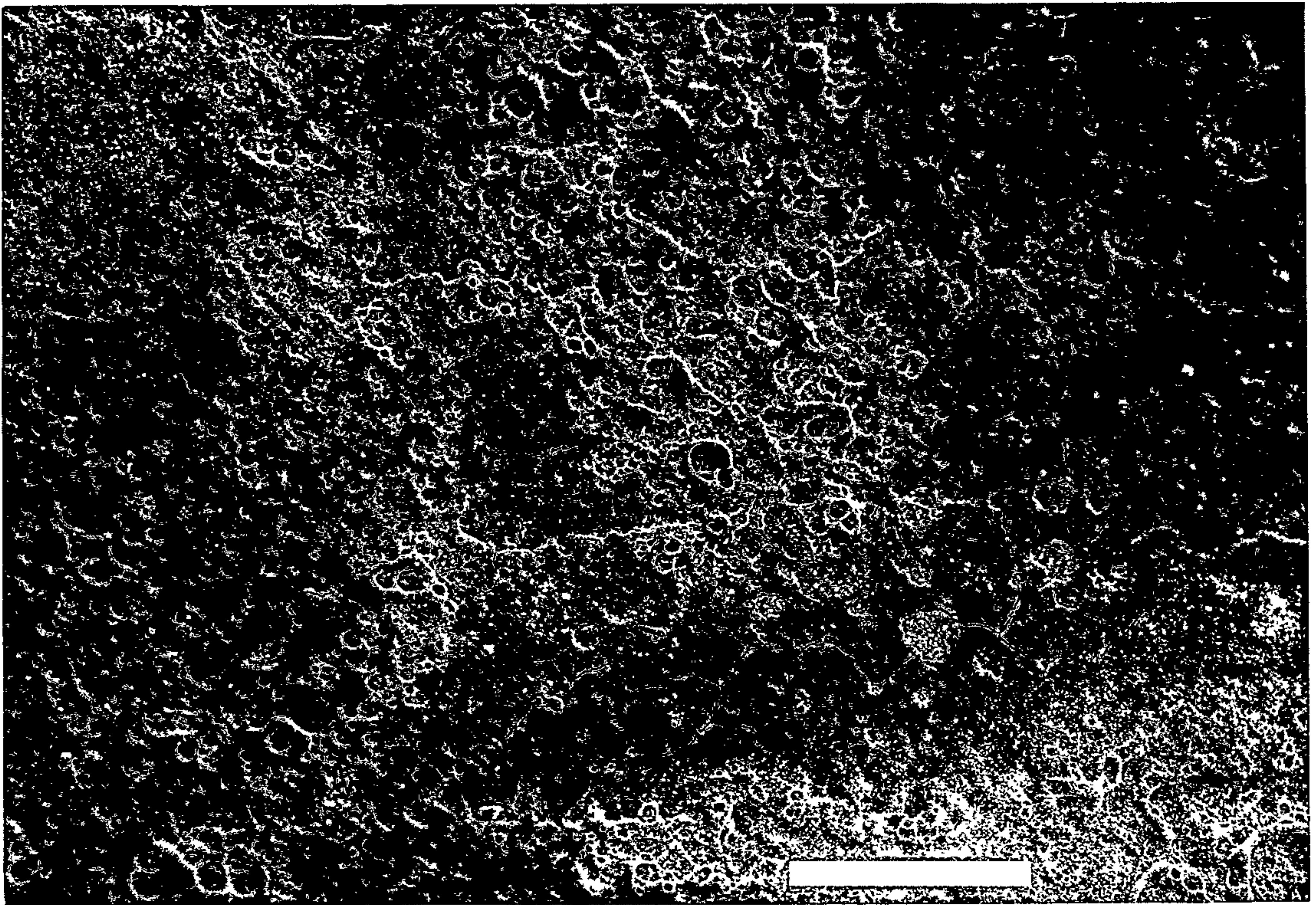


Figure 6-17 Overlapping of impact disks. Peripheral placement of cratering.

Energy density = 318.3 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

Figure 6-18 Unusual impact disk (right) beside a more typical impact disk.

Energy density = 572.9 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

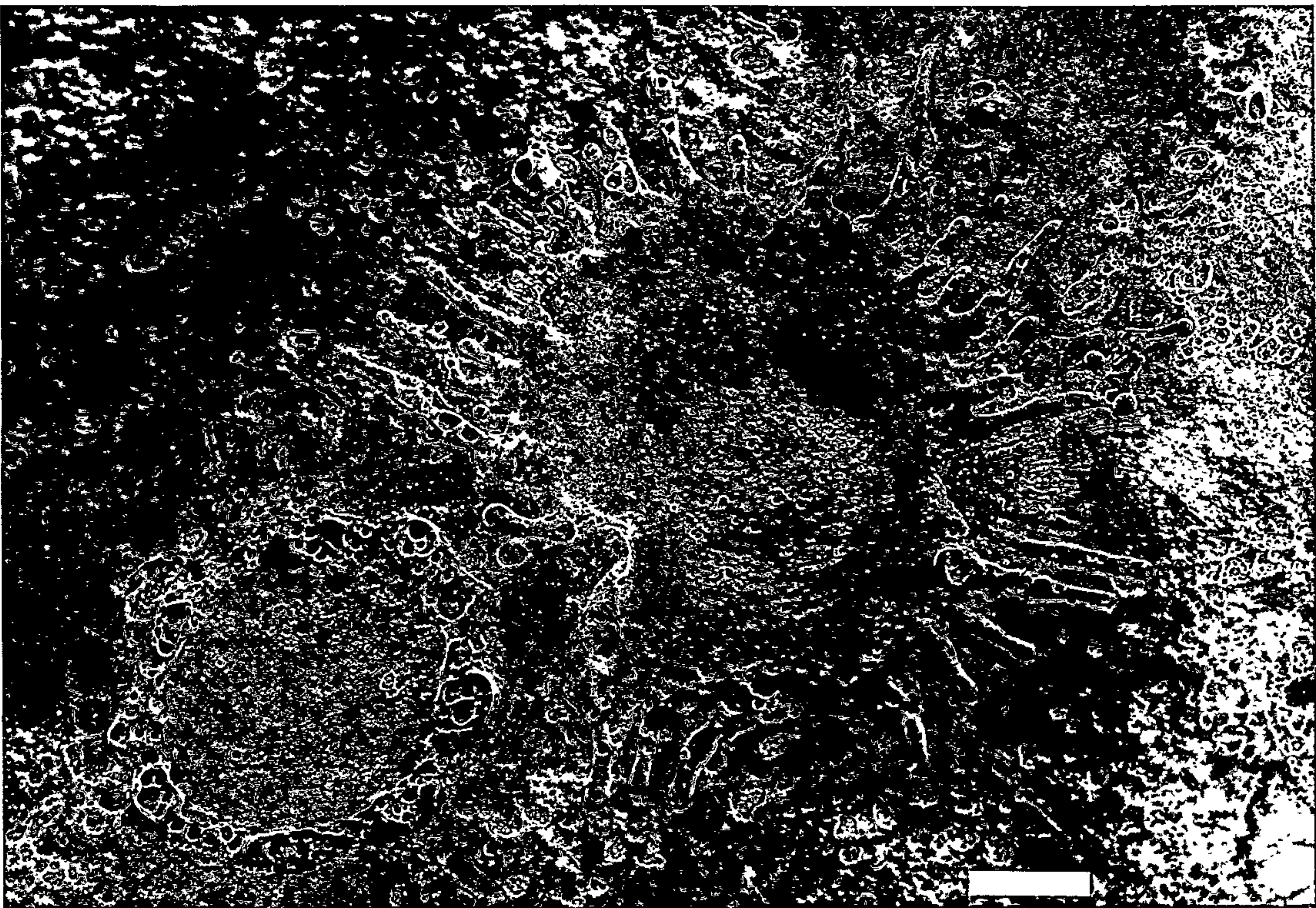
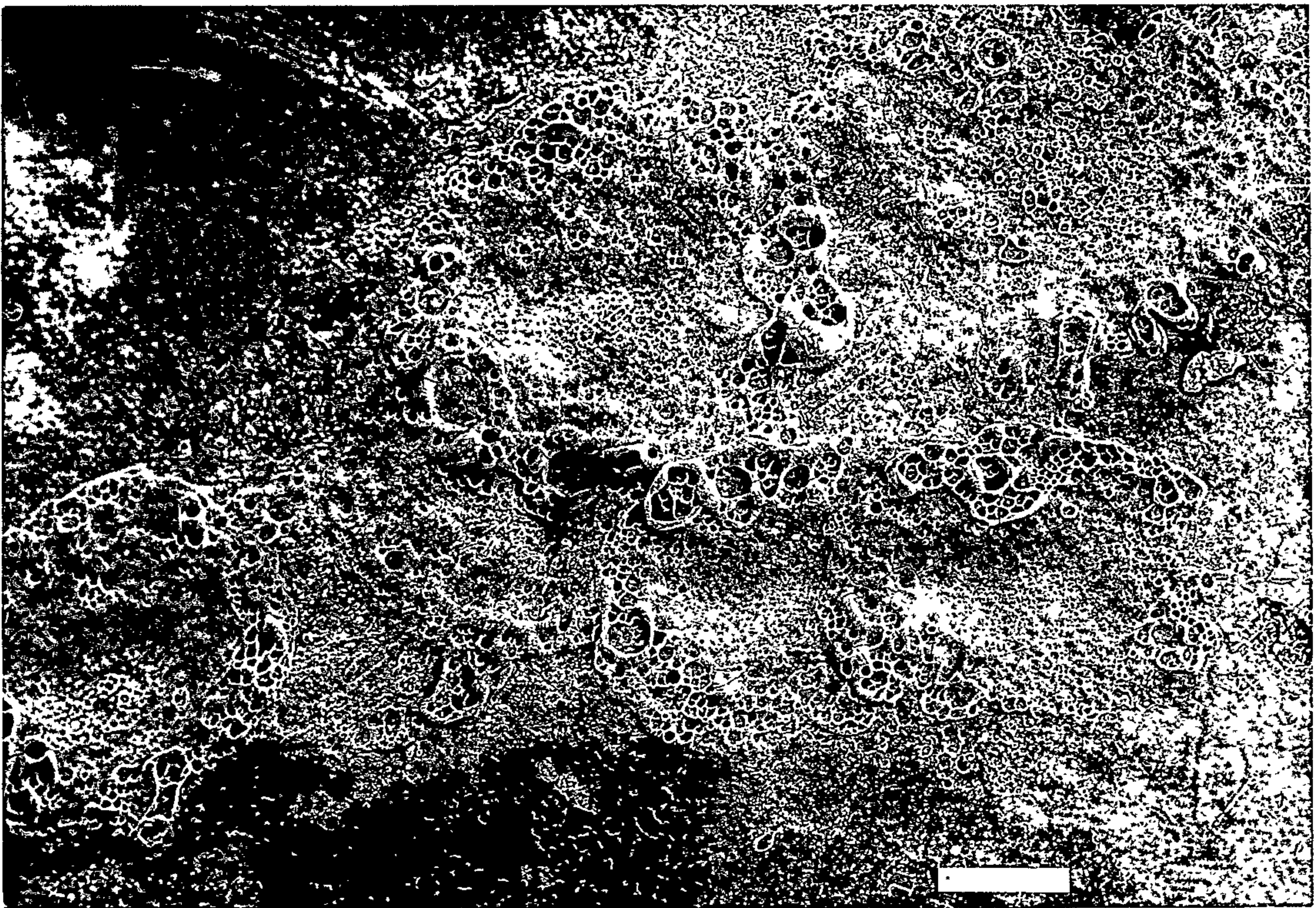


Figure 6-19 Typical impact disk.

Peripheral cratering is well demarcated from the interdisk enamel. Smaller pitting is present centrally.

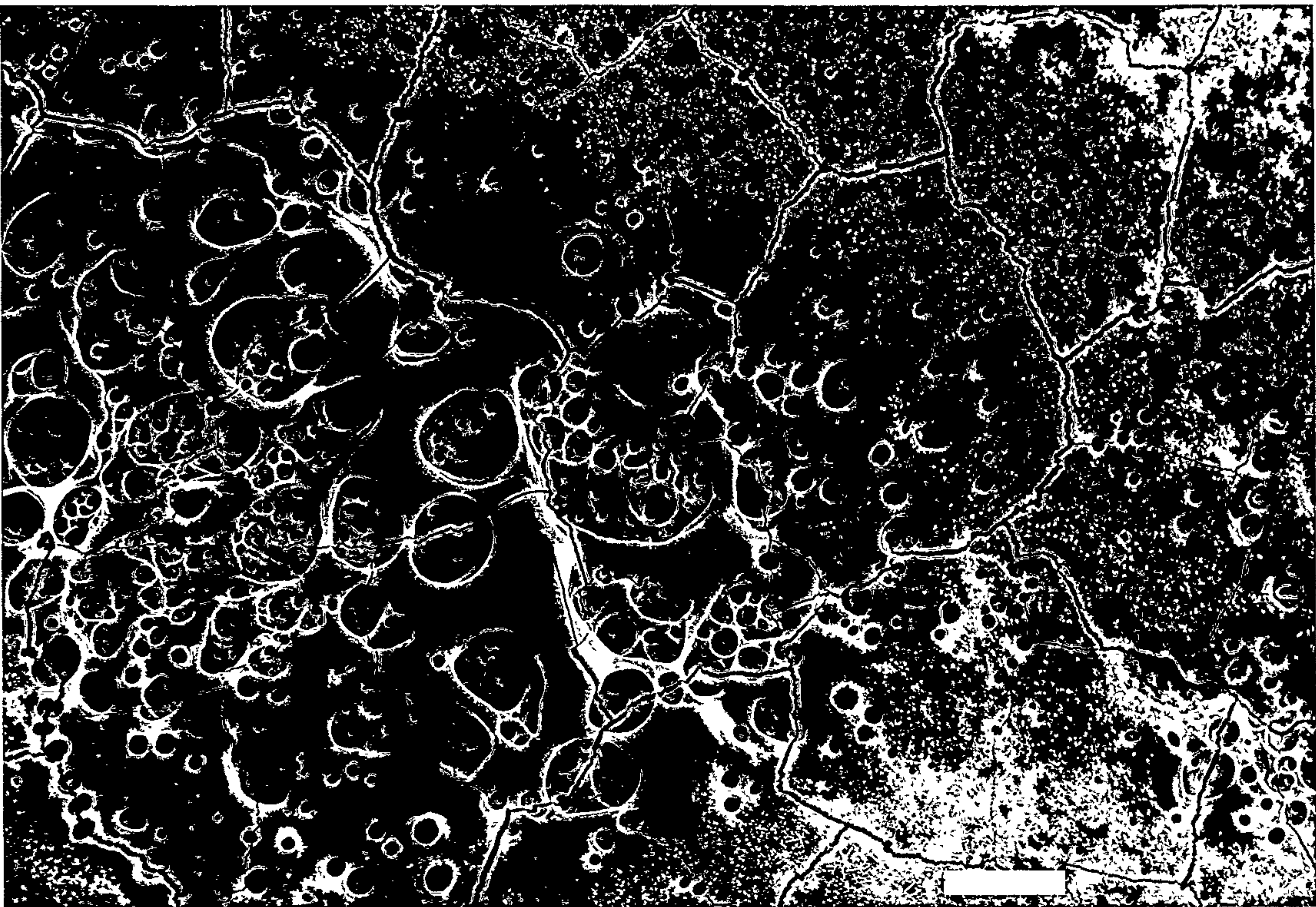
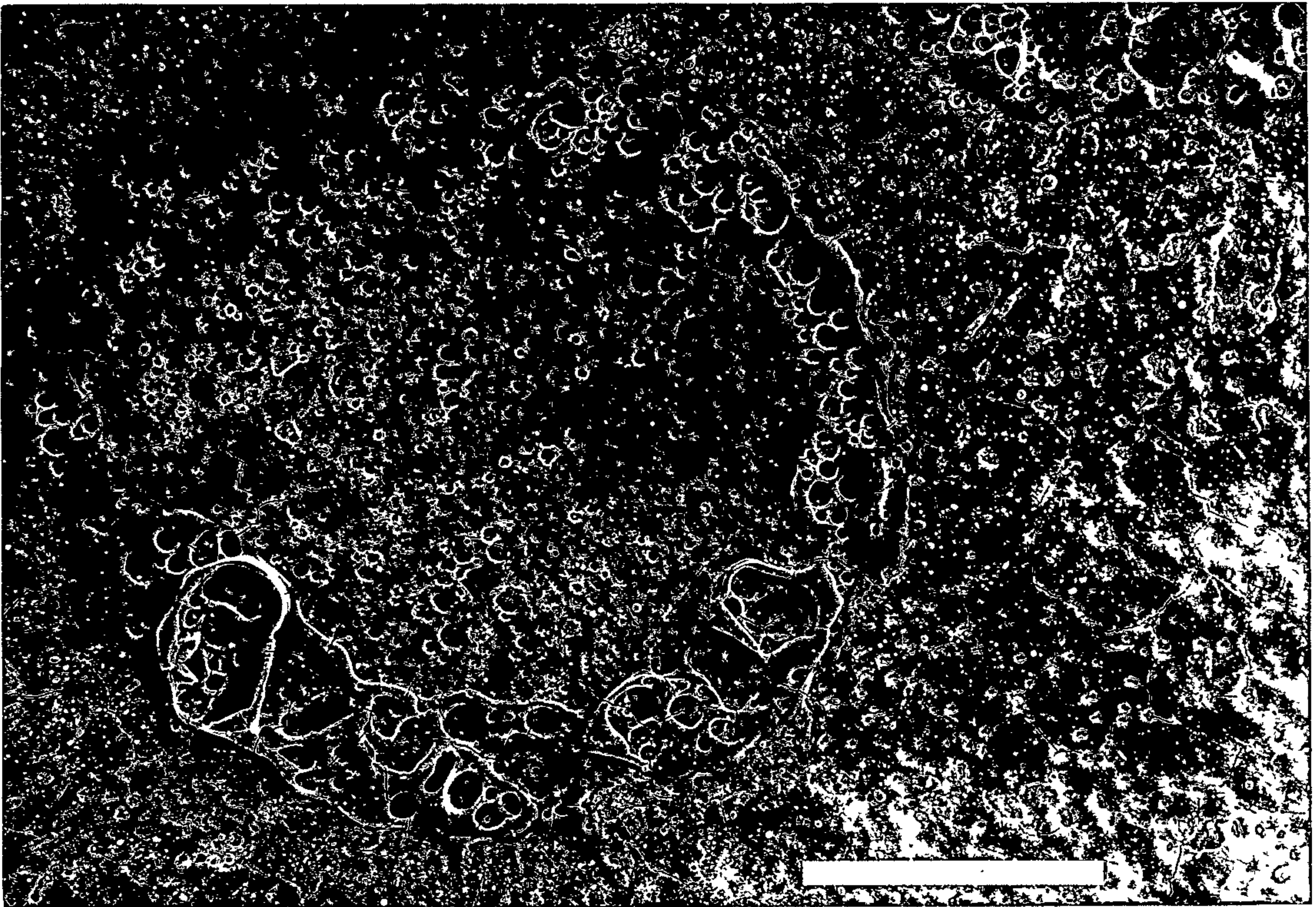
Energy density = 318.3 J/cm^2

Bar = $100 \text{ }\mu\text{m}$

Figure 6-20 Comparison between smooth, glassy surface of disk areas and relatively normal texture of interdisk areas.

Energy density = 223.8 J/cm^2

Bar = $10 \text{ }\mu\text{m}$



glassy craters of widely varying size. These multilocular formations abruptly terminate at the outer edge of the disk and diminish towards the centre to much smaller "pitting". Figure 6-20 shows the contrast of the smooth, once molten peripheral enamel with the enamel between the impact areas.

6.2.2 Pulse rate

Varying the pulse rate, if the energy density remained constant, did not alter the changes induced in the enamel surface but it did alter the density or distribution of the impact craters. Comparison between the same energy density at twice the pulse rate shows an increased overlapping of separate disks (Figs 6-21 and 6-22). This provides an increase in the proportion of disk to interdisk areas, and, consequently, a relative increase of cratered enamel compared to cracked and pitted and rough exposed enamel after flaking. The procedure would be twice as fast because the same energy can be imparted in half the time for the same coverage or disk density.

6.2.3 Initiator

Varying the amount of initiator had little effect on the surface alterations created except at the two extremes. Without initiator, there is no detectable change in the enamel surface either clinically or microscopically. Figures 6-23 and 6-24 show a comparison between unlased enamel and lased but uninitiated enamel. When pre-

Figure 6-21 Low pulse rate showing sparse distribution of impact disks with cratering visible in each.

Energy density = 159.2 J/cm^2

Pulse rate = 15 pps

Bar = 1 mm

Figure 6-22 High pulse rate with greater overlapping of disks to give a more uniform coverage of cratering.

Energy density = 159.2 J/cm^2

Pulse rate = 30 pps

Bar = 1 mm

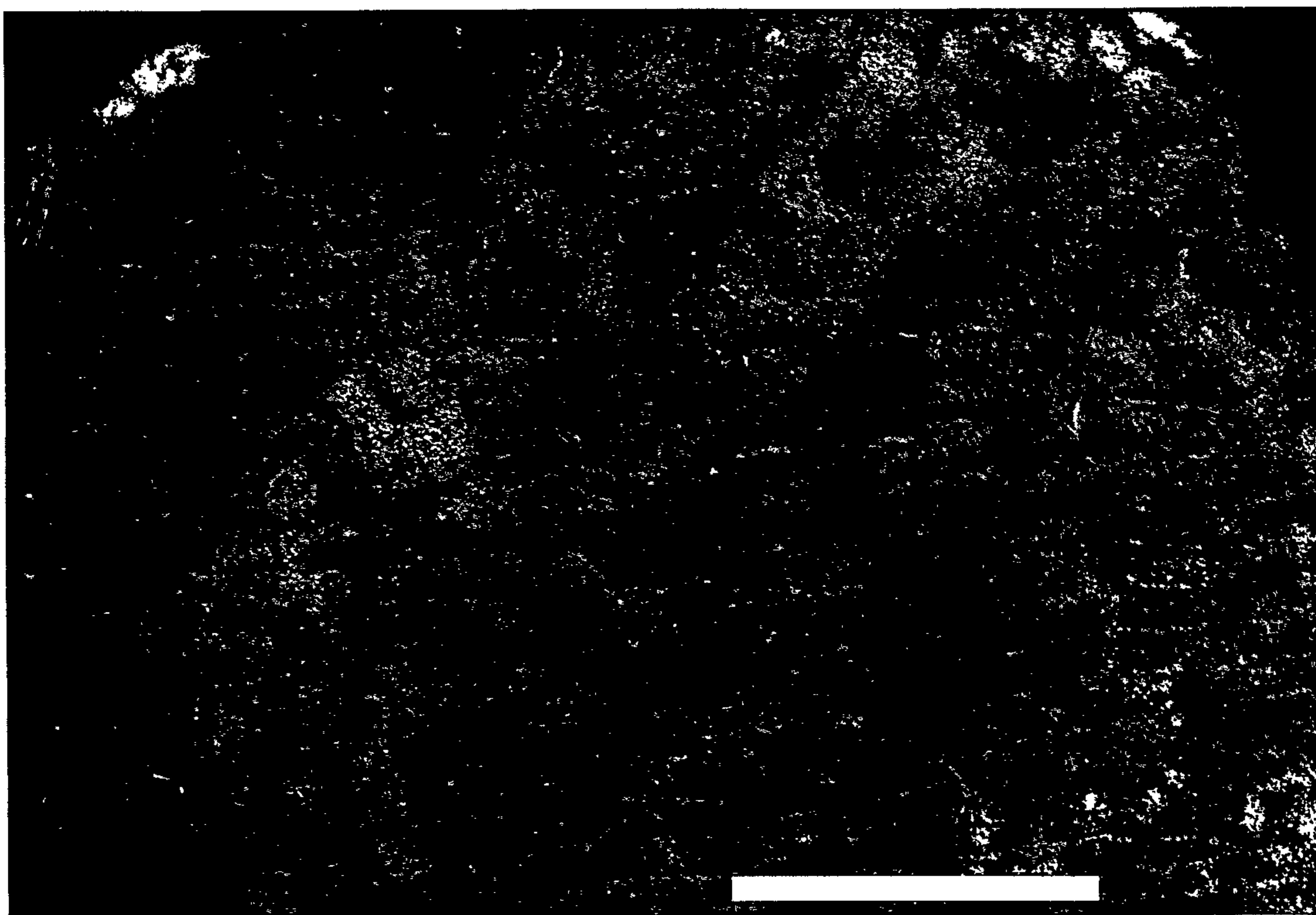
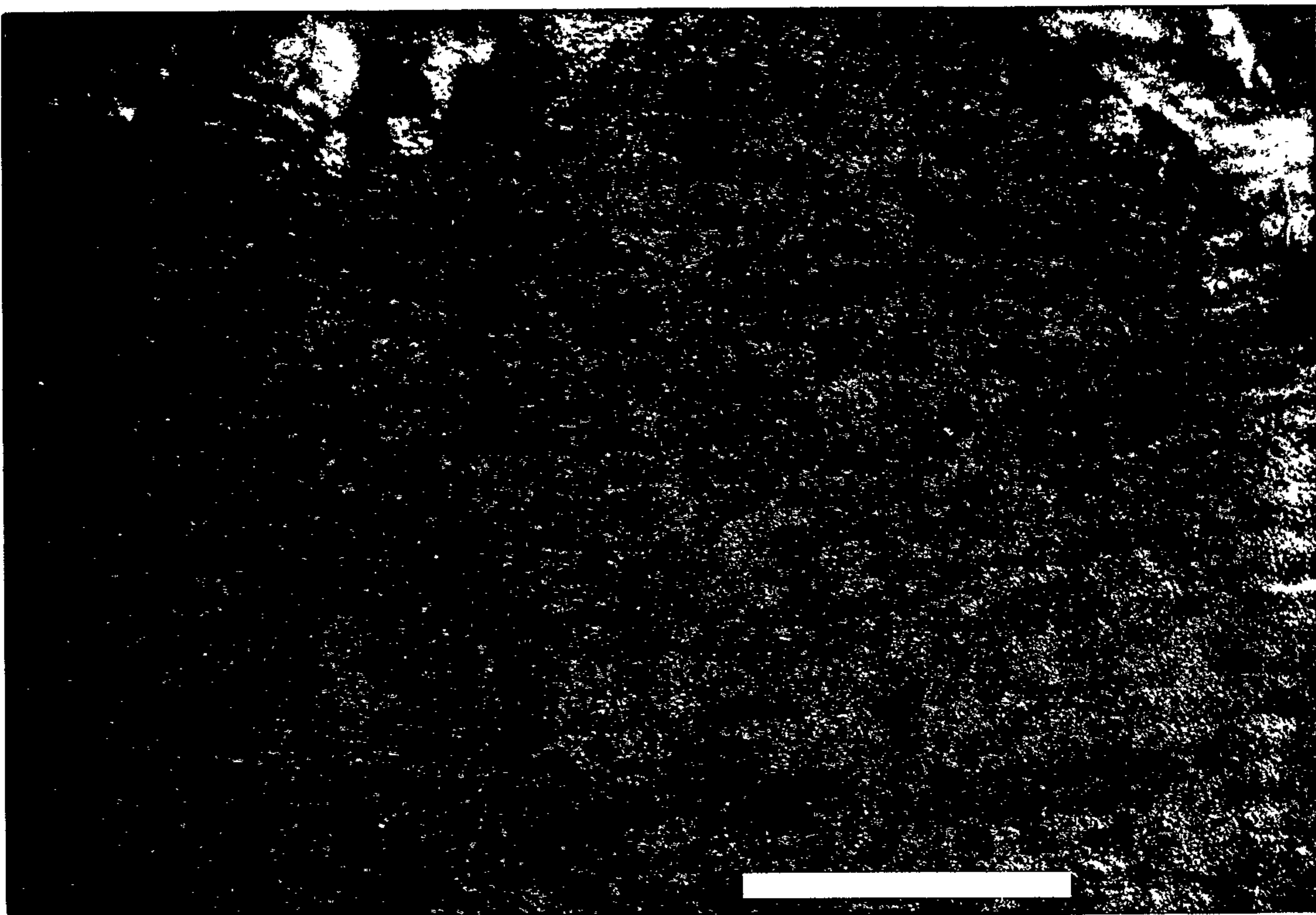


Figure 6-23 Unlased enamel surface.

Bar = 10 μm

Figure 6-24 Lased but uninitiated enamel.

Energy density = 62.2 J/cm²

Bar = 10 μm

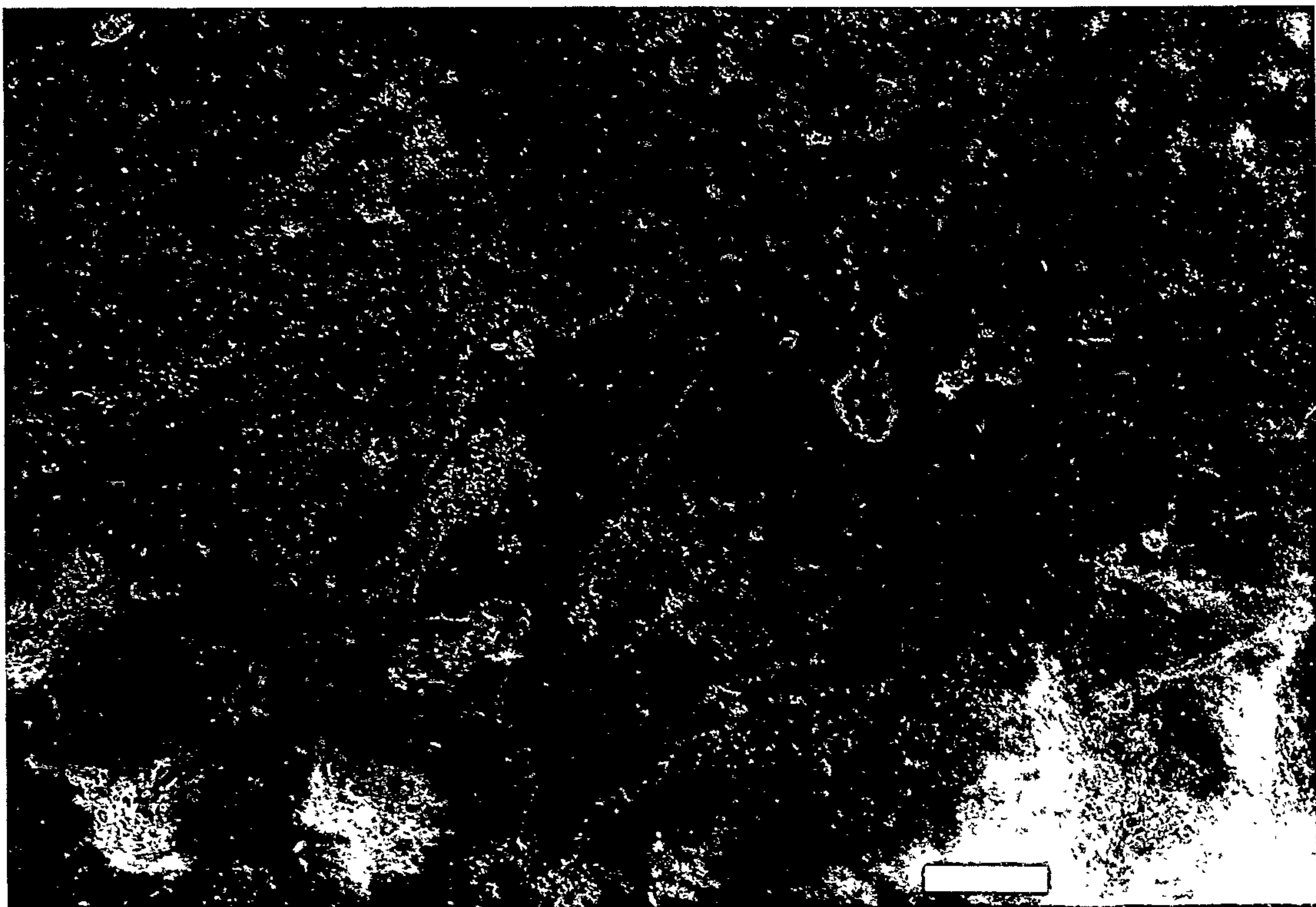
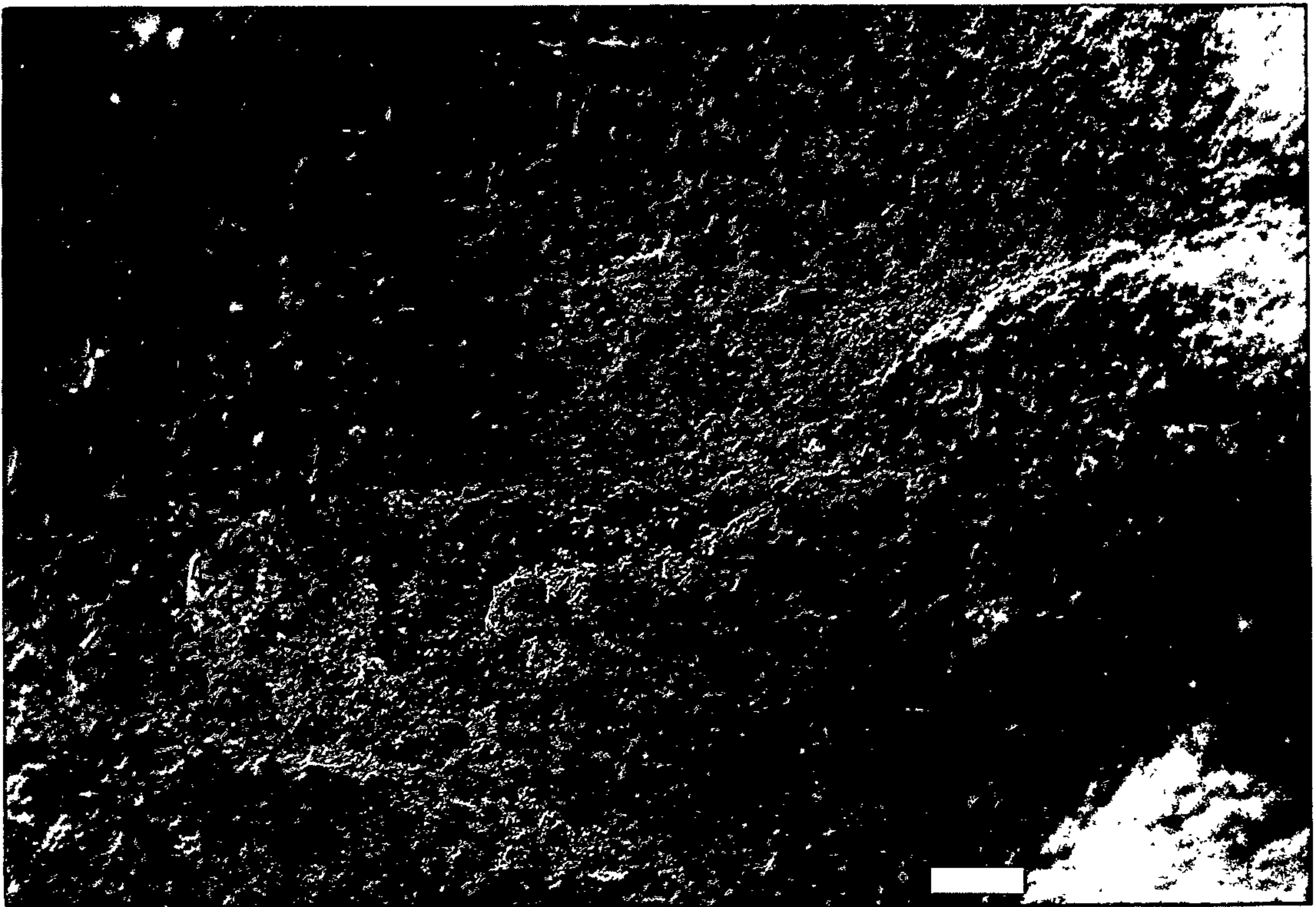


Figure 6-25 0.5 μL initiator.

Reduced surface alterations and increased surface debris.

Energy density = 62.2 J/cm^2

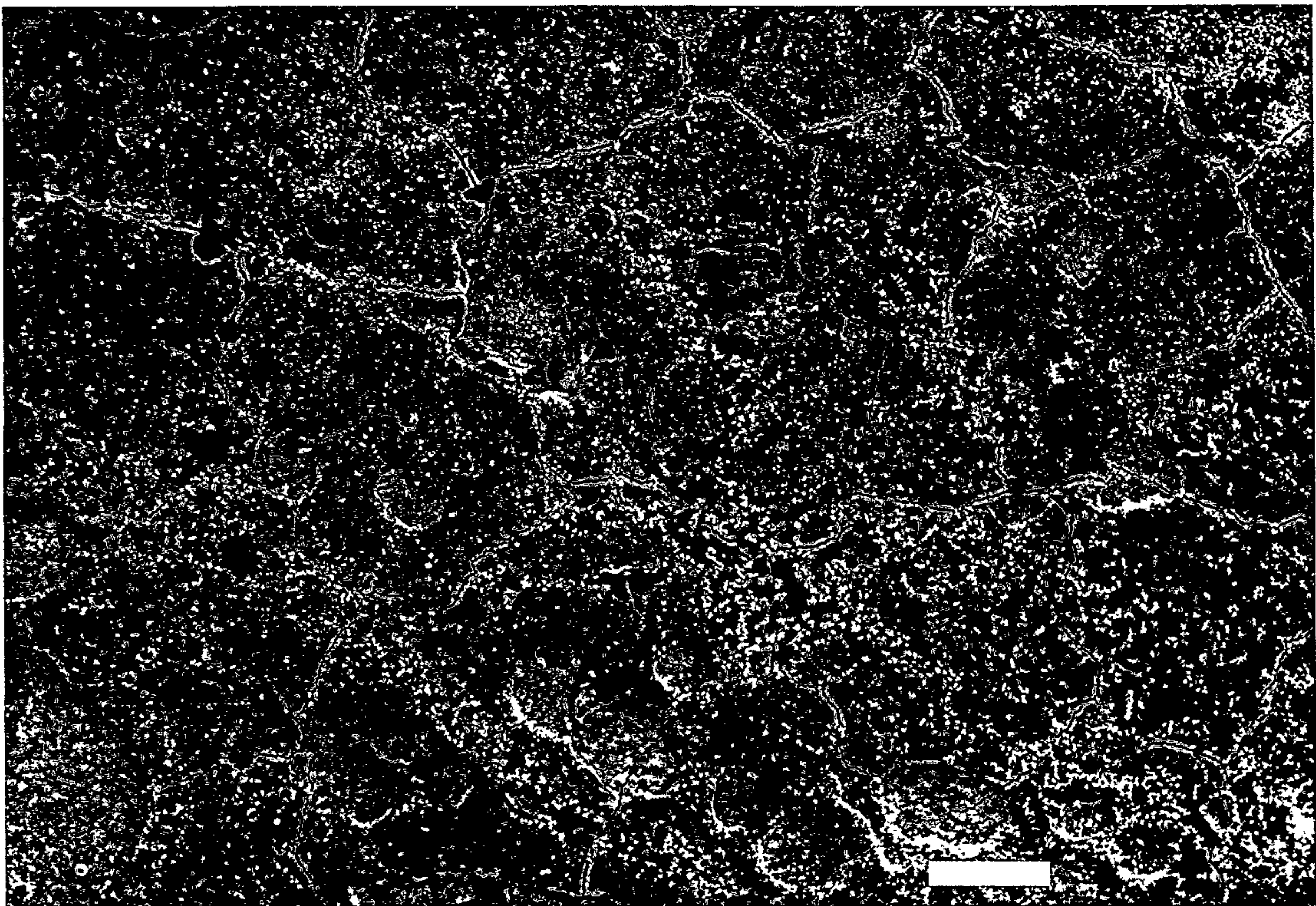
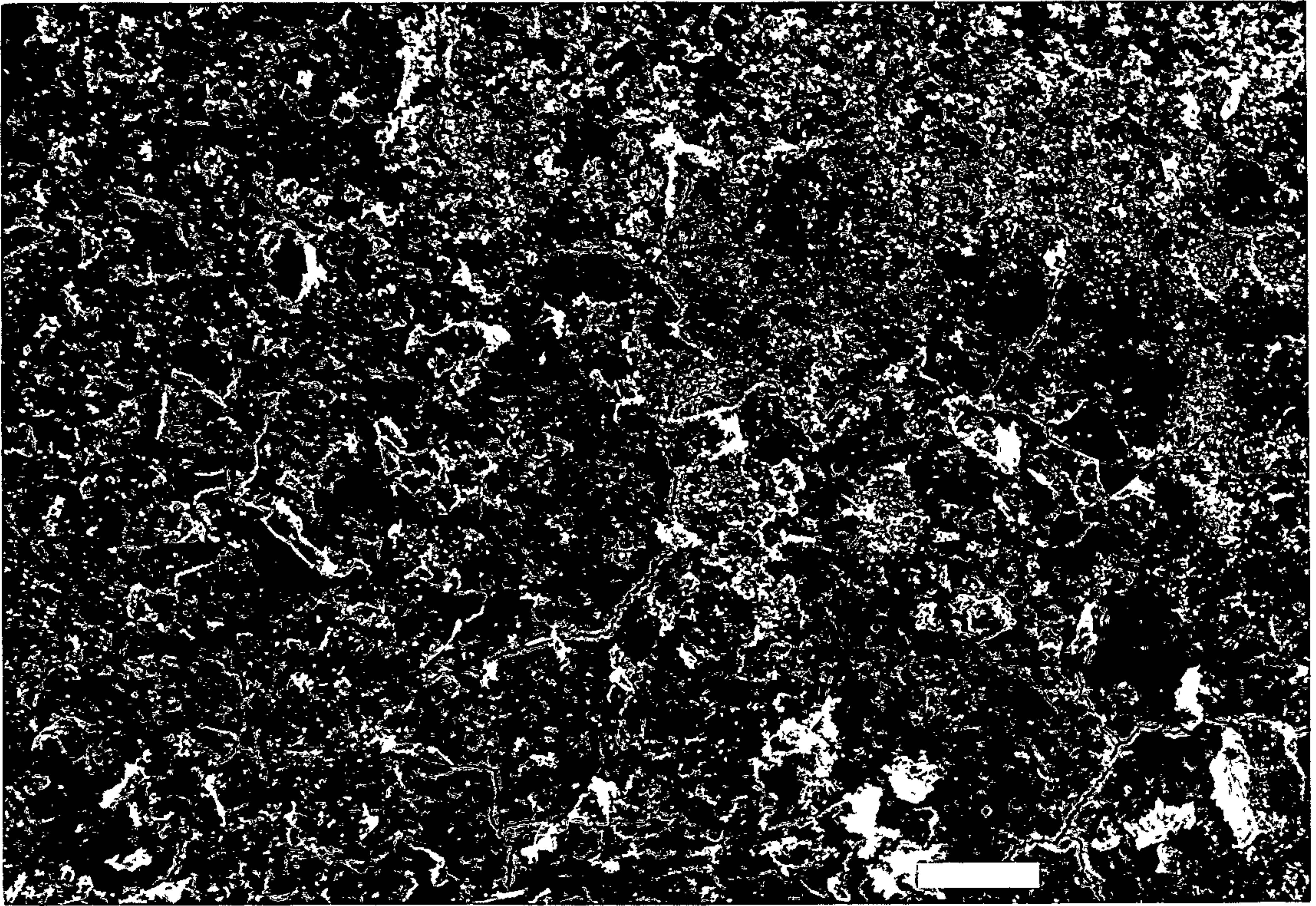
Bar = $10 \mu\text{m}$

Figure 6-26 50 : 50 water and initiator.

Cracking and flaking of enamel consistent with other findings at that energy density.

Energy density = 62.2 J/cm^2

Bar = $10 \mu\text{m}$



determined excessive amounts of initiator were used (0.1 μL and 0.5 μL), no benefit was gained in increasing the amount of initiator in either the quantity or the quality of the etched surface. In fact, too much initiator tended to hinder laser-induced changes in the enamel. Figure 6-25 shows only fine cracking and pitting with much debris, most likely the initiator, remaining on the surface. A ratio of 50 : 50 water to initiator did not change the induced effects in two of the specimens (Fig 6-26), but as shown previously, one sample displayed areas of pitting without cracking.

6.3 Comparison With Acid-Etch

6.3.1 Surface topography

Figure 6-27 demonstrates the gross appearance of both techniques. Homogenous coverage by the phosphoric acid contrasts with the pockmarked surface produced by the laser. Figure 6-28 shows a common border between acid-etching on the left and laser-etching on the right. The surface cracking is only apparent on the lased side, stopping abruptly where the acid-etching starts. At increasing magnification, details of each can be compared (Figures 6-29 to 6-32). The acid-etched surface presents a spicular appearance with selective dissolution of either enamel rod peripheries or cores. The increased surface area required to increase mechanical interlocking and, therefore, enhance bonding is clearly apparent. On the other hand, the lased surface,

Figure 6-27 Half acid-etched / half laser-etched crown.

Demarcation line separates acid-etching on left and laser-etching on right.

Acid etch = 37% phosphoric acid for 30 seconds

Laser etch = Energy density of 318.3 J/cm²

Bar = 1 mm

Figure 6-28 Boundary between acid-etching and laser-etching.

Cracking and cratering present on the lased side stops abruptly at the acid-etched side.

Acid etch = 37% phosphoric acid for 30 seconds

Laser etch = Energy density of 318.3 J/cm²

Bar = 10 μm

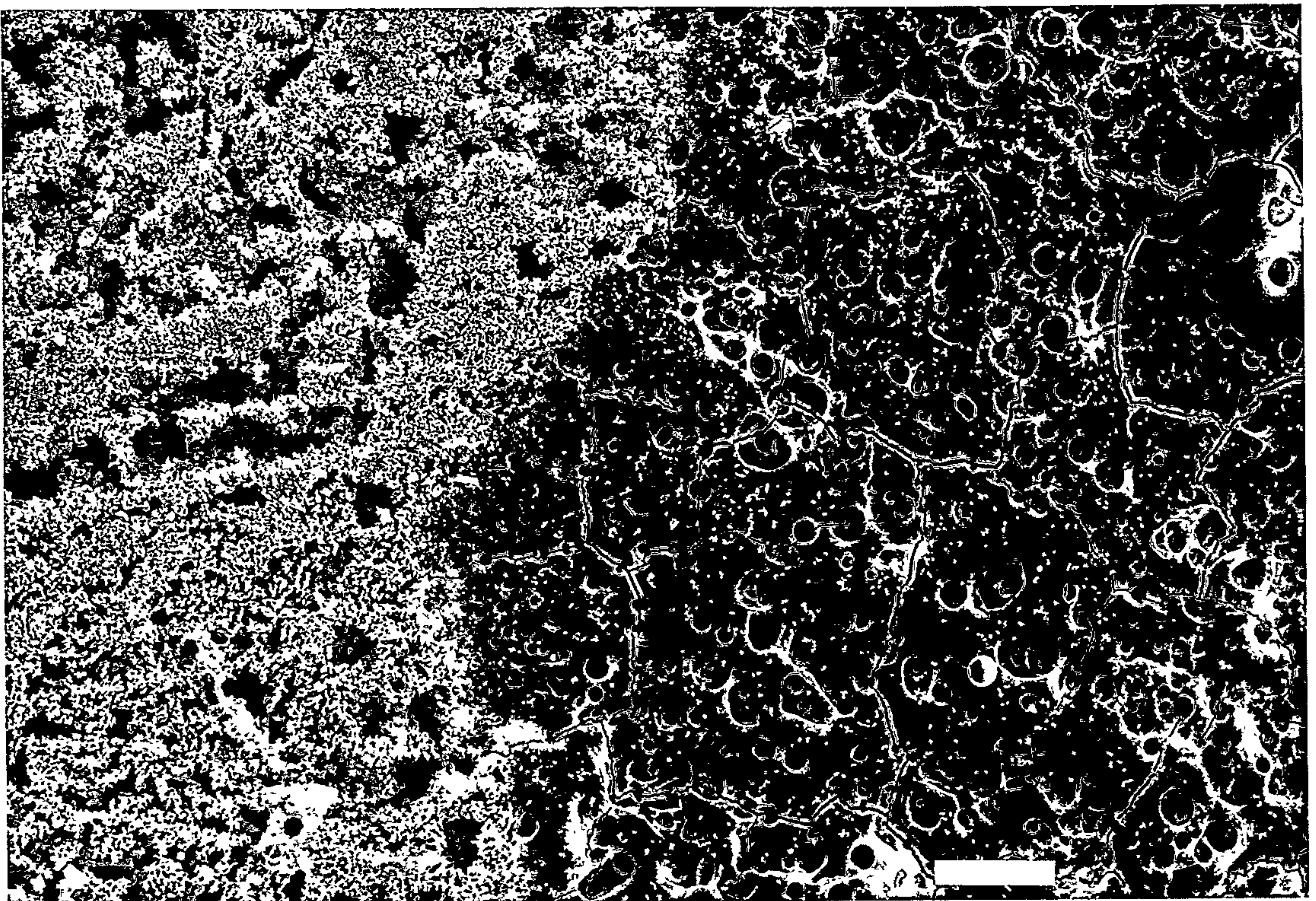


Figure 6-29 Low magnification acid-etch showing spicular appearance with preferential dissolution of enamel rod cores.

Acid etch = 37% phosphoric acid for 30 seconds

Bar = 10 μm

Figure 6-30 Low magnification of laser-etch showing cratering and pitting of enamel with microcracking.

Laser etch = Energy density of 318.3 J/cm²

Bar = 10 μm

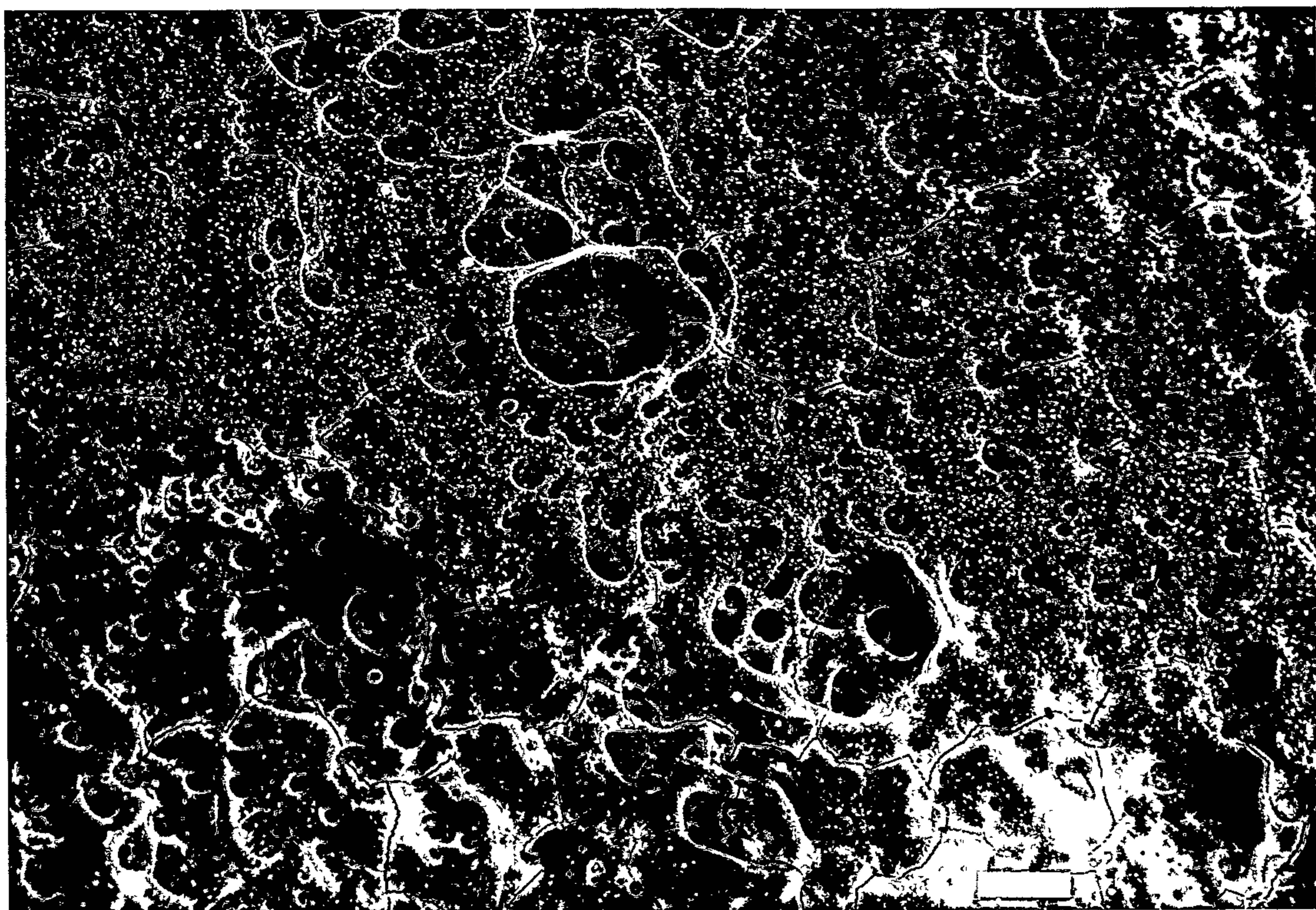
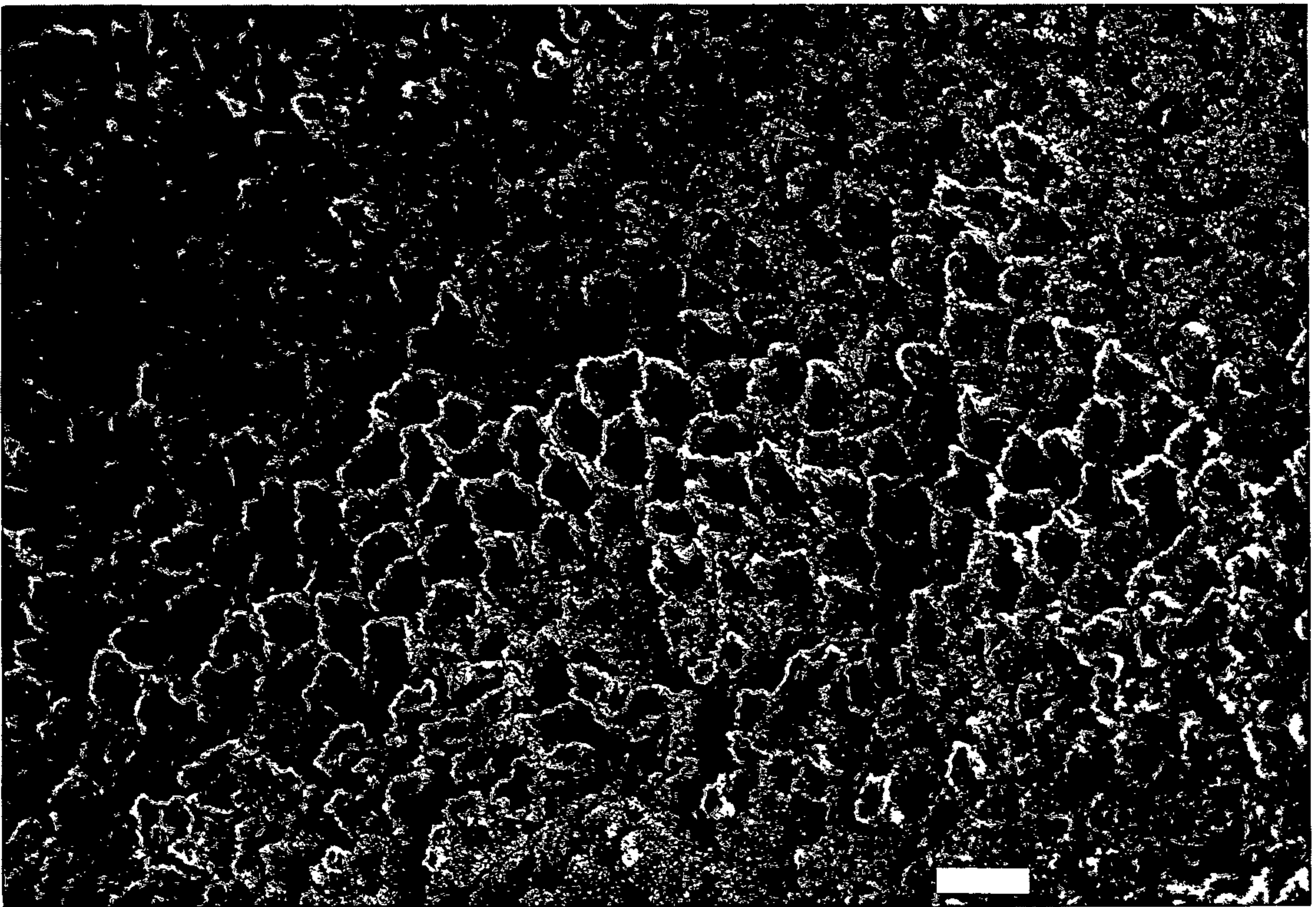


Figure 6-31 High magnification of acid-etch in Figure 6-29.

Bar = 1 μm

Figure 6-32 High magnification of laser-etch in Figure 6-30.

Bar = 1 μm

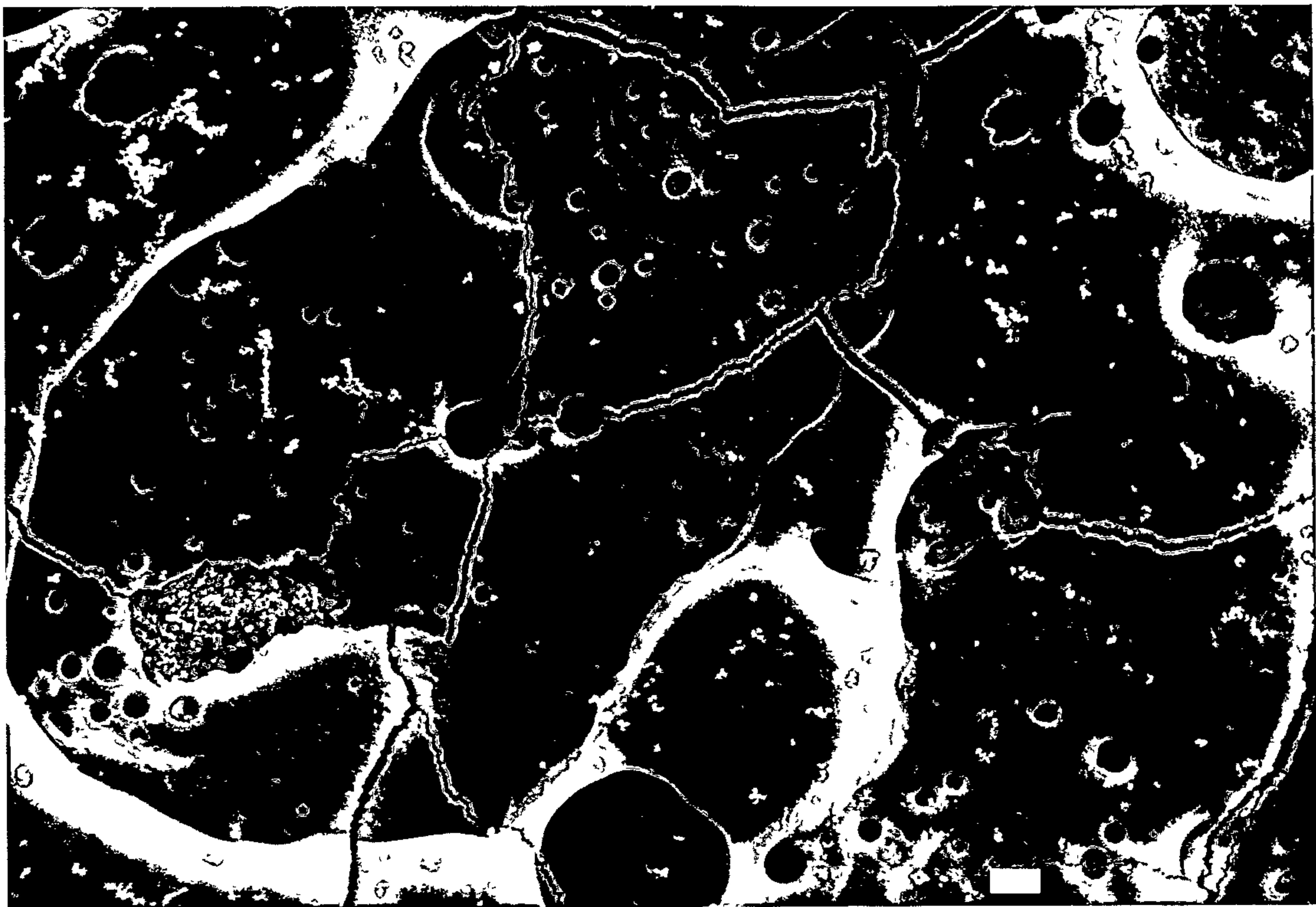


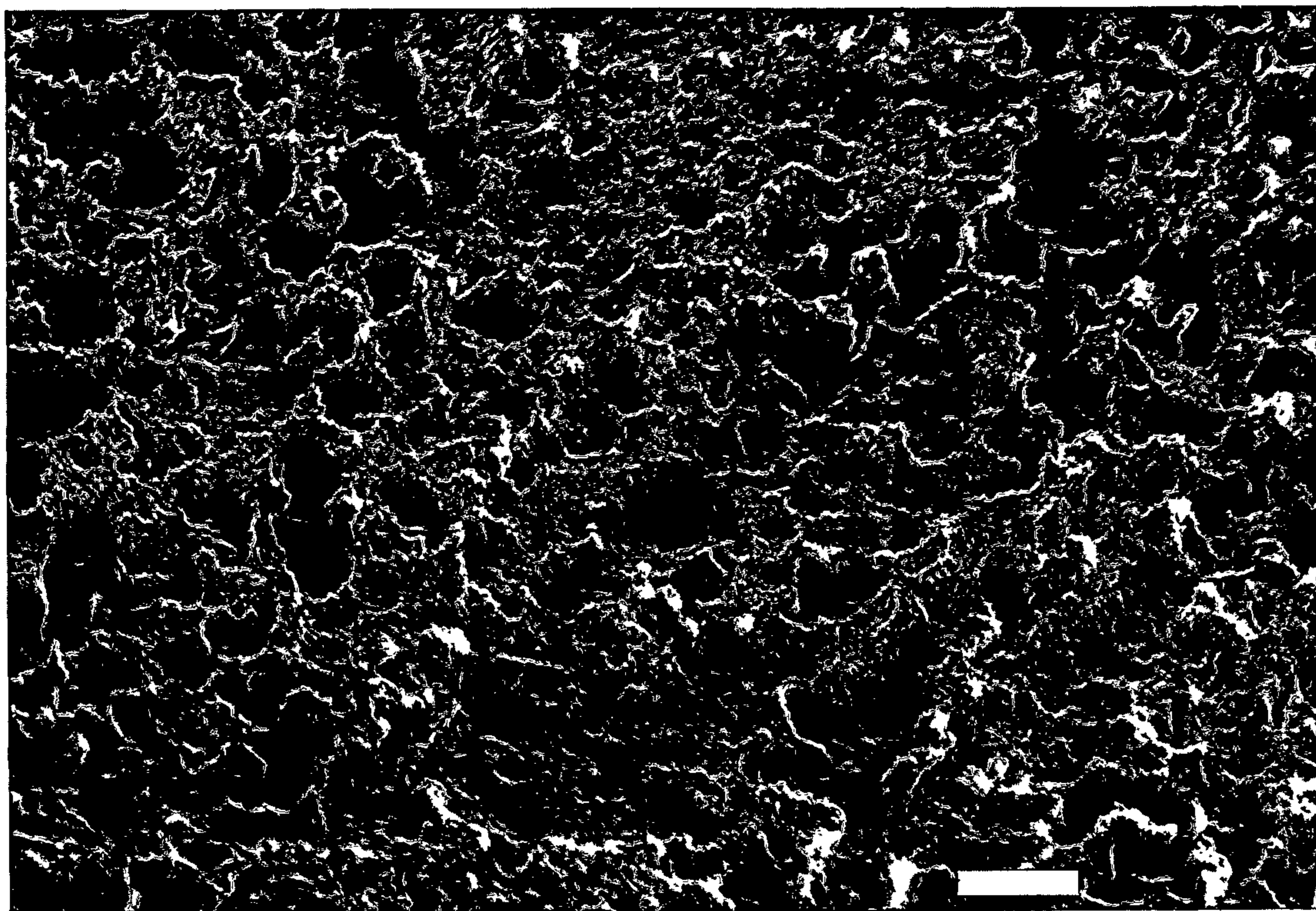
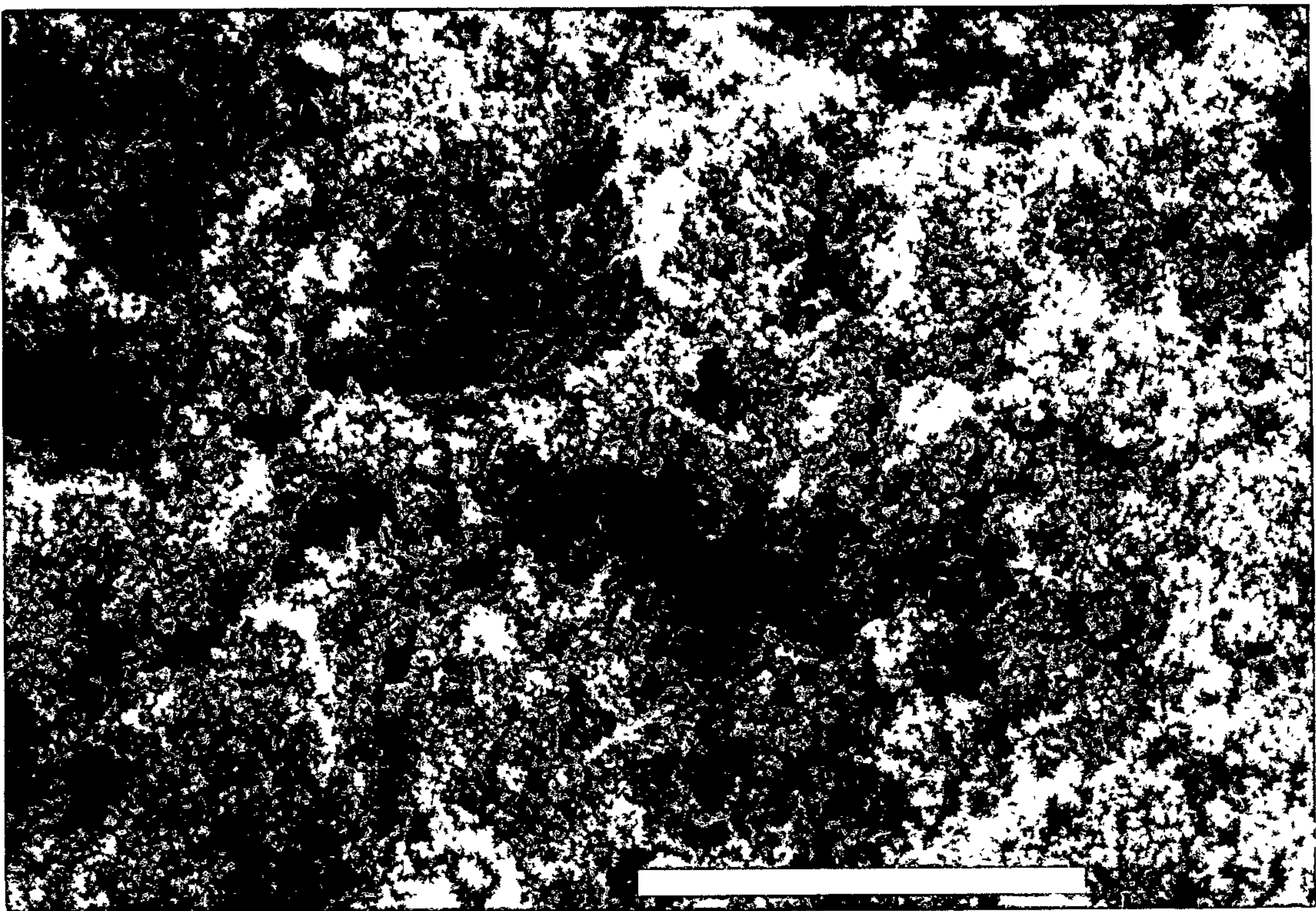
Figure 6-33 Area of rough exposed enamel at very high magnification. This presents a very similar appearance to the acid etched-enamel shown below.

Energy density = 124.3 J/cm^2

Bar = $10 \text{ }\mu\text{m}$

Figure 6-34 Acid-etched enamel for comparison with Figure 6-33.

Bar = $10 \text{ }\mu\text{m}$



as previously mentioned, is made up of a combination of surface changes involving cracking, rough exposed enamel, pitting and cratering. The proportion of each to the total area is dependent on the energy density with the pulse rate determining the distribution of the impact disks.

High magnification of areas of rough exposed enamel present an appearance remarkably similar to an acid-etched surface (Fig 6-33 and 6-34), although the distribution would be insufficient to enhance bonding. Similarly, the pitting and cracking are unlikely to improve bonding to the enamel surface, and the latter may actually reduce the strength of the surface.

Cratered areas would, however, provide a dramatic increase in surface area. The multilocular formations associated with the peripheries of the impact disks could conceivably give the necessary mechanical interlocking for a satisfactory resin to tooth bond if there was a sufficiently dense distribution.

6.3.2 Cross-sectional changes

Four teeth from Group 3, that had previously viewed topographically, were sectioned to give a comparison between laser-etched, acid-etched and normal surfaces. The energy densities used were 62.2 J/cm^2 and 159.2 J/cm^2 . Figure 6-35 is a thousand times magnification of normal enamel which shows a characteristically smooth surface. The longitudinal cracking is due to preparation of the specimen. Compared

Figure 6-35 Cross-sectional view of unlased enamel.

Longitudinal cracks present are due to the preparation.

Bar = 10 μm

Figure 6-36 Cross-sectional view of low energy density laser-etched enamel.

Horizontal cracking present 10 - 20 μm below the surface may be due to effects of laser or the actual preparation.

Energy density = 62.2 J/cm²

Bar = 10 μm

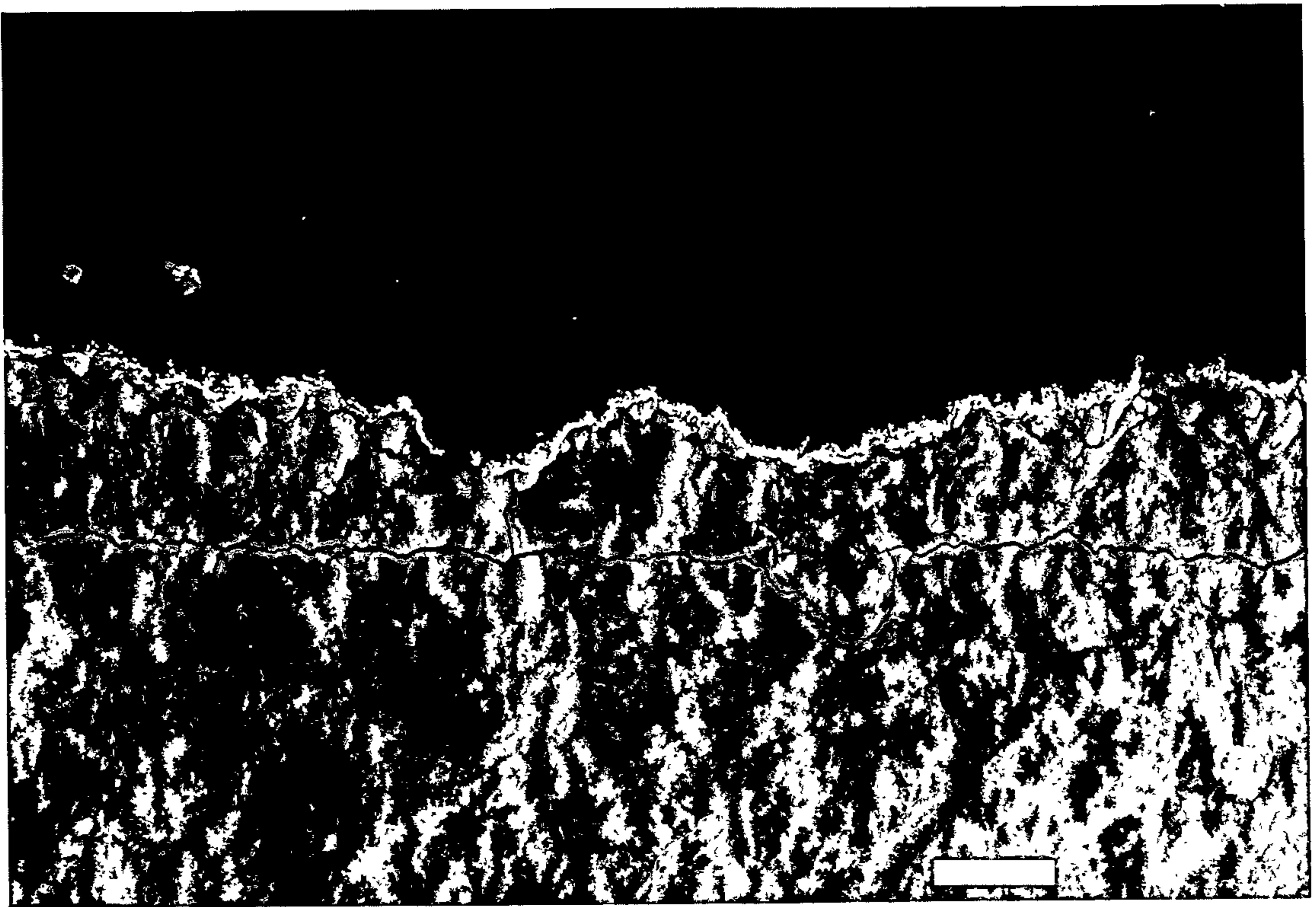
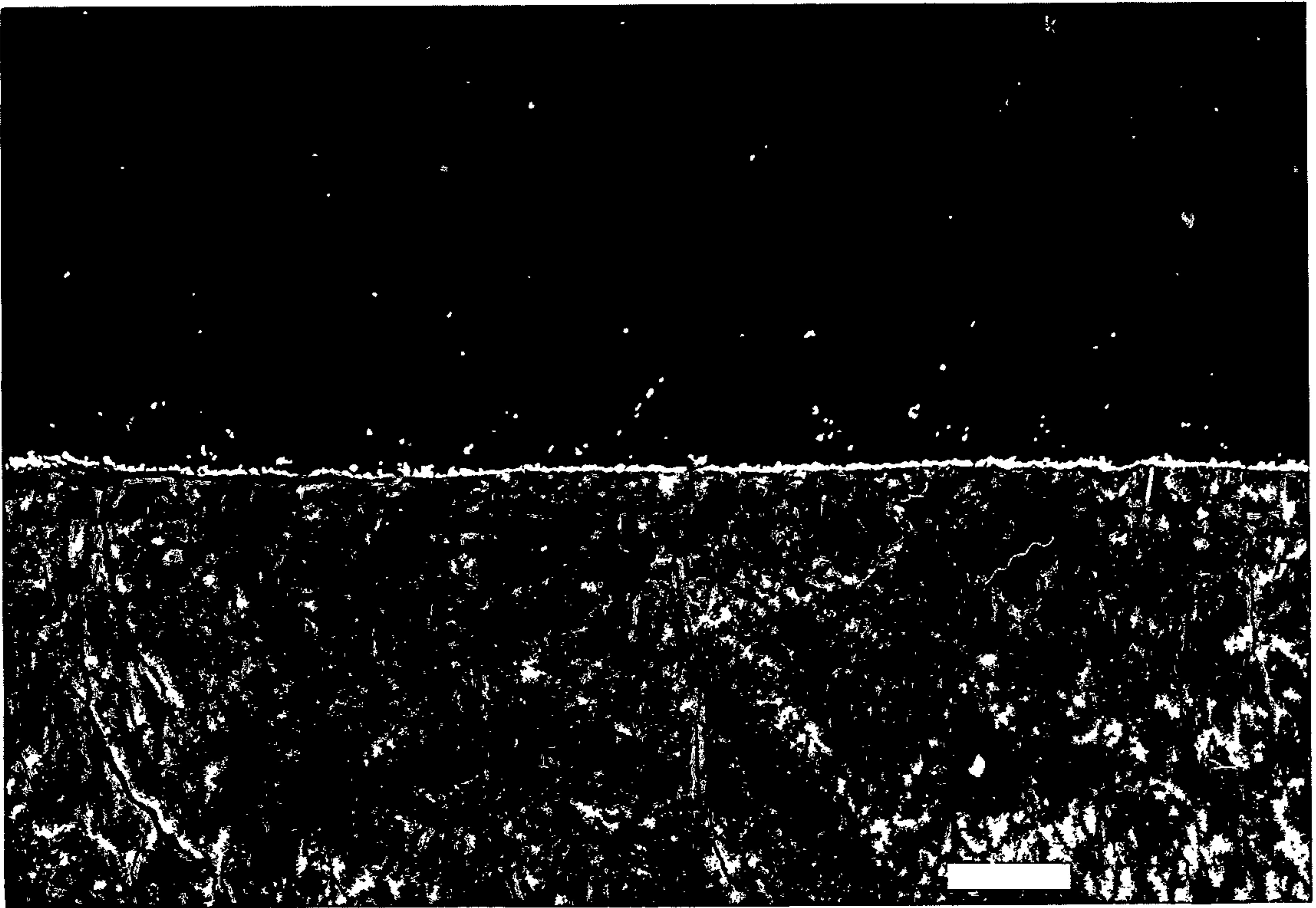


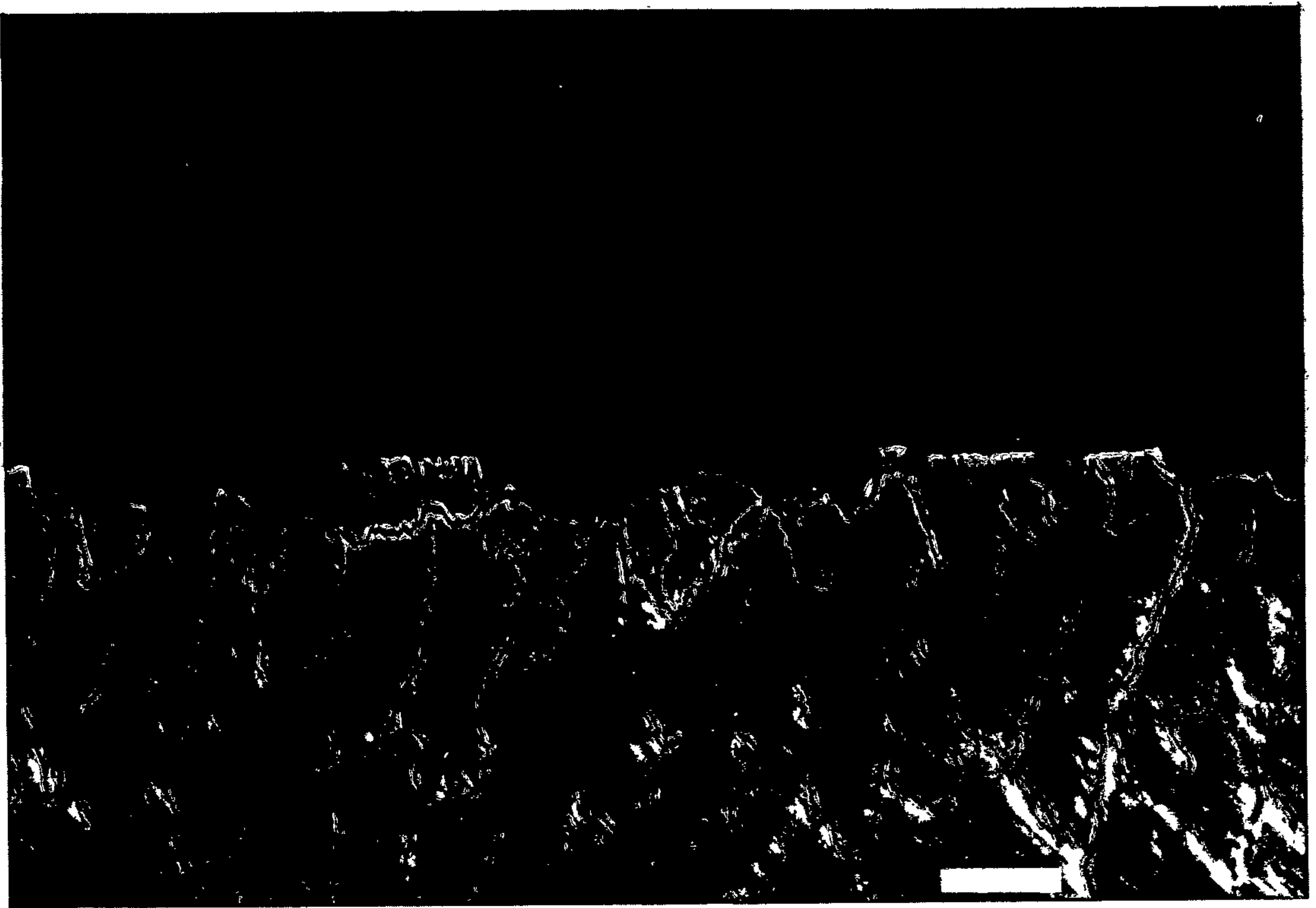
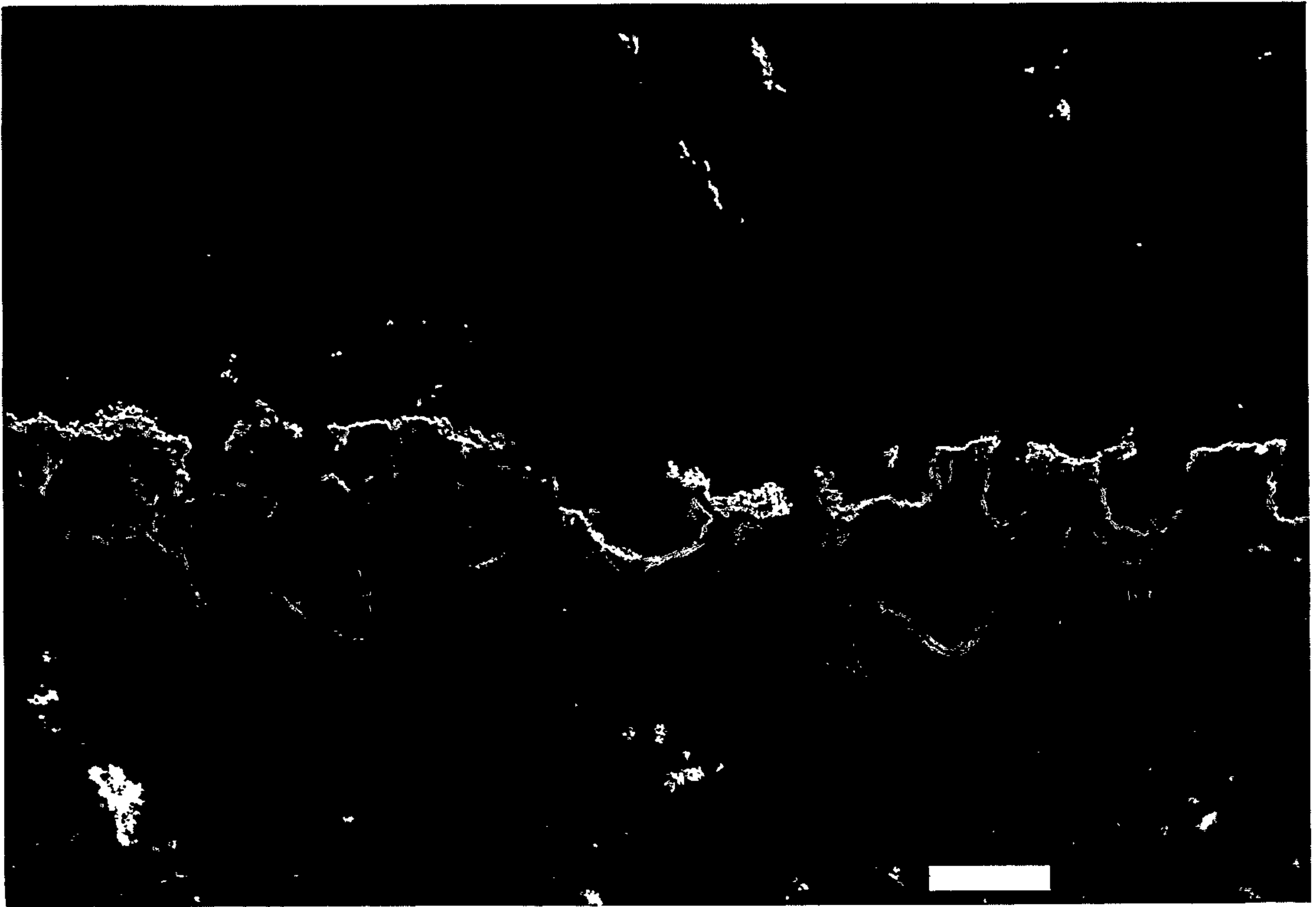
Figure 6-37 Cross-sectional view of laser-etching at higher power.

Energy density = 159.2 J/cm^2

Bar = $10 \text{ }\mu\text{m}$

Figure 6-38 Cross-sectional view of acid-etched enamel.

Bar = $10 \text{ }\mu\text{m}$



to this is the angular surface produced with acid-etching (Fig 6-38). The depth of enamel rod dissolution appears to be less than 10 μm . At 62.2 J/cm², the cross-sectional appearance of laser-etching is not unlike acid-etching (Fig 6-36). Obvious cratering is not visible. The overall changes give the impression of a smoother surface without much scope for mechanical interlocking.

Figure 6-37 is a view of laser-etching at a higher energy density. Three distinct layers are evident.

1. The first layer shows actual structural alteration in the enamel due to cratering. The crater width is variable but the depth seems to be fairly uniform at approximately 8 μm .
2. The second layer is a band of what appears to be thermally damaged enamel without actual structural alteration. It initially appears dark in colour following the contour of the craters but becomes progressively lighter with increasing depth. This layer extends approximately 2 μm to 10 μm below the base of the cratering into the enamel.
3. The deepest layer is normal enamel showing rod orientation.

CHAPTER 7

DISCUSSION

7.1 Introduction

After the first functioning laser was built in 1960, research began to find medical applications for this new light source. Initially it was thought lasers were "a solution looking for a problem" because many applications were proposed but few were economically justifiable (Johnson, 1990). As the technology has evolved some medical specialties such as ophthalmology and dermatology have made more extensive clinical use of them, particularly ruby and argon lasers.

The development of the CO₂ laser proposed a new and different scalpel ("optical knife", "light scalpel") as an adjunct to standard surgical systems. Treatment of lesions which could not be accomplished by standard surgery was made possible. With the advent of the Nd:YAG laser, energy could be transmitted through small fibre-optic cables which opened new avenues for many specialties. It was this feature that led to the development of a commercially available laser system for use in dentistry. It has been argued that the CO₂ laser is preferable for hard tissue dental

therapy due to its favourable absorption characteristics in enamel (Zakariasen *et al*, 1991), but the problem of transmission for intraoral use is still being overcome. Even though the Nd:YAG laser used in this study is rated as Class IV (energy density exceeding 10 J/cm^2) the energy levels recommended are continually being reduced as further research is done to minimise the chance of possible skin and eye injuries.

A variety of treatment modalities have been studied with the Nd:YAG laser including caries removal, caries prevention, dentine desensitisation, endodontic and periodontic procedures and soft tissue removal and recontouring. The substitution of laser-etching for acid-etching had also been proposed after a roughening of the enamel surface was noticed whilst examining scanning electron micrographs of laser-etched enamel. Hess, (1990) thought that this cratering of the enamel may facilitate the bond strength of dental materials to enamel and ultimately replace the conventional method of enamel etching. Research in this particular field has been aimed at four major areas;

1. The pulpal effects of laser energy for the purpose of enamel etching.
2. The types of surface alterations produced by different laser exposures for etching and whether these exposures are detrimental to the enamel surface.
3. The beneficial effects of laser interaction with dental enamel,

mainly with respect to increased caries resistance.

4. The bond strengths obtained with laser-etching compared to those with acid-etching.

This study concentrated on the second area by using scanning electron microscopy to compare and contrast the effects of Nd:YAG laser versus acid-etching. The clinician applying treatment should be able to expect consistently better results than with acid-etching without the problems of safety to the patient and operator or adverse effects on the underlying enamel, pulp or soft tissues.

7.2 Macroscopic Appearance

To the naked eye, the enamel that has been laser-etched is readily discernible from the unaltered surface. Between 95.5 J/cm^2 and 124.3 J/cm^2 there is a transition from the residual initiator indicating the location of the laser beam impact sites to the sites themselves being more opaque than the normal enamel. This tends to give the impression that the enamel is "etched" because a white, frosty surface is the clinical indication of a microscopically satisfactory etch produced by phosphoric acid. From these results, an assessment of the amount of initiator remaining indicates the level of energy reaching the tooth surface. With increasing energy density it is more likely that the enamel will be cratered rather than just cracked or pitted. This cratering

offers the best opportunity for increased retention through mechanical interlocking. Thus a satisfactorily laser etched surface could be distinguished clinically in much the same manner as acid etched surface. The key signs are a lack of residual initiator as well as relatively opaque impact sites. This seems to occur around an energy density of 124 J/cm^2 .

The accuracy of the laser can only be fully appreciated with use. The HeNe aiming beam and initiator use allows "etching" to be precisely contained. There is no risk of an increased potential for demineralisation interproximally and cervically due to unwanted placement of the etching medium. However, the retention of initiator in the lasered surface prior to the bonding of orthodontic brackets also raises the problem of a potentially unaesthetic appearance after removal of these brackets. Will it be removed with standard debonding procedures? The amount of residual initiator decreases with increasing energy density, therefore making it essential to use a more powerful setting from an aesthetic as well as a functional perspective. Research in this area is needed to assess this issue.

7.3 Microscopic Appearance

The three main variables involved in etching a tooth with an Nd:YAG laser are the energy density of the beam, the rapidity with which the beam is pulsed and the type and amount of initiator used. The effects of each will be discussed in turn.

7.3.1 Energy density

The energy densities used in this study ranged from half the recommended value to about nine times the value. This provided evidence for a continuum of changes in the enamel surface although not all were entirely energy dependent. There were four alterations observed:

1. Cracking

Surface microcracking appeared in all the samples except one. The exception was with the use of a diluted solution of the initiator. Energy transfer to the enamel surface had still occurred due to the presence of small pitting. The other two teeth prepared in the same way both demonstrated microcracking which suggests that this one result was probably an experimental fluke.

Cracking was absent from adjacent unlased areas as well as acid-etched areas which supports the contention by Hess, (1991) that it is not merely an artefact produced by the preparation process. The actual depth of the microcracking could not be measured. However, the roughened enamel exposed after flaking was typically free of cracking, suggesting that it is only superficial.

As previously mentioned, the laser produces its effect mostly by generating heat. The result of this may be restricted to a modest rise in temperature or the material concerned may burn, melt or vaporise. In the latter cases the

sudden generation of gases may produce the effect of a miniature explosion which directly ruptures the immediate structure (Jeffrey *et al*, 1990). Alternatively, it could be a result of either thermal stresses in the enamel due to expansion and contraction through rapid heating of a localised area or shock waves associated with the beam's interaction with the tooth.

Cracking remained constant in quantity and width despite increasing energy density. This suggests that a minimum temperature is required for the cracking to occur yet it is independent of further changes in temperature. Boehm *et al* (1977) found that at energy densities above 23.5 J/cm^2 microcracks on the enamel surface were produced. The cracks are also probably related to flaking of enamel in areas of weakness leading to ejection of this top layer. This is supported by the finding that these areas were concentrated over the imbrication lines. The cracking may contribute to an increased mechanical interlocking. However, this is balanced by the possibility that it may also reduce the strength of the enamel in this layer.

2. Flaking

Flaking of enamel in some areas appears to be related to cracking. This corresponds to the findings of Ferreira *et al*, (1989). They found that the ejection of material was more common in areas having longitudinal rather than transverse prism orientation, that is, over the imbrication lines. Like cracking, it is also found at all energy densities to roughly the same extent.

The appearance of this enamel at high magnification is remarkably similar to acid-etched enamel and may be the cause of claims of similarity between the two techniques. Thus if both cracking and the rough exposed enamel created after flaking were sufficient to enhance bonding then bond strength would conceivably be the same despite the energy density used. This contention cannot be supported because of two factors, namely, the physical size of these areas and the non-uniform coverage by the flaked enamel.

3. Pitting and cratering.

These changes, although noted as being separate, are very likely to be related, and will thus be discussed together. In contrast to cracking and flaking, they do seem to be energy-dependent with pitting appearing as a form of "low energy" cratering. Pitting was most common at lower energy densities. Increases in size and concentration occurs to the 124 J/cm^2 level. Cratering then seems to predominate at the impact sites, increasing in definition and complexity of the multilocular formations at 159.2 J/cm^2 . A doubling or tripling of the energy density to 318.3 J/cm^2 or 572.9 J/cm^2 , however, does not proportionally increase the quality or quantity of cratering produced.

The alterations created in the enamel surface are mainly due to the heat produced by absorption of the laser. Assuming that one pulse of the laser beam causes one impact crater, then the surface alterations created relate to that energy density for that particular setting.

As the energy density increases the impact disks became more distinct and the proportion of cratering increased.

The appearance of a "typical" impact disk affords an indication of the cross-sectional intensity of the beam. According to Apfelberg (1987), the beam profile is likely to be of a basic Gaussian distribution giving maximum intensity at the centre and diminishing intensity at the periphery. The highest intensity of the beam would thus be directed towards the centre causing an ablation of the initiator which then transfers the heat to the enamel. A shallow depression would result with the molten enamel pushed away from the centre which then solidifies forming the rim. It is the "bubbling" of the enamel which creates the dramatic increase in surface area in these areas on cooling.

The results found in this study are very similar to those previously reported with CO₂ and argon lasers even though the Nd:YAG laser uses a different active medium and a different wavelength. Ferreira *et al* (1989) described two different surface changes created by the CO₂ laser, crazed and cratered enamel and crazed enamel. They also found associated areas of the enamel surface which had lifted off to give a rough exposed surface. Patel *et al* (1988), also using a continuous wave CO₂ laser at irradiances of 90 W/cm² - 15000 W/cm², demonstrated four distinct surface changes in lased enamel; ablation, melting, resolidification and cracking and flaking. Perhaps, whilst the absorption is variable depending on the wavelength of the laser being used, the effects of the heat produced at comparable energy densities once imparted, causes

the characteristic changes seen in the enamel. Thus the type of laser is not critical if the heat can be transferred efficiently to the enamel via a laser initiator.

The surface area increase appears comparable, although entirely different, from the surface area increase with acid-etching. The mechanical interlocking required for bonding is afforded by the cratering. As it is not present in sufficient quantities below 124.3 J/cm², and its quality does not improve after 159.2 J/cm², an energy density between these two values would produce the maximum etch for the minimum energy outlay. Contrary to the manufacturer's recommended setting of 62.2 J/cm² (0.75 watts at 15 pps with a 320 µm fibre), the results of this study suggest an energy density of 124.3 J/cm² (1.5 watts at 15 pps with a 320 µm fibre) or 159.2 J/cm² (0.75 watts at 15 pps with a 200 µm fibre) would produce the most uniformly retentive surface. The laser-etching of selected areas only contrasts with uniform coverage obtained with the conventional method.

7.3.2 Pulse rate

Because cratering is mainly restricted to the periphery of the impact disks, the density of these sites will increase the total area that is cratered. However, cratering cannot be increased indefinitely. The two factors involved are power density which increases as pulse rate increases for the same period of time, therefore increasing the total energy received by the tooth. The second is that heat transfer is dependent on the

presence of the initiator (see later discussion). The energy density used determines the extent of pitted and cratered enamel but does not seem to affect the cracking or flaking. The pulse rate essentially determines the proportion of the enamel surface covered by this cratered enamel which, in turn, is determined by the presence of initiator. A surface with minimal remaining initiator and a high percentage of overlapping cratering will undoubtedly provide a stronger bond than a less uniformly covered surface with a lot of residual initiator. Whilst overlapping of craters is possible with faster pulse rates, the effects in these regions are not additive. The same result can be gained with the same energy density and power density in half the time with double the pulse rate.

7.3.3 Initiator

Use of initiator is absolutely essential to induce surface alterations. A major benefit from this is that only those areas designated by the initiator will be etched, ensuring maximum accuracy. Thin, uniform coverage seems to give the most efficient transfer of energy to the enamel surface. Excess initiator reduces the changes observed perhaps by dissipating the energy within itself rather than transferring it to the tooth.

From the surface results, alterations can be obtained microscopically which give the impression that a suitable mechanical interlocking could be obtained and hence a satisfactory bond between enamel and restorative materials. The production of this

surface can also be determined macroscopically which is important for clinical application.

7.4 Comparison with Acid-Etch

7.4.1 Surface topography

There is no doubt that the "etched" surface obtained by lasing is entirely different to that obtained with acid on a microscopic level. Clinically, the production of a satisfactory etch can be determined much more reliably with acid but with further research this may also be possible with lasing. Both take about the same amount of time to complete. Even though pumicing and washing of the tooth surface is not required with lasing, a dry tooth is necessary for accurate placement of the initiator. The issue of residual initiator is still unresolved. Accuracy was far better with the laser, which would be very beneficial in reducing unwanted etching. Lasing of an area suitable for bonding of orthodontic brackets took approximately thirty seconds per tooth at the recommended setting. With a faster pulse rate this time was reduced.

The only surface alteration induced by the laser which would give a substantial increase in mechanical interlocking was cratering. This was in stark contrast to the spicular surface produced with the phosphoric acid. The distribution of cratering was restricted mainly to the periphery of the impact sites which cannot give the uniform

coverage obtained with acid. Nevertheless, from the bonding studies it appears that this non uniform-distribution is sufficient to provide an adequate bond for the retention of orthodontic brackets. A quantitative assessment of the various surfaces with respect to bonding strength is beyond the scope of this study but would be the logical follow-up.

The extent to which the enamel is affected by the universal cracking would be important at the time of debonding. However, if they are as superficial as the uncracked underlying enamel suggests then the integrity of the enamel is probably not in jeopardy. A microscopic analysis of both acid and lased surfaces would determine if any permanent damage had been done.

7.4.2 Cross-sectional changes

There have been limited cross-sectional studies done on the effects of the Nd:YAG laser. However the results obtained are very similar to those reported with the use of the CO₂ laser. Nelson *et al* (1986a) found that lasing of human dental enamel produces a localised increase in temperature with rapid melting and cooling of apatite crystallites. By keeping the energy density constant whilst increasing the peak power density by pulsing the laser, the melting and recrystallisation is confined to a thin surface region (approximately 5 µm) without affecting the immediate underlying surface enamel.

Reports that enamel penetration by acid and laser energy are approximately the same in depth of 20 μm - 40 μm (Myers and Riddle, 1990; Myers, 1990) intimate that the level of mechanical retention is also similar. The results of this study support the contention that cratering is basically restricted to the first 10 μm . However, the depth of the acid-etch in cross-section obtained in this study was also only about 10 μm . Thermal effects are visible for the same depth again under this melt region which Nelson *et al* (1986b) believe is large enough to affect the organic matrix by altering the permeability and hydration properties of the enamel and therefore its solubility behaviour. An actual assessment of the thermal changes elicited is not possible purely on visual inspection, so more comprehensive testing would need to be done.

CHAPTER 8

SUMMARY AND CONCLUSION

8.1 Summary

The ability to increase the adhesion of restorative materials to enamel has revolutionised many aspects of dentistry. For orthodontics, bonding, rather than banding of attachments has meant a significant improvement in aesthetics and the maintenance of dental and periodontal health. The currently accepted method is acid etching which was introduced by Michael Buonocore in 1955.

The fluoride-rich surface enamel adjacent to orthodontic brackets may be inadvertently removed with this etching technique. This has led to the concern that these areas may be predisposed to decalcification during treatment. Improvements in both materials and techniques have therefore been the source of continued research. Recently, Nd:YAG laser technology has been developed specifically for dental applications in soft and hard tissue procedures. One of the proposals suggested is that laser etching, with its speed, ease and accuracy of placement, could be an alternative

to acid etching. However, relatively little information is known about the effects of an Nd:YAG laser beam on human dental enamel. Bonding strength studies have demonstrated favourable results in comparing laser etching with acid etching, yet the literature is inconclusive as to the method by which this is achieved.

This experiment used extracted human premolar teeth, which were laser-etched with varying energy densities, pulse rates and initiator volumes, to determine the similarities and differences in enamel surface alterations compared to conventional etching. A number of trends emerged.

1. No alterations in the enamel surface were observed with lasing when the initiator was not used. This ensures that only those areas marked with initiator will be etched. Accuracy is much greater with lasing than acid-etching.
2. Increasing the amount of initiator beyond a surface smear impeded the quality and quantity of the changes induced. The initiator may actually absorb the laser energy internally without conducting it to the surface as heat.
3. Varying the energy density between 37.3 J/cm^2 and 572.9 J/cm^2 gave four different surface alterations. These were cracking, flaking, pitting and cratering. The first two changes were found in all specimens and were not dependent on the energy density. Pitting appeared to be a low energy form of cratering which increased in definition with increasing energy density. Cratering of enamel is the only alteration which would increase the surface

area of the enamel sufficiently to enhance bonding. As it is only found in direct association with the impact craters, there would not be the same uniform coverage as provided by acid-etching.

4. The surface alterations obtained in this study are very similar to those previously reported with both CO₂ and Nd:YAG lasers. Differences with respect to wavelength are not apparent in the manner that they are with soft tissues. The energy density of the beam may therefore be the major, if not the single, determinant of enamel changes in laser-etching.
5. The manufacturer's recommendation of an energy density of 62.2 J/cm² would not provide enough cratering to increase available surface area. An energy density between 124.3 J/cm² and 159.2 J/cm² gives adequate cratering for each impact disk. Doubling or tripling the power does not proportionally increase the quantity or quality of the surface cratering.
6. Increasing the pulse rate does not alter the observable changes for a particular energy density. However, it does increase the proportion of the tooth surface covered by the impact disks. Therefore the highest pulse rate possible will give the most complete coverage. This will also increase the power received by the tooth.
7. A direct comparison of acid-etching with laser-etching demonstrated the dissimilarities between the two techniques. Cross-sectionally, the effects of

lasing appear to be restricted to approximately the first 10 μm . It is unknown whether this is sufficient to provide tag lengths comparable to acid-etching. A separate study specifically looking at this aspect is required.

8.2 Conclusion

The Nd:YAG laser used at an energy density above 124.3 J/cm^2 produces cratering in the enamel surface which would provide a means of mechanical interlocking for bonding materials. However, whilst it does not give the complete coverage that acid-etching does, its placement is far more accurate. Superficial cracking of the enamel is always present and tends to weaken the enamel leading to areas of flaking. The roughened enamel exposed by this appears similar to the surface obtained with acid-etching but is not present in sufficient quantities to enhance bonding. The question of residual initiator may preclude its use for the pre-treatment of the enamel prior to orthodontic bracket placement due to aesthetic concerns. At this stage, etching with an Nd:YAG laser would not be a suitable alternative to acid-etching as used in orthodontics.

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