# Therapeutic Ultrasound

Ultrasound is a form of MECHANICAL energy, not electrical energy and therefore strictly speaking, not really electrotherapy at all. Mechanical vibration at increasing frequencies is known as sound energy. Below about 16Hz, these vibrations are not recognisable as sound, and the normal human sound range is from 16Hz to something approaching 15-20,000 Hz (in children and young adults). Beyond this upper limit, the mechanical vibration is known as ULTRASOUND. The frequencies used in therapy are typically between 1.0 and 3.0 MHz (1MHz = 1 million cycles per second).

Ultrasound Waveform with Compression and Rarefaction
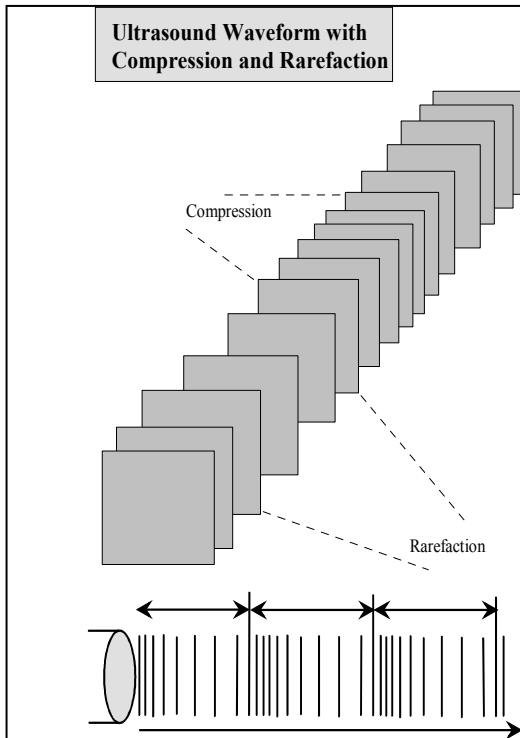
Sound waves are LONGITUDINAL waves consisting of areas of COMPRESSION and RAREFACTION. Particles of a material, when exposed to a sound wave will oscillate about a fixed point rather than move with the wave itself. As the energy within the sound wave is passed to the material, it will cause oscillation of the particles of that material. Clearly any increase in the molecular vibration in the tissue can result in heat generation, and ultrasound (US) can be used to produce thermal changes in the tissues, though current usage in therapy does not focus on this phenomenon.
(Williams 1987, Baker et al 2001, ter Haar 1999, Nussbaum 1997)
In addition to thermal changes, the vibration of the tissues appears to have effects which are generally considered to be non thermal in nature, though, as with other modalities (e.g. Pulsed Shortwave) there must be a thermal component however small. As the US wave passes through a material (the tissues), the energy levels within the wave will diminish as energy is transferred to the material. The energy absorption and attenuation characteristics of US waves have been documented for several types of tissue.

Ultrasound Waves :

FREQUENCY - the number of times a particle experiences a complete compression/rarefaction cycle in 1 second. Typically 1 or 3 MHz.

WAVELENGTH - the distance between two equivalent points on the waveform in the particular medium. In an 'average tissue' the wavelength @ 1MHz would be 1.5mm and @ 3 MHz would be 0.5 mm.

VELOCITY - the velocity at which the wave (disturbance) travels through the medium. In a saline solution, the velocity of US is approximately 1500 m sec^{-1} compared with approximately 350 m sec^{-1} in air (sound waves can travel more rapidly in a more dense medium). The velocity of US in most tissues is thought to be similar to that in saline.

These three factors are related, but are not constant for all types of tissue. Average figures are most commonly used to represent the passage of US in the tissues. Typical US frequencies from therapeutic equipment are 1 and 3 MHz though some machines produce additional frequencies (e.g. 0.75 and 1.5 MHz).

The mathematical representation of the relationship is **V = F.λ**
where **V** = velocity, **F** = frequency and λ is the wavelength.

US Waveform

The US beam is not uniform and changes in its nature with distance from the transducer. The US beam nearest the treatment head is called the NEAR field, the INTERFERENCE field or the Frenzel zone. The behaviour of the US in this field is far from regular, with areas of significant interference. The US energy in parts of this field can be many times greater than the output set on the machine (possibly as much as 12 to 15 times greater). The size (length) of the near field can be calculated using r^2/λ where r= the radius of the transducer crystal and λ = the US wavelength according to the frequency being used (0.5mm for 3MHz and 1.5mm for 1.0 MHz).

As an example, a crystal with a diameter of 25mm operating at 1 MHz will have a near field/far field boundary at : **Boundary = 12.5mm^2/1.5mm $\approx$ 10cm** thus the near field (with greatest interference) extends for approximately 10 cm from the treatment head when using a large treatment head and 1 MHz US. When using higher frequency US, the boundary distance is even greater. Beyond this boundary lies the Far Field or the Fraunhofer zone. The US beam in this field is more uniform and gently divergent. The 'hot spots' noted in the near filed are not significant. For the purposes of therapeutic applications, the far field is effectively out of reach.

Example of an Ultrasound Beam Plot
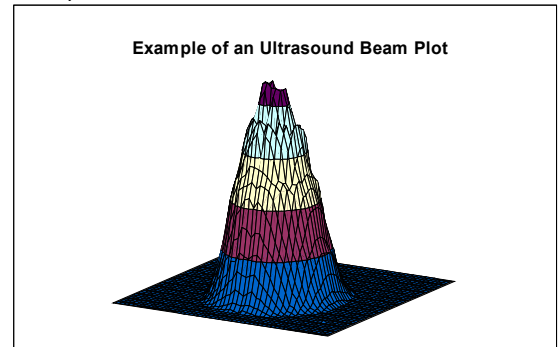
One quality indicator for US applicators (transducers) is a value attributed to the Beam Nonuniformity Ratio (BNR). This gives an indication of this near field interference. It describes numerically the ratio of the intensity peaks to the mean intensity. For most applicators, the BNR would be approximately 4 - 6 (i.e. that the peak intensity will be 4 or 6 times greater than the mean intensity). Because of the nature of US, the theoretical best value for the BNR is thought to be around 4.0 though some manufacturers claim to have overcome this limit and effectively reduced the BNR of their generators to 1.0.

Ultrasound Transmission through the Tissues

All materials (tissues) will present an impedance to the passage of sound waves. The specific impedance of a tissue will be determined by its density and elasticity. In order for the maximal transmission of energy from one medium to another, the impedance of the two media needs to be the same. Clearly in the case of US passing from the generator to the tissues and then through the different tissue types, this can not actually be achieved. The greater the difference in impedance at a boundary, the greater the reflection that will occur, and therefore, the smaller the amount of energy that will be transferred. Examples of impedance values can be found in the literature e.g. Ward 1986.

The difference in impedance is greatest for the steel/air interface which is the first one that the US has to overcome in order to reach to body. To minimise this difference, a suitable coupling medium has to be utilised. If even a small air gap exists between the transducer and the skin the proportion of US which will be reflected approaches 99.998% which in effect means that there will be no transmission.

The coupling media used in this context include water, various oils, creams and gels. Ideally, the coupling medium should be fluid so as to fill all available spaces, relatively viscous so that it stays in place (!!), have an impedance appropriate to the media it connects, and should allow transmission of US with minimal absorption, attenuation or disturbance. For a good discussion regarding coupling media, see Casarotto et al 2004, Klucinec et al 2000, Williams 1987 and Docker et al 1982. At the present time the gel based media appear to be preferable to the oils and creams. Water is a good media and can be used as an alternative but clearly it fails to meet the above criteria in terms of its viscosity.

In addition to the reflection that occurs at a boundary due to differences in impedance, there will also be some refraction if the wave does not strike the boundary surface at 90°. Essentially, the direction of the US beam through the second medium will not be the same as its path through the original medium - its pathway is angled. The critical angle for US at the skin interface appears to be about 15°. If the treatment head is at an angle of 15° or more to the plane of the skin surface, the majority of the US beam will travel through the dermal tissues (i.e. parallel to the skin surface) rather than penetrate the tissues as would be expected.

Absorption and Attenuation :

The absorption of US energy follows an exponential pattern - i.e. more energy is absorbed in the superficial tissues than in the deep tissues. In order for energy to have an effect it must be absorbed, and at some point this must be considered in relation to the US dosages applied to achieve certain effects.

Because the absorption (penetration) is exponential, there is (in theory) no point at which all the energy has been absorbed, but there is certainly a point at which the US energy levels are not sufficient to produce a therapeutic effect.

% available energy

Exponential Energy Absorption in a Medium
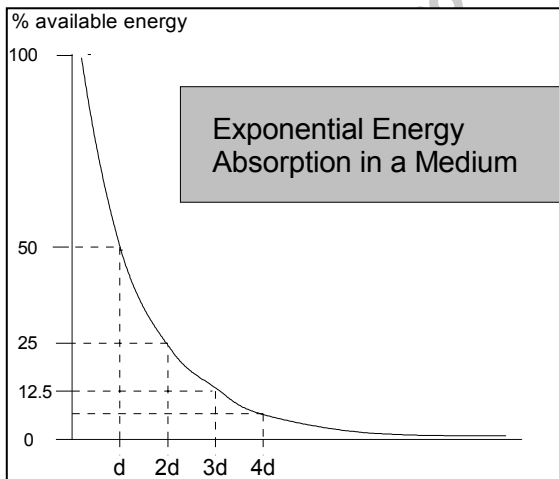
A representation of the exponential absorption is shown in the adjacent diagram.

As the US beam penetrates further into the tissues, a greater proportion of the energy will have been absorbed and therefore there is less energy available to achieve therapeutic effects. The half value depth is often quoted in relation to US and it represents the depth in the tissues at which half the surface energy is available. The will be different for each tissue and also for different US frequencies. The table below gives some indication of typical (or average) half value depths for therapeutic ultrasound. (after Hoogland 1995)

	1 MHz	3 MHz
Muscle	9.0 mm	3.0 mm
Fat	50.0 mm	16.5 mm
Tendon	6.2 mm	2.0 mm

As it is difficult, if not impossible to know the thickness of each of these layers in an individual patient, average half value depths are employed for each frequency
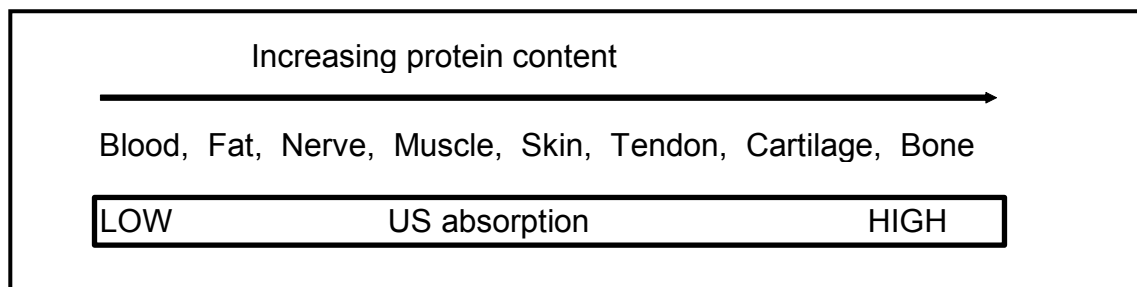
    3 MHz   2.0 - 2.5 cm
    1 MHz   4.0 cm

These values (after Low & Reed) are not universally accepted (see Ward 1986) and some current research (as yet unpublished) suggests that in the clinical (real world) environment, they may be significantly lower.

To achieve a particular US intensity at depth, account must be taken of the proportion of energy which has been absorbed by the tissues in the more superficial layers. The table on the right gives an approximate reduction in energy levels with typical tissues at two commonly used frequencies.

Depth (cm)	3 MHz	1 MHz
2	50%	
4	25%	50%
6		
8		25%

As the penetration (or transmission) of US is not the same in each tissue type, it is clear that some tissues are capable of greater absorption of US than others. Generally, the tissues with the higher protein content will absorb US to a greater extent, thus tissues with high water content and low protein content absorb little of the US energy (e.g. blood and fat) whilst those with a lower water content and a higher protein content will absorb US far more efficiently. It has been suggested that tissues can therefore be ranked according to their tissue absorption.

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┌─────────────────────────────────────────────────────────────┐
│                                                               │
│           Increasing protein content                          │
│                                                               │
│        ────────────────────────────────────────►             │
│                                                               │
│    Blood,  Fat,  Nerve,  Muscle,  Skin,  Tendon,  Cartilage,  Bone │
│                                                               │
│   ┌─────────────────────────────────────────────────────┐    │
│   │ LOW                US absorption            HIGH      │    │
│   └─────────────────────────────────────────────────────┘    │
└─────────────────────────────────────────────────────────────┘
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Although cartilage and bone are at the upper end of this scale, the problems associated with wave reflection mean that the majority of US energy striking the surface of either of these tissues is likely to be reflected. The best absorbing tissues in terms of clinical practice are those with high collagen content – LIGAMENT, TENDON, FASCIA, JOINT CAPSULE, SCAR TISSUE (Watson 2000, ter Haar 99, Nussbaum 1998, Frizzel & Dunn 1982)

The application of therapeutic US to tissues with a low energy absorption capacity is less likely to be effective than the application of the energy into a more highly absorbing material. Recent evidence of the ineffectiveness of such an intervention can be found in Wilkin et al (2004) whilst application in tissue that is a better absorber will, as expected, result in a more effective intervention (e.g. Leung et al 2004).

Pulsed Ultrasound

Most machines offer the facility for pulsed US output, and for many clinicians, this is a preferable mode of treatment. Until recently, the pulse duration (the time during which the machine is on) was almost exclusively 2ms (2 thousandths of a second) with a variable off period. Some machines now offer a variable on time. Typical pulse formats are 1:1 and 1:4 though others are available. In 1:1 mode, the machine offers an output for 2ms followed by 2ms rest. In 1:4 mode, the 2ms output is followed by an 8ms rest period. The effects of pulsed US are well documented and this type of output is preferable especially in the treatment of acute lesions. Some machines offer pulse parameters that do not appear to be supported from the literature.

The duty cycle (% of time during which the machine gives an output) will be 50% for the 1:1 mode and 20% for the 1:4 mode. This is a relevant factor in dosage calculations.

Therapeutic Ultrasound & Tissue Healing

One of the therapeutic effects for which ultrasound has been used is in relation to tissue healing. It is suggested that the application of US to injured tissues will, amongst other things, speed the rate of healing & enhance the quality of the repair. The following information is intended to provide a summary of some of the essential research in this field together with some possible mechanisms through which US treatments may achieve these changes. It is not intended to be a complete explanation of these phenomena or a comprehensive review of the current literature. It may, none the less, provide some useful basic information for clinical application.

The therapeutic effects of US are generally divided into: THERMAL & NON-THERMAL.

THERMAL:
In thermal mode, it will be most effective in heating the dense collagenous tissues and will require a relatively high intensity, preferably in continuous mode to achieve this effect.

Many papers have concentrated on the thermal effectiveness of ultrasound, and much as it can be used effectively in this way when an appropriate dose is selected (continuous mode >0.5 W cm^{-2}), the focus of this paper will be on the non thermal effects. Both Nussbaum (1998) and ter Haar (1999) have provided some useful review material with regards the thermal effects of ultrasound. Comparative studies on the thermal effects of ultrasound have been reported by several authors (e.g. Draper et al 1993, 1995a, b) with some interesting, and potentially useful results.

It is too simplistic to assume that with a particular treatment application there will either be thermal or non thermal effects. It is almost inevitable that both will occur, but it is furthermore reasonable to argue that the dominant effect will be influenced by treatment parameters, especially the mode of application i.e. pulsed or continuous. Baker et al (2001) have argued the scientific basis for this issue coherently.

Lehmann (1982) suggests that the desirable effects of therapeutic heat can be produced by US. It can be used to selectively raise the temperature of particular tissues due to its mode of action. Among the more effectively heated tissues are periosteum, collagenous tissues (ligament, tendon & fascia) & fibrotic muscle (Dyson 1981). If the temperature of the damaged tissues is raised to 40-45°C, then a hyperaemia will result, the effect of which will be therapeutic. In addition, temperatures in this range are also thought to help in initiating the resolution of chronic inflammatory states (Dyson & Suckling 1978). Having made these comments, most authorities currently attribute a greater importance to the non-thermal effects of U/S as a result of several investigative trials in the last 15 years or so.

NON-THERMAL:

The non-thermal effects of US are now attributed primarily to a combination of CAVITATION and ACOUSTIC STREAMING (te Haar 99, Baker et al 2001, Williams 1987). There appears to be little by way of convincing evidence to support the notion of MICROMASSAGE though it does sound rather appealing.

CAVITATION in its simplist sense relates to the formation of gas filled voids within the tissues & body fluids. There are 2 types of cavitation - STABLE & UNSTABLE which have very different effects. STABLE CAVITATION does seem to occur at therapeutic doses of US. This is the formation & growth of gas bubbles by accumulation of dissolved gas in the medium. They take apx. 1000 cycles to reach their maximum size. The `cavity' acts to enhance the acoustic streaming phenomena (see below) & as such would appear to be beneficial. UNSTABLE (TRANSIENT) CAVITATION is the formation of bubbles at the low pressure part of the US cycle. These bubbles then collapse very quickly releasing a large amount of energy which is detrimental to tissue viability. There is no evidence at present to suggest that this phenomenon occurs at therapeutic levels if a good technique is used.

ACOUSTIC STREAMING is described as a small scale eddying of fluids near a vibrating structure such as cell membranes & the surface of stable cavitation gas bubble (Burns 1981, Dyson &
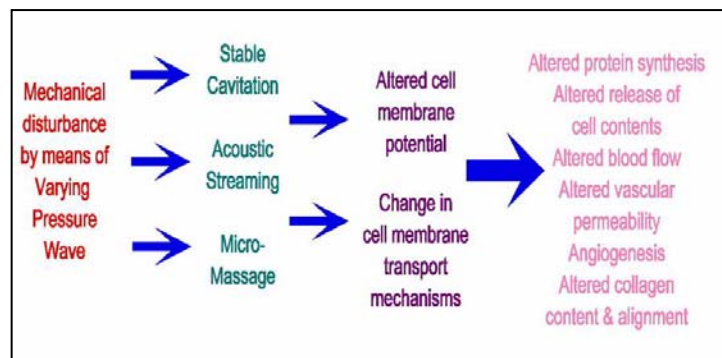
Suckling 1978). This phenomenon is known to affect diffusion rates & membrane permeability. Sodium ion permeability is altered resulting in changes in the cell membrane potential. Calcium ion transport is modified which in turn leads to an alteration in the enzyme control mechanisms of various metabolic processes, especially concerning protein synthesis & cellular secretions.

The result of the combined effects of stable cavitation and acoustic streaming is that the cell membrane becomes 'excited' (up regulate), this increasing the activity levels of the whole cell. The

US energy acts as a trigger for this process, but it is the increased cellular activity which is in effect responsible for the therapeutic benefits of the modality (Watson 2000, Dinno et al 1989, Leung et al 2004).

MICROMASSAGE is a mechanical effect which appears to have been attributed less importance in recent years. In essence, the sound wave travelling through the medium will cause the molecules to vibrate, possibly enhancing tissue fluid interchange & affecting tissue mobility.

Ultrasound Application in relation to Tissue Repair

The process of tissue repair is a complex series of cascaded, chemically mediated events that lead to the production of scar tissue that constitutes an effective material to restore the continuity of the damaged tissue. The process is more complex than be described here, but there are several interesting recent papers and reviews including (Wener & Grose 2003, Toumi & Best 2003, Watson 2003, Hill et al 2003, Neidlinger-Wilke et al 2002, Lorena et al 2002, Latey 2001).

INFLAMMATION:

During the inflammatory phase, US has a stimulating effect on the mast cells, platelets, white cells with phagocytic roles and the macrophages (Nussbaum 1997, ter Haar 1999, Fyfe & Cahal 1982, Maxwell 1992). For example, the application of ultrasound induces the degranulation of mast cells, causing the release of arachidonic acid which itself is a precursor for the synthesis of prostaglandins and leukotreine – which act as inflammatory mediators (Mortimer & Dyson 1988, Nussbaum 1997, Leung et al 2004). By increasing the activity of these cells, the overall influence of therapeutic US is certainly pro-inflammatory rather than anti-inflammatory. The benefit of this mode of action is not to 'increase' the inflammatory response as such (though if applied with too greater intensity at this stage, it is a possible outcome (Ciccone et al 1991), but rather to act as an 'inflammatory opimiser'. The inflammatory response is essential to the effective repair of tissue, and the more efficiently the process can complete, the more effectively the tissue can progress to the next phase (proliferation). Studies which have tried to demonstrate the anti inflammatory effect of ultrasound have failed to do

so (e.g.El Hag et al 1985 Hashish 1986, 1988), and have suggested that US is ineffective. It is effective at promoting the normality of the inflammatory events, and as such has a therapeutic value in promoting the overall repair events (ter Haar 99). A further benefit is that the inflammatory chemically mediated events are associated with stimulation of the next (proliferative) phase, and hence the promotion of the inflammatory phase also acts as a a promoter of the proliferative phase.

Employed at an appropriate treatment dose, with optimal treatment parameters (intensity, pulsing and time), the benefit of US is to make as efficient as possible to earliest repair phase, and thus have a promotional effect on the whole healing cascade. For tissues in which there is an inflammatory reaction, but in which there is no 'repair' to be achieved, the benefit of ultrasound is to promote the normal resolution of the inflammatory events, and hence resolve the 'problem' This will of course be most effectively achieved in the tissues that preferentially absorb ultrasound – i.e. the dense collagenous tissues.
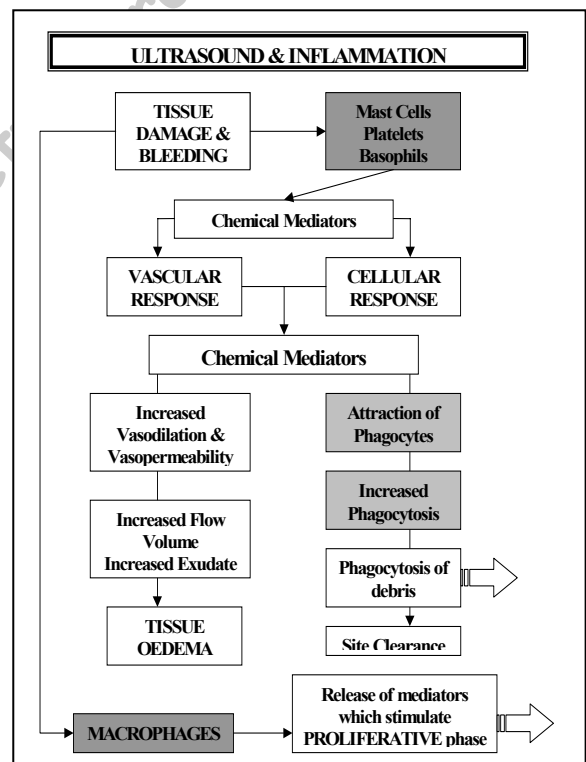
PROLFERATION:

During the proliferative phase (scar production) US also has a stimulative effect (cellular up regulation), though the primary active targets are now the fibroblasts, endothelial cells and myofibroblasts (Ramirez et al 1997, Mortimer and Dyson 1988, Young & Dyson 1990, Young & Dyson 1990b, Nussbaum 1997, 1998, Dyson & Smalley 1983, Maxwell 1992). These are all cells that are normally active during scar production and US is therefore pro-proliferative in the same way that it is pro-inflammatory – it does not change the normal proliferative phase, but maximises its efficiency – producing the required scar tissue in an optimal fashion. Harvey et al (1975) demonstrated that low dose pulsed ultrasound increases protein synthesis and several research groups have demonstrated enhanced fibroplasia and collagen synthesis (Enwemeka et al 1989, 1990, Turner et al 1989, Huys et al 1993, Ramirez et al 1997). Recent work has identified the critical role of numerous growth factors in relation to tissue repair, and some accumulating evidence has identified that therapeutic US has a positive role to play in this context (e.g. Reher et al 1999)
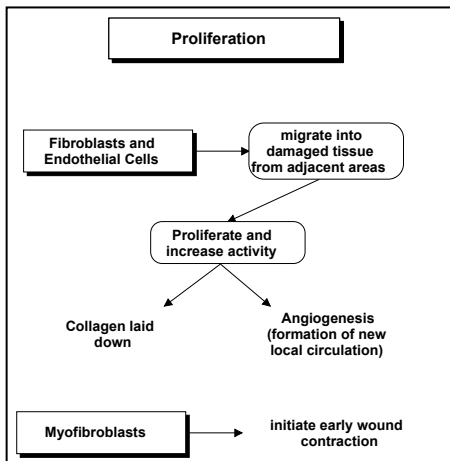
REMODELLING:

During the remodelling phase of repair, the somewhat generic scar that is produced in the initial stages is refined such that it adopts functional characteristics of the tissue that it is repairing. A scar in ligament will not 'become' ligament, but will behave more like a ligamentous tissue. This is achieved by a number of processes, but mainly related to the orientation of the collagen fibres in the developing scar (Culav et al 1999, Gomez et al 1991) and also to the change in collagen type, from predominantly Type III collagen to a more dominant Type I collagen (Vanables 1989, Forrest 1983). The remodelling process is certainly not a short duration phase – research has shown that it can last for a year or more – yet it is an essential component of quality repair (El Batouty et al 1986, ter Haar 1987)
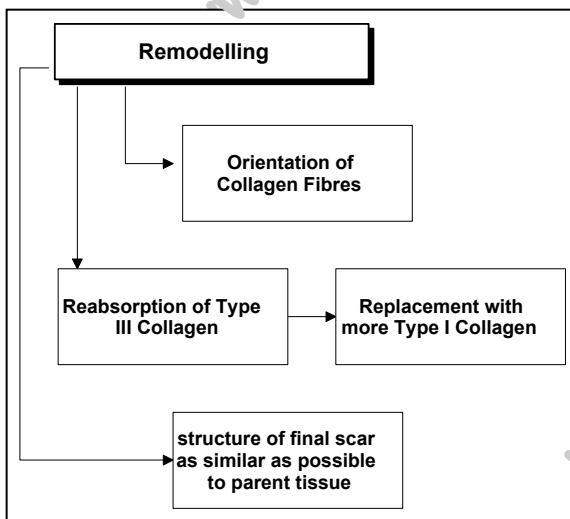
The application of therapeutic ultrasound can influence the remodelling of the scar tissue in that it appears to be capable of enhancing the appropriate orientation of the newly formed collagen fibres and also to the collagen profile change from mainly Type III to a more dominant Type I construction, thus increasing tensile strength and enhancing scar mobility (Nussbaum 1998, Wang 1998). Ultrasound applied to tissues enhances the functional capacity of the scar tissues (Nussbaum 1998, Huys et al 1993). The role of ultrasound in this phase may also have the capacity to influence collagen fibre orientation as demonstrated in an elegant study by Byl et al (1996), though their conclusions were quite reasonably somewhat tentative.

The application of ultrasound during the inflammatory, proliferative and repair phases is not of value because it changes the normal sequence of events, but because it has the capacity to stimulate or enhance these normal events and thus increase the efficiency of the repair phases (ter Haar 99). It would appear that if a tissue is repairing in a compromised or inhibited fashion, the application of therapeutic ultrasound at an appropriate dose will enhance this activity. If the tissue is healing 'normally', the application will, it would appear, speed the process and thus enable the tissue to reach its endpoint faster than would otherwise be the case. The effective application of ultrasound to achieve these aims is dose dependent.

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Therapeutic Ultrasound – Contraindications and Precautions

CONTRAINDICATIONS :

* Do not expose either the embryo or foetus to therapeutic levels of ultrasound by treating over the uterus during pregnancy
* Malignancy
* Tissues in which bleeding is occurring or could reasonably be expected
* Vascular abnormalities including deep vein thrombosis, emboli and severe arteriosclerosis / atherosclerosis
* Anaesthetic areas
* Acute infections
* Haemophiliacs not covered by factor replacement
* Application over :
    The eye

The stellate ganglion
The cardiac area in advanced heart disease
The spinal cord after laminectomy
Subcutaneous major nerves
The cranium
Subcutaneous bony protuberances
The gonads
Active epiphyses in children

PRECAUTIONS :

* Always use the lowest intensity which produces a therapeutic response
* Ensure that the applicator is moved throughout the treatment
* Ensure that the patient is aware of the nature of the treatment and its expected outcome
* If a thermal dose is intended, ensure that any contraindications that apply have been considered
* Caution is advised in the vicinity of a cardiac pacemaker or other implanted electronic device
* Continuous ultrasound is considered unwise over metal implants

HAZARDS :

Reversible blood cell stasis can occur in small blood vessels if a standing wave is produced while treating over a reflector such as an air/soft tissue interface, soft tissue/bone or soft tissue/metal interface whilst using a stationary applicator.
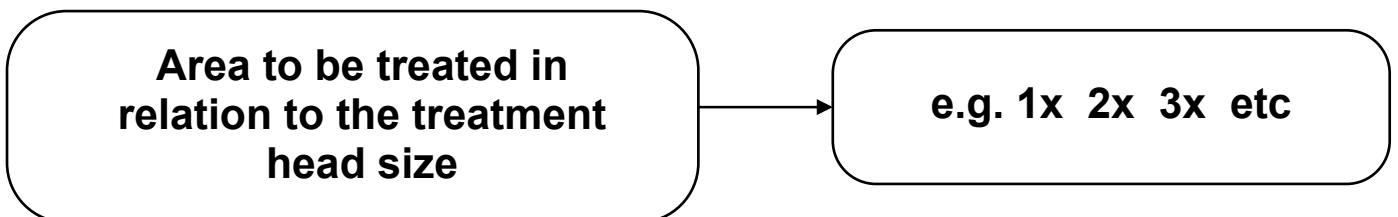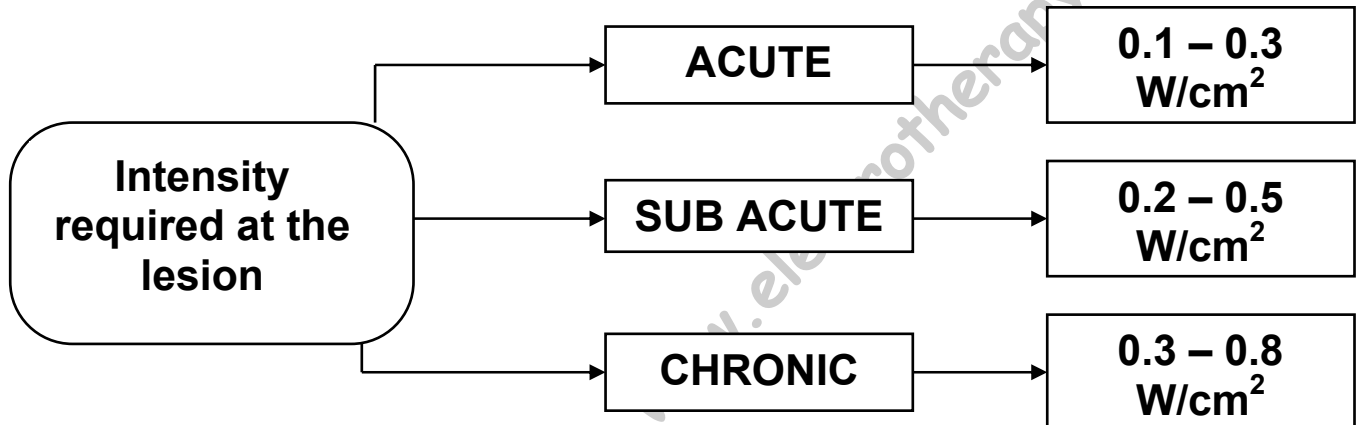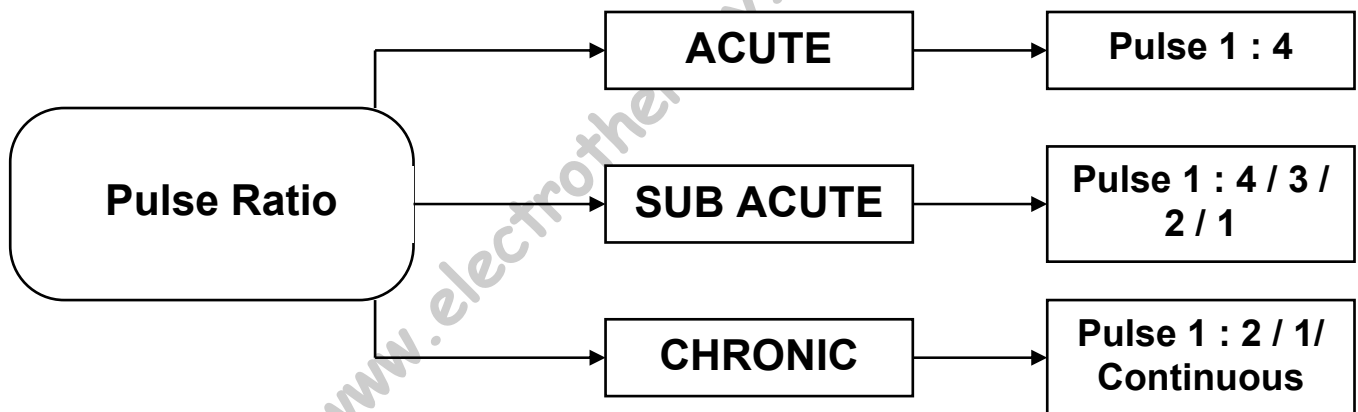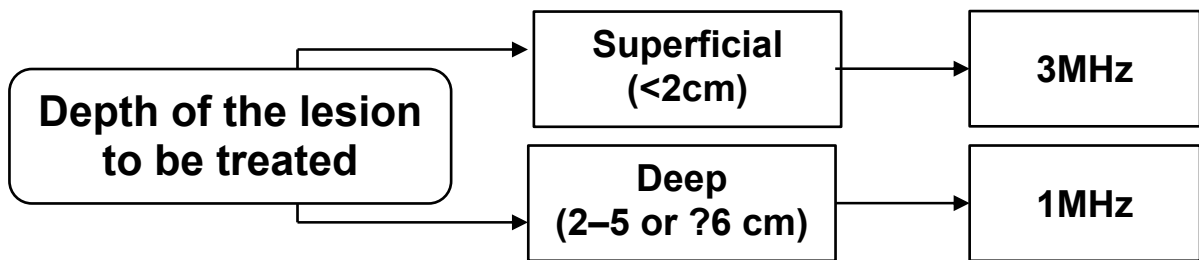
Continuous movement of the treatment head removes this hazard.

TREATMENT RECORD :

The operator should note :
    Machine
    Machine settings – :
    frequency, intensity, time,
    pulse parameters
    Area treated
    Any immediate or untoward effects

Ultrasound Dose Calculations

Depth of the lesion to be treated → **Superficial (<2cm)** → **3MHz**

Deep (2–5 or ?6 cm) → **1MHz**

Pulse Ratio

- **ACUTE** → **Pulse 1 : 4**
- **SUB ACUTE** → **Pulse 1 : 4 / 3 / 2 / 1**
- **CHRONIC** → **Pulse 1 : 2 / 1/ Continuous**

Intensity required at the lesion

- **ACUTE** → **0.1 – 0.3 W/cm^2**
- **SUB ACUTE** → **0.2 – 0.5 W/cm^2**
- **CHRONIC** → **0.3 – 0.8 W/cm^2**

Area to be treated in relation to the treatment head size → **e.g. 1x 2x 3x etc**

Ultrasound treatment principle – 1 minutes worth of ultrasound per treatment head area
Therefore longer if PULSED and longer for LARGER TREATMENT AREAS
Treatment time = 1 x (no of times treatment head fits onto tissue to treat) x (pulse ratio)