A COMPUTER SIMULATION MODEL FOR ANALYSIS OF RISE IN SKIN TEMPERATURE FOLLOWING IRRADIATION WITH THE GaAlAs DIODE LASER

Takashi Maeda¹ and Toshio Ohshiro²

¹Ohshiro Clinic and ²Japan Medical Laser Laboratory, Shinanomachi, Tokyo, Japan

A popular benchmark for defining a laser therapy system is an output power of less than 100 mW, and low reactive-level laser therapy (LLLT) is commonly regarded as 'safe' for irradiating skin. However, with very small spot sizes in the order of 200 μm , the power density at tissue of a 100 mW laser therapy system is in the order of 318 W/cm². If target tissue is painted with black ink or a similar pigment, this power density is quite capable of causing a burn injury. This has obvious implications in the use of such LLLT systems in treating dark-coloured naevi such as naevus cell naevus, or in LLLT on black patients. In order to examine this problem in detail, the authors designed a computer model with a program to calculate and analyze the temperature rise in an irradiated hemispherical block of homogeneous tissue at various powers in the range of 6 mW - 60 mW, and at spot sizes of 200 μm , 400 μm , 1.0 mm and 2.0 mm following 5 seconds irradiation with an 830 nm GaAlAs diode laser, after which the laser was 'turned off'. The laser was deemed to be applied in the contact pressure mode, thereby compressing superficial blood vessels in the irradiated area so that blood flow cooling could be discounted. At maximum absorption, it was predicted that the 60 mW beam of the GaAlAs diode laser at 830 nm and a 200 μm spot size has the potential for causing thermal injury. The authors would like to stress that this is only a theoretical computer model, with hypothesized constants and a homogeneous target material. However, this model certainly demonstrates the error of applying arbitrary limits as to what constitutes the output power of an LLLT system, and would also appear to limit the application of LLLT at the above parameters on dark-colored naevi or for darker-skinned patients.

Key words: photothermal damage; power density; laser therapy

Introduction

There are a number of reported definitions as to what constitutes a laser therapy beam in terms of output power and incident energy density. Output powers of 100 mW and less are popularly regarded both by researchers and manufacturers as 'biostimulation' or 'laser therapy' levels, and available systems reflect this belief with output powers from as low as 1 mW to 100 mW. Some of these systems deliver the laser energy through a fiberoptic light guide, producing an unfocused divergent beam at the distal end of the probe or handpiece, giving a variety of spot sizes with the minimum spot size equal to the diameter of the output device. Other systems however deliver a focused beam, and even at 60 mW, a focused beam of 200 μm , such as that delivered by the Matsushita Panalas 4000 (marketed by Mochida Pharmaceutical Company, Tokyo as the Luketron) will produce an 830 nm, continuous

wave beam with a power density of just under 200 W/cm². This is quite capable of producing a photothermally destructive reaction, especially in a dark-pigmented target. Over 2,000 units of the above system have been sold in Japan alone, and there are many other laser therapy systems which deliver 60 mW or more.

The present study presents a computer simulation model for an 830 nm, 60 mW continuous wave beam incident on a simulated hemispherical block of homogeneous tissue, with a range of spot sizes and absorption rates.

Materials and Methods

Laser System and Preliminary Experiments

The laser system used in the first part of the study was the Panalas 4000 manufactured by Matsushita Electrical Company, Japan, and marketed by Mochida Pharmaceutical Company as the Luketron. This is a diode

Addressee for Correspondence:
Toshio Ohshiro MD, JMLL, Shinanomachi Station Building, 2nd Floor, 34 Shinanomachi, Shinjuku-ku, Tokyo, Japan 160.

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Fig 1: Black plastic pen beside probe of Luketron GaAlAs system probe, with thermal damage seen at arrow, following 5 s irradiation at 830 nm, 60 mW, c/w and a spot size of 200 μ m.

laser system, based on GaAlAs laser diodes, delivering convergent focused beams to a 200 μ m spot, with an output power of 60 mW, at 830 nm in continuous wave (c/w), which can be delivered continuously or chopped in a range of frequencies. In the present study the c/w setting was used.

It was noted from actual treatment sessions in the authors' clinic involving patients with dark-coloured naevi that during laser therapy with the Luketron system the patients reported feeling heat in the irradiated area. To test this, a black plastic pen was irradiated with the laser in contact mode c/w, for several seconds, and a spot of melted plastic 0.2 mm (200 μ m) in diameter was observed in the target area (Figure 1). It

Fig 2: Black-ink painted target on foot being irradiated by the Luketron diode laser therapy system.

was therefore presumed that the temperature in the irradiated area had risen to well over 100 °C.

An area in between the big toe and second toe on one of the author's (TM) feet was painted black using a black felt-tipped oil-based 'magic marker' pen, and was irradiated with the laser (Figure 2). As soon as laser radiation was started a distinctly painful burning sensation was felt. A radiant heat thermometer (CT-30, COS Corporation) was aimed at the target area, but despite the burning sensation still being felt, the overall rise in temperature was recorded as only 2 °C. The problem was the minute spot size (200 μ m) was not emitting enough radiant heat for the comparatively large thermometer detector to pick up. A computer program was therefore written to simulate the condi-

Fig 3: The hemispherical (left) and cubic (right) models for calculating heat rise and speed of conducted heat transfer in tissue volume.

Fig 4: Schematic representing application criteria of probe and tissue for this study. The probe is applied in the contact pressure technique, so as to eliminate blood system cooling from the hemispherical tissue volume of interest directly under the probe tip.

tions, which could then simulate data for a range of spot sizes, irradiation times and absorption rates.

Thermal Spread and Tissue Model

There are two models used to calculate the heat rise and spread of heat in a homogeneous material: the cubic model and the concentric hemispherical model, seen in Figure 3. In the case of the cubic model, the selected volume of tissue is divided into cubes and each individual cube is assigned 8 points, one for each corner. For each of these 8 points a program can be written to determine the rise in temperature based on the χ^3 model, and then applied to each of the cubes into which the tissue of interest has been divided. However the memory and computer speed requirements to run such a program are very large and fast, respectively, and would require the use of a supercomputer. The computer available for the present study was a standard personal computer (PC), (Sharp X68000). The hemispherical model was therefore chosen, which has a much more simple calculation requirement, as the hemisphere is divided into a series of equidistant concentric radii. The advantage of this model is the accepted assumption that the temperature at any point in any given segment is the same, and that the number of segments into which the area of interest can be divided is reduced. Once the thermal conductivity in the 'tissue' is decided, calculating the equal heat rise in each successive segment becomes somewhat less complicated and within the capabilities of a program which can be run on a PC.

Figure 4 shows how the laser probe was to be 'applied' to the 'tissue' in the chosen model. The con-

Table 1: Conditions of calculation.

Value	Conditions
Heat Conductivity of Target Tissue	The heat conductivity in tissue is unknown. The heat conductivity in water is 0.6 W/mK, and that of fat is 0.18 W/mK. For the purposes of this computer simulation, the hypothetical heat conductivity in the target tissue was thus set at 0.5 W/mK
Specific Heat of Target Tissue	As the specific heat of water is 4.18 J/gK, a hypothetical value for the specific heat of the target tissue was set at 3.5 J/gK
Target Hemisphere	Radius = 5 mm, in 50 equal divisions of 100 μ m each
Irradiation Time	5 s, divided into 5,000 equal 'shots' of 1 ms (0.001 s) for the purpose of the calculation
Heat Source	60 mW
Range of Spot Sizes	200 μ m, 100 μ m, 1.0 mm, 2.0 mm
Range of Absorption Rates	10%, 20%, 50%, 100%
Power Loss in Air	Neglected (heat conductivity is very low)
Cooling Effect of Blood	Considered only in temperature constant area
Temperature Constant Area	Radius > 5 mm from centre of target hemisphere

tact pressure method has been reported as providing maximum penetration into target tissue while minimizing losses by air diffusion and surface reflection. The hemisphere of tissue with a radius of 5 mm, seen directly under the probe tip, was the tissue volume of interest, and was divided into 50 equal concentric segments: i.e. each segment was 100 μ m thick. This was the tissue whose rise in temperature would be calculated. The tissue surrounding that hemisphere was assumed to be of constant temperature.

Calculations

The conditions of calculation are seen in Table 1.

The following equations were used in the calculation of the volume of thermal motion through the target hemisphere.

Heat Conduction per Time Segment (hc/ts)

$$hc/ts = \frac{DT \times CA \times HC \times T}{\text{Distance of Conduction}} \dots (1)$$

(where DT = difference in temperature in $^{\circ}$ K, CA = conduction area in cm^2 , HC = heat conductivity in W/mK, and T = time in seconds)

DT (°K) = Temp. of one segment (°K) - temp of next segment (°K)

$$TR = \frac{IHV - OHV}{\text{Heat Capacity}} \dots (2)$$

(where TR = temperature rise in a segment in °K, IHV = inlet heat volume in J, and OHV = outlet heat volume in J)

$$PTR = \frac{WRP \times T}{HC} \dots (3)$$

(where PTR = photoabsorptive temperature rise in °K, WRP = power of heat source in watts, T = time in s and HC = heat capacity in J/K)

$$HC = [V \bullet SG] \times SH \dots (4)$$

(where HC = heat capacity in J/K, V • SG = the product of the volume and specific gravity in g/m³, and SH = specific heat in J/gK)

$$\text{Conduction Area (m}^2\text{)} = \text{Area of Cross-Section of Hemisphere} \dots (5)$$

Computer Program

The program is shown in Figure 5. For those readers who wish to try the program, please note that 'n' must be replaced by the specific escape code characters to initiate a carriage return and line feed on the printer being used (for example in Japanese DOS-based systems the code is 'Yn';). The program was written in C language, so with the aid of a C compiler, this program will run on many other PC systems with only minute changes necessary.

Method

The program calculated and analyzed the temperature rise in an irradiated hemispherical block of homogeneous tissue (diameter of 10 mm) at various powers in the range of 6 mW - 60 mW, and at spot sizes of 200 µm, 400 µm, 1.0 mm and 2.0 mm following 5 seconds irradiation with an 830 nm GaAlAs diode laser, after which the laser was 'turned off'. The laser was deemed to be applied in the contact pressure mode, thereby compressing superficial blood vessels in the irradiated area so that blood flow cooling could be discounted. The results for all parameters were plotted together on a chart. The program then simulated the temperature distribution at the same spot sizes as above from the centre of the irradiated area to any point within 2 mm from the centre (i.e. a hemisphere with a radius of 2 mm). The simulation was then repeated with a constant spot size of 200 µm, but with absorption rates of 100%, 50%, 20% and 10%, and the results plotted as above over time and a distance of up to 1 mm from the centre of the hemisphere.

Results

Figure 6 shows the simulated temperature rise at the centre of the area of interest plotted over time (left hand graph) and temperature rise over distance from the centre of the irradiated area (right hand graph) for the range of spot sizes from 200 µm to 2.0 mm, with absorption constant at 100% in both graphs. At 60 mW and a spot size of 200 µm, a temperature rise of 200°C after 0.1 s was simulated, reaching a maximum temperature rise of approximately 272°C at the 5 s mark, at which point laser irradiation was stopped and the temperature returned rapidly to near the preirradiation level by 6 s. The larger spot sizes produced correspondingly lower rises in temperature, with a maximum rise of 18°C at the 2 mm spot. A dose-dependent temperature relationship was seen between the various spot sizes.

Figure 7 is the simulation of temperature rise plotted against absorption rates of 100%, 50%, 20% and 10% (left) and the rise in temperature within 1 mm of the centre of the irradiated area for the same absorption rates (right), with a constant irradiated spot size of 200 µm in both graphs. At 100% absorption, the temperature rise was approximately 272°C after 5 seconds, whereas at 10% absorption the temperature rise was 27°C.

According to this computer simulation, at maximum absorption, the 60 mW beam of the GaAlAs diode laser at 830 nm and at a 200 µm diameter spot size, which is similar to the spot size in contact mode of the Luketron system, has the potential for causing thermal injury.

Discussion

Reports have previously appeared on the thermal safety of prolonged LLLT levels of irradiation with a 60 mW, 830 nm c/w GaAlAs diode laser on the rat joint model *in vivo*.^(1,2) These reports examined 30 minutes irradiation of rat skin, subcutaneous tissue and exposed synovial membrane, but found no adverse effects, either in the gross morphology or the histology of the irradiated tissue. However, the spot size in these studies was approximately 6 mm in diameter, giving an incident power density of around 2 W/cm². The laser probe was thus at a set distance from the irradiated tissue, being used in the noncontact mode. Many reports have appeared where the contact mode and contact pressure method of diode laser therapy application,^(3,4,5) resulting in much smaller spot sizes from these focused systems. Because some patients reported feeling a distinct increase in temperature when the diode laser was used over a dark-coloured naevus, the authors were concerned about this aspect of LLLT

```

/*Program for calculation of temperature change in a laser-irradiated hemisphere.*/
#include <stdio.h>
#include <doslib.h>
#include <stdlib.h>
unsigned char infile[60],outfile[60],buff1[256];
double templ,SG,SH, power,heat_source_radius, dr,dt,rl;
double irradi_time,endtime,conducted_heat_rate;
double temp[250],j_rate[250],c_area[250],cap[250],om[250];
double density,time,temp_difference,energy,dr2,dr3,d[14]
int temp_const_area,finish,err=0,j,z,target_centre,heat_dist,countc,countb,i
FILE *fp;
FILE *fpo;
/***** program start *****/
void main(int,argc,char *argv[ ])
{
if(argc == 3){
printf("\nCOPYRIGHT (c) TAKASHI MAEDA MD");
setmem(infile,59,0);
setmen(outfile,59,0);
j=strlen(argv[1]);
strncpy(infile,argv[1],j);/*Set File Name*/
printf("\n");printf(infile); printf("reading \n");
j=strlen(argv[2]);
strncpy(outfile,argv[2],j);/*Set File Name*/
fp=fopen(infile,"rt"); /*read, open text mode*/
fpo=fopen(outfile,"wt"); /*open in write text mode*/
if(fp != 0 && fpo != 0){
read_parameter( );
fclose(fp);
fp=fopen(infile,"rt");
reset( );
header_write();
calculate( );
}
else printf("\n UNABLE TO READ FILE \n");
fcloseall( ); /*close all files*/
}
else {printf("\n AN ILLEGAL NUMBER HAS BEEN ENTERED FOR THE PARAMETERS \n");
Printf("\n Usage:");
Printf("\n This_program input_file output_file \n");
}
}
/*****subroutines*****/
int header_write( )
{
while( fgets(&buff1,255,fp)!=NULL){
fprintf(fpo,"%s",&buff1);
}
return(0);
}

```

Fig 5: Program used to produce simulated data.

(a) (part 1) Full program for calculation of tissue temperature rise for different spot sizes and absorption rates. Note this is in C language, and will require use of a C compiler to run on any PC.

```

}

int      read_parameter( )
{
char      L;
i=0;
while( fgets(&buff1,255,fp) !=NULL){
    L=strlen(buff1);
    buff1[L-1]=0;
    if(buff1[0] != 0X2D){      /*INITIAL CHARACTER "-" = 0x2D*/
        d[i] = atof(buff1);
        i++;
    }
    if(i==13)break;
}
temp1=d[0];
printf("\n INITIAL TEMPERATURE                =%E °C",temp1);
Conducted_heat_rate=d[1];
printf("\n CONDUCTED HEAT RATE                =%E W/mK",conducted_heat_rate);
SG=d[2]*1000000;
printf("\n SPECIFIC GRAVITY                    =%E g/m³",SG);
SH=d[3];
printf("\n SPECIFIC HEAT                        =%E J/gK",SH);
power=d[4];
printf("\n HEAT SOURCE POWER                    =%E W",power);
r1=d[5]/100;
printf("\n RADIUS OF HEAT SOURCE                =%E m",r1);
dr=d[6]/100;
printf("\n DIVISION LENGTH OF RADIUS IN CALCULATION=%E m",dr);
dt=d[7];
printf("\n DIVISION LENGTH OF TIME IN CALCULATION =%E s",dt);
irrad_time=d[8];
printf("\n IRRADIATION TIME                    =%E s",irrad_time);
endtime=d[9];
printf("\n CALCULATION END TIME                =%E s",endtime);
temp_const_area=d[10];
printf("\n CONSTANT TEMP AREA FROM CENTRE (SEGMENT NUMBER)=%d", temp_const_area);
target_centre=d[11];
printf("\n FREQUENCY OF WRITING TARGET CENTRE TEMPERATURE DATA = one per %d
TIMES\n",target_centre);
heat_dist=d[12];
printf("\n FREQUENCY OF WRITING HEAT DISTRIBUTION DATA = one per %d
TIMES\n",heat_dist);

return(0);
}

int      reset( )
{
long int      i2,i3,l;

```

Fig 5: (Continued)

(a) (part 2) Full program for calculation of tissue temperature rise for different spot sizes and absorption rates. Note this is in C language, and will require use of a C compiler to run on any PC.

```

dr2=dr*dr;
dr3=dr2*dr;
for(i=0;i;i++){
    j_rate[i]=0;
    temp[i]=temp1;
    i2=(i+1)*(i+1);
    c_area[i]=dr2*i2*6.28;
    i3=i2*(i+1);
    l=i3-i*i*i;
    cap[i]=dr3*l*2.09*SG*SH;
}
en=r1*r1*r1*6.28; /*volume*3*/
density=power*dt*3; /*energy*3*/
density=density/en; /*energy/volume*/
i=0;
while(r1 = (i+1)*dr){
    j_rate[i]=density/SH/SG; /*temperature rise in a*/
    z=i; /*period of time segment*/
    i++;
}
return(0);
int calculate( )
{
    end=temp_const_area+1;
    if(end > 240)end=240;
    && — and
    || — or

    countb=-1;
    countc=-1;
    while(time < endtime){
        countc++;
        countb++;
        for(i=0;i < 1;i++){om[i]=temp[i];
        if( target_centre==countc || heat_dist==countb){
            fprintf(fpo,"****NEXT**** \n %E seconds \n",time);}
        for (i=0;i < i++;){
            temp_difference=om[1]-om[i+1];
            if(temp_difference < 1e-100)temp_difference=0;
            en=temp_difference*c_area[i]*dt*conducted_heat_rate;
            en=en/dr;
            temp[i+1]=temp[i+1]+en/cap[i+1];
            temp[i]=temp[i]-en/cap[i];
            if(i==0 && target_centre==countc) || heat_dist==countb){
                fprintf(fpo "%E°C \n",temp[i]);}
            else {if(heat_dist==countb)fprintf(fpo,"%E°C\n",temp[i]);}
        }
        if(time
ad_time){

```

Fig 5: (Continued)

(a) (part 3) Full program for calculation of tissue temperature rise for different spot sizes and absorption rates. Note this is in C language, and will require use of a C compiler to run on any PC.

```

        for(i=0;iz+1;i++){
            temp[i]=temp[i]+j_rate[i];
        }
    }
    time=time+dt;
    temp[temp_const_area]=temp1;
    if(countc==target_centre) countc=0;
    if(countb==heat_dist) countb=0;
}
return(0);
}
/*program end*/

```

Fig 5: (Concluded)

(a) (part 4) Full program for calculation of tissue temperature rise for different spot sizes and absorption rates. Note this is in C language, and will require use of a C compiler to run on any PC.

- An input file example.
- Comment lines must have the initial character "-".
- All values must be written in the same order as below.
- There should be no changes in the number of input lines.
- Comment lines can be omitted or added as you like.
-
- initial temperature (*C)
- 0.
- heat conduction rate (W/mk) (Water = 0.6 W/mk)
- 0.5
- specific gravity (G/cm³) (Water = 1 G/cm³)
- 1
- specific heat (J/kg) (Water = 4.18 J/kg)
- 3.5
- power of heat source (W)
- 0.06
- radius of heat source (cm)
- 0.1
- segment of radius used in the calculation (cm).
- the smaller the value, the more accurate the result,
- but the longer the time required for the calculation
- 0.01
- segment of time used in the calculation (seconds)
- the smaller the value, the more accurate the result,
- but the longer the time required for the calculation
- 0.001
- exposure time to heat source (seconds)
- 5.0
- ending time for calculation (seconds)
- 6.0
- number of segments from centre of irradiated area to
- temperature constant area. (This value must be between
- 1 to 240. Any other value will require rewriting of the
- program
- 100
- frequency of writing centre temperature data to disk
- calculated per ?? times. The smaller this value,
- the greater the disk space requirement.
- 100
- frequency of writing temperature data of all segments to disk
- (distribution writing frequency)
- per ?? times calculation
- 1000

Fig 5: (continued) Program used to produce simulated data. (b) Sample input file.

application, and so they designed the study in this report.

Some authorities have stated that output powers of less than 100 mW can be classed as laser photobioactivation levels, or that such lasers can be classed as 'low power' systems. However, as has also been presented in a previous issue of this journal,⁽⁶⁾ even a 20 W CO₂ laser can be adjusted to produce 2 W/cm² in target tissue which is definitely a photobioactivative level, but the output power of the laser is still 20 W. This simply reminds the authors and hopefully the readers that any form of terminology which is based on the laser system or its output power is totally erroneous: it is the incident power at the tissue and the spot size which are of paramount importance, in other words, the power density, coupled with the irradiation time, giving the energy density. A 60 mW laser at a spot size of 200 µm gives an incident power density of approximately 191 W/cm². Given sufficient absorption, this incident energy density will produce thermal damage, as has been demonstrated both on a black plastic pen, a black-painted spot on the human foot, and in the computer simulation model.

100% absorption at 830 nm within a depth of 5 mm is almost impossible in the normal clinical setting: even the darkest naevus cell naevus will give an approximate maximum of 50% absorption at 830 nm, but over a deeper penetration pattern. Although slightly pigment specific, an 830 nm beam will usually penetrate well into target tissue, for centimetres rather than millimetres, producing deep absorption in most tissue types. This is due to this wavelength's poor absorption in water, and generalized absorption in the cell membrane, rather than in pigmented chromophores.⁽⁷⁾

Fig 6: Graphs plotted from the computed data to represent simulated temperature rise in tissue over time (left) and distance from the centre of the target tissue (right) for a range of spot sizes. At 5 s, the laser exposure was 'stopped'.

Fig 7: Graphs plotted from the computed data to represent simulated temperature rise in tissue over time (left) and distance from the centre of the target tissue (right) for a range of absorption rates.

For practical clinical indication, the first simulation (constant 100% absorption, varied spot sizes) points to the importance of limiting the incident power densities to well below 300 W/cm² in laser therapeutic applications, especially in darker pig-

mented target tissue, strictly to avoid the possibility of photothermal damage. The studies in the rat model quoted above used a power density of around 2 W/cm². The second computer simulation gave results for a range of superficial absorption rates from 10% to

100% at a constant spot size of 200 μm in diameter. However, in tests involving laser safety, it is always better to err on the side of caution. With a spot size of 1.0 mm, which is the contact spot size of the system used in the rat studies already mentioned,⁽²⁾ the incident power density at 60 mW output power is under 8 W/cm², roughly 24 times less than at a spot size of 200 μm . The ranges of tissue temperature rise will thus be proportionately much less for all absorption rates in practical clinical applications.

The authors would like stress that this is only a theoretical computer model, with hypothesized constants and a homogeneous simulated target material, but it would certainly appear to limit the application of 830 nm diode LLLT systems at the above parameters (200 μm spot size at 60 mW) on dark-colored naevi or for darker-skinned patients. With appropriate spot sizes and power densities, however, even with the strict criteria set by the simulation program, the temperature rise in irradiated tissue can be negligible. Above all, this model certainly demonstrates the error of applying arbitrary limits as to what constitutes the output power of an LLLT system, and stresses the importance of the power and energy densities in determining what is a photoactivative reaction in LLLT indications.

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