Abstract—A model to simulate heating as a result of pulse repetitions during infrared neural stimulation (INS), with both single and multiple emitters is presented. This model allows the temperature increases from pulse trains rather than single pulses to be considered. The model predicts that using a stimulation rate of 250 Hz, with typical laser parameters at a single stimulation site results in a temperature increase of 2.3 °C. When multiple stimulation sites are used in analogy to cochlear implants, the temperature increases further depending upon the spacing between emitters. However, when the light is more localised at multiple stimulation sites the temperature increase is reduced.

I. INTRODUCTION

Infrared neural stimulation (INS) is a novel technique allowing direct stimulation of neurons with infrared light [1]. It has many advantages over traditional electrical stimulation, including greater spatial resolution [2], it does not require direct contact and it does not leave an artefact on recording electrodes [3]. While there are other techniques to stimulate neurons which are not usually photosensitive, such as transfecting neurons with light sensitive channel gates, such as channelrhodopsin-2 (ChR2), to allow stimulation with light [4], [5], INS has the advantage of not requiring the modification of the target neurons. The detailed mechanism of INS is still under investigation [3], [6], [7]. However, INS appears to be mediated by water absorption of the laser light causing rapid localised heating [3], which drives a transient change in the membrane capacitance [6] and may activate the TRPV4 ion channels [7].

Cochlear implants are a successful bionic device using electrical neural stimulation and have shown the ability to restore hearing to profoundly deaf individuals [8]. The effectiveness of cochlear implants is, in part, due to the tonotopic mapping of the spiral ganglion neurons [9], which allows the targeting of neurons corresponding to high frequencies at the base of the cochlea and lower frequencies closer to the apex. While modern implants have up to 22 electrodes, most cochlear implant users respond as if only using 8 electrodes [10], [11] due to current spread between the electrodes and the targeted neurons [11]. While this response is adequate for speech perception in a quiet area, experiments with normal hearing patients have shown that up to 20 channels are required for good speech perception in noisy environments [10] and at least 32 are required for melody recognition [12]. Much research has been directed towards reducing the current spread from electrodes to focus stimulation on a smaller groups of neurons [13]. Techniques to localise the current include current steering through the use of multiple current sources and sinks such as tripolar stimulation [11], [14]. However there is little evidence that these techniques reduce the spatial extent of neural excitation [13] and it has not shown promise in clinical applications [8], [15]. Current focusing through the use of a phased array [16] is a similar technique that may overcome some of the limitations with current steering, however it still requires higher current levels compared to traditional monopolar stimulation.

Due to the limitations of current electrical stimulation techniques, there is interest in the use of INS to replace or complement traditional electrical stimulation. The spatial selectivity of INS potentially allows more stimulation sites than conventional cochlear implants, thereby improving hearing in recipients and generating a great deal of interest in INS of auditory nerves [1], [17], [18]. The safety of INS has been a topic of some discussion in the literature [19] [20] [21] [22], with work focused on studying the behaviour and histology of nerves exposed to INS at a high repetition rate. Izzo et al. [21] showed that INS at 13 Hz did not cause damage to the gerbil cochlea over 6 hours. More recently, Goyal et al. [22] demonstrated that there was no damage in a guinea pig cochlea for INS at 250 Hz to over 5 hours.

Modelling of INS has previously been performed through use of a Monte Carlo method [23] to determine the spatial distribution of absorbed light in tissue and finite element analysis [24] to investigate the temporal behaviour of the resulting heat distribution. The Monte Carlo model allows for temperatures to be predicted from different radiant exposures and fibre dimensions and suggests that the temperature resulting from single stimulation events in the cochlear can be as low as 0.1°C [23]. The analysis of temporal behaviour of INS through use of a finite element heat flow analysis suggests that there may be two broad regimes of INS [24]; further study of this aspect may help to optimise the process and reduce the energy required to stimulate neurons. The dynamic behaviour predicted from this combined Monte Carlo and finite element model of INS compares well to a range of experimental studies [24].

In this work, the finite element model presented in [24] has been extended to investigate pulse repetition rates that more closely simulate experimental use of INS, such as that of Goyal et al. [22] and future use with multiple stimulation sites.
II. MODELLING OF INFRARED NEURAL STIMULATION

The optical and thermal constants used in this work for absorption ($\mu_a$), scattering ($\mu_s$), anisotropy ($g$), refractive index ($n$), thermal diffusivity ($\alpha$), and heat capacity ($C_p$) are the same as used in [23] and [24]. The absorption coefficient ($\mu_a$) follows the absorption of water [25] as a first approximation. This is due to a lack of information regarding light absorption in tissue at the wavelengths typically used for INS. Similarly, thermal diffusivity ($\alpha$) is based on water’s value of $\alpha = 1.43 \times 10^{-6} \text{ m}^2\text{s}^{-1}$ (for all tissue and perilymph), as any reductions in thermal conductivity due to reduced water content are usually balanced by similar reductions in heat capacity [26]. The cochlea model from [23] and [24] is used, where a fibre is located in perilymph, 500 $\mu$m from the middle of a 100 $\mu$m thick nerve layer, with a 10 $\mu$m thick bone layer between the nerve and perilymph.

The Finite Element Analysis used was similar to that reported in [24], however a voxel size of 10 $\mu$m was used to allow a larger time step of $\Delta t = 100 \mu$s ($\Delta t_{\text{max}} = \frac{\Delta x^2}{6\alpha} = 116 \mu$s). This kept total computation time for 10 second simulations to practical limits and allowed a total simulation size of $2.56 \times 2.56 \times 2.56$ mm.

III. COMPARISON OF HEAT DISTRIBUTIONS

The temperature change from exposure to a short laser pulse, on the axis of the beam, can be considered though use of (1), a basic analytical solution for the temperature from a collimated beam and instantaneous pulse [20], [27]:

$$T(z, 0) = \frac{\mu_a H(z)}{\rho c}$$

where $\rho$ is the density of the medium, $c$ is the heat capacity and $H(z)$ is the radiant exposure at distance $z$. $H(z)$ is given by the Beer-Lambert law:

$$H(z) = H(0)e^{-\mu_a z}$$

where $H(0)$ is the radiant exposure at the fibre tip. The applicability of this solution for more complex geometries can be considered by comparing it to the output from the combined Monte Carlo (MC) and Finite Element (FE) heat conduction models.

Figure 1 shows the expansion of light from a fibre, in a medium with no absorption. Grey lines show the maximum angle of expansion in both directions. Near the fibre, there is a region of even energy, centred on the fibre axis where the radiant energy can be predicted by the Beer-Lambert law. In this region, the loss of light expanding in one direction is matched by light from the other. Beyond this region, the radiant energy is less than the absorption loss predicted by the Beer-Lambert law for a collimated beam. This transition point is given by $z_T = \frac{\theta_{\text{core}}}{\tan(\theta_{\text{eff}})}$, where $\theta_{\text{core}}$ is the fibre’s numerical aperture, $\theta_{\text{core}}$ is the fibre’s core diameter and $n$ is the refractive index of the media. When comparing between different fibre diameters, it is helpful to keep the transition point ($z_T$) constant. A constant transition point ensures that differences in temperature at the neurons are not due to varying radiant exposures, but are driven by changes in the localisation of light. It is also worth noting that this effect will be more pronounced in air, as the effective NA of a fibre in water is given by $N_{A,\text{eff}} = \frac{N_A}{n}$.

This localisation and transition can be seen more clearly by plotting the temperature along the fibre axis expected due to an instantaneous pulse. Figure 2 shows the temperatures predicted by Eq. (1) and for the Monte Carlo model using 100 $\mu$m and 200 $\mu$m core diameter fibres with NA of 0.11 and 0.22 and refractive index of $n = 1.33$.

The results in Figure 2 show that a fibre with a core diameter of 100 $\mu$m has similar localisation compared to a fibre with a core diameter of 200 $\mu$m and NA = 0.22 when the 100 $\mu$m core diameter fibre’s NA is reduced to 0.11. While it may not be practical to specify fibre design so precisely, similar effects to varying the NA can be achieved by use of lenses to focus or disperse light. To allow convenient comparison between different fibre core diameters, the NA can be selected to keep...
$z_T$ constant. As a $\varphi_{\text{core}} = 200 \, \mu m$, NA = 0.22 fibre been widely used in the cochlear INS literature [1], [2], [18], [21], for other fibre core diameters used in this paper the NA has been selected to keep $z_T$ constant at $z_T = 600 \, \mu m$. The NA is given by:

$$\text{NA} = n \tan^{-1} \left( \frac{\varphi_{\text{core}}}{2z_T} \right)$$ (3)

Optimising fibre design and stimulation wavelengths can assist in designing an emitter which ensures stimulation is localised in the longitudinal direction as well as the transverse, while also not requiring very small distances between the emitter and target nerves as the case when using more strongly absorbing wavelengths (e.g. 1937 nm [20]). Careful use of modelling when designing optically emitters for INS allows localisation of light regardless of the wavelength and allows the risk of cross talk between different channels and across the cochlea spiral to be evaluated.

IV. EFFECT OF STIMULATION RATE

Current cochlear implants use an electrical stimulation rate of up to 900 Hz [8]. Due to modulation, the average rate may be lower, as the full rate is only used in bursts and depends on the frequency spectrum of the audio input [8]. Studies of INS in the cochlea have used stimulation rates of between 2 Hz and 250 Hz [1], [18], [21]. Izzo et al. (2007) [21] used repetition rates of up to 13 Hz in a gerbil cochlea found no change in CAP response after 6 hours of continual exposure. Similarly, Rajguru et al. (2010) [18] excited a cat cochlea for up to 10 hours at 200 Hz and did not observe a reduction in CAP amplitude or any damage to the spiral ganglion neurons in histology samples. A recent mini-review of the literature by Goyal et al. [22], additionally shows that pulse repetitions of up to 250 Hz can be used without any reduction in response amplitude or evidence of histological damage over periods of 6 hours. Typically, in current electrical implants, stimulation rates near 900 Hz are desired, as some patients report unwanted pitch perceptions such as buzzing when using stimulation rates in the range of 100 – 250 Hz [28].

Figure 3 shows an example of the heating in the spiral ganglion neurons modelled with a 200 $\mu m$ core fibre located 500 $\mu m$ from the neurons, pulse energy ($E_{\text{pulse}}$) = 25 $\mu J$, pulse length $t_p = 100 \, \mu s$ with repetition rates of 10 Hz, 50 Hz and 100 Hz. These laser parameters have been shown to generate action potentials in a number of studies [22]. For comparison the heating from a single pulse is 0.11°C. After 1 second, the 10 Hz, 50 Hz and 100 Hz pulse rates have maximum temperatures of 0.17°C, 0.51°C and 0.95°C respectively. The increase in baseline temperature is similar to that observed by Wells et al., where a two-fold increase in the peak temperature was observed when stimulating an exposed nerve at 5 Hz from a 600 $\mu m$ core diameter fibre [3], despite the different geometry and fibre diameter modelled here. We note that INS in the cochlea requires lower radiant exposure to generate a response compared to INS in peripheral models [1]. As such, the temperatures predicted by the model here are significantly lower than would be expected for peripheral stimulation at similar rates.

Figure 4 shows the peak temperature reached after 10 seconds, normalised against the peak temperature achieved by a single pulse, in fibre core diameters of 100 $\mu m$, 200 $\mu m$ and 400 $\mu m$ with NA selected to keep the transition point distance constant at $z_T = 600 \, \mu m$. The model was allowed to run to a time of 10 seconds, as the peak temperature was found to have reached a plateau and stabilised. Stimulation pulse rates of 1, 2.5, 5, 10, 25, 50, 100, 250 and 1000 Hz were selected to show the change from the lower stimulation rates used in some studies (eg 13 Hz [2], [21]) to higher rates (eg 250 Hz [18], [22]).

Figure 4 shows that as the frequency of stimulation increases, so does the peak temperature and that smaller core diameter fibres show less of an increase in temperature when
Normalised Peak Temperature

While the wavelength used is different (Goyal et al. [22]) which were not found to cause damage. The model was modified to allow the effective separation greater. To investigate the effects of multiple stimulation sites, the model was modified to allow for stimulation at three simultaneous sites, spaced at a pitch of 250 Hz, with a single stimulation site, the peak temperature after 10 seconds is 20.7 times a single pulse’s peak temperature. When three emitters are used the ratio of peak temperatures becomes 24.9, 29.3 and 39.3 for separation distances of 750 µm, 500 µm and 250 µm respectively. Frequencies higher than 250 Hz are not shown as peak temperatures continue to increase linearly.

The temperature increase for multiple emitters can also be considered in terms of frequency. The frequencies which display the same temperature increase of 20.7 relative to a single pulse are 207 Hz, 175 Hz and 130 Hz for the respective separation distances of 750 µm, 500 µm and 250 µm. Additionally, when using a separation distance of 250 µm there is significantly more heat build up between the emitters, which may cause stimulation in regions not directly targeted by INS.

V. EFFECT OF STIMULATION SITE SPACING

Current cochlear implants use electrodes spaced at a pitch of at least 0.75 mm [8], although current spread in tissue may make the effective separation greater. To investigate the effects of multiple stimulation sites, the model was modified to allow for stimulation at three simultaneous sites, spaced at a pitch between 0.25 mm and 0.75 mm. The following parameters were used: λ = 1850 nm, NA = 0.22, \( \phi_{\text{core}} = 200 \) µm, pulse energy = 25 µJ and pulse length = 100 µs.

![Figure 5](image-url) Peak temperature after 10 s of stimulation, normalised against the peak temperature achieved by a single pulse in a single fibre and when three fibres are positioned 750 µm, 500 µm and 250 µm apart. When multiple stimulation sites are positioned closer, the peak temperature rises. At 250 Hz, with a single stimulation site, the peak temperature after 10 seconds is 20.7 times a single pulse’s peak temperature. When three emitters are used the ratio of peak temperatures becomes 24.9, 29.3 and 39.3 for separation distances of 750 µm, 500 µm and 250 µm respectively. Frequencies higher than 250 Hz are not shown as peak temperatures continue to increase linearly.

The increase in temperature due to emitter spacing can be seen more clearly by plotting the temperature ratio against the emitter spacing. Figure 6 show the ratio of peak temperature after 10 seconds of stimulation to that of a single pulse when using a fibre with a core diameter of 200 µm, λ = 1850 nm and fibre core diameter of 200 µm was used. The temperature was calculated in front of the the central emitter.

![Figure 6](image-url) Peak temperature after 10 s of stimulation, normalised against the peak temperature from a single pulse, as a function of varying separation between three emitters with a stimulation rates of 50 Hz, 75 Hz, 100 Hz and 125 Hz. A wavelength of \( \lambda = 1850 \) nm and fibre core diameter of 200 µm was used. The temperature was calculated in front of the central emitter.

Figure 7 shows the temperature at the target nerves and at sites 750 µm, 500 µm and 250 µm laterally displaced from the centre of the beam for a single emitter. The temperatures outside the directly exposed region do not display a rapid increase in temperature.

![Figure 7](image-url) Temperature at the target nerves and at sites 750 µm, 500 µm and 250 µm laterally displaced from the centre of the beam for a single emitter. The temperatures outside the directly exposed region do not display a rapid increase in temperature.
increase in temperature. As INS is thought to be dependent on a temperature gradient in time [6], this indicates that stimulation is likely to be localised to the region exposed to the laser radiation. Sites further from the stimulation show reduced temperature increases. After 10 seconds, the targeted nerves reach a peak temperature of 0.14°C, while non-stimulated sites 750 μm, 500 μm and 250 μm away increase by only 0.023°C, 0.010°C and 0.005°C respectively.

![Temperature Gradient](image)

**Fig. 7.** Temperature at nerve 500 μm from fibre (core diameter = 200 μm, NA = 0.22, \( E_{\text{pulse}} = 25 \ \mu J \)) with a pulse rate of 5 Hz. Also shown is temperatures at sites 750 μm, 500 μm and 250 μm without stimulation. Temperature increase from a single pulse = 0.11°C, peak temperature after 10 s = 0.157°C.

Figure 8 shows three different modulation cases compared to a 100% duty cycle pulse stimulation pattern, with the same parameters as used previously (\( E_{\text{pulse}} = 25 \ \mu J, \lambda = 1850 \ \mu m, t_p = 100 \ \mu s \)). Three different modulations schemes are used: one where all three emitters fire simultaneously but at a rate of one third that of the full rate, giving a 33% duty cycle (Fig. 8, 33% DC); one where each emitter fires in turn, giving an average rate duty cycle of 33% (Fig. 8, 1/3 Modulation); and a burst rate, where each emitter fires 10 times at the stimulation frequency before the next emitter, again giving an overall duty cycle of 33% (Fig. 8, 10/30 Modulation). From the results in Figure 8, there is no difference between the modulation cases where the emitters fire simultaneously or once each; for these cases the overall temperature change is the same as a stimulation rate of one third. However, the burst stimulation shows a larger temperature increase. This shows that the thermal load of both long term average and burst stimulation rates need to considered in the design of any INS based implant.

![Stimulation Frequency vs. Normalised Peak Temperature](image)

**Fig. 8.** Peak temperature after 10 s, normalised against the peak temperature from a single pulse, for different modulation schemes after 10 s, for three emitters separated by 750 μm. See text for description of modulation schemes.

**VI. DISCUSSION**

When investigating laser-tissue interactions, transfer of heat due to convection is usually not considered due to the low perfusivity in most tissues, resulting in little heat transfer during the interaction time [27]. However, as the duration of repetitive INS considered in this work (10 s) is much greater than a typical laser-tissue interaction (< 1 s), it is worth considering the impact of convection driven heat transfer during INS. The perfusion of blood and other fluids in the target, such as the spiral ganglion, is applicable to all cases of INS. However, during cochlear stimulation, the natural flow of perilymph through the scala tympani and the formation of convection cells in the scala tympani perilymph due to heating from the laser may also provide convective heat transfer.

Natural blood flow inside the cochlea could potentially cool the spiral ganglion neurons by transferring the heat to the rest of the body. While blood flow in the cochlea has been studied in some detail [29], only relative flow is typically measured [30]. Lacking blood perfusion rates for the cochlea, typical values for general tissue are used instead as a first approximation. Niemz [27] provides a range of 0.46 - 1.0 mLg\(^{-1}\)min\(^{-1}\) for the brain. Assuming a density of 1 g mL\(^{-1}\) for tissue, an irradiated volume of \(1.25 \times 10^{-4}\) mL, temperature increase of 3°C and volumetric heat capacity of 4.18 JmL\(^{-1}\)K\(^{-1}\), the potential convective energy flow rate due to perfusion can be found using \(E = \Delta T C_p F_v = 1.58 \ \text{mJmin}^{-1}\) for 25 μJ pulses at a 250 Hz repetition rate. This potential convective energy flow is low compared to the energy introduced by the laser and conducted out of the irradiated region, \(E = 600 E_{\text{pulse}} = 375 \ \text{mJmin}^{-1}\). Therefore, the perfusion of blood through the cochlea is unlikely to provide a significant cooling effect for high repetition rate stimulation.

When the cochlea is undamaged, the volume flow of fluid (i.e. perilymph) in the scala tympani is very low [31], experimental measurements report flows in the range of 1.6 nLmin\(^{-1}\) [32]. However, in a damaged cochlea, this can rise to 0.5 μLmin\(^{-1}\) [33]. The perilymph flow rate for an undamaged cochlea is too low to provide significant cooling and will have less influence on temperature achieved than blood flow (\(E \approx 13 \ \mu \text{Jmin}^{-1}\)). In a damaged cochlea, the perilymph flow can be significantly higher, so the convective cooling from this process may be greater than that due to blood flow. For a flow rate of 0.5 μLmin\(^{-1}\), the perilymph could allow heat transfer of \(E = 4.2 \ \text{mJmin}^{-1}\), which is still significantly lower than the loss due to conduction. Therefore, it is unlikely that the temperatures achieved in the experiments
discussed by Goyal et al. [22] are lower than that predicted by the model due to perilymph flow out of the cochleostomy. However, care must be taken when extrapolating the differences due to perilymph flow, as experimental arrangement may significantly change this flow rate and resultant temperatures.

Another possibility worth exploring is that heating from the laser could generate convection cells inside the scala tympani perilymph and could also assist in cooling. Convection cells are strongly dependent on geometry and orientation, as they are driven by differences in buoyancy [34]. Therefore, it is difficult to model a realistic case as the geometry will vary between experimental subjects and neuronal targets. However, the feasibility of this process can be estimated by taking a case where convection is most likely to occur. The Rayleigh number (Ra) describes the primary form of heat transfer in a fluid. When the Rayleigh number is below the critical Rayleigh number for the fluid most heat transfer will be conductive, but when it is greater than the critical value, convection can play a role in heat transfer. The Rayleigh number is given by:

\[ Ra = \frac{g \beta}{\nu \alpha} (T_s - T_\infty)L^3 \]  

where the acceleration due to gravity \( g = 9.8 \text{ m/s}^2 \), \( \beta = 3.62 \times 10^{-4} \text{ °C}^{-1} \) is the thermal expansion coefficient of water (at 37 °C) [34], \( \nu = 6.61 \times 10^{-7} \text{ m}^2\text{s}^{-1} \) is the kinematic viscosity of water (at 37 °C) [34], \( \alpha = 1.43 \times 10^{-7} \text{ m}^2\text{s}^{-1} \) is the thermal diffusivity, \( T_s - T_\infty \approx 3^\circ \text{C} \) is the temperature difference between baseline temperature and the maximum temperature due to the laser and \( L \) is the distance between the heated fluid and the boundary of the fluid and is between 200 \( \mu \text{m} \) and 500 \( \mu \text{m} \), depending on the location of the stimulation in the cochlear spiral. For these values, a Rayleigh number of \( Ra = 0.901 \) for \( L = 200 \mu \text{m} \) and \( Ra = 14.1 \) for \( L = 500 \mu \text{m} \) is found. The critical Rayleigh number for water depends on the temperature and geometry, however even a lower bound value of \( Ra_c = 1700 \) [35] is still much greater that found for than the Rayleigh number found temperatures and geometry relevant to INS in the cochlea. Therefore, thermally driven natural convection is unlikely to provide a significant cooling mechanism during INS of the cochlea.

Although the model, as presented, has focused on the cochlea, INS has also been applied to other target nerves, such as peripheral nerves [3] and the brain [36]. The investigation of high repetition rates may also be important in these other modalities, where INS typically requires larger radiant exposures to achieve stimulation [1]. However, it is expected that the trends predicted here should extend to higher radiant exposures and other geometries. The most significant difference will be due to the presence of an air-tissue interface, from the nerve being partially exposed to the air (e.g. Ref. [3]) which will reduce the conduction of heat. There may also be an increase in convective losses from both circulation of air and evaporation of water in the tissue.

To test the model for INS in cases where an air-tissue interface is present, the model was modified to simulate an optical fibre positioned 100 \( \mu \text{m} \) away from the tissue, in air. To ensure that the difference reported by the model was due to thermal conduction and not a different beam profile, the absorbed heat profile was kept the same as the one used in the water-tissue geometry. The thermal diffusivity of the glass fibre was set to \( \alpha = 3.7 \times 10^{-7} \text{ m}^2\text{s}^{-1} \) and \( \alpha = 1.0 \times 10^{-5} \text{ m}^2\text{s}^{-1} \) for air. The initial temperature of the air and fibre was assumed to have stabilised at the same temperature as the tissue. Using a 200 \( \mu \text{m} \) core diameter fibre with stimulation rate of 50 Hz, no significant difference was observed between the air-tissue and water-tissue geometries at tissue depths greater than 250 \( \mu \text{m} \). However, a slightly higher temperature was observed in the air-tissue geometry at reduced depths. At the smallest distance examined (50 \( \mu \text{m} \)), the normalised peak temperature increased by 7.2% from 3.04 to 3.26 after 10 s of stimulation.

To evaluate the effect of evaporation on the resulting temperatures, Stelling’s formula for evaporation can be used [37], [38] to find the heat loss through vapourisation. Using the formula and constants provided in Refs. [37], [38] with tissue in a 400 \( \mu \text{m} \) radius heated to 60 °C (air temperature 25 °C, RH = 25%) an evaporative cooling rate of 461 mJ/min or 27.7 mJ/min⁻¹ is found. As a laser heating rate of 375 mJ/min⁻¹ gives a temperature increase of only 2.3 °C at 500 \( \mu \text{m} \) depth), it is therefore reasonable to conclude that heat loss through evaporation makes a negligible contribution when tissue temperatures remain at safe levels.

Another source of heat loss specific to peripheral nerves and brain stimulation experiments is the loss to forced convection, either from circulation of cooler room temperature air or addition of saline solutions to prevent the tissue from dehydrating (e.g. Ref. [39]). Given the high thermal diffusivity of air relative to water and tissue, it is likely that the air temperature will stabilised close to the tissue temperature, unless there is strong airflow present. For the saline solution case, unless the solution is continuously flowing over the tissue it is unlikely to provide a major source of cooling to the tissue during stimulation. However, cooling of the tissue to below body temperature may contribute to the weaker response at the start of pulse trains as observed by Duke et al. [39].

Overall, the combination of different heating and cooling effects from an air-tissue interface suggests that, although the temperature increase may be up to 10% higher in more exposed neural tissue, other convective factors may reduce the tissue temperature over longer time periods. Despite these factors, the results suggest that the general trends observed for increasing pulse rates and different fibre diameters will be similar in both the water-tissue and the air-tissue cases.

While the 100 \( \mu \text{m} \) core diameter fibre could achieve stimulation rates around 2.5 times higher than the 200 \( \mu \text{m} \) fibre with the same thermal load, the reduction in the volume of neurons activated may impact loudness perception, as loudness may be partially encoded by the number of neurons stimulated [40], [41]. Reductions in loudness perception due to the improved spatial localisation of stimulation in an optically based cochlear implant may need further investigation.

The lack of crosstalk in the model supports the conclusions of Richter et al. [2] who found that stimulation from INS is highly localised and that the spread of activation is similar to that produced by acoustic tone pips. Therefore, an optically
based implant would not suffer from spread of excitation and with the same 750 μm emitter/electrode spacing could provide the 20 independent channels required for speech perception with a noisy background environment [10]. With a reduced 500 μm emitter spacing, 32 independent channels could be produced which could allow music perception [12]. While there are a number of technical hurdles required to produce an optically based cochlear implant, it potentially offers a significant improvement in cochlear implant performance and far greater spectral information when compared to the effective 8 channel limitation in current implants [10], [11].

Although the model presented here is focused on INS, it may also be of use for more general pulsed laser stimulation techniques, such as optogenetics, with the selection of appropriate physical constants for those wavelengths. Although optogenetics typically uses significantly less energy than INS [4], new optogenetic channels with higher maximum pulse rates are being developed [42] and thermal effects may need to be considered.

VII. CONCLUSION

The modelling presented predicts a significant increase in temperature in the tissue of the cochlea during INS with a high stimulation rate. This temperature may be detrimental during chronic usage and shows the need for more detailed studies for stimulation rate. This temperature may be detrimental during temperature in the tissue of the cochlea during INS with a high

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VII. CONCLUSION

The modelling presented predicts a significant increase in temperature in the tissue of the cochlea during INS with a high stimulation rate. This temperature may be detrimental during chronic usage and shows the need for more detailed studies for devices with multiple excitation sources. When multiple site stimulation is used, the temperature tends to increase further, with multiple emitters separated by 750 μm leading to an additional 20% increase in temperature over a single emitter. On the other hand, the model indicates strong localisation of light and resultant heat, which suggests the potential of an optically based cochlear implant to greatly improve the performance and increase the number of perceived channels. It also shows the possible need to reduce the total optical energy delivered to a neural system, either through use of a hybrid optical/electrical stimulation technique or by introducing a selective absorber of light to neurons.

Future work planned includes a more detailed validation of the model, including the differences between varying stimulation geometries.

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