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THERMAL SAFETY OF VIBRO-ACOUSTOGRAPHY USING A CONFOCAL TRANSDUCER

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Abstract

Vibro-acoustography (VA) is an imaging method that forms a two-dimensional (2-D) image by moving two cofocused ultrasound beams with slightly different frequencies over the object in a C-scan format and recording acoustic emission from the focal region at the difference frequency. This article studies tissue heating due to a VA scan using a concentric confocal transducer. The three-dimensional (3-D) ultrasound intensity field calculated by Field II is used with the bio-heat equation to estimate tissue heating due to ultrasound absorption. Results calculated with thermal conduction and with blood perfusion, with conduction and without perfusion and without conduction and without perfusion are compared. Maximum heating due to ultrasound absorption occurs in the transducer's near-field and maximum temperature rise in soft tissue during a single VA scan is below 0.05°C for all three attenuation coefficients evaluated: 0.3, 0.5 and 0.7 dB/cm/MHz. Transducer self-heating during a single VA scan measured by a thermocouple is less than 0.27°C.

Keywords

Vibro-acoustography; Tissue heating; Bio-heat equation

INTRODUCTION

Elasticity imaging allows the study of tissue mechanical properties such as stiffness, which is related to tissue pathologies. Vibro-acoustography (VA) is an imaging method that is sensitive to the elasticity of the object. This method uses two cofocused beams of ultrasound with slightly different frequencies to “probe” the object (Fatemi and Greenleaf 1998). As shown in Figure 1, a confocal transducer (a fixed focus, concentric, two-element annular array) can be used to produce these cofocused beams. The object vibrates at the difference frequency of these two ultrasound beams and the resulting acoustic emission (in kilohertz range), which contains information about tissue stiffness at the probed location, is recorded by an acoustic hydrophone. By scanning the object in a raster C-scan format, a two-dimensional (2-D) vibro-acoustic image of the object can be formed. Vibro-acoustography is useful for many imaging applications such as breast, prostate and artery (Alizad et al. 2004, 2006, 2008; Fatemi et al. 2003; Pislaru et al. 2008).

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Ultrasound absorption by tissue can lead to significant heating and therefore should be evaluated for this ultrasound imaging method. It is important to note that VA imaging involves raster C-scan and usually the imaging transducer is mechanically translated during the imaging procedure. C-scan format was not considered in the standard soft tissue scanned mode thermal index model (Curley 1993). Therefore, it is necessary to evaluate ultrasound heating specific to VA imaging by accounting for the 2-D raster scanning of the ultrasound beam during a C-scan. In this article, we simulate tissue heating due to ultrasound absorption in a typical VA scan using the confocal transducer. The 3-D intensity field of an ultrasound beam transmitted by the confocal transducer is calculated by Field II (Jensen and Svendsen 1992). This 3-D intensity field is then used with the bio-heat transfer equation (NCRP 1992; Pennes 1948) to simulate tissue heating and cooling during and after the VA scan. Tissue heating at three different attenuation levels (0.3, 0.5 and 0.7 dB/cm/MHz) is studied. Results with thermal conduction and blood perfusion, results with conduction and no perfusion and results without conduction and without perfusion are compared. In addition, transducer self-heating during a VA scan is measured experimentally with a thermocouple in this study.

It is important to note that the purpose of this study is to evaluate the thermal safety of VA. Our primary goal is to estimate the upper limit of tissue heating in the worst case scenario: the method should be safe if worst case heating is below the acceptable threshold. Precision of temperature estimation is of secondary importance here. Therefore, the method and simulation parameters used in this study are generally very conservative. For example, in simulation of tissue heating due to ultrasound absorption, insulating boundary condition is used. The extreme case of ultrasound heating without thermal conduction and blood perfusion is also evaluated in addition to the more realistic estimation with both conduction and perfusion. Although these analyses may not be the most realistic, they always overestimate temperature rise and, therefore, are consistent with the purpose of this study.

METHODS

Heating in soft tissue during a VA scan can be caused by absorption of ultrasound and by transducer self-heating. These two heating sources are considered separately in this study. Heating due to ultrasound absorption is estimated using computer simulation, while transducer self-heating is measured experimentally. Total heating can then be estimated from these two sources.

Heating due to ultrasound absorption

FIELD II (Jensen and Svendsen 1992), a linear acoustic field simulation software program, is used to calculate the ultrasound intensity field generated by the confocal transducer in VA imaging. The confocal transducer has a center frequency of 3 MHz, a diameter of 45 mm and spherically focuses at 70 mm. Because the frequency of the two ultrasound beams in VA (f_1 and f_2 in Fig. 1) only differs by about 1%, the intensity fields at frequencies f_1 and f_2 should be very similar. Therefore, we approximate both intensity fields (at f_1 and f_2) with the intensity field calculated at the mean frequency $(f_1 + f_2)/2$ (i.e., 3 MHz for the confocal transducer). Intensity fields calculated by FIELD II for all three attenuation levels evaluated (0.3, 0.5 and 0.7 dB/cm/MHz) are individually scaled to appropriate peak *in situ* intensity values derated at these attenuation levels when the confocal transducer is driven by maximum voltage of the current VA system. The spatial-peak pulse-average intensity (I_{sppa}) of the confocal transducer measured by a calibrated hydrophone (HGL-0200, Onda Corporation, Sunnyvale, CA, USA) in water at maximum system output is 162 W/cm². Assuming an uncertainty of about 25% for our intensity measurement, the I_{sppa} of the confocal transducer in water is conservatively set to 200 W/cm² in this study. Therefore, the peak *in situ* I_{sppa} is derated to be 46.9, 17.8 and 6.8 W/cm² for attenuation level of 0.3, 0.5, and 0.7 dB/cm/MHz, respectively.

Tissue heating is estimated by the bio-heat transfer equation:

$$\dot{T} = \kappa \nabla^2 T - \frac{T}{\tau} + \frac{q_v}{c_v}, \quad (1)$$

where T (°C) is temperature, $\dot{T}$ (°C/s) is the time rate change in temperature, κ (cm²/s) is the thermal diffusivity for soft tissue, τ (s) is the perfusion time constant, q_v (W/cm³) is the rate of heat production per unit volume, and c_v (4.2 J/cm³/°C) is the heat capacity per unit volume for soft tissue (NCRP 1992; Pennes 1948). The heat source function for ultrasound is (Palmeri and Nightingale 2004):

$$q_v = 2\alpha I, \quad (2)$$

where α (Np/cm) is the absorption coefficient for soft tissue and I (W/cm²) is the ultrasound intensity field calculated by FIELD II as described above. The total attenuation in tissue results from the combined losses due to absorption and scattering. In the following calculations, the absorption coefficient α is set to be equivalent to the attenuation coefficient used to derate ultrasound intensity for all three attenuation levels evaluated (*i.e.*, 0.3, 0.5 or 0.7 dB/cm/MHz). This approach assumes that all ultrasound losses turn into heating. Therefore, this is a conservative assumption that leads to overestimation of tissue heating by ultrasound.

Time evolution of the temperature field with thermal conduction and without blood perfusion is simulated by the finite element method (FEM). The thermal diffusivity for soft tissue κ is set to be 0.00143 cm²/s (Palmeri and Nightingale 2004) in these simulations. Because of the linearity of the bioheat equation, the effect of blood perfusion can be added by multiplying the finite element solution a posteriori with a factor of $\exp(-t/\tau)$ (Yeung and Atalar 2001). This approach reduces the computation load: the same FEM results can be used to calculate heating at different blood perfusion levels. The perfusion time constant τ is set to 2000 s, which is consistent with the values of fat and resting muscle (NCRP 1992; Yeung and Atalar 2001). As we will see in the simulation results, maximum tissue heating in VA occurs in the near-field close to the transducer surface. Therefore, selection of τ relevant to muscle and fat is appropriate.

A heat transfer finite element code based on the C++ numerical library Diffpack (inuTech GmbH, Nuernberg, Germany) is used in the finite element simulations. Insulated (*i.e.*, zero flux) boundary conditions are used along all boundaries in the domain. The top boundary is the transducer-tissue interface (refer to Fig. 2 for the orientation/geometry of these simulations). Since axial symmetry is assumed, only half of the 2-D axial plane (along the transducer axis and toward one side of the transducer) of the transducer is modeled and the results are rotated around the ultrasound beam axis to obtain the 3-D heating field. Bilinear quadrilateral elements are used for the spatial discretization and a backward Euler scheme is used for the time integration. The mesh discretization consists of a structured mesh with 120 elements in the direction perpendicular to the ultrasound beam axis and 380 elements along the ultrasound beam axis for a total of 45,600 elements. Each element is 0.5 mm by 0.5 mm. Therefore, the finite element simulation covers a 6 cm (perpendicular to transducer axis) by 19 cm (along transducer axis) area. Due to axial symmetry, we essentially have data for a 12 cm by 19 cm area. Simulation results indicate that heating toward the lateral and bottom boundaries is negligible. We, therefore, only use the center region (6 cm by 10 cm) of the 12 cm by 19 cm area in the final heating calculation. The lateral span of the region (6 cm) is still larger than the diameter of the confocal transducer (4.5 cm) and the axial span of the region

(10 cm) is also larger than the focal depth of the transducer (7 cm). Heat fields calculated by FEM with a spatial resolution of 0.5 mm are then interpolated to 0.25 mm resolution.

Besides the finite element analysis, we also calculate tissue heating in the absence of both conduction and perfusion, where eqn (1) simplifies to:

$$\dot{T} = \frac{q_v}{c_v}, \quad (3)$$

The solution to eqn (3) provides a linear relationship between the increase in temperature ΔT (°C) and heat application time t (s):

$$\Delta T = \frac{q_v}{c_v} t. \quad (4)$$

From eqns (2) and (4), the temperature increase can be written as (Palmeri and Nightingale 2004):

$$\Delta T = \frac{2\alpha I}{c_v} t. \quad (5)$$

Equation (5) states that heating is proportional to pulse-average ultrasound intensity and exposure time. Therefore, when thermal conduction and blood perfusion are neglected, the intensity field calculated by FIELD II can be directly used with eqn (5) to estimate tissue heating. After the ultrasound is turned off, the heat field will not change with time because there is no conduction or perfusion. Finite element analysis is not required in this approach and computation is simplified significantly. Tissue heating calculated using this simplified approach is very conservative and will be compared with results including conduction and perfusion calculated with finite element analysis.

The VA scanning parameters studied in this paper are: scanned area = 40 mm × 40 mm, scan resolution = 0.25 mm, ultrasound exposure time for each pixel = 0.1 ms, scan speed = 2 ms per pixel. With these parameters, a single VA scan lasts about 52 s. VA scanning with the confocal transducer is achieved by mechanically moving the transducer in 2-D (sequential raster scan). The total heating due to a complete VA scan is, therefore, calculated by summing heat fields associated with each ultrasound beam focusing at each pixel of the 2-D VA scanning grid. Because each VA image pixel corresponds to a different spatial location and is imaged at a different time instance, its heat field is selected from the appropriate time instance of the FEM simulation and spatially shifted by the proper amount in the summation. Heat fields calculated by FEM (with a spatial resolution of 0.5 mm) are interpolated to 0.25 mm resolution before the summation.

Heating due to transducer

A 30 gauge type T thermocouple (Omega Engineering Inc., Stamford, CT, USA) is sandwiched between the confocal transducer and a porcine muscle sample (12 cm by 29 cm by 6 cm) to evaluate transducer self-heating. A thin layer of ultrasound gel provided acoustic coupling between the transducer and the porcine muscle sample. According to the VA scanning parameter specified above, the confocal transducer is turned ON for 0.1 ms in every 2 ms, with a total of 25921 (161 × 161) transmissions. Temperature rise is recorded for 51.84 s (duration of a single VA scan) with a sampling frequency of 100 Hz. To be conservative, the confocal transducer does not scan and stays in the same location during the whole VA “scan” (51.84 s).

Therefore, this experiment overestimates transducer heating because in a real VA imaging process, heat generated by the transducer is dissipated over a larger area due to transducer scanning. Five repeated measurements are taken at different locations across the muscle sample. These experiments are conducted at room temperature (20°C) where the thermal conductivity of muscle is low compared with that at body temperature (Duck 1990). Therefore, these results are conservative.

RESULTS

Time evolution of the heat field of a single ultrasound beam (*i.e.*, imaging of one pixel in VA) simulated by FEM in a medium with an attenuation coefficient of 0.5 dB/cm/MHz is shown in Figure 2. Because these 3-D fields are axisymmetric, only the center planes along the transducer axis are presented. The confocal transducer is located above the top of these images at $z = 0$. Figure 2(a) is the temperature rise at the end of 0.1 ms ultrasound exposure. Figure 2 (b) is the heat field after 10 s, with the presence of thermal conduction but no blood perfusion.

Figure 3 shows the effects of thermal conduction and blood perfusion on final heating at the end of a single VA scan (40 mm $\times$ 40 mm C-scan, 0.25 mm resolution). Attenuation coefficient of the medium is set to 0.5 dB/cm/MHz. The confocal transducer is located above the top of these images (at $z = 0$) and scans sequentially: first left-to-right along the x axis to form one VA image line and then repeat line by line along the y axis (in-out of image plane). Only center planes of the 3-D heating volumes (at $y = 0$) are shown here. Images at orthogonal orientation (*i.e.*, y - z plane at $x = 0$) are very similar and therefore not shown here. Figure 3(a) is calculated without conduction and without perfusion and the maximum temperature rise is 0.0349°C. Figure 3(b) is calculated with conduction ($\kappa = 0.00143$ cm²/s) and without perfusion and the maximum temperature rise is 0.0316°C. Figure 3(c) is calculated with conduction ($\kappa = 0.00143$ cm²/s) and with perfusion ($\tau = 2000$ s) and the maximum temperature rise is 0.0312°C.

The spatial maximum temperature rise during and after a VA scan (40 mm $\times$ 40 mm C-scan, 0.25 mm resolution) is shown in Figure 4. The medium has an attenuation coefficient of 0.5 dB/cm/MHz. The VA scanning speed is 2 ms/pixel and the whole VA scan takes about 52 s. For each data point in Figure 4, we calculate the 3-D heating field at that time step and use the maximum of the 3-D heating field as the value for that data point. It is important to note that the location of the maximum heating point moves during the VA scan: it moves from the transducer focal plane toward the transducer near field as the VA scan progresses. This is due to the overlap of the heating contribution from multiple ultrasound beams during the raster C-scan. The impact of overlapping on spatial max temperature rise changes during the VA scan. During the scan of the first several VA image lines, heating at the transducer focus overlaps from line to line and causes the spatial max temperature rise to increase rapidly. The beamwidth with significant heating is 1.5 mm at the transducer focus. This “heating beamwidth” is larger than the -6 dB intensity beamwidth because ultrasound outside the -6 dB beamwidth can still produce non-negligible heating. By 1.932 s, the transducer has scanned 6 VA image lines at 0.25 mm resolution (6 lines $\times$ 161 pixels/line $\times$ 2 ms/pixel = 1.932 s). Therefore, the spatial max temperature rise plateaus after 1.932 s because the scan range is now greater than 1.5mm (6 $\times$ 0.25=1.5 mm) and heating at the transducer focus does not accumulate significantly. After 5.2 s, heat accumulation in the transducer near field exceeds that at the focus and the spatial max temperature rise starts to increase again appreciably with the VA scan. And the position of the spatial max temperature rise shifts from the transducer focal plane toward the near field. However, temperature now increases at a slower rate because heating in the near field is lower than at the focus for each ultrasound beam. These simulations also show that the cooling process begins at the end of the VA scan (around 52 s). The dotted line is calculated with conduction and with perfusion and, therefore, is used as the baseline for comparison here. One can see that omission of blood perfusion ($\tau = 2000$ s) changes the results very little. Further neglect of

thermal conduction slightly overestimates heating but still provides fairly good temperature estimation during the VA scan. Temperature rise of the solid line remains constant during the cooling period because there is no heat loss due to the absence of both conduction and perfusion.

Figure 5 shows the effect of tissue attenuation on the final heating field at the end of a VA scan. Thermal conduction and blood perfusion are neglected in these simulations. The transducer output level is fixed at the same level (*i.e.*, driven with the same voltage) as in Figure 3. The confocal transducer is located above the top of these images (at $z = 0$) and scans sequentially: first left-to-right along the x axis to form one VA image line and then repeat line by line along the y axis (in-out of image plane). Only center planes of the 3-D heating volumes (at $y = 0$) are shown here. Images at orthogonal orientation (*i.e.*, y - z plane at $x = 0$) are very similar and, therefore, not shown here. Figure 5(a) is calculated for 0.3 dB/cm/MHz attenuation and maximum temperature rise is 0.0226°C . Figure 5(b) is calculated for 0.7 dB/cm/MHz attenuation and maximum temperature rise is 0.0468°C . Compared with results in Figure 3 for 0.5 dB/cm/MHz attenuation, one can see that higher attenuation leads to moderately higher heating, if the acoustic output of the transducer is kept identical for all three attenuation levels.

Temperature rise at the transducer surface measured by the thermocouple is $0.27 \pm 0.03^{\circ}\text{C}$ at the end of a single VA scan. Mean and standard deviation are obtained from five repeated measurements. Figure 6 shows a typical heating curve during a single VA “scan” (transducer does not move during these measurements). To reduce measurement noise of the thermocouple, temperature data in Figure 6 have been filtered by a zero-phase digital Butterworth lowpass filter with a cutoff frequency of 1 Hz.

DISCUSSION

This study indicates that maximum heating due to ultrasound absorption in vibro-acoustography occurred close to the transducer, not at the focal plane (*i.e.*, the imaging plane). This finding is consistent with those reported by other studies where the imaging transducer is fixed in space and the ultrasound beam is steered electronically (Curley 1993; Palmeri and Nightingale 2004). This result is expected because heating at the transducer focus due to each imaging beam is confined to a small 3-D region and, therefore, does not accumulate much during the 2-D C-scan in VA. However, the near-field of each imaging beam covers a much larger region and tends to overlap as the imaging beam is scanned from one pixel to another. Therefore, heating in the near field accumulates quickly during scanning and eventually surpasses the heating at the focal plane, which has less accumulation during scanning.

Results in Figures 3 and 4 indicate that thermal conduction and blood perfusion may be neglected in calculation of ultrasound heating for VA. This neglect slightly overestimates ultrasound heating in VA but still produces fairly good estimates of the final temperature field. This simplification eliminates the need for finite element analysis and greatly reduces the computation cost. Results obtained with this simplified approach provide a less accurate but more conservative heating evaluation. Given that thermal safety is the primary focus, this simplified approach is a good choice for future simulations of other transducers and scanning parameters because it reduces computation but does not compromise safety evaluation.

Simulations in this study show that heating due to ultrasound absorption in a relatively intense VA scan (40 mm $\times$ 40 mm C-scan, 0.25 mm resolution) is below 0.05°C at the maximum output of our current VA imaging system for all three tissue attenuation levels evaluated (0.3, 0.5 and 0.7 dB/cm/MHz). This kind of heating is about an order of magnitude below the normal body temperature fluctuation and, therefore, should be safe for human application. Results presented here may also be extended to tissue not directly modeled. For example, the specific heat of adipose tissue is $2.3 \text{ J/cm}^3/^{\circ}\text{C}$ (Fahey et al. 2006; Duck 1990), which is about half the

value used in our simulation ($4.2 \text{ J/cm}^3/^{\circ}\text{C}$). Therefore, heating in adipose tissue may double to about 0.1°C during a single VA scan when conduction and perfusion are neglected. On the other hand, heating in highly perfused tissue such as liver ($\tau \sim 150 \text{ s}$, refer to Fahey 2006) should be less than 0.05°C for a single VA scan studied here. However, we expect that the reduction of heating should not be significant because maximum heating due to ultrasound absorption occurs close to the transducer surface where we typically find fat/muscle ($\tau \sim 2000 \text{ s}$).

Measurements in this study show that transducer self-heating is less than 0.27°C during a single VA scan ($40 \text{ mm} \times 40 \text{ mm}$ C-scan, 0.25 mm resolution). Because the transducer physically moves during a real VA scan, temperature rise in tissue due to transducer heating should be significantly less than 0.27°C . Nevertheless, tissue heating due to the transducer is likely to be higher than that due to ultrasound absorption. For a single VA scan evaluated in this study, total tissue heating should be no more than $0.05 + 0.27 = 0.32^{\circ}\text{C}$. For repeated VA scans over the same region, heating from multiple scans can accumulate and needs to be monitored. Because maximum temperature rise due to both transducer heating and ultrasound absorption should occur at the skin surface, experimental measurement of temperature rise at skin surface would be a practical choice to monitor heating for repeated VA scans when thermal safety may become a concern.

This study has some limitations. Nonlinear propagation of ultrasound is not considered in the simulations here because FIELD II is a linear acoustic field simulation program. However, we expect that nonlinear effects should not greatly change estimates of maximum heating occurring in the transducer near-field because nonlinear effects should be small in the near-field where the ultrasound intensity is low before the beamfocuses. In addition, because nonlinear effects are cumulative, their effect in the near field should be small due to the short propagation distance. Therefore, nonlinear effects should not significantly alter the values of maximum heating in these simulations, although it may change the heating pattern beyond the near field. In addition, VA heating in the presence of bone is not addressed by this study. Absorption of ultrasound in bone is significantly higher than that in soft tissue. Therefore, in applications where bone might be exposed, the impact of bone on heating should be analyzed.

CONCLUSIONS

Ultrasound heating in soft tissue due to a vibro-acoustography scan using a confocal transducer is simulated in this study. Blood perfusion has negligible effect on the temperature rise. Further omission of thermal conduction slightly overestimates the results but still gives fairly good estimates of final heating. Maximum heating due to ultrasound absorption occurs in the transducer near-field during a VA scan. Heating during a single VA scan is below 0.05°C in soft tissue for all three attenuation levels evaluated here (0.3 , 0.5 or 0.7 dB/cm/MHz). Transducer self-heating during a single VA scan measured by a thermocouple is less than 0.27°C .

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Fig. 1.

Diagram of the vibro-acoustography imaging system. The object vibrates at the beat frequency $\Delta f = f_1 - f_2$ and its acoustic emission is recorded by the acoustic hydrophone. The focal spot of two cofocused ultrasound beams moves over the object in a raster C-scan to form a two-dimensional (2-D) image.

Fig. 2.

Temperature rise on transducer axial plane resulting from vibro-acoustography imaging of one pixel in a medium with 0.5 dB/cm/MHz attenuation. The transducer is located above the top of these images. (a) Heat field at the end of 0.1 ms ultrasound exposure. (b) Heat field after 10 s of thermal conduction (blood perfusion is neglected here).

Fig. 3.

Center planes of the three-dimensional (3-D) heat field caused by a 40 mm $\times$ 40 mm vibro-acoustography scan in a medium of 0.5 dB/cm/MHz attenuation. The transducer is located above the top of these images. (a) Heat field calculated without conduction and without perfusion (Maximum temperature rise is 0.0349°C). (b) Heat field calculated with conduction and without perfusion (Maximum temperature rise is 0.0316°C). (c) Heat field calculated with conduction and with perfusion (Maximum temperature rise is 0.0312°C).

Fig. 4. Spatial maximum temperature rise during and after a vibro-acoustography scan ($40\text{ mm} \times 40\text{ mm}$ C-scan, 0.25 mm resolution) in a medium with 0.5 dB/cm/MHz attenuation. The VA scan is finished at about 52 s . Neglect of conduction and perfusion slightly overestimates heating during the scan.

Fig. 5.

Center planes of the three-dimensional (3-D) heat field caused by a 40 mm $\times$ 40 mm vibro-acoustography scan in a medium of (a) 0.3 dB/cm/MHz attenuation and (b) 0.7 dB/cm/MHz attenuation. The transducer is located above the top of these images. Conduction and perfusion are neglected and the acoustic output of the transducer is fixed, leading to higher in situ I_{sppa} in (a) than in (b). The maximum temperature is 0.0226°C in (a) and 0.0468°C in (b).

Fig. 6. Temperature rise at the transducer surface measured by a thermocouple during a single vibro-acoustography scan.