Generalized polynomial chaos as a uncertainty quantification method for modeling MR guided laser induced thermal therapy.

Samuel John Fahrenholtz[1,2], David Fuentes[1], John D. Hazle[1,2], and Roger Jason Stafford[1,2]

[1]*Imaging Physics, The University of Texas M.D. Anderson Cancer Center, Houston, Texas, United States,* [2]*The University of Texas Graduate School of Biomedical Science, Houston, Texas, United States*

Introduction: MR-guided laser induced thermal therapy (MRgLITT) is available as an alternative approach to surgery for lesions in the brain, among other sites. [1]. To minimize invasiveness and improve efficacy, MRgLITT in the brain may benefit from prospective treatment modeling. However, there are difficulties in modeling the thermodynamics of MRgLITT in brain, including local convective heat sinks (i.e., vessels and ventricles) as well as variable tissue perfusion. Previous investigations have demonstrated that simplifying assumptions can be inappropriate, e.g. neglecting convective blood vessels or setting tissue perfusion to be an on/off step function [2]. Furthermore, the problem is confounded by limited *in vivo* data on tissue properties pertinent to the Pennes bioheat transfer equation (BHT), e.g. variable tissue perfusion or variable optical properties. Previous studies have used a finite element model (FEM) approach to solve the nonlinear BHT partial differential equation in realistic treatment geometries. This abstract introduces FEM with generalized polynomial chaos (gPC) as a possible new technique for modeling the BHT and compares the results to proton resonance (PRF) shift MR thermal imaging (MRTI) in a canine. Compared to single deterministic simulations, gPC has the unique availability of confidence intervals (CIs). gPC uses a few model realizations to fit a polynomial to the solution's distribution. Once the polynomial is acquired, it is sampled many times. The central limit theorem ensures the sample mean approaches the polynomial mean. The polynomial mean is model the quantity of interest, e.g. temperature.

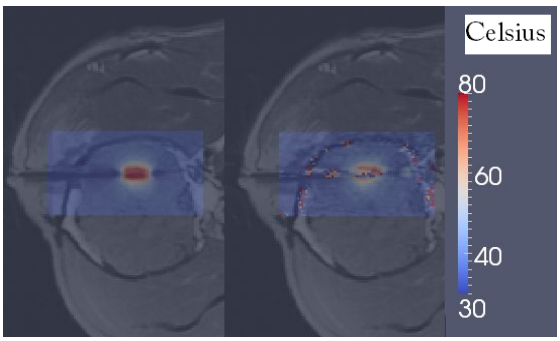

Figure 1: The left image is the simulation prediction at the peak temperature; the right image is the MRTI data.

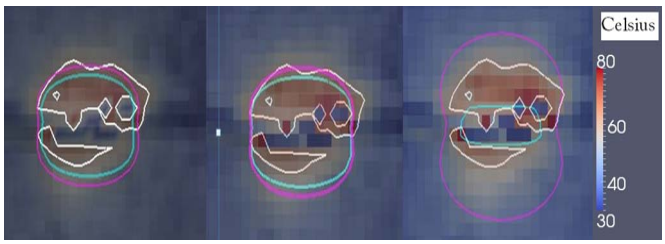

Figure 2: These three magnified images are the MRTI heated region. The contours are 60 °C isotherms for k, ω_b, and μ_a from left to right respectively. The white contour is the MRTI while the other two are the simulation mean temperature plus and minus two standard deviations. The background coloring is from MRTI.

Methods: An MRgLITT procedure performed in canine brain was analyzed retrospectively to assess the usefulness of the algorithm. The dog was treated using a water cooled 980-nm laser (Visualase, Inc., Houston, TX) placed in normal canine brain tissue with real-time MR temperature imaging (MRTI) guidance on a 1.5T clinical scanner (GE Healthcare Technologies, Waukesha, WI). A test pulse (40 s @ 4 W) was used to confirm laser fiber location prior to a therapy pulse (150 s @ 15W). During the laser fiber placement, a ventricle was errantly breached and leaked cerebral spinal fluid (CSF) near the treatment site. MRTI was acquired by complex phase subtraction from a data interleaved, multiplanar, echo-planar sequence exploiting the temperature dependent water proton frequency shift. Simulations utilized libMesh (Open Source) for FEM simulation and DAKOTA (Sandia National Laboratories) for gPC implementation. The simulation's FEM geometry was designed to model the dimensions of the laser fiber and the laser light distribution. Three constitutive values (CVs)—ω_b, k, and μ_a— were chosen to have uniform distributions in a range meant to represent the physical limits of the CVs. Thermal damage-dependent expressions were used to describe the CVs. The choice of CVs was submitted to DAKOTA. Each CV had four realizations. E.g., for the CV, tissue perfusion ω_b, the range was 3 to 9 kg/(m^3 s) and four values were chosen by DAKOTA's implementation of gPC to be the four realizations of ω_b. Once the simulation data was collected, CIs were acquired by adding and subtracting two standard deviations to the simulation mean. The CIs were processed in ParaView 3.10.1 (Sandia National Laboratories). Finally, the simulation results were numerically compared at each time point using the RMS difference, weighted by the MRTI uncertainty. The RMS difference processing is further described in [3]. RMS was processed in MATLAB 7.9.0 (The MathWorks; Natick, MA).

Results: The MRTI and simulation data are compared in the figures. The MRTI uncertainty weighted RMS difference for ω_b, k, and μ_a respectively had peak values of 2.2, 2.2, and 2.6 occurring just after the ablative laser pulse ended. The RMS error have similar magnitude as [3]. Note in Figure 1, the MRTI data has a cooler overall temperature on the lower portion of the heated region. Therefore in Figure 2, the MRTI contour is not symmetric. The gPC CIs are symmetric. All of the gPC CI contours disagree with the MRTI data on the lower side of Figure 2. In the upper portion of Figure 2, the k and ω_b contours agree with the MRTI contour while μ_a does not.

Discussion: The goal of the work presented was to investigate of gPC could produce a mean temperature with a standard deviation, which was successful. As measured by the RMS, the accuracy of the presented method is similar to [3]. In Figure 1, the MRTI temperature is not symmetric because of the CSF leak. Likewise in Figure 2, the asymmetric MRTI contours are due to the leaked CSF. Because the CSF was not modeled, the three simulations failed in that area. The ω_b and k models had smaller variance than the μ_a, explaining the relative sizes of the CIs. All three models' CIs enclose or approach the upper MRTI contour. Using gPC in BHT modeling introduces some advantages. It provides information on the distribution of possible outcomes. Assuming the conditions are correctly modeled, knowing the range of likely outcomes demonstrates the sensitivity of a treatment. If implemented clinically, gPC's CIs may provide critical information regarding laser fiber placement.

References:

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