INTRODUCTION

As a technique for noninvasive monitoring of biomolecular function, diffuse optical tomography (DOT) offers a number of practical, economic, and physiologic advantages relative to other imaging modalities [1,2]. This technology, compact, can be brought to the bedside, and is easily combined with other imaging technologies. In addition, by extending the measurement to allow for the capture of a time series, it can explore various dynamic phenomena, in particular, those associated with the hemoglobin signal [3-5].

An especially important class of dynamic phenomena are those that lead to feedback mechanisms which serve to regulate a multitude of biological processes. Among those is vascular autoregulation, which is the process by which brain regulates its blood flow in response to varying metabolic demands. In this and accompanying series of reports [218 M-M, 229 M-M, 222 M-M, 145 W-M], we explore the hypothesis that by using appropriate data sampling and analysis approaches, we can define features associated with autoregulatory processes in much more detail than previously possible.

To systematically document this, here we have applied modeling techniques to study the fidelity by which autoregulatory transitions in a linear model can be recovered using optical configuration similar to that used in clinical studies.

METHODS

FEM Model and Optical Configurations

Figure 1A shows a FEM mesh employed together with the possible epode positions. Two meshes were used in this report. One is a fine mesh, with 3742 nodes and 24,119 elements, used for simulation measurement generation. The other is a coarse mesh that has 2620 nodes and 11,565 elements for inverse calculations. These optical configurations were investigated, as shown in Figure 1C. C1 has 30 sources and 30 detectors; C2 has 22 sources and detectors, and C3 is 15 sources and detectors.

Optical Properties and Hemodynamic States

Model: A weakly absorbing homogeneous-scattering medium \( \mu_s = 0.055/100cm \) containing two regions, 1.5 cm in diameter, separated by 2.5 cm and whose center is \( 1.5 \) cm below the surface (Figure 1B). As shown in Figure 2A, also modeled are time variations in the absorption properties of these regions that correspond to the change in \( IH \) levels of a magnitude typically seen in the cortex (30%). Figure 2B shows the accompanying variations in \( HI \)-hemostatic states in the modeled inclusions.

Model and Solutions

Simulated detector readings for a specific optical configuration were acquired by using the finite element method to solve the diffusion equation, with a DC source term and Robin boundary conditions, as described in Ref. [5].

Verification

Images were reconstructed by using the previously described Normalized Difference Method (NDM) [6], which is a modified linear perturbation technique that is highly robust to many difficult-to-eliminate sources of experimental error and uncertainty associated with DOT measurements.

Detector Noise Models:

Noise-free detector data were used in the first instance, and Gaussian white noise was added to the normalized detector data in the noise. In particular, we used a model in which, for each source-detector channel, the noise level (i.e., the standard deviation of the noise distribution) is a pre-selected percentage of the noise-free detector reading and given as the fourth power of the distance between source and detector [7]. In the first noise-added trials, the minimal noise level (noise level and detector co-located) was, respectively, 50, 20, 10, 5, 0, and 0%, which are defined as the noise level (1.25, 2.5, and 5% (used in Figure 2B and Table 2)). The noise levels used for each trial were stationary, i.e., the noise levels were not time-varying.

Noise Suppression for Recovered Images

Reconstructed images retrieved from noisy detector data were subsequently treated with two noise-suppression schemes. The first method was temporal low-pass filtering (LPF) with a zero-phase FIR digital filter, using a tophat frequency-response function with threshold frequency set to 0.15 Hz and a 0.05 Hz roll-off. The second noise-suppression scheme was spatial “pillbox” filtering (sLPF), wherein the image value at each FEM node is replaced by a weighted average of the values at that node and its nearest neighbors [7].

RESULTS

Qualitative and quantitative assessments of the fidelity of our method are presented in Figures 3-4 and Table 2 and 3. Figure 3A shows recovered autoregulatory states as a function of noise level for the C1 measurement configuration. Images shown were thresholded in accordance to the expected green tissue variance for pixels containing the inclusion (normalized to 50%). Inspection shows that only nominal degradation in image quality occurs for noise level \( 5\% \).

In Figure 6 shows the spatial extent of recovery of the autoregulatory states as a function of noise level for the C1 measurement configuration. Images shown were thresholded in accordance to the expected green tissue variance for pixels containing the inclusion (normalized to 50%). Inspection shows that only nominal degradation in image quality occurs for noise level \( 5\% \).

CONCLUSIONS

Conclusions are summarized in Tables 2 and 3. As expected, as the noise levels increased, the recovered temporal variance for pixels containing the inclusion (normalized to 50%) was reduced. Inspection shows that only nominal degradation in image quality occurs for noise level \( 5\% \).

REFERENCES


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