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Towards freehand image acquisition in optical coherence tomography

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A new data-capture method could provide increased flexibility for clinical imaging applications.

Optical coherence tomography (OCT), a biomedical diagnostic-imaging technique, is currently transitioning from the research laboratory into clinical practice.[1] Providing micrometer-scale resolution over millimeter-scale fields of view, OCT is the optical analog of ultrasound. It uses low-coherence interferometry to perform optical ranging in biological tissue. Clinical applications under investigation include optical biopsy[2] and detection of tumor margins during image-guided surgery.[3] The clinical applications of OCT benefit from its real-time imaging capability and diverse sample-arm designs, including hand-held probes, rotating fiber-optic catheters, and miniature needles. The real-time diagnostic capability has been greatly enhanced by recent developments in swept-source technology that permit screening of large areas of tissue,[4] with the potential to greatly reduce tissue-sampling errors. Freehand image acquisition leverages the technique's high-speed imaging advantage to offer greater flexibility for clinical applications such as image-guided surgery or OCT-guided needle biopsy.

OCT data acquisition typically involves precision lateral scanning of a focused beam. While providing excellent accuracy, the sample-arm design generally limits the attainable field of view and scan geometry. Freehand scanning of an imaging probe offers an attractive alternative, although it can cause image artifacts because of variations in scan velocity and probe orientation. Consequently, image formation requires synchronization of the acquired A scans (depth-resolved scattering profiles) with the relative displacement between sample and probe. We developed a cross-correlation-based algorithm for OCT data acquisition that does not rely on precision beam scanning. It essentially reconstructs an image by assembling A scans that are equally sampled in distance rather than time.[5] The approach is based on motion-estimation techniques using speckle decorrelation, similar to those used in sensorless freehand 3D ultrasound imaging[6] and optical coherence elastography.[7] Our method is based on the degree of correlation between sequential A scans as measured by the Pearson coefficient, ρ. Consecutive A scans within a given resolution volume will have $\rho \approx 1$ because of a high degree of overlap. On the other hand, for a separation well

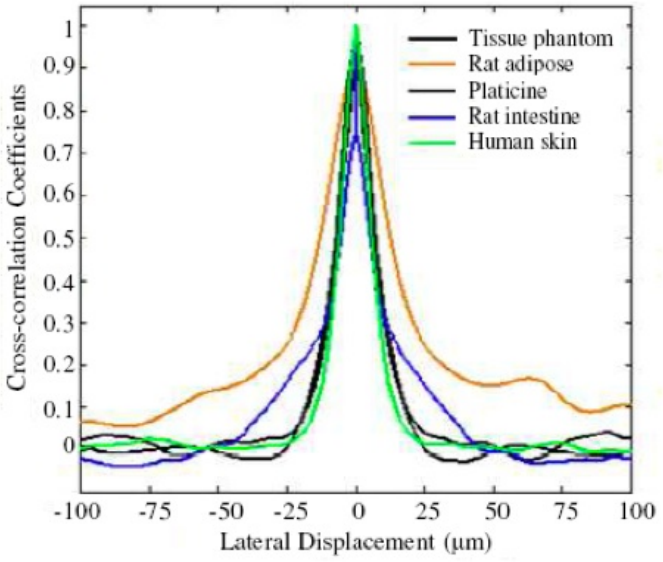

Figure 1. *Average decorrelation curves obtained from galvanometer-scanned images of several tissue-phantom samples and biological tissues (negative distance corresponds to the cross-correlation between the current and previously acquired A scans). The phantom was a silicone-based sample with titanium dioxide (TiO_2)-scattering particles. (Reproduced with permission.[5])*

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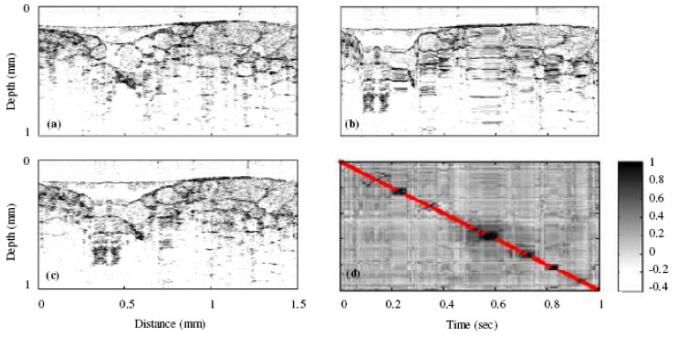

Figure 2. *Image assembly for human adipose tissue over a sample length of 1.5mm. (a) Galvanometer-scanned image. (b) Nonuniform hand-scanned image. (c) Assembled image. (d) Cross-correlation matrix. Red points show the A scans selected for image assembly. (Reproduced with permission.*[5]*)*

beyond the system's lateral resolution, A scans will become uncorrelated, with $\rho \approx 0$. By selecting an appropriate threshold, the algorithm can discard highly correlated A scans and assemble an image with the desired spatial sampling rate.

We measured decorrelation curves in a variety of phantoms and biological tissues. Figure 1 shows the cross-correlation coefficients obtained from ensemble averages of 400 A scans at each lateral displacement. The characteristic decorrelation length, measured by the decrease in the cross-correlation coefficient to $1/e$ of its maximum, was approximately equal to the system's lateral resolution, which governs the speckle size in OCT images of scattering tissues. The decorrelation lengths may be greater in samples containing prominent structural features,such as adipose tissue (body fat), but all curves converged to a low correlation value for lateral displacements beyond the system's transverse resolution. We used these decorrelation curves to determine the appropriate threshold for our cross-correlation algorithm.

We obtained assembled images for tissue phantoms and biological tissues during nonuniform scanning and compared these to uniformly scanned images acquired over the same field of view. Figure 2 presents results for human adipose tissue. The tissue sample was resting on a spring-loaded stage to guide the axis of manual translation. This ensured close alignment with the imaging plane of the galvanometer-scanned image. Significant distortions can be seen in the raw, nonuniformly scanned dataset in Figure 2(b), while good agreement is observed between the assembled galvanometer-scanned images.

In conclusion, we have developed a novel, fast cross-correlation-based approach for image acquisition using structural and speckle information from A scans. We have demonstrated its implementation in several phantom samples and biological tissues. Future work will focus on improving the robustness of the algorithm to account for inter- and intrasample variations in the decorrelation curves. One approach would be to incorporate tissue-classification algorithms to adaptively modify the threshold based on tissue type. Overall, this technique offers an attractive, computationally simple, and cost-effective method for large field-of-view imaging with user-defined scan geometry. Implementation of the algorithm during freehand scanning with hand-held probes, insertion and manipulation of needle probes, and catheter-based OCT could enable image-guided biopsy and high-resolution imaging of tissue structures over large fields of view, unrestricted by lateral beam-scanning mechanical and control systems.

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