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Yoko Miura, Wolfgang Draxinger, Christin Grill, Tom Pfeiffer, Salvatore Grisanti, Robert Huber, "MHz-OCT for low latency virtual reality guided surgery: first wet lab experiments on ex-vivo porcine eye," Proc. SPIE 11078, Optical Coherence Imaging Techniques and Imaging in Scattering Media III, 110780E (19 July 2019);
doi: 10.1117/12.2527123

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Event: European Conferences on Biomedical Optics, 2019, Munich, Germany

MHz-OCT for low latency virtual reality guided surgery: First wet lab experiments on ex-vivo porcine eye

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ABSTRACT

MHz-OCT systems based on FDML swept laser sources combined with the massive parallel processing capabilities of modern computer hardware enable volumetric imaging, processing and stereoscopic display at video rates. The increasing image quality and speed might enable new fields of application where the volumetric OCT completely replaces stereoscopic microscopes instead of being a mere supplement. Aside from the depth resolving capability, a particular advantage is the ability to display a whole image volume from arbitrary points of view without the need to move the actual microscope or to rotate the patient's eye. Purely digital microscopy is already offered as alternative to traditional through-an-eyepiece surgical microscope. We explore the use of virtual reality to present digital OCT microscopy images to a trained surgeon, carrying out a series of surgical procedures ex-vivo on a porcine eye model.

1. VIRTUAL REALITY AS A DISPLAY MODALITY FOR DIGITAL MICROSCOPY

Digital microscopy is the use of a microscope in which the traditional eyepiece is replaced by a computerized imaging system, either in the form of a full field image sensor, or a single channel light detector and flying spot scanner system. Thus in a broad sense OCT systems which image through an objective lens – possibly close to the diffraction limited optical resolution – may also be classified as a form of digital microscopy.

While mounted surgical microscopes are described as offering superior image quality and stability, they can lack the freedom and the ”felt” better structural perceptibility of head mounted loupes. This motivated us to integrate a real-time 4D-OCT imaging with a Virtual Reality system. Some first work into this direction, however at low volume rates, has been described by Draelos et. al.[1] In this work we therefore focus on the unique technical challenges that arise from highest imaging speeds, and present first reports on the user experience of such a system.

2. EXPERIMENTAL SETUP

We used a commercial ultrahigh speed OCT system (Optores OMES) consisting of a FDML swept laser light source, imaging interferometer and detector, scanning head microscope and signal acquisition and processing computer. The imaging system operates at a center wavelength of $1310\ \mathrm{nm}$ with a sweep bandwidth that is freely adjustable between $80\,\mathrm{nm}$ and $110\ \mathrm{nm}$. The fundamental FDML sweep frequency is $419\ \mathrm{kHz}$, after 4× buffering resulting in a $1.68\ \mathrm{MHz}$ A-scan rate.

The virtual reality system consists of a HTC Vive Pro head mounted display (HMD) and the Lighthouse-2.0 tracking system. An auxiliary tracking accessory was mounted to the microscope scan head (fig. 1A), to allow positioning of the reconstructed OCT volume in virtual space. The HMD operates at a constant fixed $90\ \mathrm{Hz}$ refresh frequency; volume updates are rendered for every 3rd display refresh, letting the VR drivers smooth out head movements by *temporal reprojection*. With the C-scan update frequency of approximately 15 Hz there are therefore between 4 and 6 display refreshes executed during a single C-scan. The position tracking update frequency was limited to 50 Hz.

With a lateral scanning range of $20\ \mathrm{mm}$ in the fast axis, this required at a minimum 600 A-scans per B-scan. Reducing the lateral scanning range allows an increase in B-scan frequency (and thereby C-scan update rate). Taking into account the 2048 sampling points per A-scan this gives an effective imaging volume resolution of 600×150×1024 voxels, at an imaging depth of 4 mm.

Optical Coherence Imaging Techniques and Imaging in Scattering Media III, edited by Maciej Wojtkowski, Stephen A. Boppart, Wang-Yuhl Oh, Proc. of SPIE-OSA Vol. 11078, 110780E · © 2019 SPIE-OSA · CCC code: 1605-7422/19/$21 · doi: 10.1117/12.2527123

3. USER EXPERIMENTS, OBSTACLES AND POSSIBLE IMPROVEMENTS

The first tests included cutting of the conjunctiva, incision at the corneal limbus and insertion of a needle (fig. 1B). The user experiences revealed several hurdles, which must be overcome to make a practical implementation viable in the future.

The instruments being used posed a couple of challenges. First and foremost, the instruments' surfaces were highly reflective and polished; the direct back reflections from there caused detector saturation which ultimately results in a stripe saturation artifact over the whole depth. This kind of artifact can be very confusing to the user by obscuring the true distance between tool and subject. Especially problematic are experiments or procedures that require instruments with a larger surface area like surgical scissors. Those exhibit strong reflections which result in artifacts with a visually periodic structure that completely obscured the whole OCT volume.

Even in the case that the instrument doesn't cause a saturation artifact, it still casts a shadow on the sample; this might be mitigated by developing special strategies of insertion angle of instrument into the OCT volume. The inevitable thickness of the instrument complicates the estimation of the distance between instrument and contact point, since only the top surface is actually visible in the OCT volume, and in the case of opaque instruments everything beneath it is cast into shadow.

Pronounced visual confusion results from the negative depth field flipping into the positive image field and vice versa. While these secondary images can be avoided when taking diagnostic images by careful placement of subject and choice of capture parameters, the circumstances of using a tool often don't allow to avoid intersection with the negative depth half space (fig. 1C). A couple of methods to implement full field imaging have been shown[2–4], which lends itself as a possible solution. In recent work we also demonstrated the long imaging depth ranges achievable with a dispersion compensated FDML light source[5,6]; simply elongating the imaging range too far, that the negative depth field lies outside of the working distance poses as an alternative solution.

A possible mitigation of both problems just described would be the inclusion of a virtual representation of the instrument, by tracking the tool in the same way as the HMD and the microscope. This would also allow addressing the problem of visual feedback on macro movements.

The obtainable lateral scanning range was limited by both the beam deflector capabilities and the imaging optics of the OCT system. However, even the widest possible lateral scanning range was already considered by the test user surgeon as being on the verge of being too narrow for the tasks at hand. We tested an augmented reality approach by overlaying the 3D OCT volume to the simultaneously recorded stereoscopic video stream of the HMD's internal two cameras (fig. 1D).

It is desirable to increase the imageable area for surgery of the anterior segment of the eye. Besides the virtual reality 3D reconstruction, also the En-Face representation exhibits a remarkably high image quality. Especially the contrast behavior and the level of detail recognizable are remarkable, even at a comparatively low sampling density. We therefore consider an additional En-Face representation integrated into the VR display to be desirable (fig. 1E) displayed simultaneously to the 3D volume rendering (fig. 1f).

4. CONCLUSION AND OUTLOOK

In these first trials mainly surface surgical procedures or procedures close to the surface were performed which was mainly due to the limitations in depth range. Procedures further down in the crystalline lens were not possible because the imaging range was insufficient. In future studies the system will be reconfigured to extend the range. However, the image quality is sufficiently high to display at least the surface of the crystalline lens.

Reconstruction speed at 30 volume rendering per second and head tracking with 50 Hz appear to be sufficient. The 15 Hz OCT C-scan rate however we must consider insufficient. Despite the plasticity of the rendering, the real stereoscopic and steric situation is sometimes misinterpreted. This might be caused by lack of experience with the new modality and should be further investigated. Right now, we must also consider the possibility of visual information overload. We also have to note that the sparse B-scan sampling of just 150 frames causes recognizability issues with very thin instruments and structures, falling in the gaps between the frames, and thereby becoming difficult to read.

As a consequence, the frame density and frame number must be increased which in turn requires an OCT system operating at even higher A-scan rates, which might be achieved by a 3 MHz or a 6 MHz sweep frequency light source. An increase in number of B-scans by a factor of 2 to 4 would lead to nearly isotropic sampling distance across the volume.

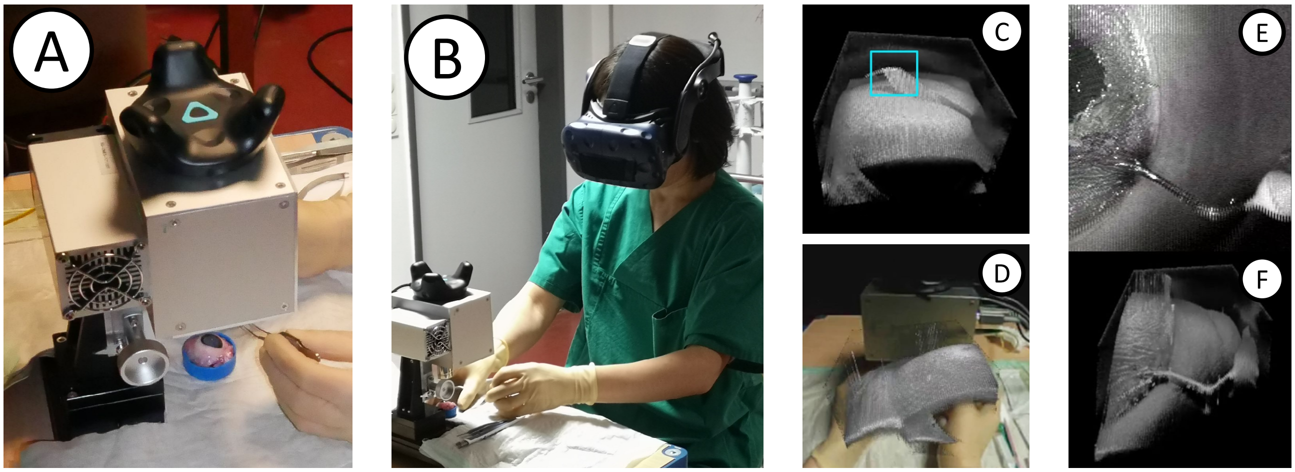

Figure 1: (A) The OCT microscope scanning head with tracker accessory mounted on top. (B) Ophthalmologist performing a series of surgical procedures ex-vivo. (C) Part of the used instrument reached into the negative depth, flipping down into the reconstruction volume. (D) Fusion display of camera feedthrough and virtual volume rendering. (E) OCT derived En-Face image with (F) corresponding VR user view.

REFERENCES

1. M. Draelos, B. Keller, C. Viehland, O. M. Carrasco-Zevallos, A. Kuo, and J. Izatt, "Real-time visualization and interaction with static and live optical coherence tomography volumes in immersive virtual reality," *Biomed. Opt. Express* 9, pp. 2825–2843, Jun 2018.
2. M. A. Choma, C. Yang, and J. A. Izatt, "Instantaneous quadrature low-coherence interferometry with 3×3 fiber-optic couplers," *Opt. Lett.* 28, pp. 2162–2164, Nov 2003.
3. M. V. Sarunic, M. A. Choma, C. Yang, and J. A. Izatt, "Instantaneous complex conjugate resolved spectral domain and swept-source oct using 3×3 fiber couplers," *Opt. Express* 13, pp. 957–967, Feb 2005.
4. B. Hofer, B. Považay, B. Hermann, A. Unterhuber, G. Matz, and W. Drexler, "Dispersion encoded full range frequency domain optical coherence tomography," *Opt. Express* 17, pp. 7–24, Jan 2009.
5. T. Pfeiffer, W. Draxinger, W. Wieser, T. Klein, M. Petermann, and R. Huber, "Analysis of FDML lasers with meter range coherence," *Proc. SPIE* 10053, Optical Coherence Tomography and Coherence Domain Optical Methods in Biomedicine XXI, 100531T, Feb 2017.
6. T. Pfeiffer, W. Draxinger, C. Grill, and R. Huber, "Long-range live 3d-oct at different spectral zoom levels," *Proc. SPIE* 10416, Optical Coherence Imaging Techniques and Imaging in Scattering Media II, 104160L, Aug 2017.