High-speed OCT light sources and systems [Invited]

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Abstract: Imaging speed is one of the most important parameters that define the performance of optical coherence tomography (OCT) systems. During the last two decades, OCT speed has increased by over three orders of magnitude. New developments in wavelength-swept lasers have repeatedly been crucial for this development. In this review, we discuss the historical evolution and current state of the art of high-speed OCT systems, with focus on wavelength swept light sources and swept source OCT systems.

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References and links

1. D. Huang, E. A. Swanson, C. P. Lin, J. S. Schuman, W. G. Stinson, W. Chang, M. R. Hee, T. Flotte, K. Gregory, C. A. Puliafito, and J. G. Fujimoto, “Optical coherence tomography,” Science 254(5035), 1178–1181 (1991).
2. M. Wojtkowski, “High-speed optical coherence tomography: basics and applications,” Appl. Opt. 49(16), D30–D61 (2010).
3. W. Drexler, M. Liu, A. Kumar, T. Kamali, A. Unterhuber, and R. A. Leitgeb, “Optical coherence tomography today: speed, contrast, and multimodality,” J. Biomed. Opt. 19(7), 071412 (2014).
4. A. M. Zysk, F. T. Nguyen, A. L. Oldenburg, D. L. Marks, and S. A. Boppart, “Optical coherence tomography: a review of clinical development from bench to bedside,” J. Biomed. Opt. 12, 051403 (2007).
5. W. Wieser, B. R. Biedermann, T. Klein, C. M. Eigenwillig, and R. Huber, “Multi-megahertz OCT: High quality 3D imaging at 20 million A-scans and 4.5 GVoxels per second,” Opt. Express 18(14), 14685–14704 (2010).
6. D. H. Choi, H. Hiro-Oka, K. Shimizu, and K. Ohbayashi, “Spectral domain optical coherence tomography of multi-MHz A-scan rates at 1310 nm range and real-time 4D-display up to 41 volumes/second,” Biomed. Opt. Express 3(12), 3067–3086 (2012).
7. J. Xu, X. Wei, L. Yu, C. Zhang, J. Xu, K. K. Y. Wong, and K. K. Tsia, “High-performance multi-megahertz optical coherence tomography based on amplified optical time-stretch,” Biomed. Opt. Express 6(4), 1340–1350 (2015).
8. D. J. Fechtig, B. Grajciar, T. Schmoll, C. Blatter, R. M. Werkmeister, W. Drexler, and R. A. Leitgeb, “Line-field parallel swept source MHz OCT for structural and functional retinal imaging,” Biomed. Opt. Express 6(3), 716–735 (2015).
9. T. Klein, W. Wieser, L. Reznicek, A. Neubauer, A. Kampik, and R. Huber, “Multi-MHz retinal OCT,” Biomed. Opt. Express 4(10), 1890–1908 (2013).
10. E. A. Swanson, D. Huang, M. R. Hee, J. G. Fujimoto, C. P. Lin, and C. A. Puliafito, “High-speed optical coherence domain reflectometry,” Opt. Lett. 17(2), 151–153 (1992).
11. G. J. Tearney, B. E. Bouna, S. A. Boppart, B. Golubovic, E. A. Swanson, and J. G. Fujimoto, “Rapid acquisition of in vivo biological images by use of optical coherence tomographic,” Opt. Lett. 21(17), 1408–1410 (1996).
12. G. J. Tearney, B. E. Bouna, and J. G. Fujimoto, “High-speed phase- and group-delay scanning with a grating-based phase control delay line,” Opt. Lett. 22(23), 1811–1813 (1997).
13. A. Rollins, S. Yazdanfar, M. Kulkarni, R. Ung-Arunyawee, and J. Izatt, “In vivo video rate optical coherence tomography,” Opt. Express 3(6), 219–229 (1998).
14. G. Häusler and M. W. Lindner, “Coherence radar and ‘spectral radar’-new tools for dermatological diagnosis,” J. Biomed. Opt. 3(1), 21–31 (1998).
15. B. Golubovic, B. E. Bouna, G. J. Tearney, and J. G. Fujimoto, “Optical frequency-domain reflectometry using rapid wavelength tuning of a Cr4+:forsterite laser,” Opt. Lett. 22(22), 1704–1706 (1997).
16. S. R. Chinn, E. A. Swanson, and J. G. Fujimoto, “Optical coherence tomography using a frequency-tunable optical source,” Opt. Lett. 22(5), 340–342 (1997).
17. M. Wojtkowski, R. Leitgeb, A. Kowalczyk, T. Bajraszewski, and A. F. Fercher, “In vivo human retinal imaging by Fourier domain optical coherence tomography,” J. Biomed. Opt. 7(3), 457–463 (2002).
18. L. Kranendonk, R. Bartula, and S. Sanders, “Modeless operation of a wavelength-agile laser by high-speed cavity length changes,” Opt. Express 13(5), 1498–1507 (2005).
46. C. H. Henry and R. F. Kazarinov, “Quantum noise in photonics,” Rev. Mod. Phys. 48(2), 899–902 (1976).
47. K. H. Y. Cheng, B. A. Standish, V. X. D. Yang, K. K. Y. Cheung, X. Gu, E. Y. Lam, and K. K. Y. Wong, “Sensitivity advantage of swept source and Fourier domain optical coherence tomography,” Opt. Express 11(18), 2183–2189 (2003).
48. J. F. de Boer, B. Cense, B. H. Park, M. C. Pierce, G. J. Tearney, and B. E. Bouma, “Improved signal-to-noise ratio in spectral-domain compared with time-domain optical coherence tomography,” Opt. Lett. 28(21), 2067–2069 (2003).
49. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, T. Klein, and R. Huber, “Dispersion, coherence and noise of Fourier domain mode locked lasers,” Opt. Express 17(12), 9947–9961 (2009).
50. W. Wieser, G. Palte, C. M. Eigenwillig, B. R. Biedermann, T. Pfeiffer, and R. Huber, “Chromatic polarization effects of swept waveforms in FDML lasers and fiber spools,” Opt. Express 20(9), 9819–9832 (2012).
51. I. A. Walmsley and C. Dorrer, “Characterization of ultrashort electromagnetic pulses,” Adv. Opt. Photonics 1(2), 308–437 (2009).
52. T. Butler, S. Stepniewa, B. O’Shaughnessy, B. Kelleher, D. Goulding, S. P. Hegarty, H. C. Lyu, K. Karmowski, M. Wojtkowski, and G. Huyet, “Single shot, time-resolved measurement of the coherence properties of OCT swept source lasers,” Opt. Lett. 40(10), 2277–2280 (2015).
53. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, T. Klein, and R. Huber, “Direct measurement of the instantaneous linewidth of rapidly wavelength-swept lasers,” Opt. Express 35(22), 3733–3735 (2010).
54. A. Bilanca, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Numerical study of wavelength-swept semiconductor ring lasers: the role of refractive-index nonlinearity in semiconductor optical amplifiers and implications for biomedical imaging applications,” Opt. Lett. 31(6), 760–762 (2006).
55. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, T. Klein, and R. Huber, “Dispersion, coherence and noise of Fourier domain mode locked lasers,” J. Opt. Soc. Am. B. 29(4), 656–664 (2012).
56. T. Klein, W. Wieser, C. M. Eigenwillig, B. R. Biedermann, and R. Huber, “Megahertz OCT for ultrawide-field retinal imaging with a 1050 nm Fourier domain mode-locked laser,” Opt. Express 19(4), 3044–3062 (2011).
57. M. Y. Jeon, J. Zhang, Q. Wang, and Z. Chen, “High-speed and wide bandwidth Fourier domain mode-locked wavelength swept laser with multiple SOAs,” Opt. Express 16(4), 2547–2554 (2008).
58. P. J. A. Thiis, L. F. Tiemeijer, J. J. M. Binsma, and T. Van Dongen, “Progress in long-wavelength strained-layer InGaAsP quantum-well semiconductor lasers and amplifiers,” IEEE J. Quantum Electron. 33(7), 1049–1056 (1997).
59. P. C. Becker, N. A. Olsson, and J. R. Simpson, Erbium-Doped Fiber Amplifiers (Academic Press, San Diego, 1999).
60. S. Marschall, T. Klein, W. Wieser, B. R. Biedermann, K. Hsu, K. P. Hansen, B. Sumpf, K.-H. Hasler, G. Erbert, O. B. Jensen, C. Pedersen, R. Huber, and P. E. Andersen, “Fourier domain mode-locked swept source at 1050 nm based on a tapered amplifier,” Opt. Express 18(15), 15820–15831 (2010).
61. T. Klein, W. Wieser, C. M. Eigenwillig, B. R. Biedermann, and R. Huber, “Megahertz OCT for ultrawide-field retinal imaging with a 1050 nm Fourier domain mode-locked laser,” Opt. Express 19(4), 3044–3062 (2011).
62. M. Y. Jeon, J. Zhang, Q. Wang, and Z. Chen, “High-speed and wide bandwidth Fourier domain mode-locked wavelength swept laser with multiple SOAs,” Opt. Express 16(4), 2547–2554 (2008).
63. P. J. A. Thiis, L. F. Tiemeijer, J. J. M. Binsma, and T. Van Dongen, “Progress in long-wavelength strained-layer InGaAs(P) quantum-well semiconductor lasers and amplifiers,” IEEE J. Quantum Electron. 33(2), 477–499 (1994).
64. P. S. Bobeck, B. Biedermann, R. Huber, and C. Jirasaheb, “Balance of physical effects causing stationary operation of Fourier domain mode locked lasers,” J. Opt. Soc. Am. B. 29(4), 656–664 (2012).
65. C. Henry, “Theory of the linewidth of semiconductor lasers,” IEEE J. Quantum Electron. 18(2), 259–264 (1982).
66. J. Peng, Q. Wang, J. Wang, S. Jiang, and K. Hsu, “All-fiber wavelength-swept laser near 2 μm,” Opt. Lett. 36(19), 3771–3773 (2011).
67. F. Nielsen, L. Thrane, J. Black, A. Bjerkevik, and P. Andersen, “Swept wavelength-swept source in the 1 microm range,” Opt. Express 13(11), 4096–4106 (2005).
68. R. Leonhardt, B. R. Biedermann, W. Wieser, and R. Huber, “Nonlinear optical frequency conversion of an amplified Fourier Domain Mode Locked (FDML) laser,” Opt. Express 17(19), 16801–16808 (2009).
69. S. Stepniewa, B. O’Shaughnessy, B. Kelleher, S. P. Hegarty, A. Vladimirov, H. C. Lyu, K. Karmowski, M. Wojtkowski, and G. Huyet, “Dynamics of a short cavity swept source OCT laser,” Opt. Express 22(15), 18177–18185 (2014).
70. G. Agrawal, Nonlinear Fiber Optics, 5th ed. (Academic Press, Boston, 2013), pp. 1–25.
71. R. H. Stolen, J. P. Gordon, W. J. Tomlinson, and H. A. Haus, “Raman response function of silica-core fibers,” J. Opt. Soc. Am. B 6(6), 1159–1166 (1989).
72. X. Jiang, H. Ma, X. Zeng, Y. Victor Xiao Dong, C. Po Ching, and K. K.-Y. Wong, “Wideband Raman-Pumped Wavelength-Swept Laser for Optical Coherence Tomography Application,” Appl. Phys. Express 6(6), 062701 (2013).
73. J. Bromage, “Raman Amplification for Fiber Communications Systems,” J. Lightwave Technol. 22(1), 79–93 (2004).
74. N. Olsson and J. Hegarty, “Noise properties of a Raman amplifier,” J. Lightwave Technol. 4(4), 396–399 (1986).
75. C. H. Henry and R. F. Kazarian, “Quantum noise in photonics,” Rev. Mod. Phys. 68(3), 801–853 (1996).
76. T. Klein, W. Wieser, B. R. Biedermann, C. M. Eigenwillig, G. Palte, and R. Huber, “Raman-pumped Fourier-domain mode-locked laser: analysis of operation and application for optical coherence tomography,” Opt. Lett. 33(23), 2815–2817 (2008).
77. K. H. Y. Cheng, B. A. Standish, V. X. D. Yang, K. K. Y. Cheung, X. Gu, E. Y. Lam, and K. K. Y. Wong, “Wavelength-swept spectral and pulse shaping utilizing hybrid Fourier domain mode locking by fiber optical parametric and erbium-doped fiber amplifiers,” Opt. Express 18(3), 1909–1915 (2010).
48. C. Chong, T. Suzuki, K. Totsuka, A. Morosawa, and T. Sakai, “Large coherence length swept source for axial
length measurement of the eye,” Appl. Opt. 37(25), 5524–5530 (1998).
49. Y. Okabe, Y. Sasaki, M. Ueno, T. Sakamoto, S. Toyoda, S. Yagi, K. Nagamura, K. Fujiura, Y. Sakai, J.
Kobayashi, K. Omiya, M. Ohmi, and M. Haruna, “200 kHz swept light source equipped with KTN deflectors for optical coherence tomography,” Electron. Lett. 48(4), 201–202 (2012).
50. J. R. Andrews, “Electronically tunable single-mode external-cavity diode laser,” Opt. Lett. 16(10), 732–734 (1991).
51. Y. H. Ja, “Optical vernier filter with fiber grating Fabry-Perot resonators,” Appl. Opt. 34(27), 6164–6167 (1995).
52. J. R. Andrews, “Electronically tunable single-mode external-cavity diode laser,” Opt. Lett. 34(27), 6164–6167 (1995).
53. Y. Yasuno, V. D. Madjarova, S. Makita, M. Akiba, A. Morosawa, C. Chong, T. Sakai, K. P. Chan, M. Itoh, and T. Yataki, “Three-dimensional and high-speed swept-source optical coherence tomography for in vivo investigation of human anterior eye segments,” Opt. Express 13(26), 10652–10664 (2005).
54. C. M. Eigenwillig, B. R. Biedermann, W. Wieser, and R. Huber, “Wavelength swept ASE source,” in (SPIE, 2009), 737200.
55. R. Huber, M. Wojtkowski, K. Taira, J. Fujimoto, and K. Hsu, “Amplified, frequency swept lasers for frequency domain reflectometry and OCT imaging: design and scaling principles,” Opt. Express 13(9), 3513–3528 (2005).
56. E. C. W. Lee, J. F. de Boer, M. Mujat, H. Lim, and S. H. Yun, “In vivo optical frequency domain imaging of human retina and choroid,” Opt. Express 14(10), 4403–4411 (2006).
57. C. M. Eigenwillig, B. R. Biedermann, W. Wieser, and R. Huber, “Wavelength swept amplified spontaneous emission source,” Opt. Express 17(21), 18794–18807 (2009).
58. Y. Yasuno, V. D. Madjarova, S. Makita, M. Akiba, A. Morosawa, C. Chong, T. Sakai, K. P. Chan, M. Itoh, and T. Yataki, “Three-dimensional and high-speed swept-source optical coherence tomography for in vivo investigation of human anterior eye segments,” Opt. Express 13(26), 10652–10664 (2005).
59. W.-Y. Oh, B. J. Vakoc, M. Shishkov, G. J. Tearney, and B. E. Bouma, “400 kHz repetition rate wavelength-swept laser and application to high-speed optical frequency domain imaging,” Opt. Lett. 35(17), 2919–2921 (2010).
60. M. Bonesi, M. P. Minneman, J. Ensher, B. Zabihian, H. Sattmann, P. Boschert, E. Hoover, R. A. Leitgeb, M. Crawford, and W. Drexler, “Akinetic all-semiconductor programmable swept-source at 1550 nm and 1310 nm with centimeters coherence length,” Opt. Express 22(3), 2632–2655 (2014).
61. A.-H. Dhalla, D. Nankivil, and J. A. Izatt, “Complex conjugate resolved heterodyne swept source optical coherence tomography using coherence revival,” Biomed. Opt. Express 3(3), 633–649 (2012).
62. C. Chong, T. Suzuki, A. Morosawa, and T. Sakai, “Spectral narrowing effect by quasi-phase continuous tuning of intra-cavity photons: Adiabatic wavelength tuning in rapidly wavelength-swept lasers,” Biomed. Opt. Express 6(7), 2448–2465 (2015).
63. C. Chong, T. Suzuki, A. Morosawa, and T. Sakai, “Spectral narrowing effect by quasi-phase continuous tuning of high-speed wavelength-swept light source,” Opt. Express 16(25), 21105–21118 (2008).
64. C. Chong, T. Suzuki, K. Totsuka, A. Morosawa, and T. Sakai, “Large coherence length swept source for axial length measurement of the eye,” Appl. Opt. 48(10), D144–D150 (2009).
65. S. H. Yun, D. J. Richardson, D. O. Culverhouse, and B. Y. Kim, “Wavelength-swept fiber laser with frequency shifted feedback and resonantly swept intra-cavity acousto-optic tunable filter,” IEEE J. Sel. Top. Quantum Electron. 3(4), 1087–1096 (1997).
66. S. Song, J. Xu, and R. K. Wang, “Long-range and wide field of view optical coherence tomography for in vivo 3D imaging of large volume object based on akinetic programmable swept source,” Biomed. Opt. Express 7(11), 4734–4748 (2016).
67. B. Petsdorff, I. Gorczynska, V. J. Srinivasan, Y. Chen, J. Jiang, A. Cable, and J. G. Fujimoto, “Ultrahigh speed spectral / Fourier domain OCT ophthalmic imaging at 70,000 to 312,500 axial scans per second,” Opt. Express 16(19), 15149–15169 (2008).
68. J. Jayaraman, G. D. Cole, M. Robertson, A. Uddin, and A. Cable, “High-sweep-rate 1310 nm MEMS-VCSEL with 150 nm continuous tuning range,” Electron. Lett. 48(14), 867–869 (2012).
69. E. C. Vail, G. S. Li, W. Yuen, and C. J. Chang-Hasnas, “High performance micromechanical tunable vertical cavity surface emitting lasers,” Electron. Lett. 32(20), 1888–1889 (1996).
70. Siegelman, Lasers (University Science Books, Mill Valley, 1986).
71. I. Grulkowski, J. J. Liu, B. Petsdorff, V. Jayaraman, J. Jiang, J. G. Fujimoto, and A. E. Cable, “High-precision, high-accuracy ultralong-range swept-source optical coherence tomography using vertical cavity surface emitting laser light source,” Opt. Lett. 38(5), 673–675 (2013).
72. Z. Wang, B. Petsdorff, L. Chen, C. Doerr, H.-C. Lee, T. Nielson, V. Jayaraman, A. E. Cable, E. Swanson, and J. G. Fujimoto, “Cubic meter volume optical coherence tomography,” Optica 3(12), 1496–1503 (2016).
91. C. Jirauschek and R. Huber, “Modeling and analysis of polarization effects in Fourier domain mode-locked lasers,” Biomed. Opt. Express 3(11), 2733–2751 (2012).

90. S. Slepneva, B. Kelleher, B. O'Shaughnessy, S. P. Hegarty, A. G. Vladimirov, and G. Huyet, “Dynamics of multifrequency operation in dual-frequency supercontinuum generation,” Opt. Express 21(15), 17368–17377 (2013).

89. S. Todor, B. Biedermann, W. Wieser, R. Huber, and C. Jirauschek, “Instantaneous lineshape analysis of Fourier domain mode locked lasers,” Opt. Express 19(21), 19240–19251 (2011).

88. C. Jirauschek, B. Biedermann, and R. Huber, “A theoretical description of Fourier domain mode locked lasers,” Opt. Express 17(26), 24013–24019 (2009).

87. C. Jirauschek, C. Eigenwillig, B. Biedermann, and R. Huber, “Fourier Domain Mode Locking Theory,” in Lasers and Electro-Optics (CLEO), 2011 Conference on, (2011), 1–2.

86. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, G. Palte, D. C. Adler, V. J. Srinivasan, J. G. Fujimoto, and R. Huber, “Real time en face Fourier-domain optical coherence tomography with direct hardware frequency demodulation,” Opt. Lett. 33(21), 2556–2558 (2008).

85. D. C. Adler, Y. Chen, R. Huber, J. Schmitt, J. Connolly, and J. G. Fujimoto, “Three-dimensional endomicroscopy using optical coherence tomography,” Nat. Photonics 1(2), 709–716 (2007).

84. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, G. Palte, D. C. Adler, J. G. Fujimoto, and R. Huber, “FDML laser for megahertz retinal OCT imaging,” Biomed. Opt. Express 2(7), 2005–2011 (2011).

83. T. Klein, W. Wieser, B. R. Biedermann, C. M. Eigenwillig, and R. Huber, “FDML laser for megahertz retinal OCT imaging,” in Lasers and Electro-Optics (CLEO), 2011 Conference on, (2011), 1–2.

82. K. Murari, J. Mavadia, J. Xi, and X. Li, “Self-starting, self-regulating Fourier domain mode locked fiber laser for OCT imaging,” Opt. Express 14(8), 3225–3237 (2006).

81. J. M. Telle and C. L. Tang, “Very rapid tuning of cw dye laser,” Appl. Phys. Lett. 26(10), 572–574 (1975).

80. R. Huber, M. Wojtkowski, and J. G. Fujimoto, “Fourier Domain Mode Locking (FDML): A new laser operating regime and applications for optical coherence tomography,” Opt. Express 14(8), 3225–3237 (2006).

79. C. Jirauschek, B. Biedermann, W. Draxinger, R. Huber, and M. Wojtkowski, “Fully automated 1.5 MHz FDML laser with 100 mW output power at 1310 nm,” in Optical Coherence Imaging Techniques and Imaging in Scattering Media (SPIE, 2013), 954116.

78. B. R. Biedermann, W. Wieser, C. M. Eigenwillig, G. Palte, D. C. Adler, V. J. Srinivasan, J. G. Fujimoto, and R. Huber, “Real time en face Fourier-domain optical coherence tomography with direct hardware frequency demodulation,” Opt. Lett. 33(21), 2556–2558 (2008).

77. I. Grulkowski, J. J. Liu, B. Potsaid, V. Jayaraman, C. D. Lu, J. Jiang, A. E. Cable, J. S. Duker, and J. G. Fujimoto, “Retinal, anterior segment and full eye imaging using ultrahigh speed swept source OCT with vertical-cavity surface emitting lasers,” Biomed. Opt. Express 3(11), 2733–2751 (2012).

76. T.-H. Tsai, B. Potsaid, Y. K. Tao, V. Jayaraman, J. Jiang, P. J. S. Heim, M. F. Kraus, C. Zhou, J. Honegger, H. Mashimo, A. E. Cable, and J. G. Fujimoto, “Ultra-high speed endoscopic optical coherence tomography using micromotor imaging catheter and VCSEL technology,” Biomed. Opt. Express 4(7), 1119–1132 (2013).

75. V. Jayaraman, J. Jiang, B. Potsaid, M. Robertson, P. J. S. Heim, C. Burgner, D. John, G. D. Cole, I. Grulkowski, J. G. Fujimoto, A. M. Davis, and A. E. Cable, “VCSEL Swept Light Sources,” in Optical Coherence Tomography: Technology and Applications, W. Drexler and J. G. Fujimoto, eds. (Springer International Publishing, Cham, 2015), pp. 659–686.
104. X. Wei, C. Kong, G. K. Samanta, K. K. Tsia, and K. K. Y. Wong, “Self-healing highly-chirped fiber laser at 1.0 μm,” Opt. Express 24(24), 27577–27586 (2016).

105. J. Xu, X. Wei, L. Yu, C. Zhang, J. Xu, K. K. Y. Wong, and K. K. Tsia, “Performance of megahertz amplified optical time-stretch optical coherence tomography (OCT-OST),” Opt. Express 22(19), 22498–22512 (2014).

106. J. Xu, C. Zhang, J. Xu, K. K. Y. Wong, and K. K. Tsia, “Megahertz all-optical swept-source optical coherence tomography based on broadband amplified optical time-stretch,” Opt. Lett. 39(3), 622–625 (2014).

107. S. Yamashita and M. Asano, “Wide and fast wavelength-tunable mode-locked fiber laser based on dispersion tuning,” Opt. Express 14(20), 9299–9306 (2006).

108. Y. Takubo and S. Yamashita, “High-speed dispersion-tuned wavelength-swept fiber laser using a reflective SOA and a chirped FBG,” Opt. Express 21(4), 5130–5139 (2013).

109. T. Kraetschmer, C. Lan, and S. T. Sanders, “Multiwavelength Frequency-Division-Multiplexed Light Source Based on Dispersion-Mode-Locking,” IEEE Photonics Technol. Lett. 19(20), 1607–1609 (2007).

110. B. Potsaid, B. Baumann, D. Huang, S. Barry, A. E. Cable, J. S. Schuman, J. S. Duker, and J. G. Fujimoto, “Ultrahigh speed 1050nm swept source/Fourier domain OCT retinal and anterior segment imaging at 100,000 to 400,000 axial scans per second,” Opt. Express 18(19), 20029–20048 (2010).

111. D. C. Adler, R. Huber, and J. G. Fujimoto, “Phase-sensitive optical coherence tomography at up to 370,000 lines per second using buffered Fourier domain mode-locked lasers,” Opt. Lett. 32(6), 626–628 (2007).

112. D. C. Adler, S. W. Huang, R. Huber, and J. G. Fujimoto, “Photothermal detection of gold nanoparticles using phase-sensitive optical coherence tomography,” Opt. Express 16(7), 4376–4393 (2008).

113. G. Palte, W. Wieser, B. R. Biedermann, C. M. Eigenwillig, and R. Huber, “Fourier domain mode locked (FDML) lasers for polarization sensitive OCT,” in (SPIE, 2009), 73720M.

114. A. H. Dulla, K. Shi, and J. A. Isatt, “Efficient sweep buffering in swept source optical coherence tomography using a fast optical switch,” Biomed. Opt. Express 3(12), 3054–3066 (2012).

115. T.-H. Tsai, C. Zhou, D. C. Adler, and J. G. Fujimoto, “Frequency comb swept lasers,” Opt. Express 17(23), 21257–21270 (2009).

116. W. Y. Oh, S. H. Yun, G. J. Tearney, and B. E. Bouma, “115 kHz tuning repetition rate ultrahigh-speed wavelength-swept semiconductor laser,” Opt. Lett. 30(23), 3159–3161 (2005).

117. W. Wieser, T. Klein, D. C. Adler, F. Trépanier, C. M. Eigenwillig, S. Karpf, J. M. Schmitt, and R. Huber, “Extended coherence length megahertz FDML and its application for anterior segment imaging,” Biomed. Opt. Express 3(10), 2647–2657 (2012).

118. B. Povazur, M. Siddiqui, and B. J. Vakoc, “A rapid, dispersion-based wavelength-swept laser for optical coherence tomography,” Opt. Express 23(7), 8437–8454 (2015).

119. T. Kraetschmer, C. Lan, and S. T. Sanders, “Photothermal detection of gold nanoparticles using phase-sensitive optical coherence tomography,” Opt. Express 16(7), 4376–4393 (2008).

120. C. Akcay, P. Parrein, and J. P. Rolland, “Estimation of longitudinal resolution in optical coherence imaging,” Appl. Opt. 41(25), 5256–5262 (2002).

121. Y. Chen, D. L. Burnes, M. de Bruin, M. Mujat, and J. F. de Boer, “Three-dimensional pointwise comparison of human retinal optical property at 845 and 1060 nm using optical frequency domain imaging,” J. Biomed. Opt. 14(2), 024016 (2009).

122. A. G. Podoleanu, “Fiber Optics, From Sensing to Non Invasive High Resolution Medical Imaging,” J. Lightwave Technol. 28(4), 624–640 (2010).

123. R. L. Shelton, W. Jung, S. I. Sayegh, D. T. McCormick, J. Kim, and S. A. Boppart, “Optical coherence tomography for advanced screening in the primary care office,” J. Biophotonics 2(7), 525–533 (2014).

124. C. D. Lu, M. F. Kraus, B. Potsaid, J. J. Liu, W. Choi, V. Jayaraman, A. E. Cable, J. Hornegger, J. S. Duker, and J. G. Fujimoto, “Handheld ultrahigh speed swept source optical coherence tomography instrument using a MEMS scanning mirror,” Biomed. Opt. Express 5(1), 293–311 (2014).

125. W. Wieser, W. Draxinger, T. Klein, S. Karpf, T. Pfeiffer, and R. Huber, “High definition live 3D-OCT in vivo: Heartbeat OCT: in vivo intravascular megahertz-optical coherence tomography,” Opt. Lett. 35(3), 622–625 (2014).

126. P. Herz, Y. Chen, A. Aguirre, J. Fujimoto, H. Mashimo, J. Schmitt, A. Koski, J. Goodnow, and C. Petersen, “Ultrahigh resolution optical biopsy with endoscopic optical coherence tomography,” Opt. Express 12(15), 3532–3542 (2004).

127. A. D. Aguirre, P. R. Hertz, Y. Chen, J. G. Fujimoto, W. Piyawattanametha, L. Fan, and M. C. Wu, “Two-axis MEMS Scanning Catheter for Ultrahigh Resolution Three-dimensional and En Face Imaging,” Opt. Express 15(5), 2445–2453 (2007).

128. K. H. Kim, B. H. Park, G. N. Maglulri, T. W. Lee, F. J. Rogomontich, M. G. Bancu, B. E. Bouma, J. F. de Boer, and J. J. Bernstein, “Two-axis magnetically-driven MEMS scanning catheter for endoscopic high-speed optical coherence tomography,” Opt. Express 15(26), 18130–18140 (2007).

129. T. Wang, T. Pfeiffer, E. Regar, W. Wieser, H. van Busekom, C. T. Lanec, G. Springeling, I. Krabbendam, A. F. W. van der Steen, R. Huber, and G. van Soest, “Heartbeat OCT: in vivo intravascular megahertz-optical coherence tomography,” Biomed. Opt. Express 6(12), 5021–5032 (2015).

130. K. L. Lurie, A. A. Gurjarpadhye, E. J. Seibel, and A. K. Ellerbee, “Rapid scanning catheterscope for expanded forward-view volumetric imaging with optical coherence tomography,” Opt. Lett. 40(13), 3165–3168 (2015).
131. B. D. Goldberg, B. J. Vakoc, W.-Y. Oh, M. J. Suter, S. Waxman, M. I. Freilich, B. E. Bouma, and G. J. Tearney, “Performance of reduced bit-depth acquisition for optical frequency domain imaging,” Opt. Express 17(9), 16957–16968 (2009).
132. A. G. Podoleanu, “Unbalanced versus balanced operation in an optical coherence tomography system,” Appl. Opt. 39(1), 173–182 (2000).
133. Y. Chen, D. M. de Bruin, C. Kerbage, and J. F. de Boer, “Spectrally balanced detection for optical frequency domain imaging,” Opt. Express 15(25), 16390–16399 (2007).
134. Y. Watanabe and T. Itagaki, “Real-time display on Fourier domain optical coherence tomography system using a graphics processing unit,” J. Biomed. Opt. 14, 060506 (2009).
135. K. Zhang and J. U. Kang, “Graphics processing unit accelerated non-uniform fast Fourier transform for ultra-high-speed, real-time Fourier-domain OCT,” Opt. Express 18(22), 23472–23487 (2010).
136. Y. Jian, K. Wong, and M. V. Sarunic, “Graphics processing unit accelerated optical coherence tomography processing at megahertz axial scan rate and high resolution video rate volumetric rendering,” J. Biomed. Opt. 18(2), 026002 (2013).
137. J. Rasakanthan, K. Sugden, and P. H. Tomlins, “Processing and rendering of Fourier domain optical coherence tomography images at a line rate over 524 kHz using a graphics processing unit,” J. Biomed. Opt. 16, 020505 (2011).
138. J. Probst, D. Hillmann, E. Lankena, C. Winter, S. Oelckers, P. Koch, and G. Hüttmann, “Optical coherence tomography with online visualization of more than seven rendered volumes per second,” J. Biomed. Opt. 15(2), 026014 (2010).
139. M. Sylwestrzak, D. Szał, M. Szkulmowski, I. Gorczynska, D. Bukowska, M. Wojtkowski, and P. Targowski, “Four-dimensional structural and Doppler optical coherence tomography imaging on graphics processing units,” J. Biomed. Opt. 17(10), 100502 (2012).
140. Y. Huang, Z. Ibrahim, D. Tong, S. Zhu, Q. Mao, J. Pang, W. P. Andree Lee, G. Brandacher, and J. U. Kang, “Microvascular anastomosis guidance and evaluation using real-time three-dimensional Fourier-domain Doppler optical coherence tomography,” J. Biomed. Opt. 18(11), 111404 (2013).
141. T. E. Ustun, N. V. Htinia, R. D. Ferguson, and D. X. Hammer, “Real-time processing for Fourier domain optical coherence tomography using a field programmable gate array,” Rev. Sci. Instrum. 79(11), 114301 (2008).
142. A. E. Desjardins, B. J. Vakoc, M. J. Suter, S. H. Yun, G. J. Tearney, and B. E. Bouma, “Real-time FPGA processing for high-speed optical frequency domain imaging,” IEEE Trans. Med. Imaging 28(9), 1468–1472 (2009).
143. A. Bradu, K. Kapinchev, F. Barnes, and A. Podoleanu, “Master slave en-face OCT/SLO,” Biomed. Opt. Express 6(9), 3655–3669 (2015).
144. M. K. K. Leung, A. Mariampillai, B. A. Standish, K. K. C. Lee, N. R. Munce, I. A. Vitkin, and V. X. D. Yang, “High-power wavelength-swept laser in Littman telescope-less polygon filter and dual-amplifier configuration for multichannel optical coherence tomography,” Opt. Lett. 34(18), 2814–2816 (2009).
145. C. Zhou, A. Alex, J. Rasakanthan, and Y. Ma, “Space-division multiplexing optical coherence tomography,” Opt. Express 21(16), 19219–19227 (2013).
146. H. Y. Lee, H. Sudkamp, T. Marvdashi, and A. K. Ellerbee, “Interleaved optical coherence tomography,” Opt. Express 21(22), 26542–26556 (2013).
147. H. Y. Lee, T. Marvdashi, L. Duan, S. A. Khan, and A. K. Ellerbee, “Scalable multiplexing for parallel imaging with interleaved optical coherence tomography,” Biomed. Opt. Express 5(9), 3192–3203 (2014).
148. D. Nankivil, A.-H. Dhalla, N. Gahm, K. Shia, S. Farsiu, and J. A. Izatt, “Coherence revival multiplexed, buffered swept source optical coherence tomography: 400 kHz imaging with a 100 kHz source,” Opt. Lett. 39(13), 3740–3743 (2014).
149. T. Klein, R. André, W. Wieser, T. Pfeiffer, and R. Huber, “Joint aperture detection for speckle reduction and increased collection efficiency in ophthalmic MHz OCT,” Biomed. Opt. Express 4(4), 619–634 (2013).
150. A. Wattak, R. Haindl, W. Trassitscher, B. Baumann, M. Pircher, and C. K. Hitzenberger, “Active-passive path-length encoded (APPLE) Doppler OCT,” Biomed. Opt. Express 7(12), 5233–5251 (2016).
151. D. Choi, H. Hiro-Oka, H. Furukawa, R. Yoshimura, M. Nakanishi, K. Shimizu, and K. Ohbayashi, “Fourier domain optical coherence tomography using optical demultiplexers imaging at 60,000,000 lines/s,” Opt. Lett. 33(12), 1318–1320 (2008).
152. O. P. Kocaglu, T. L. Turner, Z. Liu, and D. T. Miller, “Adaptive optics optical coherence tomography at 1 MHz,” Biomed. Opt. Express 5(12), 4186–4200 (2014).
153. A. F. Zuluaga and R. Richards-Kortum, “Spatially resolved spectral interferometry for determination of subsurface structure,” Opt. Lett. 24(8), 519–521 (1999).
154. Y. Yasuno, T. Endo, S. Makita, G. Aoki, M. Itoh, and T. Yatagai, “Three-dimensional line-field Fourier domain optical coherence tomography for in vivo dermatological investigation,” J. Biomed. Opt. 11, 014014 (2006).
155. B. Považay, A. Unterhuber, B. Hermann, H. Sattmann, H. Arthaber, and W. Drexler, “Full-field time-encoded frequency-domain optical coherence tomography,” Opt. Express 14(17), 7661–7669 (2006).
156. Y. Nakamura, S. Makita, M. Yamanari, M. Itoh, T. Yatagai, and Y. Yasuno, “High-speed three-dimensional human retinal imaging by line-field spectral domain optical coherence tomography,” Opt. Express 15(12), 7103–7116 (2007).
157. E. Beurepaire, A. C. Boccara, M. Lebec, L. Blanchot, and H. Saint-Jalmes, “Full-field optical coherence microscopy,” Opt. Lett. 23(4), 244–246 (1998).
158. A. F. Fercher, C. K. Hitzenberger, M. Sticker, E. Moreno-Barriuso, R. Leitgeb, W. Drexler, and H. Sattmann, “A thermal light source technique for optical coherence tomography,” Opt. Commun. 185(1-3), 57–64 (2000).

159. A. Dubois, L. Vabre, A.-C. Boccara, and E. Beaurepaire, “High-resolution full-field optical coherence tomography with a Linnik microscope,” Appl. Opt. 41(4), 805–812 (2002).

160. W. Y. Oh, B. E. Bouna, N. Iftimia, S. H. Yun, R. Yelin, and G. J. Tearney, “Ultrahigh-resolution full-field optical coherence microscopy using InGaAs camera,” Opt. Express 14(2), 726–735 (2006).

161. T. Bonin, G. Franke, M. Hagen-Eggert, P. Koch, and G. Hüttmann, “In vivo Fourier-domain full-field OCT of the human retina with 1.5 million A-lines/s,” Opt. Lett. 35(20), 3432–3434 (2010).

162. D. Hillmann, C. Lührs, T. Bonin, P. Koch, and G. Hüttmann, “Holoscopy—holographic optical coherence tomography,” Opt. Lett. 36(13), 2390–2392 (2011).

163. D. Hillmann, T. Bonin, C. Lührs, G. Franke, M. Hagen-Eggert, P. Koch, and G. Hüttmann, “Common approach for compensation of axial motion artifacts in swept-source OCT and dispersion in Fourier-domain OCT,” Opt. Express 20(6), 6761–6776 (2012).

164. R. Wang, J. X. Yun, X. Yuan, R. Goodwin, R. R. Markwald, and B. Z. Gao, “Megahertz streak-mode Fourier domain optical coherence tomography,” J. Biomed. Opt. 16, 066016 (2011).

165. D. Hillmann, H. Spahr, C. Hain, H. Sudkamp, G. Franke, C. Pfaffle, C. Winter, and G. Hüttmann, “Aberration-free volumetric high-speed imaging of in vivo retina,” Sci. Rep. 6, 35209 (2016).

166. M. F. Kraus, B. Potsaid, M. A. Mayer, R. Bock, B. Baumann, J. J. Liu, J. Hornegger, and J. G. Fujimoto, “Motion correction in optical coherence tomography volumes on a per A-scan basis using orthogonal scan patterns,” Biomed. Opt. Express 3(6), 1182–1199 (2012).

167. K. J. Mohler, W. Draxinger, T. Klein, J. P. Kolb, W. Wieser, C. Haritoglu, A. Kampil, J. G. Fujimoto, A. S. Neubauer, R. Huber, and A. Wolf, “Combined 60° Wide-Field Choroidal Thickness Maps and High-Definition En Face Vasculature Visualization Using Swept-Source Megahertz OCT at 1050 nm,” Invest. Ophthalmol. Vis. Sci. 56(11), 6284–6293 (2015).

168. R. J. Zawadzki, S. M. Jones, S. S. Olivier, M. Zhao, B. A. Bower, J. A. Izatt, S. Choi, S. Laut, and J. S. Werner, “Adaptive-optics optical coherence tomography for high-resolution and high-speed 3D retinal imaging in vivo,” Opt. Express 13(21), 8532–8546 (2005).

169. M. Wojtkowski, B. Kaluzny, and R. J. Zawadzki, “New Directions in Ophthalmic Optical Coherence tomography,” Optom. Vis. Sci. 89(5), 524–542 (2012).

170. V. J. Srinivasan, D. C. Adler, Y. Chen, I. Gorczynska, R. Huber, J. S. Duker, J. J. Schuman, and J. G. Fujimoto, “Ultrahigh-Speed Optical Coherence Tomography for Three-Dimensional and En Face Imaging of The Retina with a Linnik microscope,” Invest. Ophthalmol. Vis. Sci. 49(11), 5103–5110 (2008).

171. T.-H. Tsai, H.-C. Lee, O. O. Ahsen, K. Liang, M. G. Giacomelli, B. M. Potsaid, Y. K. Tao, V. Jayaraman, M. Figueiredo, Q. Huang, A. E. Cable, J. Fujimoto, and H. Mashino, “Ultrahigh speed endoscopic optical coherence tomography for gastroenterology,” Biomed. Opt. Express 5(12), 4387–4404 (2014).

172. T. Yonetsu, B. E. Bouna, K. Kato, J. G. Fujimoto, and I.-K. Jang, “Optical Coherence Tomography 15 Years in CT the electric field cardiology,” Circulation Journal 77, 1933–1940 (2013).

173. C. Blüter, T. Klein, B. Grajcicar, T. Schmoll, W. Wieser, R. Andre, R. Huber, and R. A. Leitgeb, “Ultrahigh-speed non-invasive widefield angiography,” J. Biomed. Opt. 17(7), 070505 (2012).

174. W. Wei, J. Xu, U. Baran, S. Song, W. Qin, X. Qi, and R. K. Wang, “Intervolume analysis to achieve four-dimensional optical microangiography for observation of dynamic blood flow,” J. Biomed. Opt. 21(3), 036005 (2016).

175. B. Baumann, B. Potsaid, M. F. Kraus, J. J. Liu, D. Huang, J. Hornegger, A. E. Cable, J. S. Duker, and J. G. Fujimoto, “Total retinal blood flow measurement with ultrahigh speed swept source/Fourier domain OCT,” Biomed. Opt. Express 2(6), 1539–1552 (2011).

176. Y. Wang, B. A. Bower, J. A. Izatt, O. A. Tan, and D. Huang, “Retinal blood flow measurement by circumpapillary Fourier domain Doppler optical coherence tomography,” J. Biomed. Opt. 13, 064003 (2008).

177. L. Ambrozinski, S. Song, S. J. Yoon, I. Pelivanov, D. Li, L. Gao, T. T. Shen, R. K. Wang, and M. O’Donnell, “Acoustic micro-tapping for non-contact 4D imaging of tissue elasticity,” Sci. Rep. 6, 38967 (2016).

178. S. Wang, M. Singh, A. L. Lopez 3rd, C. Wu, R. Raghunathan, A. Schill, J. Li, K. V. Larin, and I. V. Larinina, “Direct four-dimensional structural and functional imaging of cardiovascular dynamics in mouse embryos with 1.5 MHz optical coherence tomography,” Opt. Lett. 40(20), 4791–4794 (2015).

179. M. W. Jenkins, D. C. Adler, M. Gargesha, R. Huber, F. Rothenberg, J. Belding, M. Watanabe, D. L. Wilson, J. G. Fujimoto, and A. M. Rollins, “Ultrahigh-speed optical coherence tomography imaging and visualization of the embryonic avian heart using a buffered Fourier Domain Mode Locked laser,” Opt. Express 15(10), 6251–6267 (2007).

180. S. A. Boppart, J. Herrmann, C. Pitris, D. L. Stamper, M. E. Brezinski, and J. G. Fujimoto, “High-Resolution Optical Coherence Tomography-Guided Laser Ablation of Surgical Tissue,” J. Surg. Res. 82(2), 275–284 (1999).

181. Y. Zhang, T. Pfeiffer, M. Weller, W. Wieser, R. Huber, J. Raczkowsky, J. Schipper, H. Wörm, and T. Klenzner, “Optical Coherence Tomography Guided Laser Cochleostomy: Towards the Accuracy on Tens of Micrometer Scale,” BioMed Res. Int. 2014, 251814 (2014).

182. M. V. Sarunic, S. Weinberg, and J. A. Izatt, “Full-field swept-source phase microscopy,” Opt. Lett. 31(10), 1462–1464 (2006).
183. H. Spahr, D. Hillmann, C. Hain, C. Pfäffle, H. Sudkamp, G. Franke, and G. Hüttmann, “Imaging pulse wave propagation in human retinal vessels using full-field swept-source optical coherence tomography,” Opt. Lett. 40(20), 4771–4774 (2015).

184. D. Hillmann, H. Spahr, C. Pfäffle, H. Sudkamp, G. Franke, and G. Hüttmann, “In vivo optical imaging of physiological responses to photostimulation in human photoreceptors,” Proc. Natl. Acad. Sci. U.S.A. 113(46), 13138–13143 (2016).

185. M. Singh, C. Wu, C.-H. Liu, J. Li, A. Schill, A. Nair, and K. V. Larin, “Phase-sensitive optical coherence elastography at 1.5 million A-Lines per second,” Opt. Lett. 40(11), 2588–2591 (2015).

186. G. Geerling, M. Müller, C. Winters, H. Hoerauf, S. Oelckers, H. Laqua, and R. Birngruber, “Intraoperative 2-dimensional optical coherence tomography as a new tool for anterior segment surgery,” Arch. Ophthalmol. 123(2), 253–257 (2005).

187. E. Lankenau, D. Klinger, C. Winter, A. Malik, H. H. Müller, S. Oelckers, H-W. Pau, T. Just, and G. Hüttmann, “Combining Optical Coherence Tomography (OCT) with an Operating Microscope,” in Advances in Medical Engineering, T. M. Buzug, D. Holz, J. Bongartz, M. Kohl-Bareis, U. Hartmann, and S. Weber, eds. (Springer Berlin Heidelberg, Berlin, Heidelberg, 2007), pp. 343–348.

188. Y. K. Tao, J. P. Ehlers, C. A. Toth, and J. A. Izatt, “Intraoperative spectral domain optical coherence tomography for vitreoretinal surgery,” Opt. Lett. 35(20), 3315–3317 (2010).

189. S. Binder, C. I. Falkner-Radler, C. Hauger, H. Matz, and C. Glittenberg, “Feasibility of intraocular spectral-domain optical coherence tomography,” Retina 31(7), 1332–1336 (2011).

190. C. Viehland, B. Keller, O. M. Carrasco-Zevallos, D. Nankivil, L. Shen, S. Mangalesh, T. Viet, A. N. Kuo, C. A. Toth, and J. A. Izatt, “Enhanced volumetric visualization for real time 4D intraoperative ophthalmic swept-source OCT,” Biomed. Opt. Express 7(5), 1815–1829 (2016).

191. P. Steven, C. Le Blanc, E. Lankenau, M. Krug, S. Oelckers, L. M. Heindl, U. Gehlsen, G. Huettmann, and C. Cursiefen, “Optimising deep anterior lamellar keratoplasty (DALK) using intraoperative online optical coherence tomography (iOCT),” Br. J. Ophthalmol. 98(7), 900–904 (2014).

192. O. M. Carrasco-Zevallos, B. Keller, C. Viehland, L. Shen, G. Waterman, B. Todorich, C. Shieh, P. Hahn, S. Farsiu, A. N. Kuo, C. A. Toth, and J. A. Izatt, “Live volumetric (4D) visualization and guidance of in vivo human ophthalmic surgery with intraoperative optical coherence tomography,” Sci. Rep. 6, 31689 (2016).

1. Introduction

Fast, ultrafast, megahertz

The first published optical coherence tomography (OCT) scan had an acquisition time of seconds – for a single B-Scan [1]. Since then, imaging speed in OCT has increased dramatically by several orders of magnitude [2–4]. In OCT, speed is usually given as depth scan rate in axial scans per second, abbreviated as A-scans/s, with the unit of Hertz (Hz). Over the last 25 years, A-scan rates increased from a few hundred Hz to many megahertz. Thus, the concept of “fast” and “ultrafast” needed repeated redefinition whenever new speed records had been set. While some tens of kHz speed may have been regarded as ultrafast in the past, this speed is now considered standard or even slow speed. Thankfully, term inflation did not progress to hyperfast, and it has become more common to avoid these relative terms altogether. Instead, the fastest systems nowadays are usually simply classified by their A-scan rate as “MHz-OCT” or “multi-MHz-OCT” [5–8]. The speed development has been largely pushed by advances in the underlying OCT hardware. Therefore, we will start this review by briefly discussing the broad technological developments that drove imaging speed in OCT. Subsequently, we focus on one technology that led to many new records in OCT speed, which is wavelength-swept lasers. Then, important system performance characteristics of high-speed OCT systems are presented and system design aspects are discussed.

Over the past, it has sometimes been debated if it makes any sense to achieve ever higher OCT speeds. In general, the answer is complex, depends on the particular application, and needs to take into account changing limits for maximum permissible exposure, see e.g [9]. Moreover, the availability of suitable and cost-effective hardware is certainly crucial for successful commercialization. However, the authors think that there is no fundamental reason, why not to go to MHz or even multi-MHz speeds, since high-quality imaging has been demonstrated in a variety of application with a range of different technologies, some of which are based on commercially available swept lasers. Since OCT image quality can be as good as for slower systems, the massive increase in acquired data per time will either lead to shorter acquisition times, or to more information and hence better value for the end user. “Better
information” can take many different forms, which we will discuss in a separate section on the applications of fast OCT. We will conclude this review with a brief overview on the past and future development of fast OCT.

**OCT implementations**

The first generation of OCT was based on mechanical path-length scanning, the so-called time-domain (TD-) OCT. In the first couple of years, considerable engineering efforts went into improvement of the scanning delay lines, which need to provide optical path length differences of a few millimeters, repeatedly scanned at very high repetition rates. While motorized linear stages where used in the beginning [10], several concepts such as piezo-coupled fiber stretchers [11] and galvanometer-based dispersive delay lines [12] levered mechanical motion to get to higher speeds. Ultimately, this enabled video-rate OCT for the first time - that is acquisition of two-dimensional images with more than 20-30 frames per second [13]. Higher speeds were difficult not only due to mechanical limits of the scanning mechanism, but also since noise becomes critical for shorter exposure times.

This is where the next generation of OCT systems came into play - Fourier-domain (FD-) OCT, which has almost instantaneously triggered new speed records that were previously unimaginable. In contrast to TD-OCT, in FD-OCT the spectrum of the inference signal is resolved ([14–17]). The additional spectral information renders the mechanical scanning unnecessary. Instead, the A-scan rate is simply given by the rate with which spectra can be acquired. FD-OCT comes in two variants. In spectral-domain (SD-) OCT, a broadband light source is used in combination with a spectrometer, which usually consists of a grating and a line scan camera. The interference spectrum is thus spatially encoded on the different detector pixels. In the second implementation, swept-source (SS-) OCT, the spectrum is encoded in time. A wavelength-tunable light source [18] scans its wavelength, and a single (or differential) photo detector records all wavelengths over time. Besides this most common nomenclature, sometimes the term SD-OCT is used to denote both FD-OCT methods, and optical frequency domain imaging (OFDI) is used synonymously for SS-OCT [15, 19]. Apart from the high speed, which became technically more feasible by FD-OCT technology, the fact, that FD-OCT is inherently more sensitive than in TD-OCT is at least as important [20–22]. An intuitive explanation for the higher sensitivity of FD-OCT is that all axial locations are recorded throughout the entire A-scan duration, whereas in TD-OCT only photons at the current axial depth contribute to the signal.

In this review, we will mainly focus on “standard” OCT implementations with confocal detection and point-scanning (flying-spot) imaging beams. In SS-OCT, speed in this kind of systems is simply given by the laser sweep rate, potentially multiplexed by the number of scanning beams. Other OCT concepts that allow for very high speed, such as line-field and full-field OCT, will only be mentioned briefly. Those techniques are not confocal and hence exhibit different image quality than standard OCT systems. Therefore, it is currently no fully established in which application scenarios these non-confocal systems may replace standard OCT systems.

**2. High-speed sources: wavelength-swept lasers**

Rapidly wavelength swept lasers or simply swept sources are a type of lasers, whose development has tightly been linked to OCT. In principle, swept sources for OCT are tunable lasers. Tunable lasers have been known for a long time, however, in application for OCT, a very “unusual” set of operating parameters is required. In the early 2000’s when the OCT community became increasingly interested in SS-OCT, there was no laser available, which could achieve the desired 100 performance. The hundred cube stands for roughly the performance most of today’s commercial SS-OCT systems achieve, representing 100 nm wavelength sweep range, at 100 kHz repetition rate and with 100 pm instantaneous linewidth. So, swept sources for OCT are tunable lasers, but usually they require different components,
dedicated cavity layouts and novel operating regimes. Depending on the various interpretations of the word “laser”, there have been discussions if some of the OCT swept sources are actually lasers, however here we will still use the terms swept laser and swept (light) source interchangeably.

**Key performance characteristics**

Swept lasers are a relatively new technology, and their applications critically depend on several of their key properties. Thus, many different performance indicators exist for swept lasers, and it is important to clearly define how they are measured. In this section, we present the most common properties of swept lasers, and indicate how they affect OCT applications where appropriate. We start with characteristics that depend on the (wavelength-resolved) intensity, and then take a more detailed look at the optical signal.

**Intensity-based characteristics**

The sweep is the optical signal generated by the laser. A sweep is defined by a monotonic increase in optical frequency from start wavelength $\lambda_{\text{start}}$ to end wavelength $\lambda_{\text{end}}$, which define the sweep range $\Delta\lambda$:

$$\Delta\lambda = |\lambda_{\text{end}} - \lambda_{\text{start}}|.$$  

The sweep range is a very important parameter, since it is inversely proportional to the theoretically achievable axial resolution in OCT, see section 4. So in general, the larger the sweep range, the better the axial resolution. The sweep range represents the total sweep width. Sometimes values like “3dB width” are cited to characterize the lasers tuning range. This may not be very useful, since the output spectra of most swept lasers are not Gaussian.

The center wavelength $\lambda_{c}$ is the mean of start and end wavelength. The center wavelength is important because it represents a compromise between water absorption and scattering. For longer wavelengths, scattering decreases, but water absorption becomes more dominant. There is less scattering at longer center wavelengths, but more water absorption. The center wavelength is usually not the center of gravity of the sweep with respect to power, but the mean of start and end wavelength $\lambda_{c} = (\lambda_{\text{end}} + \lambda_{\text{start}}) / 2$. The wavelength evolution can have one of two directions. If the wavelength changes from short (“blue”) to long (“red”) wavelengths over time, the sweep is often termed “forward sweep”, and “backward sweep” for blue to red operation. This terminology must not be confused with the terms up-chirp (red-to-blue) and down-chirp (blue-to-red). Swept lasers can produce unidirectional sweeps, or alternating forward and backward sweeps, in which case the sweep is said to be bidirectional. For application in OCT, the laser needs to have a repetitive sweep train with period $T_{\text{rep}}$. For some swept laser types, there can be a dead time between the end of one sweep and the beginning of the next sweep. For instance, this is the case when for some laser one sweep direction is used, while the intracavity filter has some fly back time. In those cases, the duty cycle is less than 1. The duty cycle $D$ is defined as:

$$D = \frac{T_{\text{sweep}}}{T_{\text{rep}}},$$  

where $T_{\text{rep}}$ is the sweep repetition rate and $T_{\text{sweep}}$ the duration of the sweep. In OCT, one sweep generates a single depth profile at one sample location, the A-scan. Images are generated by stitching of those A-scans, so the OCT A-scan rate is equal or at least directly proportional to the sweep repetition rate, which is simply called sweep rate $f_{\text{sweep}} = 1/T_{\text{rep}}$. The sweep range $\Delta\lambda$ is sometimes called the (optical) bandwidth of the laser. This should not be confused with the instantaneous linewidth of the laser, which will be discussed in the next subsection.
The noise of the optical intensity of a swept laser can be characterized on different timescales, and either aggregated or wavelength-resolved [23]. Usually it is not possible to measure the noise of swept sources for OCT by RF spectrum analyzers or simple FFT of power traces, as commonly done for standard lasers. In OCT swept sources the naturally occurring output power change over the sweep would congest the noise measurement. It should be noted that for OCT lasers, such a variation is not a problem but it is actually desired, since a suitable power envelope over the sweep inherently suppresses side lobes in the image. The relative intensity noise (RIN) is the standard deviation of the respective observable vs. the mean of the observable. For instance, the RIN of the average power can be written as:

\[
RIN_{power} = \frac{\text{Stdev}(P)}{\text{Avg}(P)}. \tag{3}
\]

It should be noted that RIN measurements are strongly dependent on the measurement bandwidth. RIN measured with a slow optical power meter will be much lower than RIN measured with a high-speed photodiode and fast oscilloscope. Hence, RIN measurements without stated bandwidth are of limited value. Two methods exist to investigate the wavelength-dependent noise [23]: First, RIN can be measured within one sweep over a short time span. The measured “intra-sweep noise” can be understood as short term time fluctuations while the source is sweeping, and can also be called “sliding RIN”. Second, the fluctuations can be analyzed at the same wavelength in between different sweeps. This inter-sweep noise is called ortho-RIN, and it can be shown that it is always higher than the sliding RIN values. Both ortho- and sliding RIN are measured on the short timescales of one or neighboring sweeps. Depending on the sweep mechanism and filter type used, both center wavelength and sweep range may also undergo long-term fluctuations, which can be characterized by their standard deviation. Some swept lasers need active stabilization of both sweep range and center wavelength, at least if they are operated in a non-temperature stabilized environment. The sweep range and center wavelength stability has a critical impact on data acquisition and processing in OCT. Here, the sampled data points need to be evenly spaced in optical wavenumber \( \mathbf{k} \). This is a straightforward requirement for a wavelength-swept laser whose instantaneous optical wavenumber is a linear function of time. Most swept lasers do not exhibit this linear sweep, and hardware or software solutions need to be implemented to achieve the desired sampling. The wavelength- or wavenumber vs. time evolution can be measured by various means, for instance with the help of gas cells, etalons, fiber Bragg gratings or an interferometer.

The last parameter in this section that can be measured with (wavelength-dependent) intensity alone is the degree of polarization of the laser. Of course, for an instant in time, the optical field is well defined, and the degree of polarization will be 100%. But again, the state of polarization can change over time and hence over wavelength. For non-polarization sensitive OCT it is usually not critical if such fluctuations exist, and typical fiber-coupled OCT systems will lead to strong wavelength-dependent polarization even for a perfectly polarized laser source [24].

**Electric field characteristics**

For ultrafast pulses, it is often desirable to reconstruct the full electrical field of the pulse [25]. The electric field can either be characterized in the time-domain, or it’s Fourier equivalent, the frequency domain. In the frequency domain, the amplitude and phase of all individual frequency components of the swept laser have to be known to fully describe the electric field. This is usually neither necessary, although possible in principle [26], up to a single additive phase factor, e.g. the envelope phase. While a femtosecond laser pulse has a duration of only a few optical cycles that can be readily plotted and analyzed, a 1 \( \mu \)s long sweep has hundreds
of millions of optical cycles. Moreover, the full electric field evolution of a swept laser can be extremely complex. Thus, swept lasers are usually characterized by derived quantities that are experimentally easily accessible.

![Diagram](image)

**Fig. 1.** Left: Interference signal (fringe) of a sweep with non-linear time-wavenumber characteristic. Middle: Resampled signal. Right. Fourier-transform of the linearized fringe, showing the point-spread function (PSF).

The instantaneous linewidth characterizes the swept laser’s average coherence length. It also indicates with which resolution small spectral features can be identified with a swept laser. For instance, in fiber sensor or gas sensing applications, the instantaneous linewidth determines the measurement resolution of the system. For an idealized (non-swept) continuous wave (CW) laser, the optical waveform is a single-frequency sine. The linewidth then becomes arbitrarily small with increasing emission or measurement duration, and the words linewidth and bandwidth denote the same physical property. They are the spectral width of the single electrical field component in the frequency domain. The minimum time required to measure in order to achieve a certain spectral resolution is given by the time-bandwidth-limit. A swept laser waveform consists of many different optical frequency components. Since the optical frequency changes rapidly over time, the “allowable” measurement time is intrinsically limited. Hence, there is a fundamental limit to the measurement time and small instantaneous linewidths can’t be measured directly by analyzing a short slice in time of the laser output [27].

Instead, an “average linewidth” is usually measured via the laser’s coherence properties. Like in OCT with a single reflector in the sample arm, a beat signal (fringe) is generated by superimposing a sweep and a slightly delayed copy of it, for instance by using a Mach-Zehnder or Michelson interferometer. For a swept laser with linear time-wavenumber characteristics, the fringe pattern has only one radio-frequency component, which is proportional to the delay between the interferometer arms. For a nonlinear source, the fringe pattern can be linearized by interpolation of the evenly sampled signal, see Fig. 1. The visibility of this fringe pattern will decrease with increased delay between the sweeps due to finite linewidth. The fringe visibility can be defined as the time averaged amplitude of the interferometric fringe signal envelope. In FD-OCT, the actual OCT signal strength is given by the intensity of the Fourier-transform of the fringe signal, which represents the axial point-spread function (PSF). The decay of the PSF amplitude over depth/delay, the so-called OCT roll-off, can be faster than the decay of the fringe visibility. This can be caused by...
imperfections in the linearization of the sweep (see below) or due to phase noise in the fringe signal. Hence the roll-off should be defined by the decay of the PSF rather than by loss of fringe visibility. Usually the 6 dB drop in PSF amplitude in a Michelson interferometer is defined as “roll-off” length. It is important to note that different definitions of coherence length are used, which differ by a factor of 4! If the instantaneous linewidth has a Gaussian profile, the coherence length can be calculated from the linewidth by Eq. (7). Table 1 lists the most common key characteristics of swept lasers.

Table 1. Swept laser key characteristics

| Parameter                | Unit     | Definition                                                                 |
|--------------------------|----------|-----------------------------------------------------------------------------|
| Center wavelength        | nm       | Mean value of start and end wavelength                                       |
| Sweep range              | nm       | Difference between start and end wavelength.                               |
| Sweep rate               | Hz       | 1/sweep period, i.e. repetition rate of the sweep.                         |
| Sweep duration           | s        | “On”-time of the sweep during the sweep period.                             |
| Duty cycle               | %        | Sweep time divided by sweep period.                                         |
| Average power            | mW       | Average optical power over the sweep duration or sweep period               |
| Peak power               | mW       | Peak optical power                                                         |
| Spectral shape           | %        | Profile of the sweep spectrum.                                             |
| Linearity                | %        | Deviation from a sweep where optical wavenumber (k) is a linear function of time. |
| RIN                      | %        | Relative intensity noise, defined as optical intensity standard deviation over mean intensity. Detection bandwidth must be specified when giving RIN values. |
| Ortho-RIN                | %        | RIN at the same spectral location, but over succeeding sweeps.              |
| Sliding-RIN              | %        | RIN measured within a sweep.                                               |
| Degree of polarization   | %        | Amount of light passing through a (linear) polarizer.                      |
| Axial resolution         | µm       | PSF amplitude full-width at half maximum (FWHM). Proportional to sweep range and dependent on spectral shape. |
| Side-lobe suppression    | dB       | Power ratio of PSF main peak to largest side lobe peak.                    |
| Instantaneous linewidth  | pm / Hz  | Optical linewidth from a short temporal slice of the sweep. May change over the sweep. |
| Average linewidth        | pm       | Average of all instantaneous linewidths, weighted with intensity. Usually obtained as Fourier-transform of a roll-off measurement. |
| Coherence length         | mm       | Various definitions exist, which differ by a factor of four.               |
| Roll-off (6dB)           | nm       | OCT imaging depth in air, where the PSF has dropped by 6 dB.               |
| R-number                 | nm/dB    | Inverse of roll-off linear fit coefficient                                 |
| Phase stability          | mrad     | Standard deviation of the PSF peak phase. Depends on PSF axial position and may be limited by shot noise. |
| Sweep-to-sweep           | -        | Short-term change in any of the characteristics of the sweep over time.    |

Gain media for swept lasers

Many swept lasers are built from discrete optical components. The two most important components in those lasers are the gain medium and the wavelength-selective filter. Gain media for swept lasers will be discussed in this section, filters in the next section. Three important parameters of all gain media are the wavelength-dependent gain profile, the polarization dependence and the effective lifetime of the excited state, which is directly linked to the amount of energy that can be stored in the gain medium. In stationary lasing, gain equals loss. Since all gain media have wavelength-dependent gain, the achievable sweep range is limited by the spectral width of the gain profile. Polarization dependent gain occurs for non-isotropic gain media. If gain is dependent on the input state of polarization, usually active or passive polarization control is necessary for consistent laser performance. The “excited state lifetime” indicates the speed with which the gain medium responds to changes in optical input. Long excited state lifetimes as found in rare-earth doped fibers, can rapidly lead to undesired Q-switching. An overview of gain media used for swept lasers can be found in Table 2.
Table 2. Gain media used in swept lasers

| Gain medium | Typical bandwidth | Upper state lifetime | Polarization dependent gain |
|------------|------------------|----------------------|-----------------------------|
| SOA        | 100 nm           | 380 ps [28]          | Yes                         |
|            |                   | 700 ps               | Pol. independent models available |
| Rare earth | 50 nm            | Ytterbium: 0.8 ms [29] | No                          |
| Raman      | 30 nm, but multiplexing is straightforward | Erbium: 10 ms [30] | Yes                         |

It is tempting to increase total gain or the gain bandwidth by combination of several gain media. In general, those different gain media can be combined in series, or in parallel. Serial combinations of gain media have been used successfully, for instance with two semiconductor amplifiers [31], or a rare-earth doped amplifier in conjunction with an SOA [32]. Serial configurations will increase gain, but will only increase bandwidth considerably if both gain media show small enough absorption in the other gain medium’s gain region. In contrast, a parallel configuration of gain would increase the available bandwidth. However, parallel configurations are usually not very successful, since they inevitably lead to an intracavity Mach–Zehnder-like interferometer structure. The resulting interference in the region of spectral overlap generates loss and excess noise [33].

Today, the by far most widely employed gain medium for swept lasers are semiconductor optical amplifiers (SOAs). Compared to other gain media, they have several advantages. First, their ultrashort gain response time of a few hundred picoseconds only leads to laser intensity fluctuations of at least several GHz, which usually lies outside the OCT detection range. Moreover, the amount of stored energy in the SOA is correspondingly low, making catastrophic pulsing by spiking or Q-switching impossible. SOAs also offer broadband optical amplification with gain bandwidths (10dB) larger than 100 nm, and peak gains of more than 30 dB. This performance can nowadays be achieved in all popular wavelength ranges for OCT, especially with gains centered around 1060 nm, 1310 nm and 1550 nm. Most SOAs have polarization dependent gain, i.e. they only amplify one linear state of polarization. However, some implementations with polarization independent gain are available at the telecom wavelength bands of 1310 nm and 1550 nm. Their construction is more complicated than for polarization dependent SOAs [34], hence somewhat lower specifications in terms of gain and gain bandwidth are usually obtained. Since almost all confocal OCT systems are fiber coupled, it is very advantageous for the light source and hence the gain medium to be fiber coupled. SOA technology has been heavily influenced by the telecom industry, so high-quality and reliable fiber coupled devices are available. Novel semiconductor designs need to compete with these established solutions. For instance, tapered optical amplifiers are a special type of SOA which can provide very high gain and output power. However, the output beam has no circular symmetry, which renders fiber coupling more difficult [31]. Another important characteristic of SOAs is the linewidth-enhancement factor $\alpha$, which leads to red-shifting of the amplified output compared to the input [35, 36]. This behavior can have a strong influence on the lasers dynamics in many swept laser types.

Rare-earth doped fiber amplifiers have contributed much to the success of fiber-optic long-haul communications, and are currently extensively used in high-power fiber lasers. This makes them a well-understood and easily available option for swept lasers. The most popular rare earth dopants are Ytterbium (Yb) and Erbium (Er), which are usually used for amplification of wavelengths around 1000 nm (Yb) and 1550 nm (Er). Other dopants such as Thulium allow access to longer wavelengths around 2 µm [37], which might be interesting for some OCT applications where water absorption is not critical. Compared to the other gain media discussed in this section, rare-earth ions in fiber have a very long excited state lifetime. The host glass composition affects the excited state lifetime as well as the absorption and emission spectra. For instance, the typical lifetime of Ytterbium doped-fiber is around 0.8 ms, while Yb in some glasses can reach lifetimes of around 1.5 ms [29]. These long lifetimes
implies that large amounts of energy can be stored. The gain profile usually shows a strong wavelength dependence, and the very high peak amplification is necessary to provide sufficient gain over a wavelength range that is larger than a few tens of nanometers. The combination of high peak gain and large stored energy makes rare-earth doped fiber amplifiers susceptible to Q-switching [38]. While this can be beneficial for high-power laser applications, it may lead to unstable laser operation or even catastrophic failure in swept OCT lasers. Thus, rare-earth doped fibers have been mainly used to boost swept laser power outside the cavity [39], or in addition to other amplifiers in the cavity [32]. They were also employed in some swept lasers that rely on nonlinear phenomena [39]. In contrast, the much shorter upper state lifetime in SOAs can generate instabilities and relaxation oscillations in the GHz range [40], which is faster than the electronic detection bandwidth of today's slow OCT systems and therefore this effect is often ignored. However, for modern high speed MHz OCT, with fringe frequencies well into the GHz, this effect becomes critical.

The last gain mechanism discussed in this section is stimulated Raman scattering (SRS) [41]. In SRS, photons with two different wavelengths interact in a medium, when their energy difference falls within the Raman gain bandwidth, energy is transferred from the field with higher frequency to the field with lower frequency, the Stokes frequency. In contrast to isotropic emission in spontaneous Raman scattering, emission in stimulated Raman scattering occurs only in forward and backward directions as defined by the pump. In silica fibers, Raman gain occurs at 13 THz below the optical pump frequency [42]. For instance, with a 1457 nm pump, the peak gain is located at 1550 nm. In theory, Raman amplification has a couple of unique advantages: First, multiple pump diodes can easily be coupled into the gain fiber via a suitable wavelength division multiplexer (WDM) [43]. To a first approximation, the resulting gain spectrum is the sum of the individual gain spectra [44], and large sweep ranges might be obtained. Second, Raman amplifiers have a lower noise figure than SOAs [45]. Raman amplifiers can theoretically have noise figures close to the quantum limit of an ideal amplifier of 3 dB [46], which might reduce laser RIN. Third, the effective interaction time between pump and signal photons, given by the inverse bandwidth of the Raman gain, for the stimulated Raman effect is very short, on the order of femtoseconds. Noise frequencies might thus be shifted to regions far outside typical OCT detection bandwidths. Fourth, in Raman amplifiers, the maximum output power is only limited by pump intensity and the maximum power capacity of the fiber. Thus, extremely high powers could theoretically be obtained. Last, pumping with two orthogonally polarized pump diodes could provide polarization independent gain. With a single pump diode, orthogonal polarization of pump and signal results in gain penalty of more than 13 dB, whereas complete polarization scrambling in the fiber reduces the threshold pump power by a factor of two [41]. Despite these possible advantages, a major drawback of Raman gain is the weak interaction between pump and signal photons. Hence long fibers are necessary. Specialized Raman fiber (e.g. OFS Specialty Photonics Division) has a peak Raman gain factor of $G_R = 2.5$ W/km, about 5.7 times higher than for standard single mode fibers (SSMF) which have a peak $G_R$ of about 0.44 W/km. These gain factors still imply a fiber length of at least several hundreds of meters for reasonable pump powers of less than one Watt. This is prohibitive for most laser designs. Hence only a few swept lasers have been demonstrated that use Raman gain, in particular FDML swept laser, which already use long fiber length in their cavity [47]. Approaches exploiting various nonlinear effects and more complex gain mechanisms have been studied [48], but could not challenge the unique role of SOAs as gain medium for swept lasers.

**Wavelength-selective filters for swept lasers**

Most swept laser implementations use active tuning of the emission wavelength, which means that the instantaneous emission wavelength is directly defined by an intracavity tunable filter. For use in swept lasers, these filters need to have a broad tuning range on the order of hundred nanometers, and narrow spectral transmission of a few hundred picometers or less.
over the entire tuning range. Moreover, the filter tuning speed must be very high to enable fast sweep rates from several kHz to multiple MHz. Finally, the free spectral range has to exceed the lasing threshold bandwidth, so that only one laser wavelength band is active at a time. These requirements rule out many filter types that are used in standard tunable lasers. In this section, we present the most popular implementations: scanned grating filters and Fabry-Perot filters. The first filter type is based on optical gratings. To tune the wavelength, the angle between beam and grating needs to be tuned. For high speeds, the beam is usually deflected with a beam scanning mechanism, see Fig. 2. Polygon mirror scanners were used for some of the first filter designs that provide wide tuning range, high speed and narrow linewidth [49]. Other implementations use acousto-optic [50] or electro-optical scanners, but their number of resolvable points and hence their finesse is limited [51]. Thus large sweep range and narrow filter width is not possible simultaneously [52]. In general, scanned grating type filters with rotating polygon mirrors have the advantage of a relatively linear time-to-wavelength output. Moreover, the center wavelength and tuning range are inherently given by the optical design and scan position so drift is usually no problem and no active control of center wavelength and sweep range is necessary.

![Diagram](image)

Fig. 2. Left: Polygon scanner based filter, from [49]. Right: Electro-optic deflector tunable filter, from [51].

Due to the large form factor, potential stability issues and limited speed performance of scanned grating filters, Fabry-Perot tunable filters (FP-TF) with a very wide FSR have been used. They are more compact and add an optical leverage to the mechanical motion. In these devices, one of the two end facets can be moved very precisely and rapidly – either by electrostatic forces or by a piezo-electric actuator. The filter finesse is given by the reflectivity of the end reflectors. Very high finesse values larger than 1000 can be obtained, providing large free spectral range and narrow linewidth simultaneously. Moreover, very high tuning speeds of several hundred kHz are possible. These devices are available in a fiber-coupled package, which has the benefit of low insertion loss, see Fig. 3. Another implementation uses micro-electro-mechanical systems (MEMS) technology for miniaturized FP-TFs. A special type of FP-TF uses Vernier tuning, in which two comb-like spectra with slight detuning to each other are used to select a wavelength [53].
3. Cavity designs and laser operation modes

In this section, we will present the most common swept source designs. We start with directly swept lasers, in which the instantaneous wavelength is set by an intracavity tunable filter. This type is by far the most popular swept laser type, and includes wavelength-swept ASE sources, short cavity lasers, Fourier-domain mode-locked (FDML) lasers, and swept vertical-cavity surface-emitting lasers (VCSELs). Recently, indirectly-swept lasers have begun to attract some interest, since they are not limited by the maximum tuning speed of the filter. Instead, methods like chromatic-dispersion tuning and time-stretching of ultrashort laser pulses are used for wavelength-tuning. While these laser types may achieve very high speeds, the complete laser design is usually more complex, and sweep range, center wavelength or output power are often limited or the effective linewidth is too broad, sometimes just because the sweep is too fast to be sampled with sufficient spectral resolution.

Wavelength-swept ASE sources

Conceptually, the simplest swept source design consists of only two elements: A broadband light source and a tunable filter. If the broadband source spectrum is filtered once, the instantaneous linewidth will be very close to the filter bandwidth. Hence a narrow-band filter or multiple filtering events are necessary for long coherence lengths. Moreover, since most of the source spectrum is filtered out, the optical output power will be very low. Post-amplification is possible, but the amplifier will be operated far from saturation because of the low input power. This will lead to a large ASE background, i.e. excess noise. While the simple combination of a gain and filter element is not attractive as an OCT light source, one can take advantage of the fact that the noise background after post-amplification has a broad spectral width. Hence small instantaneous linewidth and high powers can be realized by a simple cascade of gain and filter elements. As has been shown by Eigenwillig et. al., a sequence of two gain and filter events can already lead to sufficient performance for OCT imaging [56].

This kind of laser source has several advantages compared to all other swept lasers, which rely on optical feedback. They have no fundamental speed limit, can be operated at any sweep rate with arbitrary filter drive waveforms, and have a stable wavelength evolution that is governed by the tunable bandpass filters. However, a challenge in the operation of these light sources is the very precise synchronization between the successive filter elements. This requires a phase-locked drive mechanism, where not only phase, but also amplitude is continuously adjusted to compensate inter-filter drift. Moreover, to achieve coherence lengths in excess of a few millimeters, more than two filter events would typically be necessary, since too narrow filter bandwidths result in prohibitively low powers. Another disadvantage is that
the RIN is proportional to 1/linewidth, so the longer the coherence length, the noisier the device \[55, 57\]. Wavelength swept ASE sources have been demonstrated with sweep rates of 340 kHz at center wavelengths of around 1300 nm \[56\] and 1060 nm \[58\]. Since the SOA is used like a superluminescence diode (SLED), they have also been called tunable SLED (TSLED) or swept SLED \[55\].

**Short cavity**

Currently, short cavity lasers are amongst the most popular swept laser schemes. In principle, they have a straightforward design, which adds a feedback element to a wavelength swept ASE source, so that light is filtered and amplified with every laser roundtrip. In other words, filter and gain medium are arranged in a simple linear resonator or ring resonator geometry. They are also called external or extended cavity swept lasers, since the filter element is separated from the gain element. In general, these lasers have a fundamental speed limit, since, in a simple model, lasing has to build up for every wavelength separately \[59\]. The reason for this behavior is the fixed cavity length, which results in stationary cavity modes. When the filter sweeps over those modes, mode hops between the previously active modes and new modes occur. It takes a couple of resonator roundtrips for stationary lasing to build up again for the new mode. Thus, coherence length decreases with increasing sweep rate in these lasers. Shorter cavities have a higher speed limit. However, the shorter the cavity, the larger the mode spacing. For very short cavities, the mode spacing becomes so large that mode hops lead to strongly increased laser noise, unsuitable for OCT imaging. So, in practice, this type of laser is limited to sweep speeds of a few hundred kHz, with decreasing performance for higher speeds. The first broadly employed short cavity lasers were based on polygon-grating type filters and filters using resonant galvanometer scanners. They constituted the first generation of widely employed swept lasers in OCT, at 1300 nm \[49\], 1060 nm \[60\] and 850 nm. Polygon scanner based sources were the first commercially available swept sources \[61\], and the fastest speed reported so far is 400 kHz \[62\], using optical buffering (see below). Short cavity swept lasers can also be built with FFP-TFs \[54, 59\]. A recent demonstration showed 240 kHz sweep rate with high output power and wide sweep range \[63\].

Recent research suggests that in some conditions, complex laser dynamics induced by the nonlinearity in the gain medium can occur. It has already been known for a while that forward and backward sweeps in short cavity lasers often show very different properties. Usually the forward sweep is favored, due to four-wave mixing in the SOA \[28\] and the interplay between line width broadening in the SOA and gain saturation. Interestingly, the result is that some short cavity lasers emit a periodic pulse train \[40\]. This also explains a major drawback of some of the most popular short cavity lasers: Interference is generated for structures located at integer multiples of the cavity length, which is called coherence revival. For
instance, the swept sources from Axsun (USA) generate ghost images from surfaces such as focusing lenses although they are positioned outside the OCT imaging range [65].

**Short cavity with adiabatic frequency shift**

The problem of coherence revival, linewidth broadening and limited build up time of lasing can be overcome by adiabatic tuning of the laser cavity. No laser mode hops occur, if the laser cavity mode is changed synchronously to the passband of the tunable filter in the cavity and if there is a physical mechanism which shifts the photon energy inside the laser cavity accordingly. In this case, the laser resonator is tuned adiabatically [66]. To this end, energy must be provided or removed from the light field, or more accurately from the photons in the cavity to create the desired wavelength shift. It should be underlined that changing the cavity length without shifting photon energy is not sufficient [18, 67, 68]. There are several ways to achieve adiabatic tuning: In one implementation, an acousto-optic modulator was used to actively shift the optical frequency [69]. Also, the total cavity length can be changed such that a Doppler shift is generated by a moving mirror or grating. In both cases, it is difficult to achieve high-speed sweeps over a large wavelength range.

The optical path length of the cavity can also be changed by changing the effective refractive index of the cavity. Jirauschek et al. showed that this technique automatically generated the desired wavelength-shift of the intra-cavity photons [66]. Changing the refractive index can be achieved via carrier-injection in a semiconductor material. In conjunction with a Vernier tuning mechanism, a single semiconductor chip can accommodate gain medium, the phase section for length tuning, and the Vernier filter, as shown in Fig. 4. In this Vernier-tuned distributed Bragg reflector laser (VT-DBR), adiabatic tuning over a small wavelength range is possible, before the next Vernier state must be addressed. The laser output is not an ideal continuous sweep, but a sequence of ~0.5 nm short micro-sweeps separated by transition periods. Within a micro-sweep, these lasers are mode-hop free and do not exhibit coherence revival artifacts. This is the reason why recently very good coherence properties of such sources have been reported [70]. However, the transition periods between the micro-sweeps must be removed from the sweep waveform for OCT imaging, which complicates data acquisition. Initial implementations of this technique showed low RIN, long coherence length and high phase stability, but limited tuning range [64]. An advantage is that the wavelength-vs-time output can be programmed, since the filter is not operating in resonance. Recently, high-quality OCT imaging with a VT-DBR tunable laser has been presented at 100 kHz sweep rate [70].

Despite their good coherence performance, these lasers have a fundamental sweep speed limit. After every switching to the next micro-sweep, lasing has to build up from spontaneous emission and underlies the time-bandwidth limit. To achieve a better 1 GHz linewidth, at least one nanosecond buildup-time is required after every micro step. For example Bonesi et al. reported a transition time on the order of 2.5 ns [64], so it seems unlikely that these sources are able to generate (multi-)MHz sweep rates. Song et al. reported that at 400 MHz sampling rate, the number of total sweep points is 4160, including 2878 valid points. This means that invalid states occupy 31% of the total sweep time, already at the relatively moderate speed of 100 kHz. Since the number of transitions will remain constant for constant sweep range, the 1282 transitions needed for 100 nm sweep range have a fixed duration of 3.2 µs. With the current 400 MHz clock, a 312.5 kHz VT-DBR (3.2 µs sweep duration) swept over 100 nm has 100% duty cycle of transitions, with no usable output in between. Hence, the maximum speed for these sources with reasonable tuning range for OCT might be as fast as the fastest spectrometer systems [71] today.
VCSELs

MEMS tunable vertical-cavity surface-emitting lasers (VCSELs) are essentially a Fabry-Perot filter with integrated gain medium [73], see Fig. 5. While their non-tunable counterparts are already widely used for telecom applications in data centers, considering MEMS tunable devices there is still extensive research going on since the mid 1990ties. MEMS tunable VCSELs appear very promising for OCT since they have an extremely short laser cavity, so they can be operated with one single longitudinal mode. When the emission wavelength is tuned by changing the resonator length, a Doppler shift occurs at each bounce of the moving MEMS mirror, so an adiabatic frequency shift is imposed on the photons inside the cavity. It can be shown that the Doppler shift generated by moving the end mirror of a laser, always matches the new position of the laser cavity resonances [74]. Hence, the cavity photons automatically track the resonator mode, and there is no fundamental sweep speed limit. Moreover, since there is only one mode and the number of effective roundtrips is very high, very long coherence lengths can be achieved. Typically, VCSELs used for OCT have one fixed mirror at one end of the gain medium. The other end of the cavity needs to be moved at high speeds, which necessitates low weight of the moving mass. Typically, a MEMS-based driving mechanism is used with a dielectric or micro-structured mirror. The gain medium can be either electrically pumped, like an SOA, or optically pumped, which necessitates an extra pump laser. A key advantage of VCSELs is their very narrow instantaneous linewidth, even at very high sweep speeds. Many centimeters and even meters of coherence length have been demonstrated [75, 76]. At 1060 nm center wavelength, suitable for retinal imaging, up to 580 kHz sweep rate were shown with an optically pumped VCSEL [77]. Speeds of up to 1 MHz have been demonstrated with electrically pumped VCSELs at around 1310 nm center wavelength [78]. VCSELs also have the advantage of supporting a large range of drive frequencies, up to their mechanical resonance. In principle, VCSELs can be operated at variable sweep rate, but this is complicated by the nonlinear response to the driver signal. The broad drive frequency range also opens the way for linearization of the sweep wavenumber versus time tuning characteristic.

However, VCSELs also have several specific disadvantages [79]. While VCSELs have no fundamental sweep speed limit, maximum speed is limited by the mechanical properties of the MEMS actuated filter. To achieve high speeds, these structures are very delicate, and exhibit a variety of motion patterns. While VCSELs are single longitudinal mode devices for a single transverse mode, additional higher order transverse modes can create side lobes and longitudinal multi-moding resulting in OCT imaging artifacts. Another drawback is that many current VCSEL implementation rely on optical pumping of the gain medium, which increases system complexity and size. From a laser physics standpoint, optical pumping has several advantages over electrical pumping. For instance, the lack of dopant reduces cavity loss and promotes wide tuning, while higher transverse modes are suppressed by the pump beam profile. In future, these issues might be addressed by engineering efforts. However, optical power is a critical factor in VCSEL design. Since the finesse of the resonator is relatively high, optical power in the resonator is orders of magnitude larger than the output power. This is one of the reasons while VCSELs have only been demonstrated with a few mW of output...
power. Hence booster amplifiers have to be used to increase the output power to levels suitable for MHz imaging. Polarization control is needed for the typically used polarization dependent SOAs. This is complicated by the fact that the polarization of the VCSEL output may change over time, since the commonly used laser structures have no preferred axis. In total, it is currently unclear if or when VCSELs with multi-MHz A-scan rate suitable for high quality OCT imaging will be available, but right now they already prove good performance at slower speeds.

**FDML**

While swept VCSELs overcome the speed limit of short cavity lasers with an extremely short cavity combined with inherent Doppler shift [66], Fourier-domain mode-locking is a novel laser operation regime that employs a very long cavity, on the order of hundreds of meters to kilometers [80]. More precisely, the cavity length is matched to the filter drive period, such that the resonator roundtrip frequency matches the filter frequency exactly. Hence, the entire optical sweep is stored inside the cavity, and ideally, the filter dissipates no optical power. Lasing does not need to build up for every wavelength separately, and there is no fundamental sweep speed limit for FDML lasers. A similar concept was already demonstrated in 1975 [81], but had limited impact. Moreover, it was not realized that the filter drive signal needs extremely accurate synchronization to the cavity length, to a precision of less than $10^{-6}$. Usually FDML lasers are built with SOAs and FFP-TF, but due to their modular architecture, other combinations are possible. FDML lasers have been demonstrated using a polygon filter [82], and a variety of gain media, including Raman gain [47], rare earth doped fiber [83], and tapered optical amplifiers [84]. FDML are all-fiber based lasers, built of standard telecom grade components, which provides long lifetime and high reliability. Low insertion loss leads to good saturation of the gain medium, and high output powers of more than hundred milliwatts are possible. Moreover, very wide tuning ranges of up to 160 nm have been demonstrated [85]. FDML is a truly stationary laser operation regime [35], producing highly repeatable sweeps, which reduces RIN for high-quality OCT and eliminates the need for hardware clocking [23, 86].

The detailed dynamics of FDML lasers are very complex, and subject of ongoing theoretical investigations [26, 35, 66, 87–91]. Thus, while their fundamental design is simple, FDML lasers can be more difficult to build, operate and integrate in OCT systems. First, the small amount of birefringence in the fiber creates polarization mode dispersion (PMD), which requires polarization control [24]. Fortunately, a so-called sigma-ring laser configuration eliminates most of the PMD. Second, FDML lasers are operated at the resonance frequency, so all other system components such as scanners need to be synchronized to the laser. For the common FFP-TF based FDML laser, center wavelength and tuning range need to be actively stabilized, which has been demonstrated with a few pm stability [92]. Nevertheless, with FFP-TFs the wavelength vs. time evolution can be linearized, to enable SS-OCT without resampling or special clocking schemes [93]. Third, management of chromatic dispersion in the cavity is necessary achieve very long coherence lengths [94]. FDML lasers have been demonstrated at all popular OCT wavelengths including 1550 nm [95] and they have proven good performance in many OCT and other imaging and sensing applications [96–98]. FDML lasers have repeatedly set new sweep speed records and demonstrated high-quality OCT imaging, spanning from 370 kHz in 2006 [99], to 5.2 MHz at 1310 nm in 2010 [5] and 3.3 MHz at 1060 nm in 2013 [9].

**Stretched pulse**

All previously described sources are directly swept sources, that is the instantaneous emission wavelength is governed by the intracavity tunable filter trajectory. Since a sweep is the equivalent of an extremely chirped pulse [97], they can also be generated by temporal stretching of ultrafast laser pulses, as demonstrated for the first time by Moon et. al. in 2006.
Ideally, this approach has a couple of distinct advantages: First, the sweep rate is limited only by the pulsed laser repetition rate, which can be extremely high. For instance, a titanium-sapphire laser typically runs at 80 MHz – 100 MHz repetition rate. Second, long coherence lengths are possible for a well-defined pulse. Moreover, pulsed laser can have very broad spectra of several hundred nanometers, potentially providing extremely high axial resolution in OCT. Finally, the time-stretch mechanism is passive, so that the sweep-to-sweep variability could potentially be very low. However, the initial implementations of stretched-pulse lasers showed only limited performance in OCT application, mainly due to very low sensitivity [100,101]. There are two main reasons why it is difficult to implement a high-quality sweep by stretching short pulses: First, the strict requirements for sweep-to-sweep stability and RIN can only be met with specially designed lasers [102, 103]. Second, currently the only practically feasible way to generate sufficient dispersion is to use highly dispersive optical fiber, with commercial availability only around the telecom wavelength range of 1550 nm. Hence almost all demonstrations of stretched pulse lasers have been around this wavelength range, although operation around 1000 nm has been demonstrated recently [104]. Finally, extreme amounts of dispersion are needed to stretch the pulse, which inevitably leads to high loss. Pulsed laser power must be limited to avoid nonlinear effects in the long fiber. Due to high losses, the stretched output needs to be amplified, which increases noise considerably due to the low input power [105]. Raman co-amplification can be used, at the price of increased system complexity [106]. Ironically, the disadvantages of this technology are more severe for slower sweep rates, since more stretching leads to longer fiber and more loss. Nevertheless, Xu et. al. recently demonstrated in-vivo OCT imaging at record sweep rates of up to 11.5 MHz, demonstrating that stretched pulse swept lasers might be an alternative for multi-MHz OCT imaging [7].

**Dispersion tuned swept lasers**

Dispersion tuning (DT) is another sweep mechanism that works without a bulk intracavity wavelength-selective filter [107]. Contrary to most other operation modes like FDML, chromatic dispersion is not detrimental but essential for the laser working principle. In DT, the cavity gain (or loss) is actively modulated in synchronization with the cavity round trip frequency. Due to chromatic dispersion, only a small wavelength range is active for a given modulation frequency. Both short cavity length and high chromatic dispersion improve the laser performance, which can be achieved with chirped fiber Bragg gratings [108].

A variant of DT is to employ two dispersive elements with dispersion of equal magnitude, but opposite sign, which work as pulse stretchers and compressors, respectively [109, 110]. Modulation takes place on the compressed pulse, whereas the output coupler is placed after the stretching element see Fig. 6.

![Fig. 6. Left: Swept laser based on dispersion tuning, from [109]. Right: Stretched-pulse swept laser OCT system, with Raman co-amplified fiber stretch line and booster amplifier, from [7].](image-url)
Buffering

Even in swept lasers without fundamental speed barrier, practical factors such as mechanical properties of the filter of small duty cycle can define a maximum sweep speed. An optical multiplexing technique known as time-interleaving or buffering provides a convenient way to further increase speed, first demonstrated with an FDML laser [99]. In buffering, only a fraction of the total sweep time is selected for OCT imaging, while the laser output is turned off during the rest of the time. Buffering then creates copies of this sweep via simple (fiber-optic) beam splitters, delays the copied sweeps, and then fills up the duty cycle to 100%. This is an especially useful technique in two cases: (A) Wavelength swept lasers with 50% or less duty cycle, such as the Axsun swept laser [111]. (B) for swept lasers with filters operating in resonance, producing a sinusoidal wavelength-over-time evolution, which yields reduced imaging range compared to a linear sweep, see Eq. (9). With buffering, only the most linear part of the sweep near the sweep center can be used, optimizing the OCT imaging range. Moreover, as discussed previously, forward and backward sweeps may have different coherence and noise properties, then buffering permits selection of the sweep with the best properties for OCT imaging. Since only a part of the sweep is used the filter amplitude needs to be increased to maintain the same sweep range as for the full duty-cycle sweep. Note that for a mechanical filter, increasing the amplitude induces less material stress on the filter than increasing the frequency, which would be the more obvious way to increase speed. While acceleration increases with frequency squared, it only increases linearly with amplitude. In FFP-TF based FDML lasers, up to 16x buffering has been demonstrated [5], proving the very good amplitude performance of PZT driven FP-filters. MEMS actuators seem to be more critical with respect to amplitude. In VCSELs, the maximum amplitude is additionally restricted since the gain medium is located inside the Fabry-Perot resonator structure. To generate sufficient free spectral range, the remaining gap must be very small, limiting the maximum amplitude of the wavelength tuning. The high amplitude supported by FFP-TFs seems to be one of the primary reasons why highest sweep speeds have been demonstrated with FDML lasers. Buffering has also the advantage, that successive wavelength sweeps which have been generated by copying one primary sweep have a very stable optical phase, since they originate from passive splitting. This can be a great advantage for high speed phase sensitive measurements [112, 113].

There are two main drawbacks of buffering. First, polarization effects in the long delay fiber require polarization management when a polarization dependent booster amplifier is used or for polarization-sensitive OCT [24]. Second, with beam splitters, half the power is lost upon recombination of the delayed copies. More precisely, the power is split between two outputs, which can be exploited if more than one imaging beam is used (see below). For single-beam imaging, power can be conserved by using polarization techniques, which might also be beneficial for PS-OCT [114]. At least for slower sweep speeds, direct optical switching was demonstrated to increase power efficiency of buffering [115]. However, the intrinsic insertion loss of the switches and polarization effects yield only a marginal increase in power, while inducing several other problems. Another potential drawback for commercial applications is the cost of the long delay fiber in the buffer stage. Standard single-mode fiber is extremely cheap, on the order of one cent per meter. Hence couple of hundred meter fiber length constitutes a negligible fraction of the total laser cost. At shorter wavelengths below the single-mode cut-off wavelength, it was initially believed that specialized and expensive fiber has to be used, in order to maintain single-mode operation. Fortunately, it was shown that standard and cheap single-mode fiber can still be used at these short wavelengths [32].

Additional filter elements

Whereas buffering can be used to increase the sweep rate, etalons can be used to increase the coherence length of swept lasers [116]. The resulting sweep consists of discrete wavelengths, like a frequency comb. The additional spectral filtering by the high-finesse etalon narrows the
instantaneous linewidth, but limits the imaging range due to the finite amount of sampling points in the sweep. Sample structures outside the main imaging range are aliased back, which in general is not favorable for OCT imaging. These intra cavity FPs or etalons convert the continuous wavelength sweep into a stepwise tuning.

Overview of swept lasers

In a nutshell, there are currently many interesting ideas and concepts for swept light sources intended for application in OCT (Table 3). In some cases, it is still unclear whether they can consistently achieve sufficient performance for high-quality OCT, especially at multi-MHz speeds.

Table 3. Swept lasers (research) with demonstrated high speed and high quality OCT imaging.

| Ref. | \(f_{\text{sweep}}\) | Year | Laser type | Filter type | \(\lambda_c\) | \(\Delta \lambda\) | 6 dB roll-off | Output power |
|------|-----------------|------|------------|-------------|----------------|----------------|----------------|--------------|
| [49] | 15.7            | 2003 | Short cavity | Polygon     | 1320           | 73             | 4'             | 9            |
| [54] | 0.25            | 2005 | Short cavity | FFP-TF      | 1300           | 130            | >4             | 3            |
| [61] | 20              | 2005 | Short cavity | Polygon     | 1310           | 110            | ~1             | ~5           |
| [59] | 20              | 2005 | Short cavity | FFP-TF      | 1310           | 120            | ~3             | 46           |
| [117] | 115             | 2005 | Short cavity | Polygon     | 1325           | 80             | ~2'            | 23           |
| [80] | 232             | 2006 | FDML        | FFP-TF      | 1300           | 105            | ~6             | 20           |
| [99] | 370             | 2006 | FDML        | FFP-TF      | 1310           | ~100           | ~2             | 36           |
| [107] | 200             | 2006 | Dispersion tuned | -         | 1300           | 100            | NA             | NA           |
| [85] | 100             | 2007 | FDML        | FFP-TF      | 1310           | 160            | 2              | 35           |
| [56] | 340             | 2009 | ASE sweeper | FFP-TF      | 1310           | 100            | ~2             | 50           |
| [95] | 110             | 2009 | FDML        | FFP-TF      | 1550           | 115            | NA             | 14.8         |
| [60] | 18.8            | 2009 | Short cavity | Polygon     | 1060           | 62             | ~2.5           | 2.7          |
| [68] | 2.5             | 2009 | Short cavity | Polygon     | 1060           | 18             | 14             | 3            |
| [5]  | 5200            | 2010 | FDML        | FFP-TF      | 1310           | 80             | 0.6            | 85           |
| [42] | 403             | 2010 | Short cavity | Polygon     | 1310           | 104            | 1.7 (4dB)      | 32           |
| [111] | 200             | 2010 | Short cavity | MEMS        | 1050           | 100            | 4.2            | ~18          |
| [32] | 1370            | 2011 | FDML        | FFP-TF      | 1060           | 43             | 1              | 42           |
| [58] | 340             | 2011 | ASE sweeper | FFP-TF      | 1060           | 70             | ~2             | 70           |
| [118] | 1600            | 2012 | FDML        | FFP-TF      | 1310           | 100            | 7.2            | 80           |
| [51] | 200             | 2012 | Short cavity | EO           | 1310           | 80             | 7              | 20           |
| [57] | 580             | 2012 | VCSEL       | FP-TF       | 1060           | 83             | >2.2           | >20          |
| [78] | 1000            | 2013 | VCSEL       | FP-TF       | 1310           | 107            | 24             | 40           |
| [9]  | 3350            | 2013 | FDML        | FFP-TF      | 1060           | 72             | 1.6            | NA           |
| [108] | 100             | 2013 | Dispersion tuned | -         | 1550           | 60             | ~0.8           | 8.4          |
| [64] | 102             | 2014 | Short cavity | VT-DBR      | 1310           | 30             | NA             | 1.5          |
| [64] | 200             | 2014 | Short cavity | VT-DBR      | 1550           | 40             | ~20            | 5.5          |
| [109] | 9062            | 2014 | Dispersion tuned | -         | 1550           | 84'            | ~2'            | 18'          |
| [102] | 11500           | 2014 | Stretched pulse | -         | 1557           | 58             | 26             | 3.5          |
| [7]  | 11500           | 2015 | Stretched pulse | -         | 1558           | 56             | 24             | 2            |
| [70] | 100             | 2016 | Short cavity | VT-DBR      | 1300           | 100            | 50             | >6           |
| [63] | 240             | 2016 | Short cavity | FFP-TF      | 1310           | 125            | 1.9            | 70           |

0.1 nm and 0.16 nm linewidth; ' for swept configuration, broader spectrum, better coherence and lower power for stepped configuration

Commercially available swept lasers

Swept lasers are currently available from a couple of manufacturers worldwide. Axsun (USA), Exalos (Switzerland), NTT (Japan), Micron Optics (USA), Superlum (Ireland) and Santec (Japan) provide short cavity lasers based on different filter designs with up to 200 kHz
sweep rate. A VT-DBR swept laser is available from Insight Photonic Solutions (USA), and
Thorlabs (USA) and Santec offer swept VCSELs at up to 200 kHz sweep rate. FDML lasers
from Optores (Germany) are currently the only commercially available lasers with MHz
sweep rate. All these lasers use a semiconductor gain medium, and all but two (NTT, Insight)
products rely on a mechanically tunable filter element., which could rise the question about
reliability. However, the good long-term stability despite mechanical motion is due to the
fact, that the filters and the oscillation amplitudes are very small, on the order of 1 µm. Hence
the devices are compact and rigid and their mechanical eigenfrequencies are far above
environmental acoustic waves and vibrations. So the devices may be compared to quartz
oscillators, as extensively used in electronic circuitry or to MEMS mirrors in DPL projectors
with proven long term stability. Table 4 gives an overview over the most popular swept lasers
for OCT, based on company websites as accessed on November 16, 2016.

| Company          | f_sweep | Laser type | Filter type | λc | Δλ | 6 dB roll-off | Output power |
|------------------|---------|------------|-------------|----|----|---------------|--------------|
| Axsun1           | 200     | Short cavity | MEMS FP-TF | 1060 | 1310 | 100          | 15           |
|                  | 100     |             |             | 100  | 140 | 5             | 15           |
| Exalos           | 20      | Short cavity | MEMS        | 850  | 1050 | 60           | 3.5          |
|                  | 50      |             |             | 90   | 150 | 10           | 7            |
|                  | 20      |             |             | 150  |     | 10           | 15           |
| NTT              | 200     | Short cavity | Electro-optic | 1310  | 100 | 3.5          | 17           |
| Micron Optics    | 100     | Short cavity | FFP-TF      | 1310  | 150 | 1.5          | >35          |
| Santec1          | 20      | Short cavity | Polygon     | 1060  | 1310 | 120          | 1.5          |
| Santec           | 100     | Short cavity | MEMS FP-TF | 1060  | 1310 | >100         | 15           |
| Insight photonics2 | 100   | VT-DBR Vernier | FFP-TF      | 1310  | 1550 | >50          | >6           |
| Thorlabs         | 100, 200| VCSEL MEMS  | FFP-TF      | 1060  | 1330 | 100          | 5.5          |
| Optores          | 750, 1500| FDML MEMS  | FFP-TF      | 1060  | 1330 | >100         | >100         |

1Wide range and / or high speed variants; 2No data sheet available on website, data taken from [70] for 1310 nm,
and [64] for 1550nm. Note: 6 dB roll-off was calculated as half the coherence length.

4. High-speed OCT systems

Basics

We briefly review some key performance parameters of OCT systems that will be useful in
the next sections. OCT sensitivity $S$ is defined as the weakest sample reflectivity that can still
be measured. Even in ideal OCT systems, sensitivity is limited by shot noise. This shot noise
limited sensitivity $S$ can be calculated by [22]:

$$ S = 10 \log \left( \frac{P T}{e} \right) - IL. $$

where $\rho$ is typical detector responsivity (e.g. 0.7A/W at 1060 nm), $P$ is the optical power
incident on the sample, $T$ is the sweep duration, $e$ is the elementary charge of the electron,
and IL is loss on the way back from the sample to the detector. The formula assumes that
apart from IL, all power returning from the sample is incident on the photodetector. Thus,
increased speed goes in hand with decreased sensitivity. This may limit imaging speed for
some medical and biological applications which have a limited permitted sample exposure,
see e.g. the discussions in [9, 119]. However, it can be shown that even for sensitive
applications, MHz and even multi-MHz OCT imaging can be performed with reasonable sensitivity, if well designed OCT systems are used. Additionally, current research on MHz-OCT did not take into account that the laser beam is scanned on the sample, which might relax safety considerations regarding light exposure. So, it can be expected that for all of today’s OCT applications, sufficient sensitivity can be achieved well up to MHz A-scan rates. An important feature of OCT is that transverse resolution is decoupled from axial resolution. The latter depends on the optical spectrum of the OCT signal only. In the standard formula, axial resolution $\Delta z$ for a source with Gaussian spectrum is given by

$$\Delta z = \frac{2 \ln(2)}{\pi} \frac{\lambda^2}{\Delta \lambda_{\text{FWHM}}},$$

(5)

where $\lambda_c$ is the center wavelength, and $\Delta \lambda_{\text{FWHM}}$ is the full-width at half maximum of the spectrum. Note that the same relation holds for coherence length $\Delta z$ and instantaneous linewidth $\Delta \lambda_{\text{FWHM}}$. We want to stress that this formula is only valid for a Gaussian spectrum or linewidth, which has, considering its exact definition, infinite spectral bandwidth [120]. It is of course not possible for a physical Fourier-domain OCT system to detect the entire bandwidth, so the actual axial resolution must always be somewhat lower than the theoretical value in the equation above. Nevertheless, in can be a good estimate for spectral domain systems, because the spectral output from the often used superluminescent diode has a shape similar to a Gauss distribution.

However, swept sources usually do not exhibit a Gauss-type spectrum, unless they are actively shaped [32, 64, 86]. Depending on the degree of saturation, the output is often more triangular, with the peak emission shifted away from the sweep center towards longer wavelengths. Thus, replacing the FWHM $\Delta \lambda_{\text{FWHM}}$ with the full sweep range $\Delta \lambda$ of the laser in SS-OCT is not correct. Even in cases with rectangular spectrum [64], where the FWHM of the PSF corresponds well with the value calculated by using the 3dB bandwidth in Eq. (5), numerical apodizing is used in post processing to suppress side lobes, which then again negatively effects axial resolution. In general, the precise axial resolution should always be measured, or should be calculated from the actual interference spectrum. In SS-OCT, it can be shown that for a large variety of spectral shapes, it is usually close to the following approximation:

$$\Delta z = \frac{\lambda^2}{\Delta \lambda},$$

(6)

where $\Delta \lambda$ is the sweep range of the swept laser as defined in Eq. (1). Another very important parameter for high-speed OCT is the maximum ranging depth $z_{\text{max}}$. Even though the OCT penetration into scattering tissue is ~2mm only, for clinical applications and profilometry it can be very important, to have some overhead range either to simplify patient alignment or to measure very curved surfaces.

In SS-OCT, $z_{\text{max}}$ is directly proportional to the sampling rate of the analog-to-digital converter (AD-card, digitizer). It can be calculated as:

$$z_{\text{max}} = \alpha \frac{\lambda^2}{\Delta \lambda} \frac{f_s}{4f_{\text{sweep}}},$$

(7)

with previously defined center wavelength $\lambda_c$, sweep range $\Delta \lambda$, sweep rate $f_{\text{sweep}}$ and sampling rate $f_s$. The prefactor $\alpha$ depends on the linearity of the swept laser. For a perfectly linear laser it is equal to unity. For a single sided sinusoidal sweep, as produced from filter operated in mechanical resonance, $\alpha = 2 / \pi = 0.64$. Another important parameter is dynamic range, defined as the ratio in signal strength between strongest and weakest reflection which can be measured simultaneously within one A-scan. The desired dynamic range depends on the
application. For instance, biomedical OCT images often span a reflectivity range of 25 dB to 35 dB [121]. To provide some margin for roll-off and specular reflections, an OCT system should typically provide 40 dB to 50 dB dynamic range [95].

Finally, there is another useful metric for imaging speed going beyond line rates. A-scan rate alone is not always suitable to characterize the imaging speed in the sense of information flow, since it does not consider the actual number of resolvable volume elements (three-dimensional pixels = voxels) that were recorded. For example, per Eq. (7), increased sweep rate will lead to shallower imaging range, if the detection bandwidth is kept constant. Hence the total amount of image voxels stays constant. The actual speed can be defined by the number of acquired voxels per second [122]. Care has to be taken when defining the amount of voxels, since roll-off performance limits the number of voxels that actually carry information [5].

**Scanners for high-speed OCT**

Imaging at several MHz line rate quickly leads to several kHz B-scan rate. With a single-beam flying-spot configuration, the beam scanner needs to be able to provide a scan speed as fast as the B-scan rate for unidirectional scanning, or half the frequency for bidirectional scanning. Standard galvanometer scanners have limited speed, especially for large scan angles [32]. MEMS scanners are attractive due to their small size, and since two-axis scanning is possible with a single scanner. This makes them especially attractive for handheld devices [123, 124]. However, for all scanners, standard galvanometer type or MEMS actuated ones, high speeds of a few kHz go hand in hand with small mirror sizes and / or limited scan angle. A scanning galvanometer scanners can provide very high scan speeds with large apertures and scan angles, so they are used for many multi-MHz applications [125]. Their main drawbacks are the fixed frequency, and limited phase stability, which usually requires active control of the driving waveform or data acquisition. While at least resonant scanners are readily available for frame rates of several kHz, it is challenging to implement catheter-based scanners that provide speeds suitable for (multi-) MHz OCT. If only two-dimensional imaging is required, scanning can be performed by simply pulling the endoscope [126]. For three-dimensional scanning, the fiber inside the catheter can be turned directly to provide a circumferential scanning mechanism. Higher speeds of several hundred Hz can be obtained with MEMS scanners [127, 128]. To date, highest speeds have been obtained with micromotor-based scanners. A side-viewing catheter with an ultrafast micromotor scanner was used for intravascular OCT imaging at 3 MHz, using an FDML laser [129]. Recently, a forward-viewing catheter with small outer diameter was demonstrated at 2 kHz frame rate [130].

**Data acquisition and processing**

The requirements for (multi-) MHz OCT currently also push the limits of the detection system. In swept-source OCT, the interferometric signal is digitized with a fixed digital resolution. It has been demonstrated that 8 bit A/D resolution can be sufficient for high-quality OCT imaging [5]. However, the effective number of bits (ENOB) of an 8 bit A/D is always lower than 8 bit, and usually decreases for faster sampling rates. Hence, it is more convenient and usually leads to better results to use somewhat better resolutions on the order of 10 to 12 bits [5, 131]. In any case, the detection for swept source OCT should use spectrally balanced detection, which can be achieved with readily available dual-balanced photoreceivers [132]. Alternatively, the balancing scheme can also be carried out in post-processing, which opens the way for better RIN suppression, but comes at the cost of another A/D channel, doubling the data rate [133]. For instance, a 3 MHz 1310 nm swept laser with 100 nm sweep range requires a 4 GS/s sampling rate for an imaging range of 5.7 mm. This generates a data stream of 6GB/s for 12 bit A/D rate. It is currently not possible to process these huge amounts of data in real time with the main processor(s) in a standard desktop...
computer. Fortunately, each A-scan can be processed independently, opening the possibility of fully exploiting the parallel processing power of graphics processing units (GPUs). Since MHz OCT systems were not commercially available until recently, the data processing rate was much higher than the data acquisition rate, and processing was applied to prerecorded data sets or at relatively slow speeds [134–136]. Apart from data processing, visualization of the results requires a lot of computational power [137, 138], especially when combined with additional processing, such as for Doppler-OCT [139, 140]. Additionally, the data transfer rate of current PCI express digitizers is pushed to the limit. A combination of data acquisition, transfer, processing and visualization at MHz speeds has been demonstrated recently with GPU computing [125]. Alternatively, field-programmable gate arrays (FPGAs) also offer high computational power [6, 141, 142]. If only certain subsets of the entire volume are of interest, approaches like master-slave interferometry may be used [143] or direct analog hardware processing [86].

**Multiplexing**

Various multiplexing techniques have been employed to increase speed in OCT. In the most straightforward implementation of multiplexing, multiple beams are used to scan over the sample in parallel, which directly increases imaging speed by the number of beams used. Imaging with six beam was used to generate 258 kHz A-scan rate [144]. With four beams in parallel and a laser sweep rate of 5 MHz, 20 MHz imaging rate has been demonstrated by Wieser et al. in 2010 at 1310 nm [5]. At 1060 nm, multi-beam imaging was first demonstrated by Potsaid et al. to yield 400 kHz speed [111], and the fastest speed of 6.7 MHz was obtained with an FDML laser [9]. More beams have been demonstrated. For instance, an 8x increase in imaging speed using a 1x8 fiber array has been demonstrated with effective speed of 800 kHz [145]. In swept-source OCT, each beam needs another reference arm, interferometer, receiver and A/D channel. Since the main cost factor in SS-OCT is the laser, multiplexing schemes can be cost-effective. Moreover, they do not only increase A-scan rate, but also voxel rate. However, each additional channel makes the system increasingly more difficult to produce, align and operate.

Other multiplexing schemes can be implemented with air-spaced virtually-imaged phased arrays (VIPA), which laterally disperses different wavelengths sets from one sweep on the sample [146]. Each A-scan consists of a subset of discrete wavelengths that cover the entire sweep range, similar to frequency comb swept lasers that employ an etalon to increase coherence length [116]. Each comb spectrum is reconstructed by suitable decimation of the single acquired data trace. Hence, imaging range is traded in for imaging speed. So, this scheme increases A-scan rate at constant voxel rate. 3.2 MHz line rate were achieved with a 200 kHz swept laser, using 16 lateral points [147]. Unfortunately, sensitivity is affected by loss in the VIPA. With long-coherence length swept lasers, a different multi-beam approach with constant voxel rate can be used. Here, images from each spot are in different depths within the total imaging range. A main drawback of this technique is the intrinsic loss, which occurs when the two beams from the sample arm are combined. For two beams only, this drawback can be overcome by using two polarization states and a polarization beam combiner [148]. However, even in this case, only half of the reference power contributes to the OCT signal, which again leads to a 3 dB sensitivity penalty. In joint-aperture (JA) OCT, multiple passive channels detect light backscattered from outside the illumination aperture [149, 150], which increases sensitivity, data rate and image quality, but not A-scan rate.

Another type of multiplexing can overcome limits of the data acquisition hardware. In spectral domain OCT, the speed limitation arises from the line camera. A discrete spectrometer consisting of hundreds of discrete photodiodes was constructed to enable SD-OCT imaging at several MHz line rate [151]. Four line cameras have been used with sequential readout in order to achieve line rates of one MHz in SD-OCT [152]. Another
bottleneck in data transfer was addressed by employing two analog-to-digital converters in sequence for a SS-OCT setup, to enable sustained real-time imaging [125].

**Line-field and full-field OCT**

Line-field and full-field OCT are multiplexing approaches that give up the confocal detection scheme of OCT to various extents, by using a line or 2D camera with corresponding sample illumination, instead of a single detector and point illumination. Hence multiply scattered light is not rejected as in confocal OCT, and image quality is in general inferior to standard OCT systems, especially for highly scattering samples. However, the parallelized illumination and detection generates very high speeds, and more optical power may be applied to the sample. In principle, this is similar to using an imaging spectrograph [153]. Some of the first line-field implementations had a rather slow acquisition time of 10s for a volume, corresponding to about 14 kHz A-scan rate [154]. Higher speeds were obtained by Graciar et. al. [155], and Nakamura et. al. with 823,200 A-lines/s for single frame imaging [156]. Recently, a line-field setup for high-quality retinal imaging was demonstrated by Fechtig et. al. at 1 MHz A-scan rate [8]. Full-field OCT has first been implemented in TD-OCT [157], and has the advantage that spatially incoherent, very wideband thermal light sources can be used [158]. Hence very high resolution is obtained [159]. As with SD-OCT, full-field OCT is usually implemented around 800 nm wavelength, where cheap silicon-based cameras are available. At longer wavelengths, InGaAs cameras were used [160]. These implementations where focused on resolution, not speed. Recent research demonstrated that high effective line rates are possible [155]. Bonin et. al. performed imaging at 1.5 MHz [161]. Holoscopy is a full-field technique that works without objective lens, and generates high-resolution images by numerical processing [162]. While line- and full-field OCT can generate very high effective line rates, the acquisition time for a single-line or volume is relatively long. Hence, they are much more prone to motion artifacts as very slow OCT systems, although numerical correction schemes exist [163]. Streak mode OCT also uses a 2D camera, but maintains confocal scanning by recording the spectrum in streak mode, demonstrated at up to 2 MHz [164]. However, the prolonged integration time induces a larger SNR drop when samples are measured with axial or transverse motion. Motion artifacts were almost eliminated with even higher record speeds of up to 38.6 MHz [165] for in-vivo retinal imaging. Besides the potential for very high speed, line- and full field approaches have the advantage of inherent phase-stability in transverse direction, greatly simplifying OCT-phase imaging and Doppler-type OCT.

5. Applications

There are many applications of fastest OCT systems, and we will only provide a very brief overview over some applications which show the particular class of benefit. Image-degrading motion artifacts are inevitable for many samples such as the human retina [119]. Complex hardware and numerical correction schemes have been employed [166], but they are difficult to implement and need to be tailored to the specific application. High speed reduces motion artifacts in any application, eliminating the need for complicated correction schemes. Since more data is acquired in the same time, high speed enables wide-area scanning of motion-sensitive samples [32, 167]. At the same time, high transverse resolution can be maintained, which is especially important for adaptive-optics OCT [168]. With high speed dense / isotropic sampling is possible over wide fields of view, providing a straightforward means for speckle reduction, which obstructs fine anatomical detail [169]. Moreover, arbitrary scan paths can be generated from the densely sampled volumetric data set [32, 170]. For instance, arbitrary en-face planes can be visualized with very high resolution in post-processing, see Fig. 7(A), 7(B) showing segmented layers of the human retina. This is not possible with any other imaging technique. Endoscopic imaging is another area where motion artifacts are critical. An ultrahigh speed endoscopic swept source optical coherence tomography (OCT)
A system for clinical gastroenterology was constructed, using a vertical-cavity surface-emitting laser (VCSEL) and micromotor imaging catheter. The system had a 600 kHz A-scan rate [171]. In intravascular OCT [172], motion artifacts are reduced by imaging within one heart beat OCT [129], see Fig. 7(C). Apart from motion artifacts, some applications offer only a limited time slot for data acquisition. For instance, in industrial inspection, samples move on conveyor belts at very high speeds, giving only very limited time for imaging. Very high speeds are thus essential to obtain good sample coverage and high image quality. For instance, Fig. 7(D) shows an averaged cross-section of a food container with glass splitter, acquired with MHz-OCT in a few milliseconds.

Various functional OCT imaging techniques have been envisioned to provide additional image contrast on top of the OCT structural images. In OCT angiography, multiple scans are taken from the same position, which can visualize blood flow in contrast to the static remaining sample structures. Multiple scanning strongly favors fast systems, especially for in-vivo and wide area applications [173]. With MHz A-scan rate, whole volumes instead of single frames can be analyzed to provide flow contrast [174]. Moreover, repetitive imaging of flow can yield not only velocity, but also the total flow [175, 176]. Another functional extension allows for determination of tissue elasticity [177]. In developmental biology, high speed imaging with FDML lasers has demonstrated time-resolved imaging of heart motion [178, 179]. OCT can also be used for process control in laser welding or laser ablation [180, 181]. The small single-volume acquisition time is also beneficial for phase-sensitive OCT, in which tiny displacements are captured via the OCT signals changing phase. In 2D-mode, high-speed profilometry can be achieved [182]. The high phase stability permits numerical refocusing and aberration control even for in-vivo applications [165], allowing for tracking of pulse waves on the retina [183]. Recently, Hillmann et. al. demonstrated the nanometer-scale length change of photoreceptors in the living human retina upon light stimulation [184], see Fig. 7(E). In elastography, tissue discrimination is performed by changes in sample structure under pressure, using phase-sensitive OCT. High speed helps to differentiate these changes from motion artifacts, as demonstrated by Singh et. al. at 1.5 MHz A-scan rate [185].

A very exciting application of high speed is 4D-OCT (3D + time), the continuous real-time visualization of the three-dimensional sample structure [125]. First, this provides a means to study the time-evolution of the sample structure, for instance in biology. Second, the real-time 3D coverage can have similar field-of-view and resolution as (scanning-laser) microscopy, while additionally providing views of layers inside the sample. Hence 4D-OCT may replace or augment existing microscopy solutions. One promising application is intraoperative OCT for surgical guidance, which aims to improve surgical imaging by providing depth-resolved views to the surgeon [186–189]. Currently, this technology is commercially available as ad-on to surgical microscopes, such as from Leica, Haag-Streit or Carl Zeiss Meditec. These solutions only provide 2D cross-sectional images due to limited speed. With faster systems, 4D visualization provides a wealth of additional data [190], supporting surgeons [191] and showing structures invisible with the standard operating microscope [192]. In future, those systems could expand the surgeon’s capabilities and the quality of the intervention. Figure 7(F) depicts a screenshot from a 4D-OCT system (Optores), showing an en-face view, 3D view and cross-sectional view, all from the same data set, updated live about 20 times per second in real-time.
Fig. 7. (A) Wide field angiography of human retina, from [173]. (B) Choroid, segmented from 3D OCT data set, from [167]. (C) Coronary artery, captured within one heartbeat, from [129]. (D) Glass splitter in food container. (E) Photoreceptor response, from [184]. (F) 4D-OCT of human fingertip (Optores GmbH).

6. Conclusion and outlook

25 years after the first demonstration of OCT, there is still a vibrant and active technology research community. This is very much in line with the history of OCT itself, which was always a very technology driven one. OCT development successfully brought concepts from electrical engineering, like coherent heterodyne detection, advanced spectral signal manipulation, and coherent detection technique to optical imaging. OCT research also transferred approaches from other medical imaging technologies like ultra-sound and x-ray imaging in form of Doppler sensing or catheter based probes to optical imaging. Most remarkably, many of the OCT technology developments which might have appeared very remote future technology at the time, already made it to clinical application. OCT enhancements like phase sensitive OCT angiography and intravascular catheters are already in clinical use and very widely deployed.

The speed revolution in OCT clearly triggered another paradigm shift. When commercial systems increased their performance by shifting from time-domain to Fourier domain technology and from 400 A-scans per second to 30 kHz in the early 2000’s, that was when the retinal OCT system market sky-rocketed. Improved image quality, simpler handling and many more suppliers all that can more or less directly be traced back to the technology shift and the increased speed performance. Besides the commercial success, the higher speed enabled many new OCT imaging approaches which are not possible in vivo at lower speed due to sample or patient motion. Doppler-OCT, OCT angiography, elastography, wide field polarization sensitive OCT, 2D OCT surgical guidance, all need several 10 kHz A-scan rate if used in vivo. This is why the technology change from TD-OCT to FD-OCT happened in just a few years.
The next step and the second phase in the speed revolution is the transition towards MHz-class systems. As described in this paper, there are many interesting applications that require OCT A-scan rates between 0.5 and 5 MHz. Many approaches, like “protocol free” ultra-widefield retinal imaging, heart-beat OCT, 4D surgical guidance, optophysiology, numerical aberration correction all require high speed if done in vivo. Considering the massive impact previously caused by the transition from 400 Hz TD-OCT to 30 kHz FD-OCT, which is a factor of 100, a similar disruption can be expected by going from today’s standard of ~50kHz to 5MHz – another factor of 100.

Today, there are several obstacles on the way to widespread deployment of MHz OCT: (1) there is a lack of commercially available system manufacturers, only one company offers MHz-OCT; (2) there is a wealth of technological challenges connected with going to higher speed, in the case of SS-OCT this is mainly linked to the higher RF-frequencies which go well into the GHz range by now; (3) fast data streaming becomes more and more challenging, transfer rates of more than 6 Gbytes/s have to be handled; (4) MHz-OCT systems generate a large amount of data, and it makes sense to store most of them, especially in cases where follow up imaging should be performed; (5) the wealth of information asks for new ways how to present data to the doctor or investigator; (6) and finally, right now MHz-OCT is more expensive than standard OCT, partly due to the points listed above.

We expect, that most of these issues will be solved, some even independently from OCT. Higher speed electronics and data streaming will be developed and be more widely available as telecom technology shifts more and more to coherent receiver technology, which can be used for SS-OCT. Data storage and visualization is a common problem also in other medical imaging systems. Modern CTs, ultrasound systems and even MRI machines can generate hundreds of images from a patient, rising the same problem of large data and proper visualization. For MHz SD-OCT, line scan cameras will become faster, even without the demand from OCT; the technology development there will be driven mainly by industrial inspection demand. (6) the last point, OCT systems price, is probably the most critical one – representing the typical chicken-egg dilemma; MHz-OCT system cost will come down for sustained large sales number, however many high-volume applications require low cost systems beforehand. So it can be expected that the point of MHz-OCT being more expensive will only be solved over time. This might give research some more time to identify even more applications where MHz-speed is mandatory.

To summarize, today in the year 2016, 25 years after the introduction of OCT, it looks like there will be a great diversity of OCT-technology platforms for at least another 25 years. SD-OCT is preferred when high resolution is required, full-field for phase stable measurements and optophysiology, SS- OCT when long ranging depths matter. And, since clearly not all applications require super high speed, many of the slow 100kHz class OCT approaches of today will stay for now. For applications where speed is no concern at all, we will even see TD-OCT again. In general, the advice for the next 25 years of OCT is: “Use the right tool for your problem – and some problems clearly require high speed”.

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