Since its invention in the late 1980s [1–4] and early 1990s [5–7], the original idea of optical coherence tomography (OCT) was to enable noninvasive optical biopsy, i.e., the in situ imaging of tissue microstructure with a resolution approaching that of histology, but without the need for tissue excision and postprocessing. An important advance toward this goal was the introduction of ultrahigh resolution OCT (UHR OCT). By improving axial OCT resolution by one order of magnitude from the 10–15\(\mu\)m to the submicrometer region [8–11], UHR OCT enables superior visualization of tissue microstructure, including all major intraretinal layers in ophthalmic applications as well as cellular resolution OCT imaging in nontransparent tissue. This chapter reviews the state-of-the-art technology that enables UHR OCT covering the entire wavelength region from 500 to 1,600nm and discusses fundamental limitations of OCT image resolution.

The performance of an OCT system is mainly determined by its longitudinal (axial) resolution, transverse resolution, dynamic range (i.e., sensitivity), and data acquisition specifications, including digitization resolution and speed [11]. For application in medical diagnosis, additional factors, e.g., noncontact vs. contact applicability, possible penetration into the investigated tissue, image contrast as well as extraction of functional or biochemical information in addition to the visualization of micro-structural morphology have to be considered. In addition, for clinical applications, compactness, user-friendliness, robustness, flexibility, overall costs of the OCT system, as well as the possibility to interface it to existing diagnostic technology are decisive factors.

### 8.1 Longitudinal and Transverse Resolution in OCT

In contrast to conventional and confocal microscopy, OCT achieves very high axial image resolutions independent of focusing conditions (Fig. 8.1). The axial and transverse resolutions of OCT are decoupled:
Axial (depth) resolution – defined by the coherence length of the light source (rather than the depth of field as in microscopy)

Transverse resolution – defined by the focal spot size

As in conventional microscopy, the transverse resolution and the depth of focus are determined by the focused transversal spot size, defined as the $1/e^2$ beam waist of a Gaussian beam (Fig. 8.1). Assuming Gaussian rays and only taking into account Gaussian optics, the transverse resolution can be defined by

$$\Delta x = \frac{4\lambda f}{\pi d},$$

(8.1)

where $f$ is the focal length of the lens and $d$ is the spot size on the objective lens. Increasing the numerical aperture of the objective increases the transverse resolution by reducing the focal spot size, but it decreases the depth of field, quantified by the confocal parameter $b$, which is $2z_R$ or twice the Rayleigh length $2z_R = b = \pi \Delta x^2 / 2\lambda$. Thus, improving the transverse resolution can be accomplished by increasing the numerical aperture (NA) of the objective, but at the same time decreasing $b$. A solution to this limitation is the use of a dynamic focus tracking system. Especially for ophthalmic retinal OCT imaging, low numerical aperture focusing is employed, because it is desirable to have a large depth of field and to use OCT to achieve high axial resolution.

**Fig. 8.1.** Resolution limits of OCT. OCT can achieve high axial resolutions independent of numerical aperture. Using low coherence interferometry, the axial resolution is inversely proportional to the bandwidth of the light source. The transverse resolution is given by the focus spot size. The depth of field is determined by the confocal parameter of the focused beam.