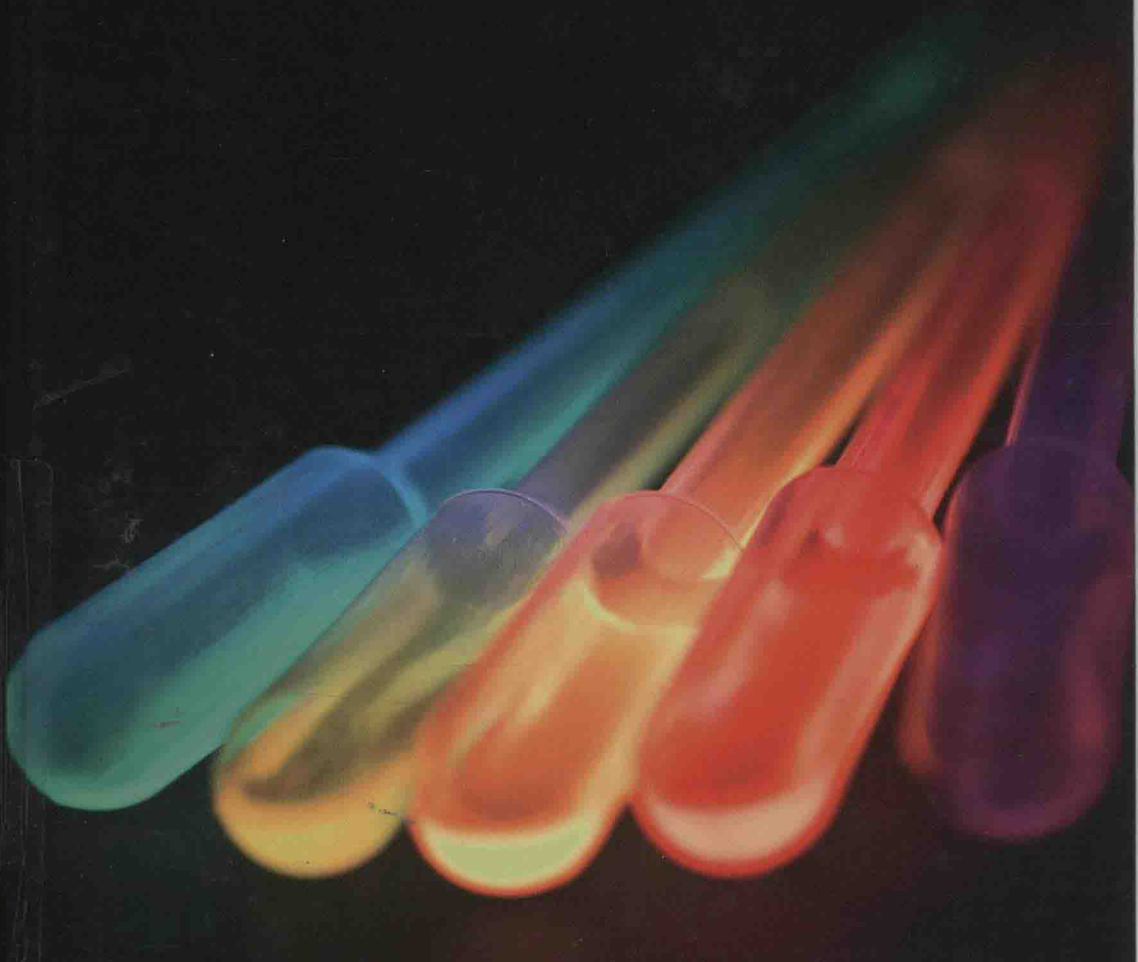
# Quantum Dots

## Optical Properties

Eva Murphy



# Quantum Dots: Optical Properties

Edited by **Eva Murphy**

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# Quantum Dots: Optical Properties

# Preface

In my initial years as a student, I used to run to the library at every possible instance to grab a book and learn something new. Books were my primary source of knowledge and I would not have come such a long way without all that I learnt from them. Thus, when I was approached to edit this book; I became understandably nostalgic. It was an absolute honor to be considered worthy of guiding the current generation as well as those to come. I put all my knowledge and hard work into making this book most beneficial for its readers.

This book gives innovative and resourceful techniques for calculating the optical and transport characteristics of quantum dot structures. The book has important chapters which discuss the novel optical properties of quantum dot structures. This is a collaborative book, providing primary research such as the ones conducted in physics, chemistry and material science. This book serves as an important source of reference for this field.

I wish to thank my publisher for supporting me at every step. I would also like to thank all the authors who have contributed their researches in this book. I hope this book will be a valuable contribution to the progress of the field.

**Editor**

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# Optical Properties of Quantum Dot Systems

# 1

# InAs Quantum Dots of Engineered Height for Fabrication of Broadband Superluminescent Diodes

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## 1. Introduction

Superluminescent diodes (SLDs) are of great interest as optical sources for various field applications like fibre-optic gyroscopes (Culter et al, 1980), optical time-domain reflectometry (Takada et al, 1987), sensing systems (Burns et al, 1983) (such as Faraday-effect electric current sensors and distributed Bragg-grating sensor systems) and short and medium distance optical communication systems (Friebele & Kersey, 1994). One of the most attractive applications of SLDs has emerged after the successful demonstration of the optical coherence tomography (OCT) technique, and identification of its advantages compared to other imaging techniques in medical research and clinical practices. OCT is a real time and non-invasive imaging technique that uses low-coherence light to generate resolution down to the sub-micron-level, two- or three-dimensional cross-sectional images of materials and biological tissues. The earliest version of the OCT imaging technique was demonstrated in 1991 by Huang and co-workers (Huang et al, 1991), by probing the human retina *ex vivo*. Imaging was performed with 15μm axial resolution in tissue using a light source with a central wavelength of 830nm. Two years later, *in vivo* retinal images were reported independently by Fercher et al. (Fercher et al, 1993) and Swanson et al (Swanson et al, 1993). Although 800nm OCT systems can resolve all major microstructural layers of tissues, image quality can be severally degraded by light scattering phenomena. In low-coherence interferometry, the axial resolution is given by the width of the field autocorrelation function, which is inversely proportional to the bandwidth of the light source. In other words, light sources with broadband spectra are required to achieve high axial resolution. Although at longer wavelengths the bandwidth requirement increases, there is a significant advantage in using light sources of longer central wavelengths for which the light scattering is significantly reduced.

In recent few years, broadband light sources around 1μm have received considerable attention for their use in medical imaging technologies. It is due to the optimal compromise between water absorption and human tissue scattering that the 1000-1100 nm wavelength range has been proposed, and demonstrated, to be more suitable for OCT applications as compared to those that use a light source with a central wavelength of 800nm (Pavazay et al, 2003; Pavazay et al, 2007). There are a myriad of choices in selecting such OCT light sources i) femtosecond or fiber lasers that are dispersed to produce super-continuum light and swept source lasers (Hartl et al, 2001; Wang et al, 2003), ii) thermal sources, and iii) superluminescent diodes (Sun et al, 1999; Liu et al, 2005; Lv et al, 2008; Haffouz et al, 2010). Although the reported OCT tomograms with the highest axial resolution (1.8μm) were so far achieved in research laboratories with a photonic crystal fibre based source (Wang et al, 2003), superluminescent diodes are considerably lower in cost and complexity as well as being smaller in size, which makes them more attractive for mass production. Superluminescent diodes utilizing quantum-dots (QDs) in the active region are considered to be excellent candidates as light source for an OCT systems. The naturally wide dimensional fluctuations of the self-assembled quantum dots, grown by the Stranski-Krastanow mode, are very beneficial for broadening the gain spectra which enhances the spectral width of the SLDs. On the other hand, the three-dimensional carrier confinement provided by the dots' shape results in high radiative efficiency required for the OCT applications.

In this chapter the main governing factors to demonstrate ultrahigh-resolution OCT-based imaging tomographs will be reviewed in the second section. Research advances in the growth processes for engineering the gain spectrum of the quantum dots-based superluminescent diodes will be summarized in the third section of this chapter. Our approach for engineering the bandwidth of multiple stacks of InAs/GaAs QDs will be presented in the fourth section and demonstration of an ultra wide broadband InAs/GaAs quantum-dot superluminescent diodes (QD-SLDs) will be then reported in the last section of this chapter. Our approach is based on the use of SLDs where the broad spectrum is obtained by a combination of slightly shifted amplified spontaneous emission (ASE) spectra of few layers of dots of different heights. Spectral shaping and bandwidth optimization have been achieved and resulted in 3dB-bandwidth as high as ~190nm at central wavelength of 1020nm. An axial resolution of 2.4μm is calculated from our QD-SLDs.

## 2. Superluminescent diodes for ultrahigh-resolution optical coherence tomography (UHR-OCT)

Since its invention in the early 1990s (Huang et al, 1991), OCT enables non-invasive optical biopsy. OCT is a technique that provides *in-situ* imaging of biological tissue with a resolution approaching that of histology but without the need to excise and process specimens. OCT has had the most clinical impact in ophthalmology, where it provides structural and quantitative information that can not be obtained by any other modality. Cross-sectional images are generated by measuring the magnitude and echo time delay of backscattered light using the low-coherence interferometry technique. The earliest versions of OCT have provided images with an axial resolution of 10-15μm. OCT has then evolved very quickly, with two-dimensional (2D) and three-dimensional (3D) microstructural images of considerably improved axial resolution being reported (Drexler et al, 1999). These ultrahigh-resolution OCT systems (UHR-OCT) enable superior visualization of tissue microstructure, including all intraretinal layers in ophthalmic applications as well as cellular resolution OCT imaging in nontransparent tissues. The performance of an OCT system is mainly determined by its longitudinal (axial) resolution, transverse resolution, dynamic range (sensitivity) and data acquisition speed. Other decisive factors like depth penetration into the investigated tissue (governed by scattering, water absorption) and image contrast need to be carefully addressed. In addition, for field application, compactness, stability, and overall cost of the OCT system should be considered.

### 2.1 Factors governing OCT imaging performance

In this section we will review the key parameters that are directly or closely related to the light source used in the OCT technique. Other limiting factors, related to other optical, electronic and/or mechanical components can affect the resolution in OCT system when not properly addressed. For more details regarding OCT technology and applications, please refer to the book edited by Drexler and Fujimoto (Drexler & Fujimoto, 2008).

#### 2.1.1 Transverse and axial resolution

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

$$\Delta x = \frac{4\lambda}{\pi}\frac{f}{d} \tag{1}$$

where $f$ is the focal length of the lens, $d$ is the spot size of the objective lens and $\lambda$ is the central wavelength of the light source. Finer transverse resolution can be achieved by increasing the numerical aperture that focuses the beam to a small spot size. At the same time, the transverse resolution is also related to the depth of the field or the confocal parameter $b$, which is $2z_R$, or two times the Rayleigh range:

$$b = 2z_R = \frac{\pi \Delta x^2}{\lambda} \tag{2}$$

Therefore, increasing the transverse resolution produces a decrease in the depth of the field, similar to that observed in conventional microscopy. Given the fact that the improvement of the transverse resolution involves a trade-off in depth of field, OCT imaging is typically performed with low numerical aperture focusing to have a large depth of field. To date, the majority of early studies have rather focused on improving the axial resolution.

Contrary to standard microscopy, the axial image resolution in OCT is independent of focusing conditions. In low-coherence interferometry, the axial resolution is given by the width of the field autocorrelation function, which is inversely proportional to the bandwidth of the light source. For a Gaussian spectrum, the axial (lateral) resolution is given by:

$$\Delta z = \frac{2Ln(2)}{\pi}\frac{\lambda^2}{\Delta\lambda} \tag{3}$$

where $\Delta z$ is the full-width-at-half-maximum (FWHM) of the autocorrelation function, , and $\Delta\lambda$ is the FWHM of the power spectrum.

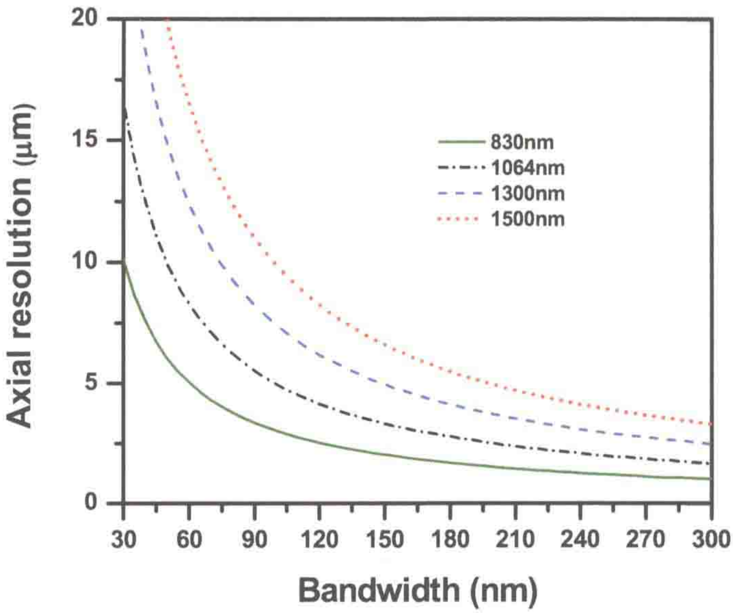


Fig. 1. Axial resolution versus bandwidth of light sources for central wavelengths of 830, 1064, 1300 and 1500nm.

Since the axial resolution is inversely proportional to the bandwidth of the light source, broadband light sources are required to achieve high axial resolution. For a given bandwidth, improving the axial OCT resolution can be also achieved by reducing the central wavelength of the light source (c.f. Figure 1). It should also be noticed that to achieve a given axial resolution the bandwidth requirement is increased at longer wavelengths. For example, to achieve an axial resolution of 5μm, the bandwidth required is only 50nm at central wavelength of 830nm, and three times higher when a light source of central wavelength of 1300nm is chosen.

### 2.1.2 Imaging speed-sensitivity in OCT

Detection sensitivity (detectable reflectivity) has a significant impact on the imaging speed capabilities of an OCT system. As the scan speed increases, the detection bandwidth should be increased proportionally, and therefore the sensitivity drops. The sensitivity of state-of-the-art time-domain OCT systems that operate at relatively low imaging speed (~2kHz A-line rate), ranges between -105 and -110dB. Increasing the optical power of the light source should in principle improve the sensitivity; however, the available sources and maximum permissible exposure levels of tissue represent significant practical limitations. The potential alternative technique for high-imaging speed is the use of Fourier/spectral domain detection (SD-OCT) or Fourier/swept source domain detection (SS-OCT) also known as optical frequency domain imaging (OFDI). The first approach, SD-OCT, uses an interferometer with a low-coherence light source (superluminescent diodes) and measures the interference spectrum using a spectrometer and a high-speed, line scan camera. The second approach, SS-OCT, uses an interferometer with a narrow-bandwidth, frequency-swept light source (swept laser sources) and detectors, which measure the interference output as a function of time. Fourier domain detection has a higher sensitivity as compared to time domain detection, since Fourier domain detection essentially measures all of the echoes of light simultaneously, improving sensitivity by a factor of 50-100 times (enabling a significant increase in the imaging speeds).

### 2.1.3 Image contrast and penetration depth in OCT

Tissue scattering and absorption are the main limiting factors for image contrast and penetration depth in OCT technology. Indeed, OCT penetration depth is significantly affected by light scattering within biological tissue, which scales as $1/\lambda^k$, where the coefficient $k$ is dependent on the size, shape, and relative refractive index of the scattering particles. The difference in tissue scattering and absorption provides structural contrast for OCT. Since scattering depends strongly on wavelength and decreases for longer wavelengths, significantly larger image penetration depth can be achieved with light centered at 1300nm rather than 800nm. However, above 1300nm the water absorption becomes a problem. So far, the majority of clinical ophthalmic OCT studies have been performed in the 800 nm wavelength region. Excellent contrast, especially when sufficient axial resolution is accomplished, enables visualization of all major intraretinal layers, but only limited penetration beyond the retina. This limitation is mainly due to significant scattering and absorption phenomena.

Water is the most abundant chemical substance in the human body, accounting for up to 90% of most soft tissues. The most commonly used wavelength window of low water absorption ($\mu_a$<0.1cm$^{-1}$) for OCT imaging is lying in the 200-900 nm range. Above 900 nm the absorption coefficient increases fairly rapidly to reach $\mu_a$~ 0.5cm$^{-1}$ at ~970 nm, drops back to ~0.13cm$^{-1}$ at 1064nm, and then continues to increase at longer wavelengths into the mid-infrared. The region of low absorption around 1060nm acts as a 'window' of transparency, allowing near infrared spectroscopic measurements through several centimeters of tissue to be made. For this reason, OCT imaging at 1060 nm can achieve deeper tissue penetration into structures beneath the retinal pigment epithelium, as well as better delineation of choroidal structure.

## 2.2 Light source for ultrahigh resolution OCT

The light source is the key technological parameter of an OCT system. The performance characteristic of the light source, such as central wavelength, bandwidth, output power, spectral shape, and stability will directly affect the OCT image resolution. For this reason, a proper choice of the light source for optimized performance OCT system is imperative. In the recent years, there has been considerable interest in the use of broadband light sources around 1064nm for use in ophthalmic OCT applications. It is due to the optimal compromise between water absorption and human tissue scattering that the 1064nm wavelength '*window*' has been proposed, and demonstrated, to be more suitable for OCT applications as compared to those that use a light source with a central wavelength of 800nm (Povazay et al, 2007). There are a myriad of choices in selecting such OCT light sources i) femtosecond or fiber lasers that are dispersed to produce super-continuum light and swept source lasers, and ii) superluminescent diodes. Highly non-linear air-silica microstructure fibers and photonic crystal fibers (PCFs) can generate an extremely broadband continuous light spectrum from the visible to the near infrared by use of low-energy femtosecond pulses (Wang et al, 2003; Hartl et al, 2011). Spectral bandwidth up to 372nm was achieved at 1.1μm central wavelength. The super-continuum light source also has the advantage of achieving faster imaging speed with higher signal-to-noise ratio.

Although the reported OCT tomograms with the highest axial resolution (1.8μm) were so far achieved in research laboratories with a photonic crystal fibre based source (Wang et al, 2003), superluminescent diodes are considerably lower in cost and complexity as well as being smaller in size, which makes them more attractive for mass production. Superluminescent diodes utilizing quantum-dots in the active region are considered to be an excellent candidate as a light source for an OCT system. The naturally wide dimensional fluctuations of the self-assembled quantum dots, grown by the Stranski-Krastanow mode, are very beneficial for broadening the gain spectra which enhances the spectral width of the SLDs. On the other hand, the three-dimensional carrier confinement provided by the dots' shape results in high radiative efficiency required for the OCT applications.

## 3. Reported superluminescent diodes for bandwidth widening and their performance parameters

Since the first report in 1993 (Leonard et al, 1993), the formation of strained self-assembled quantum dots by heteroepitaxial growth in the Stranski–Krastanow mode has been studied extensively for their fundamental properties and applications in optoelectronics. Significant breakthroughs occurred over the last two decades with the fundamental understanding of the QDs systems and the demonstration of zero-dimensional novel devices. These achievements are directly related to the noticeable advances in the epitaxial materials deposition. With self-assembled QDs growth process, a certain size inhomogeneity is common and typically not less than 10%. It has been predicted (Sun & Ding, 1999) that the full width at half maximum of the SLDs output spectrum of the $In_{0.7}Ga_{0.3}As/GaAs$ quantum dot system, with a standard deviation in the average size of the QD ensemble of 10%, can be as high as 140nm. Increasing further the size variation of the dots to 30% should result in bandwidth as high as 160nm. The confinement potential between the dots and the barriers is another important factor for modifying the spectral width. With only 10% size variation increasing the potential confinement by using higher indium composition in the dots a spectral width of 230nm was predicted in the $In_{0.9}Ga_{0.1}As/GaAs$ quantum dot system (Sun & Ding, 1999). In general, such inhomogeneous size distribution of self-assembled QDs in the active region is disadvantageous for achieving lasing of QD-lasers. However, for the designed wide spectrum QD-SLDs it becomes an effective intrinsic advantage for broadening the emission spectrum. Experimentally, using five layers of InAs/GaAs QDs grown under identical growth conditions in a molecular beam epitaxy system (Liu et al., 2005), SLDs with full width at half maximum of ~110nm at a central wavelength of 1.1μm have been made. For high resolution optical coherence tomography applications around 1060nm an even wider broadband spectrum is required. Increasing further the bandwidth of the emission spectrum of the SLDs is a complicated process and requires more than just optimization of the growth conditions of the active region of the device. The precise control of the average size distribution of the dots within one layer is a very challenging process and is very difficult to reproduce. Very practical and successful ideas based on engineering the matrix surrounding the QDs have been also proposed and applied to the fabrication of