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Microscopic Imaging and Spectroscopy with Scattered Light

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Abstract

Optical contrast based on elastic scattering interactions between light and matter can be used to probe cellular structure and dynamics, and image tissue architecture. The quantitative nature and high sensitivity of light scattering signals to subtle alterations in tissue morphology, as well as the ability to visualize unstained tissue in vivo, has recently generated significant interest in optical scatter based biosensing and imaging. Here we review the fundamental methodologies used to acquire and interpret optical scatter data. We report on recent findings in this field and present current advances in optical scatter techniques and computational methods. Cellular and tissue data enabled by current advances in optical scatter spectroscopy and imaging stand to impact a variety of biomedical applications including clinical tissue diagnosis, in vivo imaging, drug discovery and basic cell biology.

Keywords

Light scattering; optical imaging; microscopy; tissue diagnosis; cell analysis

1. Introduction

In contrast with X-rays, which travel with relatively little scatter through biological tissues, electromagnetic waves at optical wavelengths are significantly scattered at biological interfaces. This scattering of light by optically inhomogeneous biological tissues has posed a limitation on high resolution three-dimensional imaging of thick biological tissues and in general has prevented whole body optical tomographic imaging on the scale of X-ray computed tomography. Nonetheless, elastic scattering of light at tissue, cellular and subcellular interfaces provides an important source of natural biological contrast which can be harnessed to provide quantitative measurements of cellular and tissue states. Thus, significant advances in optical technology have recently been focused on utilizing this fundamental form of light-tissue interaction to enable optical-scatter-based imaging and sensing. High resolution optical scatter sensing is usually limited to tissue surfaces, thin specimens, or weakly scattering tissues. However important biomedical and biological

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applications ranging from cancer diagnosis to cell and developmental biology have been achieved even in this limited scattering regime. This review presents recent advances in optical scatter imaging and spectroscopy and includes a description of novel methodologies and the biological data that can be derived from them.

2. Elastic light scattering as a source of optical contrast

Aside from absorption by blood and certain tissue pigments (e.g. melanin) in specific wavelength ranges (1), elastic scattering is by far the most prevalent form of light interaction with biological tissue, with elastic scattering cross sections that are several orders of magnitude larger than fluorescence or inelastic Raman scattering.

In contrast with fluorescence, which carries information about the excited energy state of a specific biochemical, elastic scattering does not involve photon absorption and the atoms involved do not make a transition to an excited state before the scattering process takes place. Since the process is elastic and does not involve absorption (Fig 1A), the scattered light has the same frequency as the incident light, and elastic scattering can occur for a broad range of incident wavelengths. A detailed quantum description of the scattering process may be found in (2). In a typical measurement, light incident on a scattering source is elastically scattered in different directions upon interaction with the scattering source. By measuring the magnitude, phase, and the angular- or wavelength-dependence of the scattered far field, one can retrieve information about the scattering sources' cross sections (or extinction due to scattering), structure, and refractive index (3). The dependence of scattering on the polarization of the incident field can also be used to study the birefringent or dichroic properties of the scattering source (3, 4). Fig. 1C shows an illustration of light scatter irradiance as a function of angle and size for two spheres or different diameter (Fig. 1B). As noted in the figure, the angular intensity depends on the ratio of the particles' refractive index to the surrounding medium's refractive index, and on the ratio of particle diameter to the wavelength of the incident light. The scattering signal may be measured spectroscopically as a function of angle and wavelength as illustrated in Fig. 1C. However, if the data are acquired with an imaging system (Fig. 2), the elastic scattering signal can be spatially registered with the positions of the corresponding scattering sources, resulting in optical scatter maps (images) of the sample. In this case, the spectral dependence of the scattering signal may be obtained by varying the wavelength of the incident light used to image the sample, while the angular scattering data may be retrieved from a conjugate transform plane of the imaging system. Finally, the light scattering information may also be obtained by interferometry (Fig. 3) to quantify phase sensitive scattering.

3. Physical interpretation of the optical scatter data

Rayleigh scattering by electrical dipoles constitutes a fundamental form of elastic scattering by individual biological molecules or biological particles with dimensions much smaller than the wavelength of the light. However, biological tissue consists of many different compartments with different indices of refraction. These compartments range in size and may have dimensions and separation distances on the order of, or smaller than, the wavelength of light. For example, organelles within cells may be as large as half a micron or as small as a few tens of nanometers, and may be separated from other organelles by distances on the order of less than one to several microns. These biological compartments are too large to scatter like individual Rayleigh scattering dipoles, and often too small to allow geometrical optics approximations of light propagation through them. Elastic scattering by these biological compartments must therefore be described using other theoretical formalisms.

When light interacts with biological tissue elastic scattering is inevitable. Any spatial variation in refractive index leads to light scattering. The tissue structure and the spatial distribution of refractive index are interlinked. A simple relationship expresses the optical refractive index n through the local molecular density: $n = n_0 + \alpha\rho$, where n_0 is the refractive index of the medium surrounding a particular scattering structure, ρ is the local concentration of solids (mostly macromolecules such as DNA, RNA, proteins, and lipids), and α is the specific refraction increment. As far back as in the 1950s, it has been shown that the majority of substances found in living cells have approximately the same $\alpha \sim 0.18$ (5). This implies that a particle comprised of 100% protein would have refractive index 0.18 higher than that of water. Despite its apparent simplicity, this linear relationship was shown to hold for ρ at least up to 50%, thus covering the physiological range. Remarkably, the value of α does not significantly vary for different types of macromolecules. Thus, optical refractive index is a measure of local density and, as such, tissue architecture. This underlines the power of light scattering as a tool in probing tissue architecture.

If the information about the spatial distribution of mass density in cells and tissue were known, one could explicitly relate light scattering observables (the distribution of intensity over angle and spectrum, phase of scattered light, etc.) to the underlying tissue structure, at least in principle. The problem is that still little is known about the spatial distribution of macromolecular density inside cells and tissues. In fact, this is one of the reasons behind interest in light scattering-based microscopic techniques such as those discussed in Section 8 (6-9).

A central question is how to treat the origin of scattering. Perhaps the most popular approach is the approximation of isolated scattering particles. In its simplest form, Mie theory, which describes scattering by homogeneous spheres of arbitrary size and refractive index, is used. In particular, Mie theory has been used to determine size distributions of tissue scatterers based on angular or spectral scattering patterns. A criticism of this approach is that, first, tissue is not comprised of spherical particles and, second, Mie theory is valid only for isolated scatters, i.e. scatterers would have to be in the far field of each other. Generally, this condition is not satisfied in tissue. On the other hand, recent studies show that Mie theory-based interpretation of light scattering may indeed help elucidate (at least qualitatively) the distribution of length scales of spatial refractive index variations (rather than real scattering particles) (8, 10, 11).

The limitation of Mie theory in respect to its inability to treat nonspherical and heterogeneous particles can be lifted by use of approximations to the integral equation of light scattering: the WKB approximation (also known as the Van de Hulst or the anomalous diffraction approximation–valid for optically soft particles much larger than the wavelength of light, such as cells and cell nuclei) and the Born approximation (valid for micron-size and submicron particles, such as small organelles) (10, 12-15). In particular, the total scattering cross section of a heterogeneous, nonspherical particle is similar to that of the equiphase sphere (EPS), i.e. a sphere that would produce the same maximal phase shift as the particle. The EPS approximation works best for particles with fine grains of either refractive index or surface perturbation (11). A combination model that utilizes the Born and the WKB approximations has recently been developed (15).

The second major limitation of the Mie theory (isolated particle scattering) can be addressed by considering tissue as a random continuum of spatial refractive index variations. A comprehensive approach to describe tissue microarchitecture is by means of the mass density spatial correlation function, which is linearly proportional to the refractive index correlation function $C(r)$. Typically, the spectral and angle-dependent properties of light scattering are sensitive to $C(r)$ for length scales r from $\sim\lambda/20$ (~20 nm) to a few microns.

Light scattering observables are then expressed in terms of the properties of $C(r)$. In a particular case of randomly homogeneous, isotropic correlation with no internal boundaries, a three-parameter model based on the Whittle–Matérn functional family has been developed that describes some of the most popular types of the correlation (16, 17). This $C(r)$ is defined by index m that defines its shape, the average amplitude and the length scale of the spatial variations of refractive index. If $m<1.5$, $C(r)$ is a mass fractal correlation with the mass fractal dimension $D=2m$; if 1.5<m<2, $C(r)$ is a stretched exponential; if m=2, it is an exponential function, and, finally, m≫1 corresponds to the Gaussian correlation. The key optical properties that can be measured in elastic scattering studies (e.g., reduced scattering coefficient, μ_s', and anisotropy factor, g) can then be expressed through these three parameters. In particular, it was shown that the spectrum of the reduced scattering coefficient (frequently measured in light scattering experiments) depends primarily on the value of index m: $\mu_s' \propto \lambda^{2m-4}$ (for $m<2$). Many scattering techniques can measure $\mu_s'(\lambda)$ and thus can be used to determine the type of the refractive index correlation function simply by determining the power-law of its wavelength dependence.

An even more rigorous solution would require taking into account the heterogeneous distribution of refractive index, including length scales as small as a few nanometers and as large as tens of microns, without making the assumptions of the random isotropy and the absence of internal boundaries. This is currently beyond analytical realm and computational methodologies must be used: the only means to accomplish this is by numerically solving Maxwell's equations. Recently, new powerful numerical approaches such as finite-difference time-domain (FDTD) and pseudo-spectral time-domain (PSTD) computations have become available. Although the use of these computational methods is currently limited by computer resources, one can expect that as more powerful computers are made available in the future we should be able to expect the development of a more robust understanding of tissue scattering.

4. Advantages of utilizing light scattering for sensing

As in any optical sensing application, the performance of sensing methodologies based on elastic scattering interactions of light with biological tissue may be assessed based on four basic parameters: speed of detection, sensitivity, spatial resolution, and specificity. Due to the large cross sections of the elastic light scattering process, optical scatter signal throughput is typically much higher than signal originating from other light-tissue interactions under similar acquisition conditions consisting of the same collection optics, illumination power and photodetector sensitivity. For example, to achieve a desired signal-to-noise ratio, a wide field microscopic image of a cell acquired in phase contrast microscopy, in which contrast is based on elastic light scattering interaction, can be acquired approximately two orders of magnitude faster than a fluorescence image of the same cell in which an exogenous high efficiency fluorophore is used to label a specific biochemical entity.

The spatial resolution of current optical scatter imaging methods is limited by diffraction. Methods such as confocal reflectance microscopy or various phase contrast methods usually afford lateral spatial resolution on the order of λ/2NA, where λ is the wavelength of the light and NA is the numerical aperture of the imaging optics. Axial resolution is on the order of 0.5 μm in confocal reflectance microscopy of tissue (18), a few microns in optical coherence tomography (19) and 0.5-1 μm in holographic interferometry (20). Image resolution based on contrast from elastic scattering is governed by principles which apply without the requirement for fluorescence emission or non-linear fluorescent probes. Thus, super-resolution approaches such as synthetic aperture methods (see also Section 9.2) and

structured illumination could in principle be applied to imaging fields, such as reflectance, formed by scattered light (21).

While image resolution may be limited by diffraction, optical elastic scattering signals remain highly sensitive to changes in the physical properties of the scattering sources that are below the diffraction limit. Thus, changes in the dimensions of the scattering sources on the order of tens of nanometers can be achieved using optical scatter measurements (22). Changes in optical pathlength on the order of nanometers have also been reported within cells (23) as well as refractive index fluctuation on nanometer length-scales (9).

Based on this discussion, it is clear that sensing based on optical scatter contrast provides superior signal throughput compared to other contrast modalities, affords very high sensitivity to changes in object structure, and provides high imaging resolution without relying on exogenous markers. However, one important attribute of bioimaging techniques is molecular and functional specificity. In this regard, optical scatter spectroscopy and imaging of unstained biological tissue is non-specific. This non-specificity is two-fold: 1) elastic scattering provides data about the structure of the biological scatterers. Thus unlike fluorescence, which provides chemically specific data, elastic scattering data is inherently structurally specific but not molecularly specific; 2) the structural data measured by light scattering may originate from multiple scattering sources within a given resolution element and does not a priori necessarily correspond to a specific biological source. By analogy with fluorescence microscopy, a recent thrust has been directed at designing optical scatter molecular probes whose scattering (rather than fluorescence) properties can be used in imaging methods based on elastic scattering contrast. These probes, which are discussed in Section 9.1, can be used to enhance the signal of specific molecular targets and thus confer specificity onto the optical scatter signals.

Different experimental approaches have also been used to correlate light scatter measurements with molecular or biological function in unlabeled samples (see Section 5). However, it is important to note that non-molecularly specific optical scatter data can still be very valuable. This is especially true for optical scatter imaging modalities (e.g. reflectance confocal microscopy and OCT, see Section 8), which can be used non-invasively in vivo for microscopic visualization of tissue structure safely, at low cost, and without the use of exogenous dyes. In addition, light scattering measurements can provide a means to track non-invasively biophysical changes within tissues and cells. While such changes may not be molecularly specific, they still result from a combination of biological functions that affect structure. Thus, these biophysical changes may be used to search for tissue alterations in cases where a priori knowledge of the molecular pathways involved does not exist or where such knowledge is difficult or too costly to obtain. In these cases, the light scattering measurements may be used as an initial screening tool to identify time points during treatment, or tissue regions that exhibit significant changes in optical scatter signal compared to an established baseline. Further biochemical analysis can then be guided to these specific timepoints or regions. This screening is non-invasive and can be accomplished rapidly with minimal sample preparation or processing.

5. Biological interpretation of the optical scatter data in unlabeled samples

Optical scatter data collected from unlabeled biological tissue provides structural information which can originate from different biological compartments and interfaces, including tissue boundaries, cells, organelles or macromolecular complexes. Since multiple genes, proteins or organelles can contribute to changing the observed scattering variables or spatial patterns, the difficulty in interpreting changes in light scattering signals from a molecular biology standpoint remains: there are several potential sources that can underlie

the biophysical structural variables measured by light scattering, and the specific direct relationship between each of these biological sources and the scattering measurement is not always known a priori. Several approaches are discussed here that have been used for correlating optical scatter measurements with molecular and biological processes.

5.1 Isolating the scattering structure

Studying the scattering properties of individual biological structures can be achieved by physically isolating a structure of interest. This is best illustrated by the very well known light scattering studies of isolated viable mitochondria that were used to understand the relationship between mitochondrial function and mitochondrial matrix structure under different bioenergetic conditions. (24-30).

On the other hand, structural information could be obtained non-invasively in an imaging regime where light scattering interactions are utilized to directly visualize the scattering particles. In this case, the scattering sources can be spatially isolated in the image and their structure determined within the resolution of the optical system. Identifying a specific scatterer within an image requires that the scattering signature and spatial pattern be specifically representative of the desired structure while excluding other structures as potential contributing sources. Large biological structures with specific morphologies, such as cells and nuclei are easily identified with these methods. However, when probing subcellular compartments, quantitative optical scatter imaging methods still do not provide the specificity needed to identify directly specific organelles and subcellular structures in unlabeled cells. For this, additional modalities may be used to confer biological specificity.

5.2 Multimodal imaging

The structural data provided by the biophysical light scattering interaction of light with tissue may be combined with biochemically specific measurements to study processes that occur at the organellar or molecular levels. This can be achieved in unlabeled tissue by combining elastic scattering data with autofluorescence (31) or Raman (32) signals. Multimodal data can also be obtained by combining the elastic scatter measurements with signal from fluorescently labeled tissue (33-36). This allows correlation of the optical scatter data with molecular components which do not normally autofluoresce, and provides more flexibility in interrogating desired molecular pathways.

While the spectroscopic elastic scattering data may be found to correlate with changes in the biochemical signals, the elastic scattering data may not necessarily originate from the same specific biological source unless the optical scatter and biochemical signals can be spatially registered. In principle this can be achieved by multimodal imaging where the same biological source can be spatially isolated and interrogated by different imaging modalities such as light scatter based optical coherence tomography and two-photon fluorescence microscopy (37, 38). The specificity of the scatter signal in multimodal approaches therefore requires that the scattering signal within a given resolution element be dominated by scattering by a single biological source, and that this specific biological source be positively identified by a biologically specific modality such as fluorescence.

5.3 Biological specificity acquired by training data

Light scattering information can be utilized to differentiate and classify a finite set of samples comprising different, but well defined (e.g. normal, dysplastic), groups of unlabeled cells or tissues with great sensitivity and specificity. Multimodal approaches are also valuable in this context by providing multi-parametric data that can improve diagnosis and tissue classification (39). However, the specificity of the light scattering measurements made in these classification contexts is “constrained” to the extent that it is still limited to the

context in which the experiments are undertaken. In particular, it is not clear whether the same positive scattering changes will always be unequivocally indicative of the biological transformations under study, or whether other physiological states can result in similar effects in a set of unknown samples. In cell biology experiments, specificity may be addressed by demonstrating that the measured positive scattering signal can be modulated or blocked by molecular or biochemical experimental treatments that affect a specific signaling pathway. Nonetheless, in cases where biochemical modulation of the scatter signal is not possible, such as in signals collected in vivo from patients, extensive validation studies are required to exclude unrelated processes that could potentially give rise to the same scattering changes. To this end, some investigators have begun to address this issue by testing sample conditions that fall outside of the initially sought class definitions (negative controls). For cancer diagnosis, conditions such as age, gender, race, demographic factors or smoking history that could confound the light scattering data were for example considered and ruled out (40). With adequate and sufficient training data, this approach could ultimately lead to the ability to screen for specific biological conditions in unknown tissue samples. While at present these optical scatter classification studies do not provide a comprehensive data set encompassing all possible cellular and tissue states, these studies are still very valuable in providing the positive scattering responses of biological conditions such as cancer. As noted above, these positive responses can be used as an initial screen to identify the presence of biological change within a given sample and guide further tissue analysis.

6. Light scatter spectroscopy to assess tissues in vivo

Imaging and spectroscopic light scattering techniques have the potential to complement current diagnostic tools. Elastic scattering can be performed *in vivo*. The obtained information is quantitative. Finally, information about cell organization at scales as small as few tens of nanometers may potentially be obtained (8, 9, 11). Visualization of such small objects with far-field optical microscopy is difficult or impossible, because the required resolution is below the diffraction limit. Light scattering spectroscopy, on the other hand, enables obtaining such information, because it does not attempt to *visualize* these small objects. Whereas the resolution of a light microscope is limited to ~$\lambda/2$, with λ~400-700 nm the wavelength of light, characteristic spectral features of light scattered by particles as small as $\lambda/20$ can be distinguished (8, 11, 13, 14).

Most biomedical applications of light scattering rely either on light transport in macroscopic tissue (e.g., elastic scattering spectroscopy) or light interaction with localized microscopic volumes of biological media (e.g., light scattering microscopy modalities). From the perspective of “macroscopic” applications, light scattering has been employed to obtain insights into tissue structure. A common strategy (pursued either explicitly or inexplicitly) is to measure the distribution of light scattering intensity as a function of either wavelength or the scattering angle (Fig. 4), translate these measures into specific optical properties of tissue, and finally, link these optical parameters to the tissue structural properties. This latter step is dependent upon the availability of the robust models of light scattering in tissue that enable solving the “inverse scattering problem”. Although the origin of light scattering in tissue has not been fully elucidated, substantial progress has recently been achieved (see Section 3). Some of the key optical properties that can be measured are the reduced scattering coefficient, μ_s' (i.e. the total scattering cross section per unit volume) and the characteristics of the scattering phase function (such as the anisotropy factor, g, the relative portion of side scattering, backscattering cross section, or the phase function for a given angular range).

Elastic scattering spectroscopy was used for tissue diagnosis and showed clear differences among healthy and precancerous colonic tissues (41, 42). Diffuse optical tomography (DOT)

to measure scattering properties of tissue (μ_S' in particular) showed that it can be used as a marker of malignancy (43-48). The angular and spectral properties of scattered light were measured by low-coherence interferometry to quantify one of the most ubiquitous markers of neoplasia, increased cell nuclear size (36, 49, 50). Light scattering methods were developed to monitor changes in mitochondria and lysosomes in response to photodynamic therapy (35, 51-53), and to investigate changes in mitochondria within cancer cells (54). The size distributions of cellular organelles in live cells were also measured with light scattering to distinguish different types of immune cells (55). A multimodal study utilizing light scattering and autofluorescence showed alterations in subcellular organization yielding larger and less organized subcellular membrane units that were accompanied by a more diffuse subcellular distribution of tryptophan, and a decreased redox ratio in normal keratynocytes compared with their human pappilloma virus transformed counterparts (56). Polarized light scattering spectroscopy was utilized to quantify the fractal properties of bioengineered tissue (57). Low-coherence enhanced backscattering (LEBS) was developed to measure depth-resolved optical properties of tissue (μ_S', g, and the index of the refractive index correlation function m; see Section 3 for detail) (58-61).

Enhanced backscattering originates from the constructive interference of photons traveling time-reversed paths in a scattering medium. This results in a peak of scattered intensity centered around the backscattering direction (Fig. 4). The width of this peak is proportional to $\lambda\mu_S'$, which, in weakly scattering media such as biological tissue, is extremely small (~0.001°). The difficulty involved in resolving these sharp peaks can be avoided by recording enhanced backscattering under low spatially coherent illumination (the spatial coherence length $Lsc \ll 1/\mu_s'$), which broadens the peak by as much as two orders of magnitude (58-62). Low spatial coherence serves as a spatial filter effectively limiting the area of light collection to $\sim Lsc^2$, and the LEBS signal is primarily due to the interference of short traveling photons that emerge from the point of entry into the tissue at distances less than Lsc. This feature enables depth-resolved assessment of tissue that can be achieved by analyzing the angular profile of a LEBS peak, its Fourier transform that contains information about a range of tissue depths, or by selecting Lsc to target a desired depth. For a given Lsc, the penetration depth is $\propto Lsc^{2/3}/\mu_s'^{1/3}$. The angular profile of LEBS depends on the tissue microarchitecture through the refractive index correlation function $C(r)$. For instance, in the three-parametric approximation for $C(r)$, $C(r)$ can be experimentally characterized if three LEBS parameters are measured, such as its amplitude (the enhancement factor), peak width, and the power of the spectral dependence of the enhancement factor, respectively in Fig. 4, with all these measurements being depth-selective. Applications of LEBS demonstrated that alterations in tissue microarchitecture are a ubiquitous phenomenon that develops in the earliest stage of carcinogenesis in a number of cancer types (63, 64). This is just one of many examples of how new light scattering techniques can lead to new frontiers in biology.

7. Imaging Based on Optical Scatter Contrast

Several well-known microscopic visualization techniques have already been developed that derive contrast from optical scatter interactions with biological tissue. These include Zernicke phase contrast, differential interference contrast and dark-field microscopy. These methods enable the visualization of non-absorbing biological specimens based on spatial variations in optical phase across the sample. Thus, one of the main advantages of optical scatter contrast is that it can be used non-invasively to image unlabeled tissue. This is especially relevant in clinical applications in which the rapid, safe and low-cost assessment of unstained tissue in vivo is highly desirable. However, to enable imaging of thick tissues in vivo, an important requirement is that the optical instrumentation allows viewing individual tissue planes with sufficient axial resolution as a function of tissue depth. Two methods are

described here that rely on different optical principles to achieve non-invasive “optical sectioning” and three-dimensional imaging in vivo. The first method utilizes confocal microscopy principles to reject out-of-focus light. The second utilizes temporal coherence gating. The image signal in both methods is given by the local reflectance of the tissue. Both methods ultimately produce a spatial tissue map depicting tissue architecture and the spatial distribution of the different scattering tissue structures. Due to their optical sectioning ability and out-of-focus background rejection, these imaging modalities provide tissue images in vivo with microscopic-scale resolution.

In addition, methods that combine light scattering spectroscopy with microscopy as well as quantitative phase imaging methods have been developed to extract the local optical properties of a microscopic biological sample. Rather than focus solely on sample visualization, the goal of these methods is to create images that directly encode quantitative light scattering data and where the value of each pixel can provide direct information about the local scattering sources' size, shape and refractive index.

7.1 In vivo confocal reflectance microscopy

Confocal microscopy (65) is a widely used method for in vitro biological imaging with commercially available bench-top instruments since the 1990's. Scanning confocal microscopy is typically used for fluorescence imaging in biological applications. However, the method can also be applied to reject out of focus light and create optical sections in a backscattering reflectance regime in systems that allow partial transmission and reflection of the light to and from the sample at the same wavelength. Since fluorescent dyes cannot always be used in vivo, clinical imaging based on confocal reflectance microscopy can provide a powerful non-invasive method to visualize tissue structure for diagnostic purposes, and portable clinical systems are now available (e.g. Lucid, Inc. confocal imagers). Typically these systems employ an epi-illumination configuration with high numerical aperture (NA) objectives. The lateral resolution of the system is set by the NA of the objective used, while the axial resolution is determined by the size of an adjustable detector pinhole similar to that in a bench-top confocal microscope (66). Resolutions of 0.5-1 μm laterally and 3-5 μm axially were achieved at 350 μm depth in human skin with 1064 nm light (18). A recent research thrust in this area has been the miniaturization of confocal microscopes used for in vivo applications. Simplified scanning methods, such as line scanning, are being investigated to this end and can greatly simplify microscope design without significantly compromising resolution (67). Fiber-optics based reflectance confocal microscopes for “endomicroscopy” are also a significant thrust (68).

A spectroscopic extension of confocal reflectance microscopy was recently developed to collect at each scanned point a scattering spectrum which is processed to yield a measure of local particle size or shape at each voxel (69). The scattering spectrum is obtained by illuminating the sample with a focused broadband spot. Passing the light scattered by the sample into a spectrometer retrieves a scattering spectrum of the particles contained within the analyzed spot. A model fit to the spectroscopic data that utilizes either Mie theory (3) or the T-matrix method (70), retrieves the size, aspect ratio and refractive index of the sample at each scanned location. This method, Confocal Light Absorption and Scattering Spectroscopy (CLASS), was demonstrated by measuring alterations in the size of organelles within living cells treated with docosahexaenoic acid (DHA) (8).

7.2 Optical coherence tomography and microscopy (OCT/OCM) and spectroscopic OCT/OCM

Optical coherence tomography (OCT) is a biomedical imaging modality that is increasingly being used for clinical imaging, primarily in the areas of ophthalmology, cardiology,

dermatology, and oncology (71-74). OCT performs imaging by interferometrically detecting singly backscattered light to render a depth-resolved image of biological tissues. Most commonly, OCT utilizes fairly low-numerical aperture optics when focusing light into specimens, so as to have a long confocal parameter that approximates the depth at which OCT can image in highly scattering tissue (typically < 2 mm), thereby maintaining a relatively uniform transverse resolution throughout the imaging depth. This long confocal parameter, however, comes at the expense of lower transverse resolution. Optical coherence microscopy (OCM) (75, 76) similarly utilizes interferometric detection, but with higher numerical aperture optics to capture *en face* optical sections at various depths, as performed in reflectance confocal microscopy or most all other high-resolution optical microscopic imaging techniques. The combined use of confocal-gated optical sectioning with coherence-gated optical ranging permits OCM to depths in highly-scattering tissues that can exceed those of standard reflectance confocal microscopy (76).

Spectroscopic OCT or OCM (SOCT/SOCM) was developed out of OCT/OCM to take advantage of the broad spectroscopic content of the light used in these techniques (77-79). The SOCT technique enables spectroscopic analysis on the backscattered signals to extract information about the scatterers and absorbers in tissue, providing new metrics for functional and molecular imaging. Advances in SOCT/SOCM have enabled discrimination of endogenous tissues and cell types, as well as the detection of exogenous contrast agents (75, 80, 81). A full theoretical and applied description of SOCT/SOCM can be found in an earlier review (78), however, the SOCT principles, and the duality between time-domain SOCT and frequency-domain SOCT are illustrated in Fig. 5.

Optical systems that utilize low NA and/or spatially incoherent beams, such as in some LSS techniques, generally average the spectroscopic response of the sample over a large number of scatterers. In OCT, however, moderate focusing is commonly used, which results in a small number of scatterers contributing to the response. Optical coherence microscopy (OCM), with high NA optics and tightly focused beams, enables one to resolve individual scatterers, however, the analysis of the backscattering response is complicated by the large angular extent of the beam delivery and collection (82). A demonstration of SOCT-based scatterer sizing was performed on tissue phantoms containing known sizes of microspheres (78). Qualitatively, there was good agreement between the SOCT-recovered spectra and predictions based on the vector Mie theory. Most importantly, the apparent modulation frequency (or pitch) of the spectra, which is directly proportional to scatterer size, was well-matched with theory.

The ability of spectroscopic OCT/OCM analysis for analyzing scattering particle size can be extended to tissue specimens, and even to individual cells (75, 78). Using a spectral-domain OCM system, spectroscopic OCM was performed on a tissue specimen consisting of regions of adipose and muscle tissue Fig. 5 **(top right set)**. In this example, two different SOCM analysis methods were performed. The first method was based on metameric imaging, where the scattering spectrum with the FWHM of the source spectrum was divided into three equally-spaced subbands, and the intensity from the low-, mid-, and high-frequency bands were assigned to the red, green, and blue channels, respectively (Fig. 5, top right (c)). This method was found to represent the spectral information in a similar manner as the traditional spectral centroid method, where shifts in the spectral centroid are mapped on to colors in the image. This metameric method was found to be more robust, and more similar to the mechanism in human vision. The second method was based on spectral analysis initially proposed in light scattering spectroscopy (LSS) (83). The backscattered spectra are first analyzed by the FFT, and the first peak of the FFT data is used for hue information in a hue-saturation-value (HSV) color scale (Fig. 5, top right (d)). This peak position is related to the physical size and interscatterer distance of the dominant scatterers, such as nuclei. These two

SOCM spectral analysis methods are only representative of many that can be developed. They do, however, provide examples of how the spectroscopic analysis of scattered light can be used to provide and enhance contrast in optical images, in physically meaningful and quantitative ways.

As nuclei are known to be dominant scatterers within cells and tissues, SOCM was used to image fibroblasts in culture to demonstrate the use of these spectral analysis methods at the cellular and sub-cellular level. A novel custom integrated microscope which enables simultaneous acquisition and registration of data from OCM and multiphoton microscopy techniques was used (75, 84). The OCM images provide information on the tissue and cell structures based on optical scattering, while the multiphoton microscopy images provide information on the function, related by either the autofluorescence from the metabolic processes, or by functionally-labeled genetically-expressed fluorescent proteins (i.e., GFP co-expressed with vinculin, a cell-surface adhesion protein) (85). Fig. 5 **(bottom right set)** shows images captured with this system, including SOCM images generated using the LSS spectral analysis method. The transfected fibroblasts expressed GFP-labeled vinculin, and were co-labeled with a DNA-specific dye (Hoechst 33342) for localization of nuclei relative to the surrounding cell structures. The simultaneous multimodality imaging afforded by this microscope provided overlays of various image channels, and the co-localization of the dye-labeled nuclei and the SOCM-identified dominant scatterers within the cell culture.

Spectroscopic OCM analysis with tight focal gating serves to decouple the inherent trade-off between spectral and spatial (depth) resolution. This enables more precise extraction of spectral changes and signatures that can be used to characterize the wavelength-dependent scattering properties of cells and tissues. Wavelength-dependent scattering and the resulting spectral modulation are information-rich processes that can be readily used to extract quantitative information about scatterer size and spacing, as well as provide novel contrast mechanisms in images. These OCT-based methods offer advantages over reflectance confocal microscopy by enabling deeper imaging penetration into highly scatter tissues, due to the added coherence-gating, and the ability to perform depth-resolved spectral analysis to enhance the information content of the images.

7.3 Optical scatter imaging based on Fourier filtering

Collecting far-field angular scatter data by analyzing the optical transform of biological specimens dates back several decades (86). The far-field angular scattering data may be imaged directly and was recently utilized for automated differentiation of bacterial colonies (87). The scatter intensity pattern may also be collected in a conjugate Fourier plane (transform plane) of a microscope system to extract the angular scattering properties of the scattering sources in the field of view (88). A recent multimodal microscopic design permits imaging the sample's transform in combination with brightfield, dark-field, and fluorescence imaging as well as collection of a wavelength dependent light scattering spectrum of the sample (52). In the transform plane of a microscopic imaging system, the scattering intensity is mapped as a function of the polar scattering angle, θ, with respect to the incident light, and azimuthal angle φ as alluded to in Fig. 2 (Optical Fourier processing inset). For weakly scattering objects the angular scatter directions mapped to positions (θ,φ) in the transform plane correspond to the object's spatial frequencies $(U,V) = (\kappa \sin\theta, \kappa \sin\varphi)$, with $\kappa = n/\lambda$ where n is the refractive index of the propagation medium (21).

Unless the incident light is directed to a specific location within the sample, the scattered light measured in the transform plane of the microscope corresponds to scattering by all the scattering sources within the field of view. However, by collecting Fourier filtered images that are formed by allowing only specific angles of scatter to pass, the angular scatter information may be remapped in the imaging plane. In this case multiple Fourier filtered

images are analyzed to extract the sample's angular scattering properties at each pixel within the full field of view. This method was first implemented using two circular Fourier filters designed to measure the intensity ratio of wide to narrow angle scatter. This ratio was shown to decrease monotonically with particles size (6) and was shown to track dynamic cellular changes during calcium injury (89) and apoptosis (33). The scatter changes were modulated by biochemical and genetic modifications affecting mitochondria (34, 89). Recent implementation of a digital micro-mirror device as a Fourier filter allows filtering any desired combination of scattering angles (or spatial frequencies) within the numerical aperture of the microscope (90). The method was demonstrated using tunable Gabor-like spatial filters, with Gaussian envelopes, which provide signal localization in both Fourier and image space (91). The Gabor filtering method is sensitive to differences in particle size on the order of 30-50 nm (92), and to changes in particle orientation and aspect ratio (90). Alterations in organelle aspect ratio were measured in isogenic Bax/Bak expressing, apoptosis-competent cells (W2), and Bax/Bak null, apoptosis-defective cells (D3) (7). As expected the Bax/Bak null cells, which are expected to be deficient for mitochondrial fusion and have shorter mitochondria (93), also had a larger number of particles with lower mean aspect ratio as measured by the optical Gabor filtering method (Fig. 6).

While based on elastic scattering contrast, the Gabor-filter based technique does not rely on a detailed inverse scattering model or on Mie theory to extract morphometric measurements (92). This technique is therefore applicable to non-spherical organelles for which a precise theoretical scatter description is not easily given, and provides distinctive morphometric parameters that can be obtained within unstained living cells to assess their function. The technique is advantageous compared with digital image processing in that it operates directly on the object's field transform rather than the discretized image of the object. Thus, it does not rely on high image sampling rates. High sensitivity at low resolution was demonstrated using aliased images of microspheres (92). The cell data is also shown after block-processing the initial Gabor-filtered images such that every adjacent 4×4 pixel region is averaged into one pixel before orientation processing, effectively demagnifying the filtered images from 512*512 to 128*128 pixels, or approximately 1 μm^2/pixel (Fig. 6D&E and (7)). Although the pixels in the resulting images are large, they still demonstrate orientation detection substantially similar to that of the high magnification (Fig. 6G). A real 4× demagnification corresponds to imaging 300-500 hundred cells instead of 20-30 at comparable seeding densities. Thus this method can ultimately be applied to high throughput single cell analysis within relatively large fields of view.

In the applications described here, the sample illumination usually consists of a collimated beam of light aligned with the microscope axis. Collection of scattered light by a microscope objective with a collimated goniometric illumination assembly has also been proposed (94). In contrast, a recent application in which the incident light is focused to a 7 μm spot size was described. Angular elastic scatter data originating from a single cell was collected in a conjugate Fourier plane of the microscopy system, while the scattered light was also analyzed spectrally to yield a Raman spectrum of the same cell. Theoretical formalisms based on prediction of angular scatter of a focused beam are used to extract structural information, which is combined with chemical data based on Raman scattering (32). This method has recently been applied to the study of immune cells with the potential for future monitoring of developmental processes in these cells (55), and provides a basis for analyzing angular back-scatter data from samples illuminated by a focused laser beam.

7.4 Quantitative phase microscopy

Quantifying the optical phase and scattering properties of the sample is difficult to achieve based directly on the intensity of images acquired by conventional phase contrast or differential interference contrast methods. Thus, interferometric methods based on the

principles of holography (95) are used for quantitative phase imaging. With advances in CCD camera design and performance, it was recently shown that holograms with sufficient spatial resolution can be recorded digitally on a CCD camera (96). This allows the digital reconstruction of the object's field amplitude and phase by numerical computation (97). This approach, now widely known as digital holographic microscopy, has gained a significant amount of interest since its first inception. For a full review of digital holography methods the reader is referred to (98) and a dedicated journal issue in (99).

Different interferometric imaging systems and methods for object reconstruction have also been proposed for quantitative phase imaging (100, 101). Interferometric quantitative phase microscopy is especially attractive for biological imaging as it provides highly sensitive and high resolution quantitative phase maps of non-absorbing biological samples. The quantitative phase information can be used to study cell function dynamics, subcellular structure and organelle distribution with very high phase and temporal sensitivity (102). In general, quantitative phase microscopy methods provide an integrated measurement of the optical phase throughout the axial thickness of the sample. The refractive index and sample thickness are therefore intertwined within the measurement of the optical path length. However, refractive index tomographic imaging of cells and tissues has been recently demonstrated (20, 103).

Another innovative approach to extract subcellular refractive index fluctuations has recently been proposed. Partial Wave Spectroscopy utilizes measurements of the backscattering spectrum to assess refractive index fluctuations in epithelial cells (9). In this case the sample is illuminated by a broadband source and a low numerical aperture objective to get light scattering spectra at each resolution element. Spectral fluctuations in light scattering are analyzed to extract a measure of refractive index fluctuations as a function of depth at each subcellular location. Partial Wave Spectroscopy captures the axial fluctuations in subcellular structure into a single parameter, which gives a measure of disorder strength over a sample depth on the order of several microns. While the size of the resolution element is determined by the numerical aperture and depth of field of the objective, the scattering data reflects structural fluctuations on the order of 10 to 100 nanometers. Recently this method was shown to reproducibly detect nanoscale subcellular structural changes that were associated with genetic changes induced by the field effect of carcinogenesis (40).

Finally, the scattering properties of a biological sample also depend on the polarization of the incident light and birefringence properties of different molecular constituents (104). Quantitative polarization microscopy has been developed to measure retardance and molecular orientation within biological tissues (105). The method provides diffraction limited microscopic resolution and reports on highly oriented biological structures, such as actin (106), microtubules (107) and collagen (108).

8. Computational methods in optical scatter imaging and spectroscopy

Computational methods are essential for quantitative optical scatter imaging and spectroscopy both for modeling and interpreting the light scatter data, and for image reconstruction. Two current examples in computational approaches are presented here: 1) applications of the finite difference time domain method to infer the scattering properties of biological samples, and 2) interferometric synthetic aperture microscopy for improvement of three-dimensional imaging.

8.1 Finite-difference time-domain methods

To various degrees, the current modeling methods used in tissue optics neglect the full-vector electromagnetic wave nature of light, especially with regard to near-field interactions

of closely spaced scatterers. Furthermore, recent progress in developing realistic light scattering models to interpret the wealth of information hidden in light scattering data (see Section 3) depends on the ability to validate these models. While experimental validation remains difficult (this would require knowing the spatial distribution of refractive index in tissue with nanoscale precision), the focus has shifted towards numerical techniques. The most rigorous approach is by solving Maxwell's equations. This can be accomplished by use of the finite-difference time-domain (FDTD) method. Originally developed for military applications, FDTD is now finding new applications in tissue optics.

With currently available computers, the FDTD numerical method for solving Maxwell's equations permits straightforward numerical modeling of full-vector electromagnetic wave interactions with inhomogeneous material structures spanning approximately 1000 wavelengths or less in two dimensions, and 100 wavelengths or less in three dimensions. This capability is appropriate for calculating optical interactions (wavelengths ~ 0.4-1 μm) with structures within biological tissue (characteristic dimensions ~ 0.001 – 50 μm). Calculations of interest include the internal optical electric field standing-wave distribution, the differential (bistatic) scattering cross-section, and the total scattering cross-section. Early applications of FDTD included modeling of light interaction with retinal cells (109) and phenomenological studies of the role of organelles in light scattering by cells, where cells were modeled as consisting of randomly distributed, isolated, nonspherical scattering particles (110-113). More recently, FDTD was used to study light scattering in media with spatially continuous refractive index variations as well as the light-scattering behavior of randomly inhomogeneous and nonspherical structures, typical for tissue optics (10, 11, 13, 14, 17, 114, 115).

Although, in principle, FDTD could be used to model light interaction with tissue spanning macroscopic dimensions, this has not been feasible because even the most powerful of the existing computers lack the capabilities to deal with the enormous database of electromagnetic field vector components mandated for FDTD. To attack this problem, pseudo-spectral time-domain (PSTD) numerical algorithm to solve the full-vector Maxwell's equations was developed. PSTD is an innovative variation of FDTD wherein the spatial derivatives in Maxwell's curl operators are implemented using fast Fourier transforms. This yields “infinite-order accuracy,” i.e., essentially exact results for electromagnetic field spatial modes sampled according to the Nyquist criterion. That is, the PSTD meshing density can approach two samples per wavelength in each spatial dimension which is about an order-of-magnitude reduction from the meshing density required by FDTD. This yields an enormous decrease of computer resources relative to FDTD, and thereby permits using PSTD to directly model realistic three dimensional tissue regions having spans as large as hundreds of microns to millimeters. Within such a volume, PSTD can provide the complete near- and far-field interactions of hundreds, even thousands of arbitrarily complex biological structures with no simplifications other than the sub-wavelength discretization of the refractive index distribution (116-118). In particular, PSTD has been used to model light transport in media with the overall sizes up to a few hundreds of microns.

At present time, the major limitation of FDTD and PSTD is the computer resources that these techniques require. While FDTD modeling of light interaction with a micron-sized three-dimensional object can be performed on almost any personal computer, a 10 micron particle would require a small computer cluster. PSTD modeling of macroscopic (millimeter scale) media is only possible with the help of state-of-the-art supercomputers. However, with more and more powerful computers and computer clusters made available, the progress in FDTD and PSTD and their application to tissue optics are expected to increase dramatically.

8.2 Interferometric synthetic aperture microscopy (ISAM)

Interferometric synthetic aperture microscopy (ISAM) (119-122) is a novel computed imaging technology that is based on the coherence ranging and imaging principles of OCT, and the use of synthetic aperture principles from synthetic aperture radar (SAR) (123). ISAM uses computational imaging to overcome the trade-off between depth of focus and transverse resolution. By accurately modeling the scattering processes and the data collection system, including the defocusing ignored by OCT image formation, the scattering properties of the object can be quantitatively estimated from the collected data. As in SAR, diffraction-limited resolution is achieved throughout the final image. For both ISAM and SAR, the key to this performance is the coherent collection of a complex data set.

Interferometric microscopes, such as OCT systems, provide holographic data. That is, the phase of the backscattered light can be recovered from the raw data. This is a significant advantage over standard non-interferometric system where the phase information is lost at detection. This holographic data collection is analogous to the coherent data collection used in SAR systems. In both ISAM and SAR, the collection of complex coherent data allows the numerical implementation of advantageous operations that would be prohibitively difficult to implement physically. In SAR, the multiple along-track range profiles collected from a small aperture can be used to synthesize an aperture corresponding to the whole along-track path. One can envision the optical analogue to this when considering a focused Gaussian beam that is transversely scanned across a point scatterer. The Gaussian beam wavefront can be decomposed into a series of plane waves, each at a different incident angle. The point scatterer will be illuminated by each of these plane waves, and from many different angles, as the focused Gaussian beam is translated across the point scatterer. In ISAM, these multiple complex OCT range profiles are collected, and can be computationally reconstructed so that all planes therefore appear simultaneously in-focus. The blurred out-of-focus regions seen in OCT can therefore be brought into focus numerically (124). Fig. 7 **(top set)** shows three-dimensional OCT and ISAM images of point-like titanium dioxide scatterers in a three-dimensional tissue phantom, where the ISAM reconstruction clearly shows the spatially-invariant focus correction throughout the volume (120).

Since the inception of ISAM, there have been studies investigating numerous practical aspects for its implementation and use. ISAM algorithms have been adapted for use not only with the Cartesian-coordinate imaging common in beam-scanned systems (120), but also for radial-coordinate imaging found in optical catheter-based imaging systems (125). ISAM has been applied to full-field optical imaging systems (126), and systems using a spatially-extended partially-coherent optical source (127), which can mitigate multiple-scattering artifacts that can arise in full-field ISAM. ISAM has also been applied to high-numerical aperture imaging systems (128), in contrast to the low-numerical aperture systems most commonly found with OCT.

The practical implementation of ISAM involves not only software algorithms, but also robust and reliable hardware that can ensure phase stability in the acquired optical signals. This phase stability requirement can be met more easily with the faster acquisition capabilities of recent OCT systems employing spectral-domain detection or fast wavelength swept-sources (129, 130). The software for ISAM, also, is amenable to more efficient code that can readily enable real-time processing and display (119). Applied to biological tissue, ISAM can perform the same out-of-focus correction as was shown with the tissue phantoms, as illustrated in Fig. 7 **(bottom set)** for human breast tissue (120).

ISAM is a computed imaging technique that quantitatively estimates the three-dimensional scattering objects imaged with broadband coherent microscopy. The solution of the inverse problem allows for the reconstruction of areas in the sample or specimen that are typically

regarded as out-of-focus and previously unusable. The result of ISAM obviates the perceived trade-off between transverse resolution and depth-of-focus in OCT. While ISAM addresses an inherent weakness in OCT, namely the need to scan the focus axially to obtain images outside of the original focal plane, ISAM is not merely a method to refocus the field computationally. Refocusing may be achieved from a single interferometric image at a fixed frequency, but the resulting image is still inherently two-dimensional, failing to unambiguously distinguish contributions to the image from various depths. As in other ranging technologies, the broadband nature of ISAM allows a true three-dimensional reconstruction.

ISAM and SAR are examples in the broad class of modalities known as computed imaging. Like almost all computed imaging modalities in common practice today, they are based on the solution of linear inverse problems. Linear inversion problems offer advantages such as the option to pre-compute and store the elements of an inversion kernel for rapid computation of images from data. Thus, the methods take advantage (are even reliant on) one of the greatest advances in applied mathematics in the last half-century, the fast Fourier transform, which enables these computed imaging algorithms to run very fast, and be amenable to parallelization.

9. Future and emerging directions

9.1 Contrast agents for optical scatter imaging

The spectroscopic nature and high-resolution imaging capabilities of light provide a means for probing biological structure and function at the cellular and molecular levels. While the use of fluorescent and bioluminescent probes has become a mainstay in optical molecular imaging, a large number of other optical imaging modalities would benefit from optical probes that are based on scattering, absorption, or modulation (131, 132). To this end, advances over the last several years have developed a wide array of contrast agents suited for scattering-based modalities such as OCT and reflectance confocal microscopy. Most generally, these can be grouped as probes based on scattering, modulation, absorption, and plasmon-resonance effects.

Scattering probes are designed for efficient light scattering and are sensed either directly by detecting their scattered light, or indirectly through their attenuation of the incident light. To alter the intensity of the backscattered light in OCT, for example, scattering probes must introduce a local region of index of refraction change. One of the first demonstrations of scattering probes was the use of gold nanoparticles in electron microscopy (133) to label specific cells. In OCT, air- or gas-filled microbubbles (134) or engineered nanoparticle-laden protein-shell microspheres (135) have been used both in cell cultures and as in vivo contrast agents introduced through the vascular circulation. Because small nanoparticles have the ability to extravasate from the vascular system out toward tumor cells, many investigators have developed agents such as nanoshells or structured nanoparticles that can either be passively or actively targeted to site-specific molecular receptors, enhancing scattering-based contrast for detection (81, 136).

Detecting small localized changes in optical scattering induced by a contrast agent or probe is frequently plagued by speckle and background scattering from the tissue medium. Higher probe imaging specificity may therefore be achieved by modifying an observable property of the probe in an external and controlled manner. This was achieved using magnetomotion, where magnetic iron-oxide nanoparticles in cells or tissues are modulated with a small external alternating magnetic field. The modulating nanoparticles, which are too small to exhibit a significant level of optical scattering when stationary, are displaced over several hundreds of nanometers, and alter the local optical scattering properties of their

microenvironment. Using a sensitive amplitude- and phase-resolved scattering-based detection technique such as OCT, these magnetomotive scattering signals can be readily detected against a highly scattering but stationary background. This technique, named magnetomotive OCT (MM-OCT), has been used to image single labeled cells (137) as well as in vivo tissue (138). Advances in phase-resolved and fast-acquisition techniques have improved sensitivity to detect ~2 nM concentrations (139). By applying magnetic fields as step-like functions, temporally transient scattered signals can undergo underdamped oscillations, which when analyzed, can yield quantitative biomechanical viscoelastic properties of the medium (140).

Because Rayleigh scattering theory predicts that the scattering cross-section scales as the sixth power of the particle diameter, it becomes challenging to design contrast agents that are small enough for potential in vivo applications, while providing sufficient scattering cross-section for detection. This has motivated the development and extensive interest in metal nanoparticles that exhibit a collective excitation of electrons at a characteristic light frequency, a property called the surface plasmon resonance (141). At optical wavelengths, this is primarily observable in silver and gold, the latter being particularly useful in the near-infrared wavelength region. A wide variety of surface plasmon resonance nanoparticle geometries have been investigated, including nanospheres (142), nanoshells (143), nanorods (144), and nanocages (145). These nanoparticles afford a large optical cross-section while maintaining a small physical size to enable high tissue mobility. Ongoing work is developing methods to site-specifically target these nanoparticles to molecular receptors, as well as use their photothermal properties to damage or kill cells for therapeutic as well as diagnostic applications.

9.2 Biological Dynamics

Dynamic time-dependent changes in optical scattering can provide a wealth of information about biological processes, in both healthy and disease states. Dynamic changes in optical scattering can be used to enhance optical contrast in images, as discussed for magnetomotive optical scattering probes (Section 9.1). In addition, there are inherent dynamic biological processes that will vary optical scattering within cells and tissues.

Apoptosis—Monitoring mechanisms of programmed cell death (apoptosis) is an area of active research, as light scattering methods could ultimately be used for rapid screening of chemotherapeutic agents which act by inducing apoptosis in cancer cell. Early scattering changes within the first hour of apoptosis were shown to be consistent with an increase in the average size of subcelullar scatterers and could be modulated by Bcl-x_L, an anti-apoptosis protein that targets mitochondria (6, 33). Forward scattering spectra and phase functions of cultured cells were also shown to change within the first hour of apoptosis in cells treated with staurosporine (42). Demonstration of scattering changes concomitant with cytochrome c release during apoptosis of photodynamically treated cells further suggest that altered light scattering by mitochondria is a significant measurable signal during apoptosis (35). Cells treated with the chemotherapeutic agent paclitaxel exhibit an increase in mass fractal dimension, which correlates with an increase in subcellular graininess or texture, within the first 1-3 hours of apoptosis and again at 12 and 24 hours compared to the initial time point. The scattering changes at 1-2 hours correlated with mitochondrial alterations assessed by fluorescence; changes after 6 hours with nuclear alterations (36). Elucidating these scattering changes may ultimately lead to optical scatter based apoptosis assays that can track the specific time course of dynamic subcellular responses and aid in the investigation and discovery of apoptosis inducers.

Neuronal depolarization: Another important dynamic process investigated by light scattering is neural activity, either in single neurons, or more generally in neural tissue. Studies using low-coherence interferometry and OCT have shown optical scattering changes in nerve fibers during the propagation of action potentials, which is understood to be the result of changes in nerve fiber swelling from water and ion redistribution (146, 147). Transient intrinsic scattering changes using optical coherence imaging have also been recorded from single *Aplysia californica* neurons during the firing and propagation of action potentials (148). At this scale, optical scattering changes are believed to be the result of rearrangement of membrane dipoles, producing the localized change in index of refraction (149). It has also been suggested that polarization-based scattering measurements may offer a higher sensitivity to these changes (150). Ongoing and new studies are needed to better define the biological origins and mechanisms that can be elucidated using these dynamic scattering signals for not only further insight into neural signaling, but also for many other basic biological processes involving cell signaling, metabolism, and responses.

10. Conclusion

Light scattering spectroscopy and microscopy methods provide very high sensitivity to subtle alterations in the structure and function of healthy and diseased biological tissues. The structural data obtained by optical scatter contrast can be combined with other optical modalities to characterize cellular state and structure-function relationships in cells and organelles. Furthermore, light scattering signals can report on specific tissue time responses that can be used to monitor the effectiveness of therapeutic agents, understand metabolism and cellular dynamics in responses to physiological and pathophysiological stimuli. Recent advances in computational methods and novel molecular probes based on optical scatter contrast have demonstrated improved optical data interpretation and image resolution. This field is advancing rapidly and promises to bring novel and effective cellular analysis and diagnostic assays.

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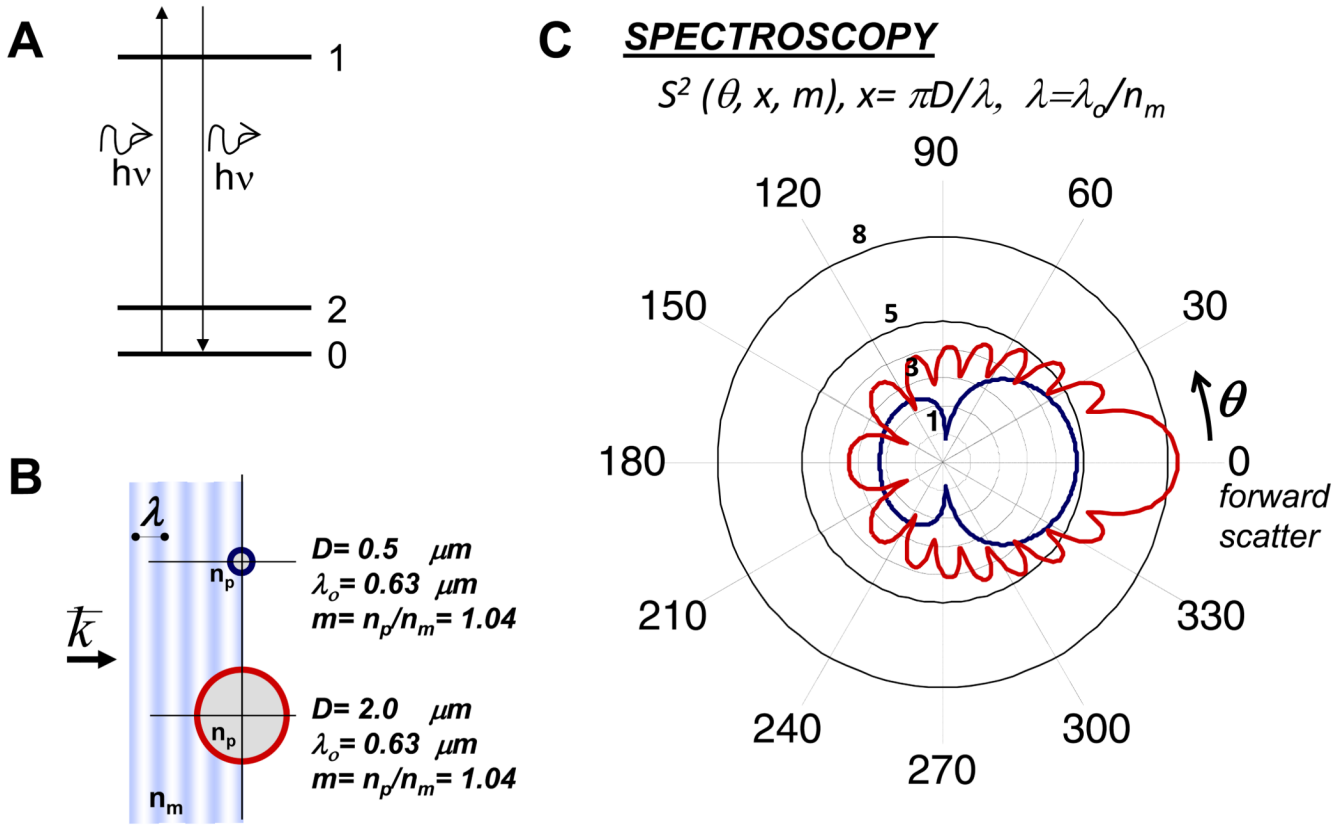

Figure 1.

Optical scatter spectroscopy. A. Energy diagram illustration of light-matter interaction based on elastic light scattering. B&C: Elastically scatterd light may be analyzed as a function of wavelength and scattering angle to extract information about particle size and refractive index. The blue scattergram corresponds to a numerical simulation of light scattering by the small blue sphere in panel B; the red scattergram to the larger red sphere in panel B. A plane wave with wavelength $\lambda=\lambda_o/n_m$ is incident on the spheres which have refractive index $n_p=1.38$ and are located in a medium with refractive index $n_m=1.33$.

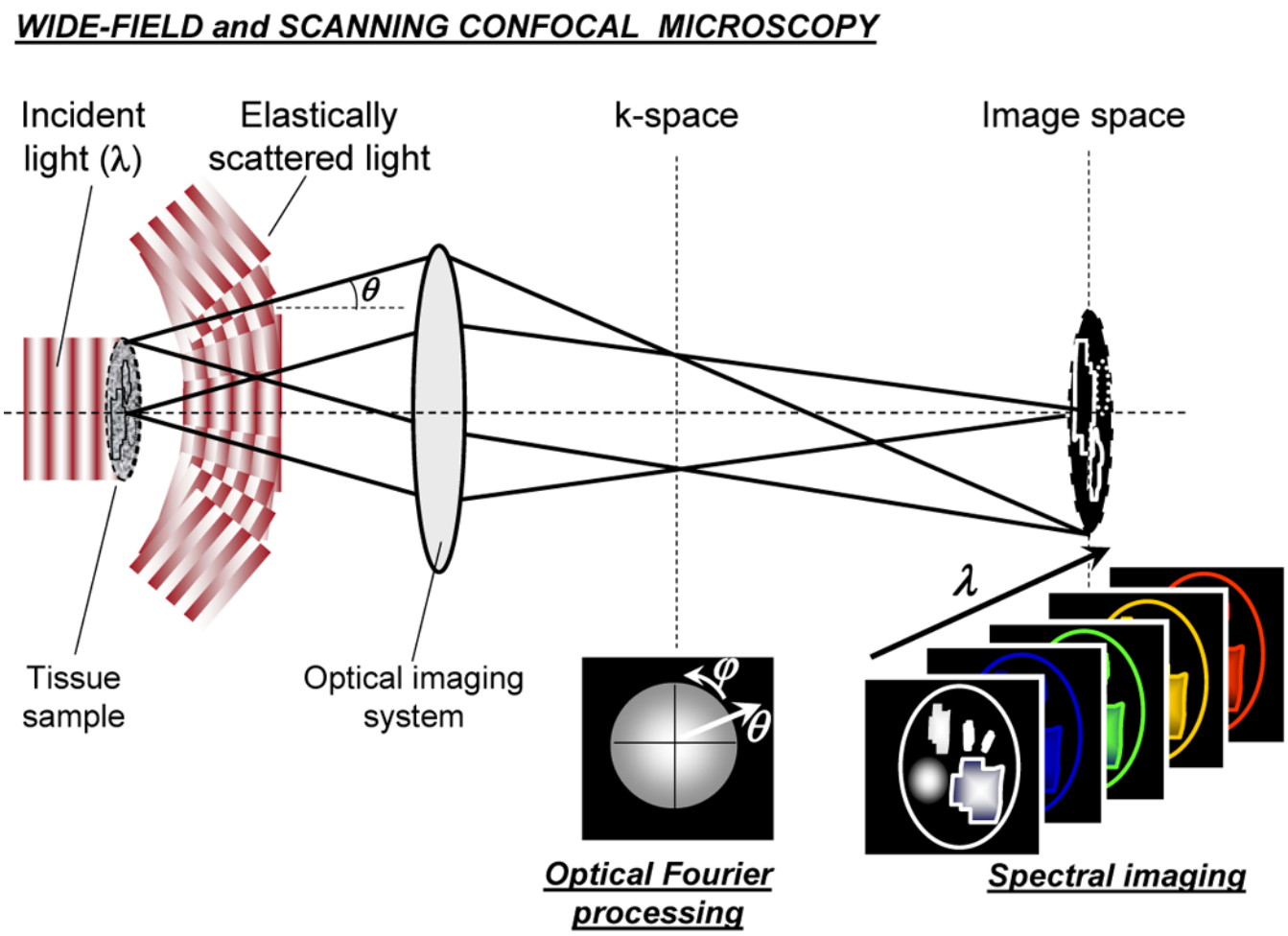

Figure 2.
Elastic light scattering measurements collected in a transmission imaging configuration. Similar data may be collected in a reflectance configuration.

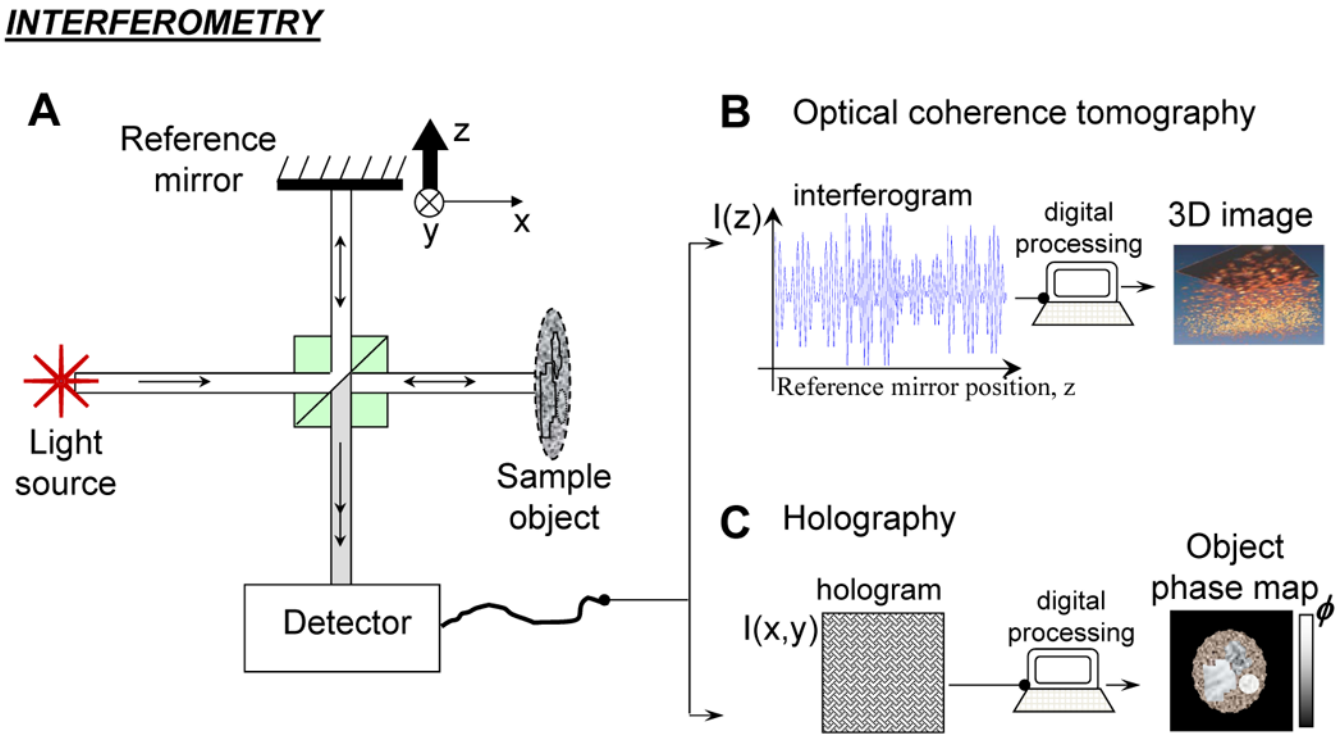

Figure 3.

A: Phase sensitive light scattering measurements collected in an interferometer. B: Low coherence interferometry utilizing a broad band source can be used to collect depth resolved reflectance data I(z) which can be processed to obtain three-dimensional images of the sample (optical coherence tomorgraphy). C: Holograms of the sample object, I(x,y), may be collected and digitally processed to obtain quantitative phase images.

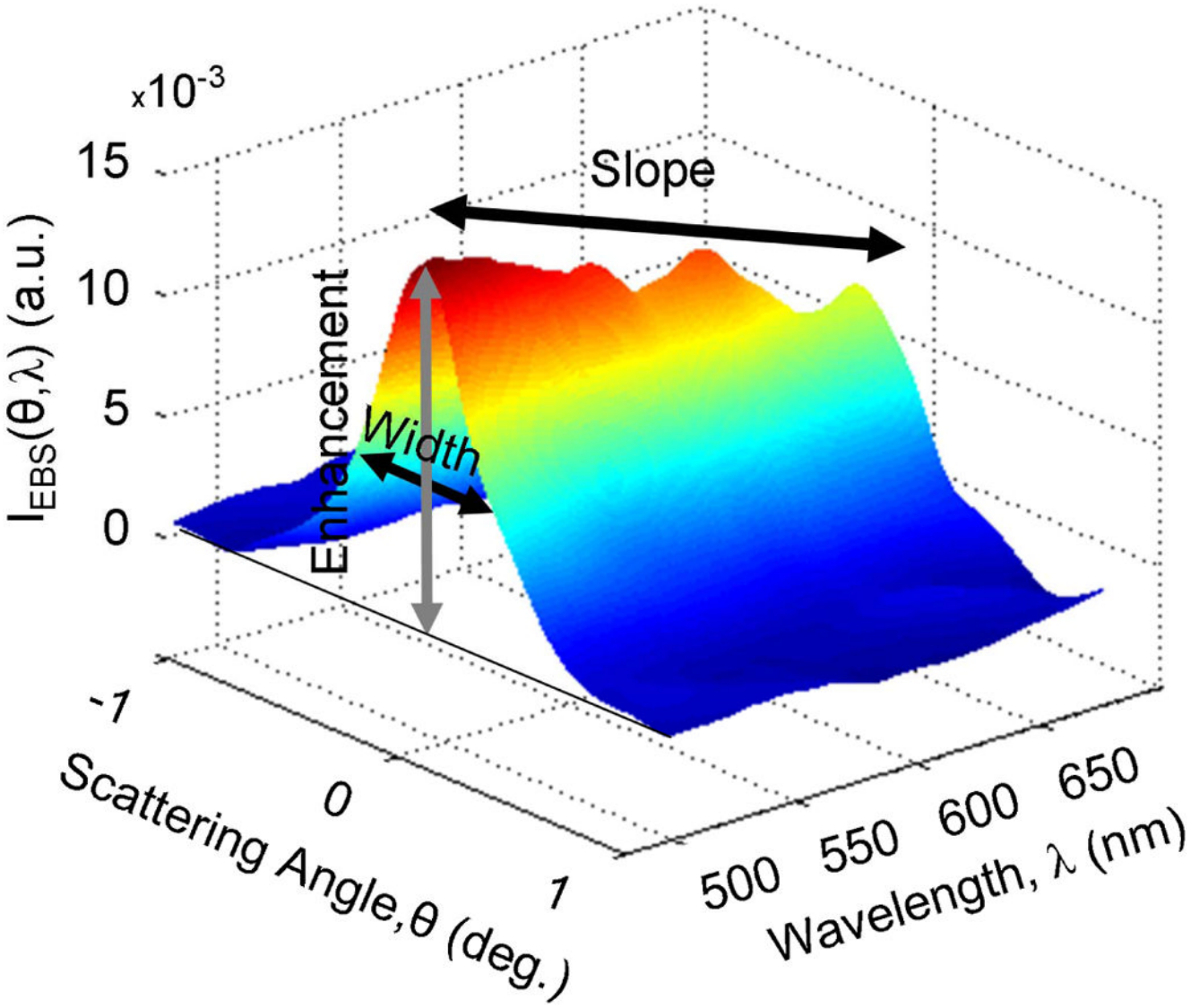

Figure 4.
LEBS recorded from human colonic mucosa as a function of angle and wavelength.

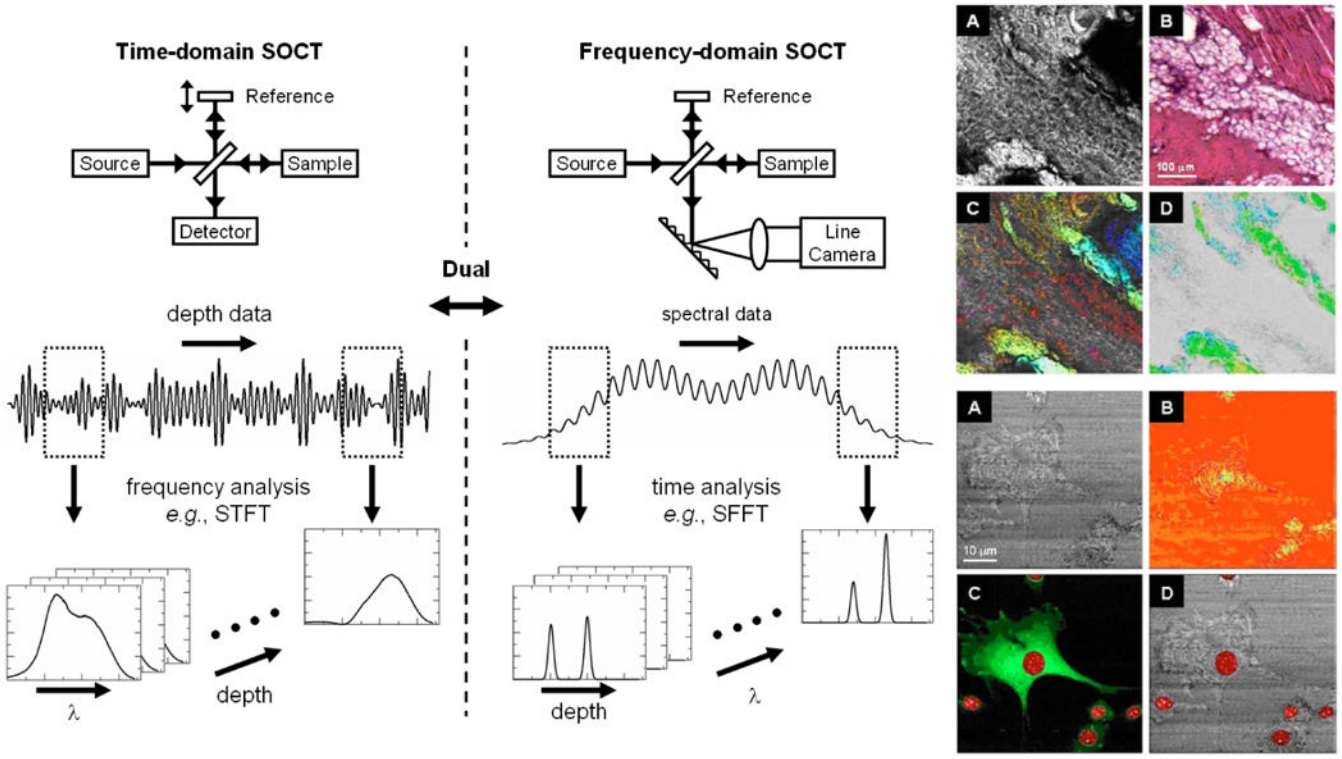

Figure 5.

Spectroscopic OCT imaging using time-domain and frequency-domain interferometry. Short-time and short-frequency Fourier transforms are performed on the acquired OCT interferograms to create 2-D SOCT signals that are indexed by wavelength and depth in the sample or specimen. Top Right Set: Spectroscopic spectral-domain OCM. A) OCM of rat tissue containing regions of adipose cells (middle) and muscle fibers (upper right, lower left). B) Corresponding histology. C) SOCM image using metameric spectral analysis. D) SOCM image using LSS spectral analysis. The scale bar is representative of all images. Bottom Right Set: Single-cell imaging with SOCM. A) Spectral-domain OCM of GFP-vinculin transfected fibroblasts in culture. B) Corresponding SOCM image showing localized regions of strong spectral scattering (yellow-green). C) Multiphoton microscopy of GFP fibroblasts co-labeled with DNA-nuclear dye. D) Overlay of multiphoton DNA-dye fluorescence and OCM images. Comparisons between (C) and (D) show validation of SOCM spectral analysis for identifying dominant scatterers. Figures adapted from (37 and 78).

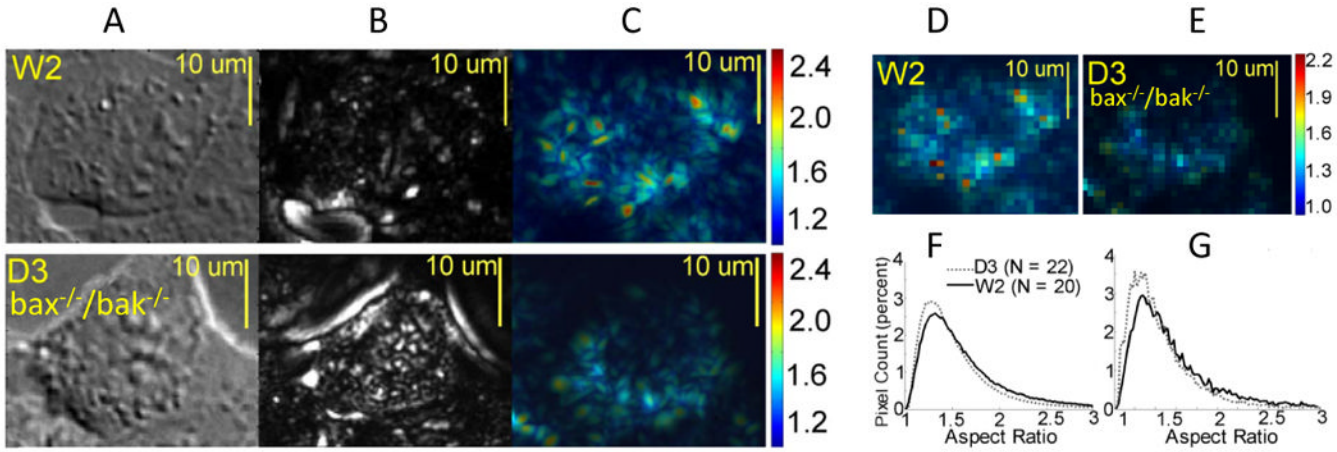

Figure 6.

Bax/Bak expressing (W2) and Bax/Bak null (D3) cells. A: Differential interference contrast, B: Dark field. C: Object orientation showing more highly oriented objects (red areas) in the W2 compared with the D3 cell. D & E: W2 and D3 orientation images after block-processing the initial Gabor filtered images to simulate a 4× demagnification. Data were collected using an orientation-sensitive Fourier filter bank consisting of nine Gabor-like filters with period S=0.95μm, Gaussian envelope standard deviation s=S/2=0.45 μm, and orientations φ=0° to φ=160° in 20° increments (See also (7)). The color intensity in panels C, D, and E represents the overall response (sum of all responses) of the pixel to the filter bank; the color hue encodes degree of orientation or "Aspect Ratio". The ratio of maximum intensity to the average of all the responses collected as a function of filter orientation was taken as a measure of aspect ratio. F&G: Pixel histograms of all W2 and D3 cells before (F) and after (G) pixel binning. In F, the mean aspect ratio parameter per cell was 1.58 for W2; 1.38 for D3 ($p<0.04$ by student t-test), and in G, 1.54 for W2; 1.35 for D3 ($p<0.05$ by student t-test).

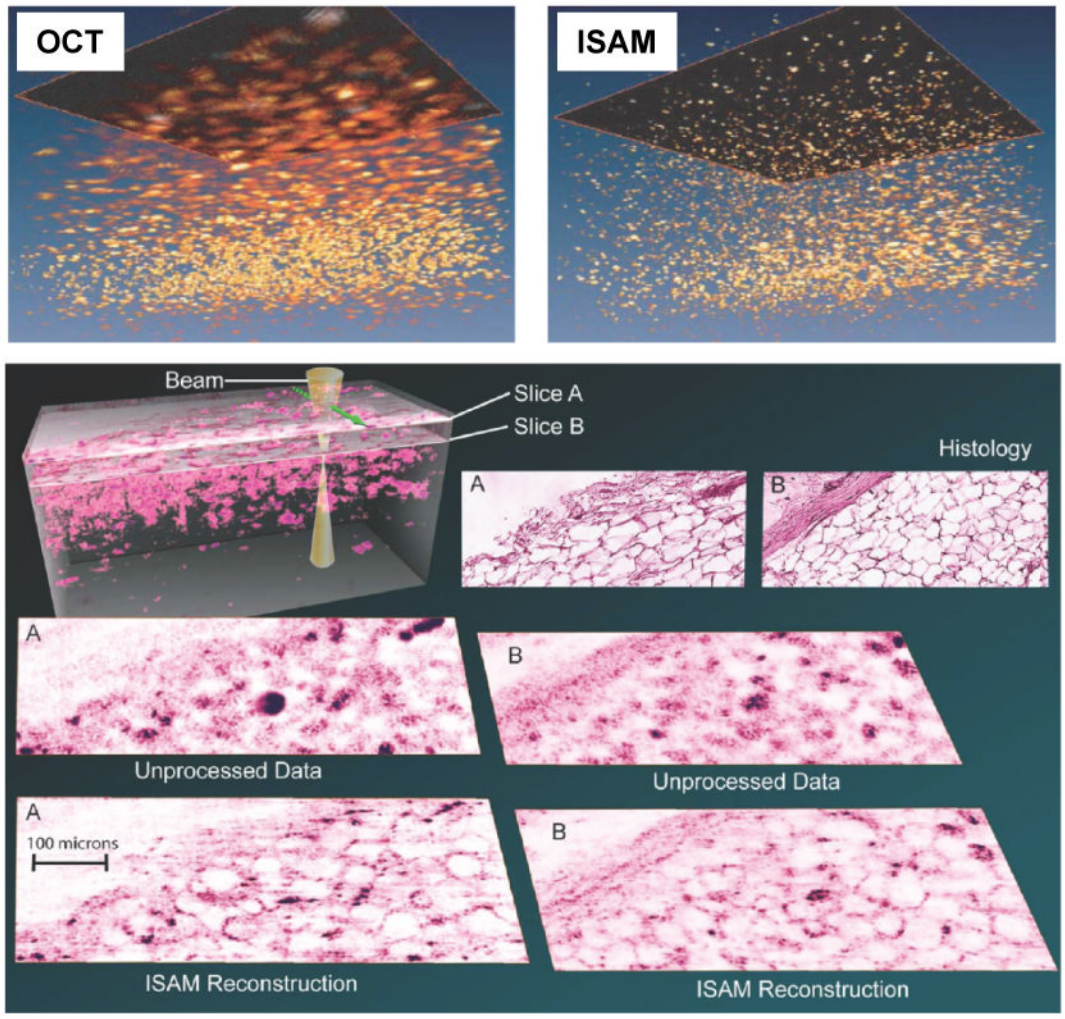

Figure 7.
Interferometric Synthetic Aperture Microscopy (ISAM). Top: Three-dimensional images of a tissue phantom composed of micron-sized point-scatterers of titanium dioxide suspended in a gel. OCT (top left) shows blurred out-of-focus scatterers near the upper surface, which are corrected to the appropriate point-like objects using ISAM (top right). Bottom: Human breast tissue. *En face* OCT images (labeled unprocessed data) and corresponding ISAM image reconstructions from two different slices/depths within a three-dimensional block of tissue. ISAM images enable fine cellular membranes and features to be resolved. Histology sections taken from approximate locations are also shown for comparison. Figure adapted from (120).