Monitoring of temperature-mediated phase transitions of adipose tissue by combined optical coherence tomography and Abbe refractometry

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Abstract. Observation of temperature-mediated phase transitions between lipid components of the adipose tissues has been performed by combined use of the Abbe refractometry and optical coherence tomography. The phase transitions of the lipid components were clearly observed in the range of temperatures from 24°C to 60°C, and assessed by quantitatively monitoring the changes of the refractive index of 1- to 2-mm-thick porcine fat tissue slices. The developed approach has a great potential as an alternative method for obtaining accurate information on the processes occurring during thermal lipolysis.

Keywords: adipose tissue; heating; refractive index; lipolysis; phase transition; Abbe refractometry; optical coherence tomography.

1 Introduction

Monitoring of temperature-mediated phase transitions in biological tissues is an acute problem in the field of laser applications in biology and medicine. This is due to the fact that most of the medical technologies rely on thermal effects. The laser-tissue thermal interaction, therefore, has been receiving more and more attention in recent years. In particular, designing of optical methods for selective non- or minimally invasive diagnostics and reduction of body fat requires robust technologies to monitor condition of adipose tissue (AT).

AT is a multicomponent tissue with constituents presented in Fig. 1. Fats within a lipid droplet in an adipocyte are typically represented by triglyceride (TG) mixtures. TGs are neutral fat molecules made up of three fatty acids connected to one glycerol molecule via the ester. Fatty acids are usually derived from TGs or phospholipids, known as “free” fatty acids (FFAs).

Measurements of the temperature-dependence of the refractive index (RI) can be used for the detection of phase transitions in the AT. The knowledge of thermal response of RI of AT including increments and phase transitions is important for getting a more accurate information on fat cell destruction pathways at laser heating or cell lipolysis induced by a low-level laser therapy. There are not many studies on RI of AT, especially in the course of tissue heating, available in the literature. Therefore, in this study, the temperature dependence of RI of AT samples has been examined utilizing a combined use of optical coherence tomography (OCT) and Abbe refractometry aiming to quantify RI temperature increment and lipid phase transitions with the goal to propose an alternative method for objective indication of processes taking place during fat cell destruction/lipolysis. In general, these studies can be useful for improvement in therapeutic protocols based on thermally induced fat cell lipolysis, including laser therapy for body contouring and spot fat reduction and photochemotherapy of cancer patients.

2 Materials and Methods

2.1 Brief Review of Adipose Tissue Properties

Table 1 shows data for RI of ATs measured by different methods and for different body sites in humans and animals.

The AT is characterized by a relatively low temperature and a few melting points that can significantly affect kinetics of the heating of the fat-bearing tissue. The lower fat melting point is caused by a higher content of unsaturated fatty acids, which includes the complex mixtures of TGs. These mixtures do not have a single melting point as the separate compounds. The fat crystals can be characterized at the microscopic level (i.e., shape, size, and orientation of crystals in lipid droplets) and at the molecular level (i.e., organization of TGs in lamellar structures of various thicknesses, correspondence to various polymorphic forms, etc.). Within the 24°C to 60°C temperature range, different lipid components of the AT undergo several phase transitions associated with the multicomponent lipid content of fat cells (Tables 2 and 3). Detection of such phase transitions is performed by combined use of the Abbe refractometry and optical coherence tomography.

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transition can be performed by measurements of the temperature dependence of the RI of the lipid-containing tissue components.

2.2 Optical Coherence Tomography and Abbe Refractometry

Structural changes in the tissue associated with the phase transitions can be assessed both in vitro and in vivo with OCT—a modern state-of-the-art noninvasive imaging modality. OCT is a well-known optical diagnostic technique to perform real-time 2-D and/or 3-D high-resolution (~5 to 8 μm) imaging of biological tissues in vivo with relatively high probing depth (up to 2 mm). Principles of this imaging modality are described in detail elsewhere. OCT has been extensively used for various applications in biomedical optics, including imaging of human skin, skin vessels, and skin blood circulation, connective tissue structure, individual vascular wall component, as well as for biotissue-mimicking phantoms and noninvasive glucose sensing. Simultaneous measurements of the RI and thickness of the sample are the advantages of OCT compared with standard techniques.

In addition to OCT, the Abbe prism classical refractometry is an attractive tool for assessment of phase transitions in in vitro studies. For comparison of RI measurements performed by OCT and Abbe refractometry, it should be considered that due to a broadband light source used in OCT it measures the group RI of a material, as a single wavelength measurements of Abbe refractometry give the phase RI. The group and phase RI are related; however, in dispersive media they are different. Simultaneous measurements of both phase and group RI and sample thickness are possible using combination of OCT and confocal microscopy or low-coherence interferometry at multiple angles of incidence enabling bulk RI measurement of scattering and soft samples.

Spectral-domain OCT has been extensively used to obtain subsurface images of ATs and for quantitative assessment of the RI. In the present study, we used a commercial OCT system (Hyperion, Thorlabs) operating at a 910-nm central wavelength with a bandwidth of 49 nm for independent measurements of optical thickness of tissue samples, from which RI can be evaluated if the sample thickness is supposed to be constant during measurements. The optical thickness measured by OCT corresponds to the group RI multiplied by the physical thickness. However, due to a low dispersion of AT on the wavelength 910 nm within a bandwidth of 49 nm, the group RI is very close to the phase RI. This OCT system...
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Table 1 RI of different ATs.

<table>
<thead>
<tr>
<th>AT type and body site</th>
<th>λ, nm</th>
<th>n, n_g (SD)</th>
<th>Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rats</td>
<td>750 to 850</td>
<td>n_g: 1.467 (0.026)</td>
<td>OCT</td>
</tr>
<tr>
<td>Mesenteric in humans</td>
<td>1300</td>
<td>n_g: 1.467 (0.008)</td>
<td>OCT</td>
</tr>
<tr>
<td>Abdominal in humans</td>
<td>930</td>
<td>n_g: 1.460 (0.002)</td>
<td>OCT</td>
</tr>
<tr>
<td></td>
<td></td>
<td>n_g: 1.386 (0.008)</td>
<td></td>
</tr>
<tr>
<td>Subcutaneous in humans</td>
<td>456 to 1064</td>
<td>1.44</td>
<td>Thin film reflectometry</td>
</tr>
<tr>
<td>Abdominal in humans</td>
<td></td>
<td>1.46</td>
<td></td>
</tr>
<tr>
<td>Bovine</td>
<td>633</td>
<td>1.455 (0.006)</td>
<td>Fiber optic refractometry</td>
</tr>
<tr>
<td>Porcine</td>
<td>488</td>
<td>1.510 (0.002)</td>
<td>Laser refractometry</td>
</tr>
<tr>
<td>Porcine</td>
<td>632.8</td>
<td>1.492 (0.003)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1079.5</td>
<td>1.482 (0.002)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1341.4</td>
<td>1.487 (0.004)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>632.8</td>
<td>1.493 (0.005)</td>
<td></td>
</tr>
</tbody>
</table>

Here n is the phase RI and n_g is the group RI. SD is standard deviation.

Table 2 The most common fatty acids of adipocytes.

<table>
<thead>
<tr>
<th>Structural formula</th>
<th>Name</th>
<th>Melting temperature, °C</th>
</tr>
</thead>
<tbody>
<tr>
<td>CH₃(CH₂)₁₆COOH</td>
<td>Lauric</td>
<td>44</td>
</tr>
<tr>
<td>CH₃(CH₂)₁₄COOH</td>
<td>Myristic</td>
<td>58</td>
</tr>
<tr>
<td>CH₃(CH₂)₁₃COOH</td>
<td>Palmitic</td>
<td>63</td>
</tr>
<tr>
<td>CH₃(CH₂)₁₂COOH</td>
<td>Stearic</td>
<td>70</td>
</tr>
<tr>
<td>CH₃(CH₂)₁₁COOH</td>
<td>Arachidic</td>
<td>77</td>
</tr>
</tbody>
</table>

Table 3 Approximate composition of porcine and human fats. FFA denotes the free fatty acids.

<table>
<thead>
<tr>
<th>Melting temperature, °C</th>
<th>AT</th>
<th>Palmitic</th>
<th>Stearic</th>
<th>Oleic</th>
<th>Linoleic</th>
<th>Linolenic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Porcine</td>
<td>36 to 45</td>
<td>27 (63)</td>
<td>14 (70)</td>
<td>45 (16)</td>
<td>5 (--5)</td>
<td>5 (--11)</td>
</tr>
<tr>
<td>Visceral human</td>
<td>30 to 35</td>
<td>25 (63)</td>
<td>8 (70)</td>
<td>46 (16)</td>
<td>10 (--5)</td>
<td>—</td>
</tr>
</tbody>
</table>

had a low spatial resolution; however, it was sufficient to provide RI measurements in different points of the samples and allowed to combine it with the AR.

For spectral-domain OCT measurements, tissue sample heating was managed by a homemade temperature control plate (based on thermoresistors) upon changing the externally applied voltage. The temperature was varied from room temperature (25°C) up to 70°C with the corresponding voltage changes in the range of 5 to 10 V. The data obtained were processed with MATLAB (MathWorks). In addition, a multiwavelength AR Atago equipped with a CMOS camera was used for the quantitative assessment of the RI, while the temperature was controlled in the range from the room temperature to 60°C by a thermostat T200 (GK Sondermaschinenbau, Germany) and was monitored remotely with an infrared camera FLIR b60 (FLIR, Sweden) providing 2% accuracy. The slices of ATs were placed on the prism of the AR.

AR employs the effect of total internal reflection to measure the RI n of a solid or liquid specimen in the visible or near-infrared (NIR) ranges. Traditionally, to obtain RIs of transparent specimens, the spatial distribution of transmitted light is analyzed. However, this method cannot be directly used for the measurement of RI of turbid biological tissues with strong scattering and absorption. Thus, in the current study, all the refractometric measurements were performed in the reflection mode.

The working principle of AR is based on the measurement of a critical angle. As it is shown in the inset of Fig. 1, the sample is placed on top of the measurement prism. A divergent light beam enters the sample through the prism, gets refracted at a critical angle on its surface, and then a telescope is used to measure position of the border between bright and dark areas. The telescope reverts the image, so the dark area is at the bottom, even if we expect it to be in the upper part of the field of view. Knowing the angle and RI of the measuring prism, it is not difficult to calculate RI of the sample

\[ n = \frac{N \sin(i_{\text{crit}})}{\sin(B)}, \]

where n and N are the RIs of the sample and the measuring prism, respectively.

Usually, the exiting angle of the critical beam from the prism to the air \( \phi \) is measured. It is easy to show by considering the refraction of light beams on the face of the prism BC plane (Fig. 1) that the RI of the sample n is related to the angle \( \phi \)

\[ n = \sin B \sqrt{N^2 - \sin^2 \phi} - \cos B \sin \phi, \]

where B is the refracting angle of the prism (the angle between the refracting faces). In fact, when measuring, there is no need to
use this equation to calculate RIs, since the reference scale of the refractometer was calibrated in the values of $n$ with regard to Eq. 1.

Calibration of the AR was carried out using 86%, 93%, and 100% glycerol–water solutions. RIs of the glycerol–water mixtures were calculated based on the corresponding data for the pure substances. All the measurements were performed at the 930-nm wavelength using the corresponding optical filter (10-nm bandwidth) supplied with the refractometer. The images from the device fed into the CMOS camera comprise two sections: the brighter part (Fig. 3) formed by light beams totally reflected from the sample, and the dark part formed by rays only partially reflected from the sample (and partially transmitted through it).

The recorded images were processed using the ImageJ software: normalization to the background and retrieval of the profile intensity curves across the screen perpendicular to the shadow-light interface [Fig. 4(a)]. The resulting calibration curve [Fig. 4(b)] shows dependence of the RI on the screen coordinate.

Fresh samples of porcine AT from food market were used in the studies. Experiments were performed for 30 samples taken from the same piece of porcine AT (10 samples were used for each series of experiments). Slicing of frozen AT samples was carried out manually with a scalpel. About 1- to 2-mm-thick tissue slices were heated from the room temperature up to 60°C by running water from a thermostat and corresponding refractometer images were automatically recorded by the CMOS camera. The RIs were retrieved during postprocessing of the obtained images: location of the light-shadow border was associated with the X-coordinate [Fig. 4(a)], and the corresponding RI was found using the calibration curve [Fig. 4(b)].

Statistical analysis of data was performed using the Statistics 6.0 software.

3 Results and Discussion

Figure 5 and Video 1 (see supplement) show the temporal evolution of OCT images of porcine fat with temperature increase from 23°C to 70°C. On high-resolution OCT images, well-seen

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**Fig. 2** Schematic illustration of combined optical coherence tomograph and Abbe refractometer system for assessing RI of AT samples. OCT system: spectral-domain (Hyperion, Thorlabs) or time-domain (Institute of Applied Physics RAS, Russia). Abbe refractometer: a multiwavelength refractometer Atago DR-M2 1550 (Atago, Japan); light source: JCR12V100W10H/G1/IR (Iwasaki, Japan); it also includes PixelINK PL-B741F CMOS camera (detector size: 8.6 mm × 6.9 mm) (PixelINK, Canada).

**Fig. 3** Refractometer screen images (used for calibration) upon application of aqueous solutions of glycerol of different concentrations: (a) 86%, (b) 93%, (c) 100%, and (d) background.

temperature-induced fat tissue optical clearing on the cellular level happened due to reduction in the light scattering. This reduction was caused by the phase transition of lipids localized in droplets from crystalline to liquid state for temperatures from about 35°C.

OCT provides accurate measurements of the RI with the approximate precision of 1% if the geometrical thickness is known. The effective RI of the tissue sample was assessed as the ratio

\[ n = \frac{z}{l}, \]  

where \( l \) is the geometrical (true) thickness of the sample and \( z \) is the OCT-observed depth, i.e., the optical thickness.

The RI of AT was assessed for the optical thickness of samples obtained with spectral- or time-domain OCT systems. The optical thickness was found from the \( A \)-scan of the OCT images as a distance between two main peaks associated with the sample-glass interfaces. To provide better localization of the boundaries, the \( A \)-scans were averaged over a certain lateral region (2 mm). This operation smooths out the random noise in the system and influence of inhomogeneity of cell structure of the tissue, while the peaks corresponding to the sample boundaries become more distinct. Processing and smoothing of OCT signals and determination of distance between peaks were performed using MathCAD (PTC, Inc.). The "medsmooth" function was used for smoothing with the help of a sliding median.

Figure 6 shows the temperature dependence of the RI of AT obtained by OCT measurements. Bars show SDs and the line connecting mean values of RI shows local alterations of the slope of RI temperature dependence.

Fig 5 OCT images of AT samples at different temperatures: (a) 23°C, (b) 35°C, (c) 45°C, and (d) 55°C. The Video file shows kinetics of the OCT images alterations (Video MP4, 2.16 MB [URL: http://dx.doi.org/10.1117/1.JBO.23.1.016003]).

Fig 4 (a) Profile intensity curves obtained from three glycerol-water solutions (black square—100%, \( n = 1.4641 \); red circle—93%, \( n = 1.4554 \); green triangle—86%, \( n = 1.4466 \)) and (b) the resulting calibration curve. The vertical arrows in (a) indicate the border between dark (to the left of the lines) and light (to the right) areas and the intersection positions (in pixels) with the X-axis correspond to refractive indices of the relevant water-glycerol solutions. The calibration did not go beyond the accuracy of the measurements provided by the Atago refractometer (10⁻⁴).
Table 4 shows mean value and SD of phase transition characteristic temperatures averaged over 10 fat samples studied by each measurement technique [AR, spectral OCT (S-OCT), and AR/time-domain OCT (AR/TD-OCT)] and comparison with the independent measurements done for 10 samples using optoacoustic technique. It is worth noting that data presented in Table 4 for all 30 samples studied in this paper demonstrate existence of the very-low-temperature transitions for 29 samples in the range from 22.3°C to 26.5°C detected by all three measurement techniques (AR, S-OCT, and AR/TD-OCT, respectively). The low-temperature transitions in the range from 30.6°C to 35.2°C were also detected by all techniques for 27 samples, whereas the middle-temperature range transitions between (39.5 ± 1.9)°C and (44.4 ± 0.6)°C were found only for AR measurements, for nine and six samples from studied 10, respectively. The high-temperature transitions, which are in the range from 50.2°C to 55.5°C, were detected for 22 samples by all three measurement techniques.

It also should be pointed out that temperature dependence observed with concurrent dual modal technique [refractometry and OCT (Fig. 9)] agrees well with the results obtained by optoacoustic measurements for ex vivo samples of porcine fat (the right column of Table 4). Data received for both techniques have a similar trend and high correlation index (Spearman’s rank correlation coefficient between the OCT and the optoacoustic measurements was 0.958; between the refractometry and the optoacoustic measurements was 0.958). Unfortunately, we could not compare the second high-temperature transition of porcine fat found in optoacoustic measurements at (65 ± 2)°C.
because of a limited working temperature range of the Atago refractometer.

The low-temperature phase transitions (22°C to 35°C) could be associated with the fusible FFA of the fat droplet like oleic acid; the middle-temperature (40°C to 44°C)—with cell membrane phospholipids, and the high-temperature (45°C to 55°C)—with less fusible FFA of the fat droplet, such as palmitic acid.

The temperature dependence of RI can be described as

\[ n = n_0 + (dn/dT)T, \]

where \( T \) is temperature [°C].

For our data (see Fig. 9), the averaged RI temperature increment \( dn/dT \) equals to \(-3.8 \times 10^{-4} \text{°C}^{-1}\) with \( n_0 = 1.466 \), corresponding well to the increment measured for the oleic acid; \(-3.8 \times 10^{-4} \text{°C}^{-1}\) (as shown in Ref. 35) with \( n_0 = 1.467 \). This result can be explained by the high content of oleic acid in porcine fat (45%) and its low melting temperature (16°C) (Table 3). Therefore, it easily and intensively leaks out from the adipose cells and accumulates on the interface between the sample and the refractometer measuring prism.

Figure 9 shows temperature dependence of the RI of AT sample calculated from the A-scans of OCT images and concurrently measured using the refractometer. As one can see, the RI data retrieved by the two independent methods are in good agreement both qualitatively and quantitatively. OCT measurements of phase transitions are prospective for in vivo studies and can be routinely used to assess the temperature-induced RI change of fat cells near the skin surface. It is important to note that phase transitions of lipids inside lipid droplets can be also detected by OCT via monitoring of reduction of light scattering (cellular optical clearing).

The presented approach has a high potential to be used for getting more valuable information about processes taking place during thermally induced fat cell lipolysis for more accurate quantification of many different therapeutic protocols, including laser therapy for body contouring and spot fat reduction and photochemotherapy of cancer patients.

Disclosures

No conflicts of interest, financial or otherwise, are declared by the authors.

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References


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