Nonlinear distortion correction for posterior eye segment optical coherence tomography with application to tree shrews

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Abstract: We propose an empirical distortion correction approach for optical coherence tomography (OCT) devices that use a fan-scanning pattern to image the posterior eye segment. Two types of reference markers were used to empirically estimate the distortion correction approach in tree shrew eyes: retinal curvature from MRI images and implanted glass beads of known diameter. Performance was tested by correcting distorted images of the optic nerve head. In small animal eyes, our purposed method effectively reduced nonlinear distortions compared to a linear scaling method. No commercial posterior segment OCT provides anatomically correct images, which may bias the 3D interpretation of these scans. Our method can effectively reduce such bias.

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1. Introduction

Optical coherence tomography (OCT) is a noninvasive imaging technique that has revolutionized eye research and the clinical practice of ophthalmology [1]. OCT devices that are designed to image the posterior segment of the eye can provide high-resolution, cross-sectional images of the retina and the optic nerve head (ONH). Multiple OCT B-scans can be acquired using raster or radial scan patterns to provide a 3-dimensional (3D) structural representation of the imaged tissues. Much interest has formed around the 3D characterization of the ONH morphology and the identification of structural biomarkers from OCT images for improved accuracy in the diagnosis of glaucoma, glaucoma risk and progression [2–7]. Glaucoma is the second leading cause of blindness in the world and ONH remodeling is a defining feature of glaucoma [8–10]. Some of the proposed disease biomarkers are 3D in nature and rely on anatomically-accurate image data for accurate diagnosis. OCT images of the eye, however, suffer from optical distortions that can bias the 3D interpretation of the actual ONH morphology and clinical biomarkers of diseases. These optical distortions depend on the OCT device (e.g. scan pattern, camera focus, camera position relative to the eye) and the optical properties of the imaged eye. Changes in camera settings (e.g. the camera position relative to the eye) between follow-up scans can lead to image distortions that can bias scientific results and their interpretation in longitudinal studies.

Clinical OCT devices used to image the posterior segment of the eye are designed and calibrated to image human eyes, and typically use an undisclosed approach to transform the OCT image into the physical object space. Clinical OCT devices are widely used for animal research because of their translational potential but often the factory settings are utilized for imaging animal eyes. However, the transformation approach used by clinical devices are often not applicable to animal eyes and leave residual distortions when applied. Garcia Garrido et al. have proposed
an empirical approach to estimate linear correction factors to overcome this issue when using a clinical spectral-domain OCT (Spectralis OCT, Heidelberg Engineering GmbH, Heidelberg, Germany) in murine eyes [11]. In this study, micro-spheres of known diameter were implanted into the retina or subretinal space and used as reference markers to calibrate the linear correction factors. Hsu et al. used a similar approach and implanted ceramic beads as reference markers to validate volumetric measurements in OCT scans [12].

To the best knowledge of the authors, nonlinear distortions are currently not corrected in clinical OCT devices used for posterior segment imaging. The two major sources of nonlinear distortions in posterior segment OCT imaging are: (i) fan-scanning, where scan rays pass through one pivot point, and (ii) refraction of the scan rays at the optical surfaces of the eye [13]. Both types of OCT nonlinear distortions can be effectively corrected by means of optical modeling [14–21]. However, optical model-based distortion correction methods require detailed knowledge of the optical properties of the OCT devices and the imaged eye. The translation of these methods to clinical OCT devices remains difficult as the optical details of commercial devices are undisclosed. Furthermore, additional assumptions or imaging modalities are required to generate the optical model of the imaged eye. Consequently, the optical model-based distortion correction method remains an exclusive solution for laboratory OCT systems.

Inspired by the work of Garcia Garriido et al. [11], we propose the first empirical nonlinear distortion correction method for posterior segment images obtained on commercial OCT devices. Our method does not require knowledge of the optical details of OCT imaging device. We have developed and tested the performance of our method against a linear distortion correction approach in tree shrews using a commercial OCT device (Spectralis OCT2, Heidelberg Engineering, Heidelberg, Germany).

2. Material and methods

2.1. Animals

The animal experiments in this study were carried out using northern tree shrews (*Tupaia belangeri*) in accordance with the ARVO Statement for the Use of Animals in Ophthalmic and Vision Research and were approved by the University of Alabama at Birmingham Institutional Animal Care and Use Committee. Five animals were utilized to establish the distortion correction methods, which included micro-bead implantation, magnetic resonance (MRI) and OCT imaging. We selected the animals to obtain measurements across a wide range of ages (80 days to 6 years of age) and axial lengths (7.37 to 8.07 mm). Ocular axial dimensions were obtained in awake animals using the LenStar Biometer LS 900 (Haag-Streit AG, Koeniz, Switzerland). The LenStar data was analyzed using tree shrew specific refractive indices [22]. Additional 10 animals were utilized for OCT imaging to benchmark our distortion correction method.

2.2. Glass bead implantation

We used glass microbeads (Thermo Scientific™ 9000 Series Glass Particle Standards; Soda Lime Glass Beads, Cat. #9100; Thermo Fisher Scientific, Waltham, MA, USA) of known diameter (98.7 ± 2.8 µm) as reference markers visible to the OCT. Anesthesia was induced in tree shrews using 4% isoflurane at 2.5 L/min. Anesthesia was maintained using xylazine (7.5 mg/kg intramuscular) and 1-2% isoflurane. A surgical microscope with OCT capability (Zeiss OPMI Lumera 700 with integrated OCT, Oberkochen, Germany) was used to guide the surgeon during microbead implantation (Fig. 1). A circular 5 mm glass coverslip (Harvard Apparatus, Small Diameter Coverslips, Cat. #1217H19; Thomas Scientific, Swedesboro, NJ, USA) was placed on top the cornea with 2.5% hypromellose ophthalmic demulcent solution (Gonak, Akorn Pharmaceuticals, Forest Lake, IL, USA) as coupling agent to allow planar ophthalmoscopy and live imaging during the injection. Glass beads were washed three times in sterile balanced salt
solution (BSS), centrifuged down, and then resuspended in BSS. Glass beads loaded into a glass capillary tube attached to a tuberculin syringe by polyethylene tubing. One drop of topical 0.5% proparacaine hydrochloride ophthalmic solution (Bausch & Lomb, Greenville, SC, USA) was placed in the eye for additional anesthetic purposes. The eyelids were cleaned with 5% betadine solution times 3 followed by washes with sterile BSS. A 18 gauge needle was used to make a sclerostomy just posterior to the limbus. The glass pipette was advanced under direct visualization of the microscope toward the retina at the posterior pole. Glass beads were slowly injected on top, into, or below the retina. The glass pipette remained in position for approximately 1 minute prior to withdrawal. An 8-0 vicryl suture was used to close the sclerostomy and the conjunctiva. Gentamycin antibiotic was placed on the operative and non-operative eyes and animals were given carprofen subcutaneously. The animals were then placed on a warming blanket and allowed to recover. Upon waking, animals were returned to their nest tube and transferred to their home cages. The animals were examined daily for 3 days post-injection for signs of inflammation or other signs of distress.

Fig. 1. Live screenshot of the surgical microscope display used during microbead implantation. The live microscope image (left-hand side), horizontal and vertical OCT scans (right-hand side) are used to guide the injection and provide a direct confirmation of the successful subretinal bead implantation. White arrows localize the same glass beads in the live microscope image and OCT scans.

2.3. Magnetic resonance imaging of posterior retinal curvature

The curvature of the posterior retina was obtained from MRI as a second reference marker. Animals were anesthetized using intramuscular injections of ketamine (100 mg/kg) and xylazine (7.5 mg/kg). Animals were placed on a heating pad with a MRI-compatible pillow to monitor the respiration rate. Isoflurane was provided as needed. A 12mm diameter coil was placed on top of the eye that was imaged. The small animal MRI system BioSpec Preclinical 9.4T (Bruker, Germany) was used to image each eye individually. 2D MRI scans were performed to obtain horizontal slices (500 µm thick) through the posterior pole with an in plane resolution of 37 µm. MRI scans were exported, and the retina at the posterior pole was manually segmented using custom MATLAB code (Fig. 2).

2.4. Optical coherence tomography of reference markers

Animals were anesthetized as described in Section 2.3 and placed on a semicircular bed with heating pad. To dilate the eyes, one drop of 2.5% Phenylephrine Hydrochloride and 1 drop of
1% tropicamide (Akorn, Lake Forest, IL, USA) were administrated to each eye. The Spectralis OCT2, a clinical spectral-domain OCT, was used to image the tree shrew eyes. A plano, hard contact lens with a 4 mm base curve and 6.3 mm diameter was placed on each cornea to provide a stable ocular surface for imaging and reduce the risk of corneal dehydration. No attachment lens was used but has been needed when imaging even smaller eyes such as murine eyes with the Spectralis [23,24]. In 5 animals, 30-degrees wide B-scans were acquired through the center of the implanted glass beads. The bead center was identified by moving the location of the B-scan on the live image until the largest bead diameter was identified. Figure 3A shows a representative B-scan. Each glass bead was imaged multiple times using a wide range of camera settings by changing the camera position relative to the eye, reference arm parameter (RAP), scan focus, and B-scan resolution (high/low). In total, 226 B-scans were collected across all five animals. Enhanced depth imaging was used for all scans. The raw OCT images were exported, and scan and imaging parameters were obtained from the meta data. Note that the RAP of the Spectralis is correlated to the reference arm length position, where RAP values are inversely proportional to the reference arm length. A custom reference target was built and imaged before and after each imaging session using the same camera settings to verify the RAP value. This validation step was necessary as we noticed inconsistent recordings of the RAP value, which may have been caused by an occasional slippage of the reference arm of the Spectralis. OCT recordings with inconsistent RAP values before and after an imaging session were disregarded. Custom MATLAB code was used to manually segment the glass bead and retinal pigment epithelium in each B-scan.

2.5. Nonlinear distortion correction approach

Posterior segment OCT devices such as the Spectralis typically use a fan-scanning pattern, where multiple A-scans are swept across a scan angle to form one B-scan. Due to this scan pattern and relatively constant optical path lengths between A-scans, posterior eye tissues such as the retina often appear flat and not curved in posterior segment OCT images [15]. A simple nonlinear image transformation can be applied to correct for the fan-scanning distortion [13], which represents the basis of our purposed method. Figure 4A,C illustrates the distorted images space and the nonlinear distortion corrected object space, respectively, where OCT scan rays (A-scans) are assumed to pass through one pivot point. Let \( P \) be an image point defined by its pixel coordinates \( u, w \), where \( u_{\text{max}}, w_{\text{max}} \) represent the pixel dimensions of the image. The point \( P \) can be mapped into its object space location \( P' \) defined by the Cartesian coordinates \( x, z \) using the following
Fig. 3. Distortion correction of one representative OCT B-scan obtained at the posterior pole with manually segmented glass bead (yellow ellipse) and RPE (red line). Shown are the original OCT image with the original scaling used by the Spectralis software (A) and the corrected images using the linear (B) and nonlinear approach (C). The spherical glass bead appears elliptical due to distortions in the original OCT images. Both distortion correction methods were able to recover the known bead diameter (98.7 ± 2.8 µm), but only the nonlinear approach was able to match the RPE curvature to the retinal curvature of the MRI image (Fig. 2).

transformation:

\[ x = r \sin(\alpha), \quad z = r_P - r \cos(\alpha), \]  

where \( r_P \) is the distance between the pivot point and the center of the distortion corrected image, \( r \) and \( \alpha \) are polar coordinates (see Fig. 4) defined as

\[ \alpha = \left( u - \frac{u_{\text{max}}}{2} \right) \frac{\alpha_{\text{tot}}}{u_{\text{max}}} k_u, \quad r = r_p + \left( w - \frac{w_{\text{max}}}{2} \right) k_w. \]  

In the above equation, \( \alpha_{\text{tot}} \) represents the total scan angle of a B-scan. The term \( \alpha_{\text{tot}}/u_{\text{max}} \) generalizes the distortion correction method to arbitrary angular resolutions, where the same \( k_u \) value applies to B-scans with different scan angles (\( \alpha_{\text{tot}} \)) and/or number of A-scans used to compose the B-scan (\( u_{\text{max}} \)). The nonlinear distortion correction consists of three unknown parameters:

- \( k_u \): the angular scaling factor
- \( k_w \): the axial scaling factor
- \( r_P \): the pivot point location

In contrast to previous studies [15–17,20], these unknown parameters are not obtained from optical modeling. Instead, we estimated these parameters empirically by matching the mapped bead diameter and RPE curvature for each B-scan to the known diameter and retinal curvature obtained from MRI images of the same eye, respectively. The following objective function was
minimized to identify the three transformation parameters:

$$
\text{cost} = \left( \frac{d_{B1}}{d_{\text{ref}}} - 1 \right)^2 + \left( \frac{d_{B2}}{d_{\text{ref}}} - 1 \right)^2 + \left( \frac{R_{\text{OCT}}}{R_{\text{MRI}}} - 1 \right)^2 ,
$$

where $d_{B1}, d_{B2}$ represent the major and minor diameter of the segmented glass bead after OCT distortion correction; $d_{\text{ref}}$ is the known glass bead diameter; $R_{\text{OCT}}$ is the RPE radius of curvature after OCT distortion correction; and $R_{\text{MRI}}$ is the retinal radius of curvature obtained from the matching MRI image. Note, that both reference markers (glass bead diameter and retinal curvature) were needed to obtain a unique minimum.

2.6. Linear distortion correction approach

To benchmark our nonlinear distortion correction method, we established a linear distortion correction approach for tree shrew eyes based on the empirical method proposed by Garcia Garrido et al. [11]. We used the following linear transformation illustrated in Fig. 4A,B to map an OCT B-scan from the image to the object space:

$$
x = \left( u - \frac{u_{\max}}{2} \right) \mu_{\text{tot}} k_{u}^{\text{lin}}, \quad z = -w k_{w}^{\text{lin}} .
$$
Similar to our nonlinear approach (1, 2), the term $\alpha_{\text{tot}}/u_{\text{max}}$ is used to generalize the linear distortion correction method for arbitrary angular scan resolutions. The linear distortion correction approach consists of two unknown parameters:

- $k_{\text{lin}}^u$: the lateral scaling factor
- $k_{\text{lin}}^w$: the axial scaling factor

Both unknown parameters were obtained for each B-scan by matching the bead diameter of the transformed OCT image to the known diameter. This was accomplished by minimizing the first part of the cost function (3). The RPE curvature was not used to determine the linear transformation parameters.

2.7. Benchmark

The performance of the linear and the nonlinear distortion correction method was evaluated against distorted images of the ONH in ten animals (20 eyes) that were not used to estimate the transformation parameters. Each eye was imaged using 48 radial scans centered at ONH. Each radial B-scan was acquired using a 30 deg scan angle, enhanced depth imaging, and the high resolution setting of the Spectralis. Repeated scans were performed at baseline position and after moving the OCT camera 3mm towards and 3mm away from the eye. The follow-up imaging mode of the Spectralis was used with the baseline scan as reference. The option to “align follow-up to reference” was deactivated during exporting the raw images. Two morphological parameters were extracted for each OCT scan: the anterior lamina cribrosa (LC) depth and peripheral RPE position using an automated segmentation method described in the next subsection. Paired t-tests were used to identify significant differences in the morphological parameters between camera positions.

2.8. Auto-segmentation

A deep learning algorithm (Reflectivity, Abyss Processing Pte Ltd, Singapore) was trained with 489 manually segmented OCT B-scans. The trained deep learning algorithm was used for

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**Fig. 5.** Auto-segmentation and 3D reconstruction of the ONH morphology from OCT radial scans. A: Representative radial OCT scan through the center of the tree shrew ONH with segmented tissue interfaces. B: 3D reconstruction of the segmented tissue interfaces of 48 radial scans after nonlinear distortion correction.
auto-segmentation of the anterior ONH tissues of each radial OCT scan. The Bruch’s membrane opening (BMO) and anterior scleral canal opening (ASCO) were defined as the end points of the auto-segmented RPE and anterior sclera, respectively. The segmented points and interfaces of all 48 radial scans were transformed into the distortion corrected object space and used for 3D reconstruction of the ONH morphology. Figure 5 shows a representative uncorrected OCT scan of the tree shrew ONH with segmented tissue interfaces and the 3D reconstruction of the ONH after nonlinear distortion correction. Two parameters were extracted from the reconstructed 3D ONH morphologies. The LC depth was computed as the normal distance between the best fit ASCO plane and the anterior LC surface averaged over the area enclosed by ASCO. The peripheral RPE position was computed as the normal distance between the best fit ASCO plane and RPE and averaged over a 50 $\mu$m band at a distance of 1000 to 1050 $\mu$m away from the ASCO centroid.

3. Results

3.1. Linear distortion correction

The linear approach was able to match the bead diameter to the known glass bead diameter of each B-scan (Fig. 3B). However, linear scaling proved inadequate for correcting the RPE curvature. The optimal lateral and axial scaling factors were determined as $k_u^{lin} = 5167 \pm 361$ rad$^{-1}$ and $k_w^{lin} = 3.912 \pm 0.062$ $\mu$m/pixel (mean $\pm$ standard deviation), respectively. The optimal axial scaling factor was similar to the value used internally by the Spectralis (3.87 $\mu$m/pixel) [25].

3.2. Nonlinear distortion correction

In all B-scans, the nonlinear distortion correction method was able to match both reference markers well (Fig. 3C), the bead diameter to the known glass diameter (mean absolute error = 0.64 $\pm$ 0.66$\mu$m) and the mean RPE radius of curvature to the retinal radius of curvature of the matching MRI image (mean absolute error = 38.4 $\pm$ 59.7$\mu$m). The optimal axial scaling factor was identified as $k_w = 3.911 \pm 0.092$ $\mu$m/pixel similar to the value obtained with the linear approach. Two nonlinear correlations were identified between (i) the angular scaling factor $k_u$ and the pivot point location $r_P$ ($R^2 = 0.9993$), and (ii) $k_u$ and two scan parameters: the scan focus and RAP value ($R^2 = 0.9185$):

$$r_P[mm] = 5.2448/k_u - 0.2418,$$

$$k_u[-] = 0.1449\ \text{RAP} + 0.1382\ SF - 3.9927,$$ (5)

where SF and RAP represents the scan focus in Diopters and position of the reference arm in mm, respectively. The two correlations were obtained as curve and surface fits, and are illustrated in Fig. 6.

3.3. Benchmark

The identified mean scaling factors and the correlations (5) were used to perform the linear and nonlinear distortion correction of the ONH scans at baseline position and after moving the OCT camera by 3mm. Moving the camera by 3mm induced distortions and changes in the imaged ONH morphology, which were subjectively restored after nonlinear distortion correction but not after linear correction. Figure 7 shows one representative ONH scan at each camera position and after applying both distortion correction methods. When using the linear correction method, the peripheral sclera appeared bowed upwards and downwards at the -3 mm and +3 mm camera position compared to baseline. Similar distortions were subjectively not seen after nonlinear distortion correction.
Fig. 6. Identified correlations between transformation and scan parameters. **A**: A nonlinear correlation was identified between two transformation parameters: the pivot point location \( r_p \) and the angular scaling factor \( k_u \). **B**: A second correlation was identified between the angular scaling factor \( k_u \) and two imaging parameters: RAP and scan focus.

Fig. 7. **A**: OCT B-Scans of the same tree shrew ONH at baseline (center row) and after moving the camera 3 mm away (top row) or towards the eye (bottom row) from baseline position. OCT distortions were corrected using the linear (left column) and nonlinear approach (right column). While obvious distortions remained visible when using the linear scaling approach (bowing of the peripheral sclera), no apparent distortions were subjectively visible after nonlinear distortion correction. **B**: Linear (left column) and nonlinear (right column) corrected OCT images at \( \pm 3 \) mm camera positions fused with baseline scans. Gray, green and magenta colors represent regions where the OCT scan match, only the baseline and \( \pm 3 \) mm camera position image intensity is present, respectively. Images were translated and rotated to match ASCO centroids and normals.
Highly significant changes ($p<0.00001$, paired t-test) in the peripheral RPE position (Fig. 8) and anterior LC depth (Fig. 9) were found when moving the camera by ±3mm relative to baseline and using the linear correction method. The difference from baseline at +3mm and -3mm camera positions of the peripheral RPE position was drastically reduced from −33.9 ± 4.4µm and 33.6 ± 4.1µm using the linear correction to 0.6 ± 3.3µm and −0.7 ± 3.6µm when using the nonlinear correction, respectively. Similar, the anterior LC depth difference from baseline at +3mm and -3mm camera positions was reduced from −3.1 ± 2µm and 4.2 ± 2µm using the linear correction to 0.7 ± 1.9µm and −0.2 ± 2µm when using the nonlinear correction. No significant differences ($p>0.05$) were found between camera positions when using the nonlinear distortion correction method except for the difference from baseline of the anterior LC depth ($p = 0.036$, paired t-test, Fig. 9). However, this difference was small (0.9µ) compared to the axial resolution of the OCT device (3.9µ).
Fig. 9. A: Illustration of the anterior LC depth relative to the best fit ASCO plane. The anterior LC depth was averaged over the projected area enclosed by the ASCO points. B, C: Plots showing the anterior LC depth (B) and its difference from baseline (C) of repeated OCT scans of 20 tree shrew eyes at different camera positions and after linear and nonlinear distortion correction. Horizontal line, box, and vertical lines represent the mean, 95th confidence intervals for the mean, and standard deviation, respectively. * and ** represent $p<0.05$ and $p<0.00001$ from paired t-test receptively.

4. Discussion

No commercial posterior segment OCT provides anatomically correct images, which may affect the 3D interpretation of these images. We have proposed and tested an empirical nonlinear distortion correction approach for OCT devices that use a fan-scanning pattern. Our method does not require knowledge of the device’s optical system. We used two reference markers to establish our distortion correction method: the known diameter of implanted glass beads and the retinal curvature obtained by MRI, an imaging technique that is not impacted by optical distortions.

Our method is based on the assumption that all A-scans of an OCT scan pass through one pivot point. While the refraction of scan rays at the optical surfaces of the eye and contact lens (if used) may introduce additional distortions, our approach can indirectly account for these distortions as long a unique pivot point is maintained. The transformation parameters were empirically obtained by minimizing the distortions of two reference markers (the bead diameter and RPE curvature) irrespectively of the underlying cause (fan-scanning, refraction, or other effects). However, our method cannot account for more complex distortions when scan rays deviate largely from a unique pivot point.
While existing distortion correction methods based on optical modeling can account for similar nonlinear distortions [14–21], these methods require detailed knowledge of the optical imaging device, the distance of the OCT device to the eye, and the optical details of the imaged eye. No simple technique exists to easily measure the distance between the eye and the OCT camera with enough accuracy to inform an optical model. The optical model of the eye will require additional measurements of the eye’s optical surfaces and axial length. Finally, the optical details of commercial OCT devices are typically not disclosed for proprietary reasons. Consequently, optical modeling based approaches remain primary an option for custom built posterior segment OCT systems and laboratory settings, where all these parameters can be controlled and/or measured. Recently, Zhou et al. have introduced optical coherence refraction tomography (OCRT) [26], a new imaging modality that acquires OCT images with angular diversity to increase lateral resolution and correct refraction-induced distortions. However, the clinical application of OCRT will require specialized OCT devices [27], which are currently unavailable. Our proposed correction method is based on an empirical approach and does not require a specialized OCRT device or knowledge of the optical system of the OCT device. In a previous study, Garcia Garrido et al. [11] also used an empirical approach, but this approach was based on linear scaling of OCT images obtained in mice. In contrast, our approach accounts for nonlinear distortions that better account for the fan scanning used by posterior segment OCT devices. While Garcia Garrido et al. were able to derive the linear scale factors from implanted glass beads with known diameters, our approach required an additional reference marker (retinal curvature) to determine the transformation parameters.

In studies that investigate ONH morphology, segmented BMO points are often used to define a reference plane, which are then used to measure changes at the ONH such as LC depth. Johnstone et al. [28] have illustrated that LC depth metrics relative to a BMO plane can confounded by changes in choroidal thickness. To overcome this issue, we use ASCO points to define a reference plane in our tree shrew OCT scans, where ASCO is visible. In some human studies an alternative reference plane based on peripheral anterior scleral points has been utilized [6,28,29]. Our results show that peripheral depth or position metrics are increasingly sensitive to nonlinear distortions caused by fan-scanning the further away you are from the scan center. Consequently, using the peripheral anterior scleral plane as reference plane can remove bias due to choroidal thickness changes, but it may increase bias due to nonlinear distortions caused by fan-scanning. Our nonlinear distortion correction method can reduce this potential bias.

It is typically believed that nonlinear distortions in posterior segment OCT imaging are negligible at the center of the B-scan [15]. Our results in tree shrews show that small changes in camera position (3 mm) can significantly alter the anterior LC depth (Fig. 9). This morphological parameter is commonly investigated in glaucoma studies [2,6,29–39] and obtained near the center of the B-scans. Our studies show that linear distortion correction may under- or overestimate the lamina cribrosa depth changes from baseline depending on the camera position. Consequently, nonlinear distortions may bias morphological studies of the ONH including parameters that are obtained at the center of the scan. Our nonlinear distortion correction method was able to reduce this bias in the case of the LC depth metric. This finding underlines the need for nonlinear distortion correction to avoid bias in studies that investigate the 3D morphology of the ONH in small animal eyes. A similar bias is likely to occur in human posterior segment OCT, but the magnitude of this effect remains to be investigated.

Previous studies have shown that posterior eye shape measurements such as retinal curvature can be significantly biased when nonlinear optical distortions remain uncorrected in posterior segment OCT scans of human eyes [15–18]. No human study has carefully evaluated the effect of optical distortions on 3D morphological metrics of the ONH or fovea in posterior segment OCT. Morphological parameters that incorporate peripheral image information or that are sensitive to the “bending” of the B-scan OCT image are likely impacted by the here reported nonlinear
optical distortions. Our results are alarming and urge for a careful investigation of such potential bias in posterior segment OCT studies of human eyes. Note that our distortion correction method can be applied to OCT angiography scans, where nonlinear optical distortions may bias retinal vasculature metrics in currently unknown ways.

We were able to identify clear correlations between two scan parameters (RAP and scan focus) and the nonlinear transformation parameters \( \kappa_u \) and \( r_P \). With these correlations in hand, the meta data of the OCT scans alone, without the need to implant additional glass beads or to use additional MRI scans, can be used to inform our nonlinear distortion correction method. Further, with these correlations established, our nonlinear distortion correction method can be used retrospective and be applied to existing OCT data. These correlations were obtained from a tree shrew population with a wide range of axial lengths (7.37 to 8.07 mm). However, no obvious correlation was found between axial length and the nonlinear transformation parameters in our study. This observation needs to be reevaluated in humans, where axial length differences are much higher across individuals than in tree shrews.

At \( \kappa_u = 0 \), our approach predicts that the pivot point position \( r_P \) moves to infinity representing a special case in our nonlinear approach \( 5 \). At \( \kappa_u = 0, r_P = \infty \), our approach predicts that the OCT camera is positioned such that the pivot point is outside the eye and the refractive interfaces of the ocular tissues and the contact lens act as a collimating lens. If the camera is moved even further way from the eye, our model predicts that \( \kappa_u \) and \( r_P \) become negative suggesting that the "theoretical" pivot point is behind the eye and the image should be "bent backward" to recover the correct retinal curvature. This situation is illustrated in Fig. 10C. We were able to reproduce this extreme case as shown in Fig. 10A,B, where the RPE curvature of the linear scaled OCT image was higher than the retinal curvature of the matching MRI scan. Our nonlinear approach predicted negative values for \( \kappa_u, r_P \) for this scan and was able to produce a realistic RPE curvature that matched the MRI curvature.

![Fig. 10. Extreme case with predicted pivot point behind the eye. A: OCT B-scan with sub-retinal glass bead after linear distortion correction showing a segmented RPE curvature (broken red line) with higher curvature compared to the retinal curvature obtained from the matching MRI scan (broken yellow line). B: The nonlinear distortion correction method predicted a "backward bending" of the image and produced a realistic RPE curvature that matched the MRI-based curvature. The scan parameters of this image were RAP = 6.541 mm, scan focus = 18.95 D resulting in negative transformation parameters \( \kappa_u = -0.426 \) and \( r_P = -12.56 \) mm. C: Illustration of the extreme case, where the true pivot point was placed in front of the eye, but the location of the "theoretical" pivot point was predicted to be behind the eye.](image-url)
Our nonlinear distortion correction method was derived for single B-scans and radial scan patterns. However, the method can be translated to cube scans if two scan angles can be extracted from the OCT data. In the case of the Spectralis OCT2, two scan angle locations can be obtained for each B-scan in pixel coordinates \((u, v)\) from the calibrated scanning laser ophthalmoscope image that is taken simultaneously during each OCT scan. Let’s assume that the \(u\)-axis represents the scan direction within each B-scan and \(v\)-axis represents the perpendicular direction, where B-scans are recorded at equidistant increments \(\Delta v\) apart from each other (Fig. 11A). Consequently, a point \(P\) within a B-scan can now be described with three pixel coordinates: \(u, v,\) and \(w\). Assuming axisymmetric distortions around the center of the cube scan grid \(O\), the nonlinear distortion correction (1) and (2) can be rewritten for cube scans into

\[
\begin{align*}
    u' &= u - \frac{u_{\text{max}}}{2}, \quad v' = v - \frac{v_{\text{max}}}{2}, \quad w' = w - \frac{w_{\text{max}}}{2} \\
    \alpha &= \sqrt{u'^2 + v'^2} \frac{\alpha_{\text{tot}}}{u_{\text{max}}} k_u, \quad \beta = \arctan\left(\frac{u'}{v'}\right), \quad r = r_p + w' k_w \\
    x &= r \sin(\alpha) \sin(\beta), \quad y = r \sin(\alpha) \cos(\beta), \quad z = r - r \cos(\alpha).
\end{align*}
\]

(6)

**Fig. 11.** Nonlinear distortion correction of an OCT cube scan centered at the ONH. A: Scanning laser ophthalmoscope image showing the 30°×25° scan region (green box). 241 B-scans were obtained in equidistant increments \(\Delta v\) along the \(v\)-axis. \(P\) and \(O\) represent a point within a B-scan and the center of the cube scan pattern, respectively. B: 3D reconstruction of the auto-segmented tissue interfaces after nonlinear distortion correction.

Figure 11B illustrates the application of our nonlinear distortion correction to a cube scan consisting of 241 B-scans centered at the ONH. Note that the image points of one B-scan are no longer aligned within a 2D plane after nonlinear distortion correction using (6) making the reconstruction of distortion corrected OCT images more difficult. It should be noted that many conventional OCT systems utilize two off-set scanning mirrors to generate volumetric image data. When this configuration is combined with an imaging telescope, each mirror is imaged to its own distinct pivot point along the imaging optical axis. Unfortunately, system design details such as distance between galvo mirrors, the imaging telescope magnification, and the resulting pivot point distance differential are typically unavailable to users of a commercial OCT system. While many OCT systems minimize this pivot distance and our technique optimizes the pivot point to be between these two points, this effect was not investigated here, and its consideration would require a modification of (6).

While our distortion correction approach was developed and tested in tree shrews using the Spectralis OCT, it is applicable to other species and OCT devices. When using our nonlinear distortion correction method in another species or with a different OCT device, the distortion correction parameters have to be reevaluated due to the empirical nature of the approach. We used implanted glass beads and retinal curvature as reference markers. While glass bead implantation
is not feasible in human eyes, alternative anatomical reference marks could be used to derive the transformation parameters. The distance between the fovea and ONH could replace the glass bead diameter as reference marker if this distance can be measured reliably and without distortions using an alternative imaging modality. Note that the tree shrew retina has an area centralis without a pit-like anatomy [40] and, therefore, the fovea could not be utilized as a reference marker in this study.

Within this study, we have discovered two specific findings related to the Spectralis OCT. First, the Spectralis records the scan parameters as meta data, which allowed us to discover the correlations (5). We have noticed during extensive imaging sessions that the RAP value can change from one scan to another without changing any other parameter including camera and eye position. We assume that a slippage of the reference arm causes this inconsistent recording of the RAP value. We imaged a custom reference target before and after each imaging session to account for a potential slippage of the reference arm in our study. Prior to the discovery, we imaged a different cohort of animals with implanted glass beads and matching MRI, but without the validation scan of our reference target. These data set resulted in a poor correlation ($R^2 = 0.535$) between $k_u$ and the scan parameters caused by the inconsistent recording of RAP. Second, the Spectralis can perform relative nonlinear transformation of follow-up scans by matching them to a baseline scan. This nonlinear transformation is unlikely based on an optical distortion correction approach. While this nonlinear transformation may simplify the subjective identification of local changes in the follow-up scan, it can introduce additional bias when studying 3D morphological parameters. An accurate nonlinear distortion correction strategy is needed to avoid unintended bias in longitudinal studies that investigate morphological changes.

In summary, we have presented the first empirical approach to correct for nonlinear distortions in posterior segment OCT devices using a fan-scanning technique. Our method does not require knowledge of optical details of the OCT device and, therefore, can be utilized for commercial OCT systems with undisclosed technical details. Nonlinear distortions may bias the 3D interpretation of posterior segment OCT scans, but the correction of these distortions has been widely neglected due to a lack of feasible methods. Our method can effectively reduce this potential source of bias.

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**Data availability.** Data underlying the results presented in this paper are not publicly available at this time but may be obtained from the authors upon reasonable request.

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