

Calibration-free retinal-anterior eye-tracker for high-speed, high-accuracy, and high-precision eye movement estimation during fixations, saccades, and smooth pursuit

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Abstract

Eye-tracking is increasingly prevalent across various aspects of our lives. It is currently used for research, education, diagnostics, and entertainment, for example, in virtual reality goggles. However, widely used video-based eye-trackers require subject-dependent calibration, which makes them unsuitable for examining individuals with unstable fixation, such as patients with neurodegenerative diseases. We present a novel calibration-free retinal-anterior eye tracker that uses retinal images to track changes in eye orientation. On an artificial eye, the eye tracker achieves an accuracy better than $0.6'$ (0.01°) and a sample-to-sample precision of $0.09'$ (0.0015°) for the reference-frame signal, improving to $0.05'$ (0.0008°) after Kalman fusion. Under similar conditions, this precision exceeds that reported for video-based eye trackers. After a one-point trajectory offset correction, the eye tracker enables determination of the absolute gaze direction. The eye-tracker was tested with an artificial eye over a traverse of -8° to $+8^\circ$. In human recordings we typically work within approximately $\pm 4^\circ$, a practical limit based on operating experience rather than on a systematic characterization. The eye-tracker is capable of measuring saccadic velocities of up to $300^\circ/\text{s}$, incorporates an integrated display for dynamic stimulus presentation, and requires only a simple chin- and forehead rest for subject stabilization. The potential of this eye-tracker is demonstrated in healthy subjects and in participants who are unable to undergo a conventional calibration procedure. Additionally, the device is equipped with a synchronized video-based eye-tracker, making it a powerful research platform for developing eye-tracking technologies.

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Introduction

Humans, as well as many other animals, have foveated visual systems, which means that spatial resolution varies across the visual field¹. Maximum visual acuity is achieved only in a small, approximately one-to-two degree central retinal area known as the fovea^{2,3}. The direction of sharp central vision can be consciously shifted between objects by voluntary eye movements, including smooth pursuit eye movements that allow tracking moving objects, or saccades that allow for quickly changing the gaze direction⁴. However, even during fixation, involuntary drifts and microsaccades constantly occur, which move stimuli across photoreceptors' receptive fields, preventing retinal image fading^{5,6} and providing visual hyperacuity⁷.

The neural control of eye movements is a complex process that involves numerous brain structures, including the cerebellum, basal ganglia, brainstem, and various cortical areas⁸. The measurement and analysis of eye movements can therefore provide insight into human cognition and central nervous system function⁹, enabling a better understanding of the principles of operation of the entire visual system, including oculomotor control mechanisms and the sources of their disorders, such as strabismus^{10,11}. Eye movement analysis is also widely used in psychological and cognitive sciences in research on, for instance, visual attention^{12,13}, perception^{14,15}, information processing^{16,17}, and consumer search-choice processes^{18,19}. Eye tracking can also be used in medicine as a tool for diagnosing neurodegenerative diseases such as glaucoma^{20,21}, Parkinson's disease^{22,23}, and Alzheimer's disease^{24,25}, as well as mental diseases such as depression^{26,27} and autism spectrum disorders^{28,29}. Importantly, eye movement measurement is much quicker and easier to perform than techniques such as electroencephalography or functional magnetic resonance imaging, which are traditionally used to assess brain activity.

Several methodologies have been developed to measure eye movements, which can be broadly categorized into invasive and non-invasive techniques. Invasive methods, such as the optical lever^{30,31} and search coils^{32,33}, offer high precision and accuracy but require direct contact with the eye. Non-invasive techniques primarily rely on tracking Purkinje reflections, which are light reflections from the optical surfaces of the eye. A commonly used method is the Pupil-Corneal Reflection (P-CR) tracker, which employs the pupil and the first Purkinje reflection (P1, from the anterior corneal surface)^{34,35}. Another example is the Dual Purkinje Images (DPI) tracker, which utilizes the P1 reflection, and the fourth Purkinje reflection (P4, from the posterior lens surface)^{36,37}. These DPI and P-CR eye-trackers provide measurements of eye movements over a relatively large angular range; however, they also require subject-dependent calibration, during which the subject is asked to look at visual targets presented at known locations^{38,39}. This procedure allows transforming the observed position of features in the eye images to a physical eye orientation and must be performed before each measurement session. Subject-dependent calibration is influenced by changes in pupil size or lens shape⁴⁰, and is not constant over time⁴¹. This, in turn, fundamentally limits the accuracy of anterior eye-trackers to approximately 30°⁴² and 10°³⁷ for P-CR and DPI eye-trackers, respectively. Additionally, the calibration procedure is based on the assumption that the subject is able to accurately fixate the calibration targets, which cannot be expected to be achieved in people with vision and eye movement disorders, as well as in people with communication difficulties due to, for example, neurodegenerative diseases.

Another family of non-invasive eye-tracking systems, retinal eye-trackers, measures eye movement through imaging the displacement of the retina⁴³. Retinal images have fixed angular sizes independent of geometrical eye properties and head motion, so the translation of retinal features in the image is directly proportional to eye-in-head rotation. As a result, to estimate eye rotation, they do not require subject-dependent calibration, thus offering superior accuracy in tracking gaze displacements (even less than 1°⁴⁴ to anterior trackers. One possible implementation of a retinal eye-tracker is the tracking scanning laser ophthalmoscope (TSLO)⁴⁵ or adaptive optics scanning laser ophthalmoscope (AOSLO)⁴⁶. In this approach, each SLO frame is divided into smaller strips that are parallel to the scanning beam. Eye movements are determined based on the shifts of individual strips relative to a retinal image used as a reference frame. The main constraint of this techniques is the low maximum velocity ($\sim 7^\circ/\text{s}$ for AOSLO and $\sim 28^\circ/\text{s}$ for TSLO) that can be tracked, which is strictly dependent on the system's field

of view⁴⁵. Beyond this limit, the correlation between the strip details and the reference frame is insufficient to reliably ascertain eye movement. Due to this limit, these eye-trackers are only suitable for accurately tracking the low-velocity parts of fixational eye movements (i.e. drift) and cannot be used for complex visually-guided tasks, such as saccades with greater amplitude and consequently higher velocity.

Another approach to retinal eye-tracking is the FreezEye tracker (FET) presented earlier by our group⁴³. It is also based on the SLO principle but, in contrast to the TSLO and AOSLO devices, it images a part of the retina with a relatively small field of view ($3.37^\circ \times 3.24^\circ$) at a high frequency of 1240 Hz. The FreezEye tracker utilized a back-illuminated static stimulus to present tasks to the user, which, due to the inability to present dynamic stimuli, effectively restricted the evaluation of eye movements solely to fixation. Moreover, the FreezEye tracker's scanning beam was visible ($\lambda=785$ nm), which could interfere with the presented visual stimuli. These two constraints significantly hindered the device's application in large-scale clinical measurements.

To address these limitations, we modified and improved the FreezEye tracker. Our present eye-tracker, neuroFET, scans $5.3^\circ \times 2.8^\circ$ portion of the retina at a frequency of 620 Hz. The device utilizes an imaging light beam invisible to the human eye ($\lambda=920$ nm), which, combined with an internal digital display, allows for the presentation of moving stimuli undisturbed by the scanning beam. Furthermore, neuroFET features an integrated video-based eye-tracker, permitting the concurrent capture of signals from three types of eye-tracking technologies: DPI, P-CR and retinal.

In this paper, we present an evaluation of the neuroFET device, a novel calibration-free retinal-anterior eye-tracker. First, we show that the retinal part of the device allows for tracking eye movement with high precision ($0.09' = 0.0015^\circ$) and accuracy ($0.6' = 0.01^\circ$) as determined by a laboratory tool simulating a rotating eye. Second, we showcase its ability to track eye movements across a wide range of tasks by conducting recordings with human participants using visual tasks designed to elicit saccades and smooth pursuit. Third, since the retinal eye-tracker does not require calibration, the device is suitable for recording from participants who are unable to complete traditional calibration procedures, such as patients with neurodegenerative disorders. To show the suitability of the neuroFET for studying such populations, we present the trajectory of eye movements recorded by our eye-tracker in a subject affected by Parkinson's disease. Finally, we also demonstrate that simultaneously recording eye movement data from the anterior surfaces of the eye and the retina within a single device enables calibrating video-based eye-trackers (P-CR and DPI) without requiring the participant to perform any specific task, or even requiring the ability to stably fixate, by using the retinal eye-tracker's signal to inform how the participant's eyes moved.

The neuroFET offers broad possibilities for applications in both research studies and clinical use. The application of this device extends beyond mere assessment of fixation stability, a notable limitation of previous retinal eye-tracking technologies. It also allows for the examination of more complex eye movements, such as saccades and smooth pursuit movements. The tracking range was tested with an artificial eye over a traverse of -8° to $+8^\circ$. However, in human recordings we work within approximately $\pm 4^\circ$, a practical limit arising primarily from the diameter of the undilated pupil and based on our operating experience rather than on a systematic characterization. Given that its calibration-free design makes it possible to rapidly acquire data from a various groups of participants, whether healthy or not, we think that this novel technology opens new possibilities for developing rapid, non-invasive methods to diagnose and monitor the progression of central nervous system diseases, including Parkinson's disease.

Results

Determination of the accuracy and precision of the neuroFET eye-tracker

The neuroFET eye-tracker was characterized using two approaches: first, by rotating an artificial eye using a rotation stage or simulating rotation by deflecting the imaging beams with galvanometric scanners, and second, by analyzing the rotations of the human eye in real-life scenarios.

Measurements conducted on an artificial eye indicated that the precision remained approximately constant throughout the measured rotation amplitude range of up to 4° , not surpassing $0.09'$ for the reference-frame signal (further reduced to below $0.05'$ by the Kalman fusion), which is superior to the precision achieved by the digital DPI eye-tracker³⁷ when recording data from an artificial eye ($0.142'$). The same experiments also showed that the accuracy with which the retinal part of the eye-tracker recovers a commanded displacement is constant across the measured range — peak excursions from the central position of up to $\pm 4^\circ$, i.e. commanded displacements up to 8° — and does not exceed $0.6'$, with a mean value of $0.2'$ (corresponding accuracy reported for the digital DPI eye-tracker reaches $10'^{37}$). In the second phase of the experiment, the accuracy of assessing the movement velocity was evaluated. To achieve this, the artificial gaze rotation followed a triangular path. The horizontal velocity estimation maintained high accuracy with deviation within $\pm 1^\circ/\text{s}$ from the true value across a broad range from $1^\circ/\text{s}$ to $237^\circ/\text{s}$, while vertical velocity estimation preserved this accuracy up to $56^\circ/\text{s}$, after which inaccuracies increased, reaching 2% deviation from the true value at $75^\circ/\text{s}$ and 12% at $177^\circ/\text{s}$ (compared to 1% at $75^\circ/\text{s}$ and 0.7% at $177^\circ/\text{s}$ for the horizontal direction). The reduced accuracy in measuring velocity along the vertical movement direction is due to the vertical operation of the slow axis of the MEMS scanners. Detailed results of the accuracy and precision assessments of the artificial eye are provided in Supplementary Materials – Supplementary Figures S1–S7.

In the second experiment, the accuracy of the retinal, P-CR, and DPI eye-trackers was evaluated using measurements from a trained, healthy human subject who performed a task of looking at a sequence of points at 1 degree eccentricity. Because the true line of sight of a human eye is not accessible to any independent ground-truth measurement, the distance between the reported gaze position and the visual target quantifies the combined error of the instrument and of the subject's own oculomotor behaviour (fixation offset relative to the target, drift, microsaccades, and the individual preferred retinal locus). The values are reported to compare the three tracking technologies recorded simultaneously from the same eye — a comparison in which the subject's oculomotor error is common to all three signals — rather than as a measure of instrument accuracy. The DPI and P-CR signals were calibrated using two approach. In the classical approach, the fragments of recordings in which the subject fixates on the visual target serve as reference points (calibration “to target”). In the novel method, we use the offset-corrected retinal signal as the reference points (calibration “to FET”). See Supplementary Material 2 section for more details. Next, for each of the obtained signals, we calculated the distance between the eye and target positions. The accuracy for each signal is defined as the most frequent value (or mode) of the calculated distance distribution.

The results for the healthy, trained subject indicate that the retinal eye-tracker achieved the highest accuracy in assessing gaze rotation, with a mode of $5.3'$. Both calibration methods produced comparable outcomes: DPI signals yielded $6.5'$ for “to FET” and $7.3'$ for “to target,” while P-CR signals showed $14.6'$ for “to FET” and $12.3'$ for “to target.” Distance distribution graphs are provided in Supplementary Figure S8. The results show that DPI and P-CR tracking can be performed without additional calibration when combined with retinal tracking, providing high accuracy and precision in measuring human eye movements. The advantage of the novel approach is that it provides calibration data for arbitrary eye rotations, as it makes no assumption that the subject is looking at the target.

Recording saccades and smooth pursuit in healthy subjects using the retinal eye-tracking technique

Figure 1 shows representative data recorded from a 40-year-old healthy participant using the retinal eye-tracker during two types of tasks: reflexive saccades and smooth pursuit, which aim to illustrate the qualitative reliability of the retinal component of the eye-tracker. Further examples of gaze trajectories captured from a 28-year-old and a 75-year-old are presented in Supplementary Figures S9 and S10. Currently, gaze trajectories have been effectively recorded in roughly 100 healthy individuals. In healthy participants, we expect to see saccades directed from the center target to the peripheral targets. From previous literature, it is known that saccades are typically curved in two-dimensional space⁴⁷, and first undershoot the target slightly before making a corrective saccade to get closer to the target location⁴⁸. We also expect to see within-participant variation in both saccade curvature and saccadic accuracy⁴⁹. During slow smooth pursuit (target velocity $< 30^\circ/\text{s}$), we expect healthy participants to accurately track the predictably moving target while performing only occasional saccades to catch up. However, at a higher target speed, smooth pursuit accuracy is expected to worsen significantly, as participants are unable to maintain gaze on the target⁵⁰.

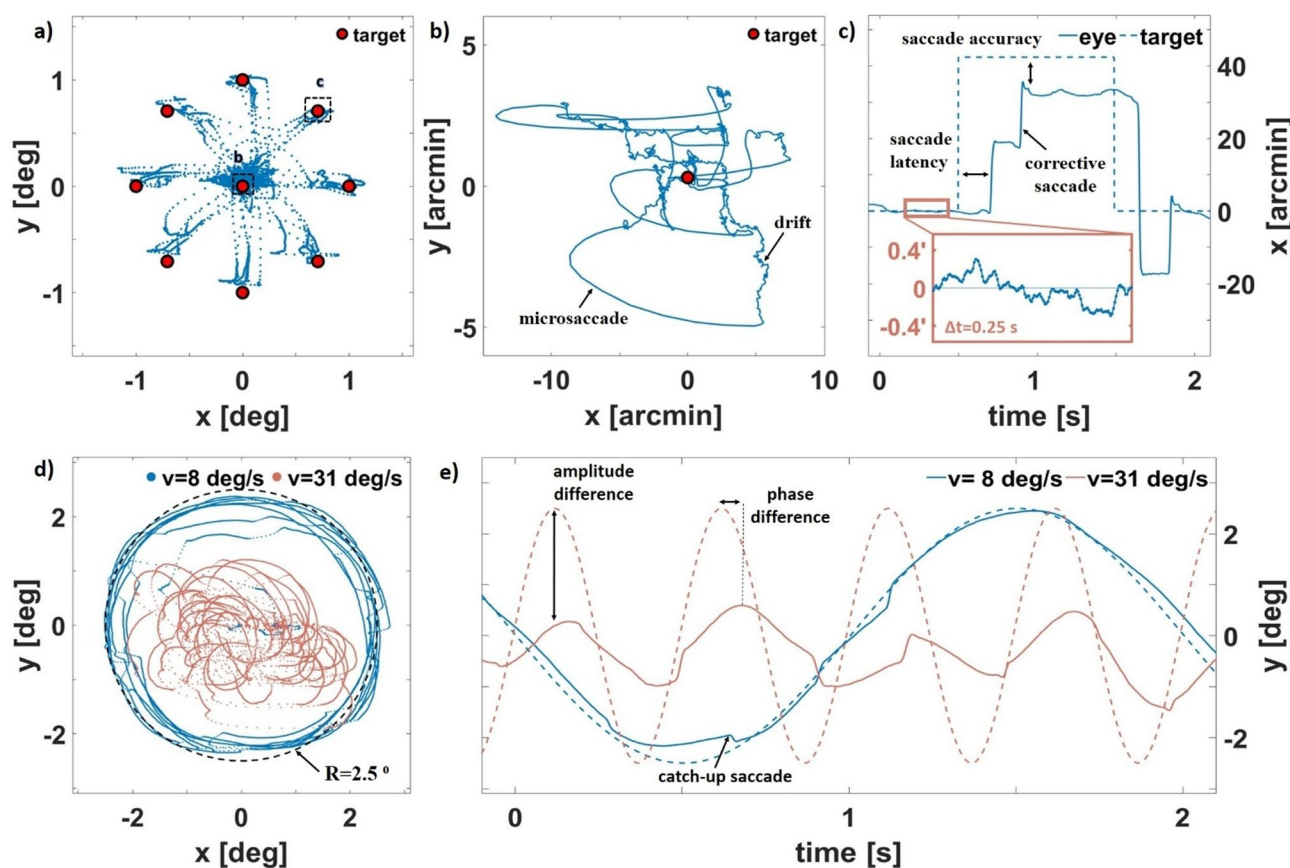


Fig. 1 Gaze position recorded at a sampling frequency of 620 Hz using a retinal eye-tracker in a healthy subject, in response to a moving stimulus. **(a)** Two-dimensional representation of stimulus locations (red points) and gaze position (blue dots) in an experiment with stimulus jumping to eccentric positions and back, resulting in 1-degree reflexive saccades in various directions. **(b)** An enlarged view of a one-second excerpt of central fixation. **(c)** Horizontal shift of the stimulus (indicated by the dashed line) and the gaze direction (represented by the solid line) during a single sequence of stimulus movement from the midpoint to the periphery and returning to the midpoint. The red inset zooms in on a fixation segment, where gaze direction varies within the 0.8° range. **(d)** Two-dimensional gaze trajectories in response to a stimulus (marked with a dashed black line) smoothly moving on a circular path with a 2.5° radius at a velocity of $8^\circ/\text{s}$ (blue dots) or $31^\circ/\text{s}$ (red dots). **(e)** A two-second fragment of the gaze's trajectory (solid lines) while pursuing the stimulus (dashed lines), moving at a velocity of $8^\circ/\text{s}$ (blue dots) or $31^\circ/\text{s}$ (red dots).

Figure 1a illustrates the two-dimensional representation of eye position (represented by blue dots) during the execution of 1-degree saccades guided by a target that jumped between the center and eccentric locations (marked by red dots). An animation depicting the images captured during this task from the pupil camera, the retinal imaging system, and the reconstructed gaze trajectory is provided in Supplementary Video S1. The gaze position direction matches the target position on the screen in each saccade direction, which is expected for a healthy, well-rested subject. The black boxes indicate the areas corresponding to the central fixation (b) and a single saccade (c), which are presented in an enlarged view in Figs. 1b–c. In Fig. 1b, micro-eye movements are shown, with amplitudes of less than $20'$ in the horizontal direction and $10'$ in the vertical direction. Example episodes corresponding to micro-saccades and drift have been marked with arrows. The change in the horizontal component of gaze direction during a single saccade is presented in Fig. 1c with a continuous line. The dashed line shows the target trajectory. An enlarged pre-saccadic fixation is shown in the box, where the gaze position varies within the $0.8'$ range. Furthermore, the diagram identifies essential parameters of the saccade, denoted by arrows: latency (the time interval from stimulus onset to saccade onset) and saccade accuracy (the discrepancy between the target position and the saccade landing point). The execution of the presented saccade demonstrated a biphasic pattern: after initially undershooting the target, a corrective saccade occurred, as indicated by an arrow.

Figure 1d illustrates the gaze position signal when tracking a target that moved along a circular path with a radius of 2.5° at linear velocities of $8^\circ/\text{s}$ (represented by blue dots) and $31^\circ/\text{s}$ (indicated by red dots). A two-second segment of the gaze and target trajectories in the vertical plane is depicted in Fig. 1e, where solid lines denote the gaze position and dashed lines represent the target position. The gaze trajectory was closely aligned with that of the target at the lower target velocity. However, small catch-up saccades made when retinal slip becomes too large are also clearly seen in the signal. On the other hand, when the target moved too fast to be followed using smooth pursuit movements, the gaze trajectory becomes irregular and jagged, revealing discrepancies in both amplitude and phase between the eye and target trajectories. This phenomenon is depicted in Fig. 1e for a target velocity of $31^\circ/\text{s}$ and is associated with the visual system's constraints on smoothly tracking objects moving at comparatively high velocities. As shown in Fig. 1, the precision of the retinal eye-tracker facilitates, on one side, a detailed examination of eye micro-movements (Figs. 1a–c), and on the other side, a detailed evaluation of the difference between the trajectory of the gaze and the target trajectory when the eye is continuously moving at a significant speed (Figs. 1d–e).

Scenarios where retinal eye-tracking enables accurate eye movement measurement, while anterior techniques face challenges

The eye movement trajectories presented in Fig. 1 were obtained from a young, well-rested, healthy subject. In eye-tracking research, this situation is considered the best-case scenario, as identifying the pupil centers or the Purkinje images P1 and P4 from the pupil camera images is relatively straightforward. However, in practical applications, particularly when assessing eye movements among elderly subjects, the frequent presence of droopy eyelids may hinder visibility of parts of the pupil and P4, thus complicating the acquisition of P-CR and/or DPI signals. Additionally, when utilizing video-based eye-trackers, it is crucial that the patient is able to precisely follow the visual target throughout the calibration process, which can also pose challenges in real-world scenario, particularly for individuals with diminished perceptual or motor skills due to illness or fatigue. Figure 2 depicts examples where video-based eye-tracking is infeasible, but gaze trajectories can still be determined using retinal eye-tracking. Figure 2a shows a single frame captured using retinal and video-based eye-tracking during measurement of a subject who has undergone cataract surgery and has an implanted artificial lens. The geometry of the intraocular lens differs from that of the natural crystalline lens, making P4 invisible within the pupil. Therefore, the required features for eye-tracking using the DPI technique are not available, even though those for the P-CR and retinal techniques are clearly detectable. Figure 2b shows pupil camera and retinal images obtained from a person with a droopy eyelid. In the presented case, the pupil is significantly obscured by the upper eyelid, preventing the use of any of the video-based eye-tracking techniques. Nevertheless, the scanning beam is able to

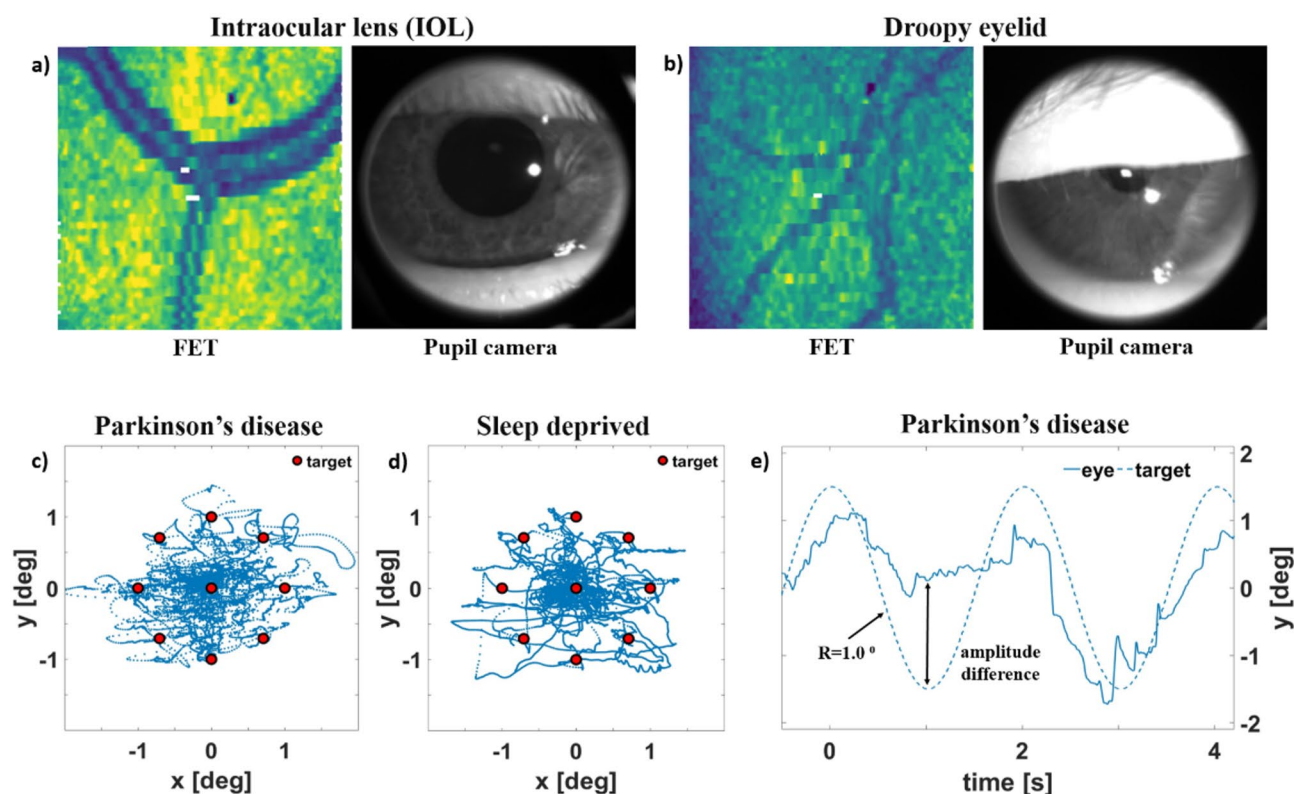


Fig. 2 Examples of retinal eye-tracking applications in circumstances that preclude the use of video-based eye-tracking techniques. **(a), (b)** Retinal and the pupil camera images obtained from subjects with an implanted intraocular lens and a droopy eyelid, where it is not feasible to determine the DPI or both DPI and P-CR signals, while still preserving the detectability of the FET signal. **(c), (d)** Two-dimensional representations of gaze direction in a patient with Parkinson's disease and a sleep-deprived subject while performing 1-degree visually-guided saccades, showcasing their inability to effectively perform the standard calibration procedure applied in video-based eye-tracking techniques. **(e)** Four seconds of vertical (y-axis) gaze direction recorded from a patient with Parkinson's disease using a retinal eye-tracking during smooth pursuit movement, demonstrating notable discrepancies between the gaze and stimulus position.

enter the eye through the remaining part of the pupil, and thus the retinal image remains detectable, permitting the utilization of retinal eye-tracking. However, this does not imply that retinal tracking will always remain possible in every case of severe ptosis. Figures 2c-d illustrate the two-dimensional gaze trajectories acquired through retinal eye-tracking for two subjects: one diagnosed with Parkinson's disease (Fig. 2c) and a sleep-deprived subject (Fig. 2d). Both individuals were asked to perform an identical task as the healthy person shown in Fig. 1, which was to look at a stimulus that moved from the center to one of the eccentric locations evenly spaced on a circle with a radius of 1 degree. At each point, the target remained stationary for 1.5 s. This task is comparable to the calibration procedure used in video-based eye-trackers. The typical concentration of gaze around fixation points, expected in healthy, rested individuals, such as the one shown in Fig. 1a, is not evident in the case presented in Fig. 2c-d. This implies that the measured subjects were unable to effectively follow the target or maintain their gaze in the intended direction. The inability of these participants to fixate on visual stimuli would preclude completion of a calibration procedure, making it impossible to track their eye movements using video-based eye-tracking techniques. Retinal eye-tracking, however, does not require patient-dependent calibration, which permits its application for evaluating gaze direction, including in subjects unable to perform calibration. Figure 2e illustrates four seconds of gaze direction variation along the vertical axis obtained utilizing retinal data for a patient with Parkinson's disease during the execution of a visually-guided smooth pursuit movement. As shown, the gaze trajectory is noticeably altered, revealing significant time-dependent differences in the amplitude

between the stimulus and the eye. Such observations are consistent with the known pathophysiology of Parkinson's disease, which is characterized by dysfunction of the basal ganglia, brain structures involved in the control of smooth pursuit movements⁵¹.

Comparison of synchronously recorded signals using the retinal, P-CR, and DPI eye-tracking techniques

The integration of retinal, P-CR, and DPI eye-tracking techniques into a single device enables simultaneous recording and direct comparison of signals from these eye-tracking technologies. Figure 3 demonstrates the gaze position and velocity for a well-rested, healthy subject during visually guided 1-degree saccades, concurrently recorded using retinal, P-CR, and DPI eye-tracking techniques. As described in the Methods section, we treat the retinal tracker signal as an indicator of the true change in gaze direction and subsequently use it to calibrate the signals from both video-based eye-tracking techniques. Figure 3a shows the gaze position recorded using each eye-tracking technique. The signal from the retinal eye-tracker data reveals a narrowly focused gaze direction around the fixation points for each target position. For both video-based eye-trackers, a wider variation in the measured instantaneous gaze direction surrounding the fixation points was observed, consistent with their diminished precision reported in previous sections. This is especially prominent with the P-CR eye-tracker, where it can be challenging to clearly differentiate the point cloud of gaze direction between central and peripheral target positions. Figure 3b depicts an alternative visualization of a 4-s segment of horizontal gaze position from data presented in Fig. 3. The enlarged fragment of the graphs shows 400 ms of data containing a single saccade. The black dashed line indicates the target location. The corresponding eye movement velocity is shown in Fig. 3c. The finest precision, observable as the smoothest representation of both the eye movement trajectory and its velocity, was obtained with the retinal eye-tracker, whereas the P-CR signal exhibited the highest level of noise. The diminished precision of the P-CR signal hinders or prevents the detection of tiny saccades. In Figs. 3b-c, the blue arrow illustrates an example of a microsaccade, detectable in the retinal and DPI signals but not in the P-CR signal. Simultaneous registration of gaze direction changes across diverse eye-tracking techniques enables direct comparison of these signals. The enlarged sections of Figs. 3b-c provide an example of using this functionality to compare the shape and key parameters of a saccade, such as amplitude and velocity, recorded with the retinal, P-CR, and DPI eye-tracking techniques. Notably, in Fig. 3b, this particular saccade captured by the DPI eye-tracker starts with an apparent reverse direction movement (backshoot), which is an artifact associated with lens lagging behind the physical rotation of the eyeball⁵², and results in overestimated saccade amplitude compared to the data from the retinal eye-tracker. The saccade velocity in the DPI signal (Fig. 3c) is also much higher. Also, in Fig. 3c, a second velocity peak (indicated by the red arrow) is distinctly observable for the DPI eye-tracker, appearing just after the maximum saccade velocity and connected with a strongly visible post-saccadic oscillation⁵³. This peak is less evident in the retinal eye-tracker data. Increased saccadic velocity relative to the retinal signal is also observable in the P-CR eye-tracker. Discrepancies in saccade characteristics across eye-trackers stem from movements of Purkinje reflections (P1 and P4), pupil center, and retina during saccades, as analyzed in our recent publication⁵⁴ using concurrent data from the device described in this article. These differences may result from DPI errors due to crystalline lens wobbling during saccades⁴⁸, which affects the iris and pupil, thereby impacting the P-CR signal.

Temporal and spectral characterization of the fused signal

The signal used in all human recordings, and the one shown in Fig. 3, is the Kalman-fused (KF) output. Because the Kalman fusion of the position and velocity estimates has intrinsic temporal-smoothing properties, the temporal behaviour of the fused signal was characterized directly (Supplementary Material 5). Direct temporal and spectral characterization of the Kalman-fused signal shows that the fusion introduces no detectable lag — on real 1° saccades the fused channel rises in 17.9 ms, against 19.0 and 17.5 ms for its pre-fusion inputs — and that it attenuates instrument noise rather than genuine movement, the KF/RF power ratio rising from 0.22 to 0.36 on

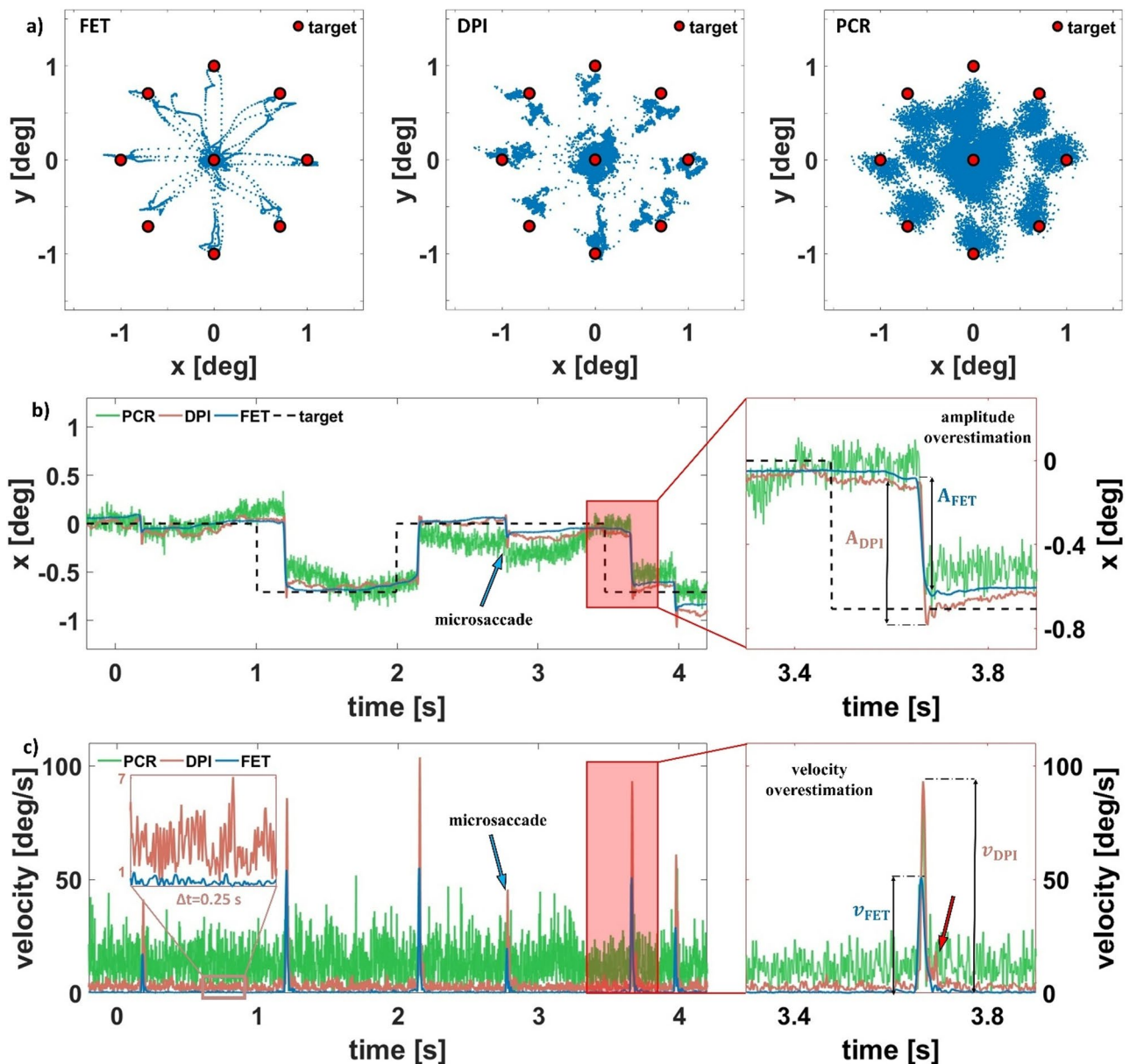


Fig. 3 Results of concurrent registration of eye rotations of a healthy, well-rested subject during performing 1-degree visually-guided saccades **(a)** Two-dimensional representations of the eye position alterations obtained utilizing FET, DPI, and P-CR eye-tracking techniques. **(b)** Horizontal component of eye rotation estimated using a retinal (blue solid line), DPI (red solid line), and P-CR (green solid line) eye-tracking techniques, with the enlarged area of a single saccade. **(c)** Corresponding eye movement velocity. In panels **(b)** and **(c)**, the differences in the determination of the eye movement amplitude and velocity values using the eye-tracker FET and DPI are highlighted.

a stationary artificial eye (pure noise) to 1.03–1.08 during human fixation with microsaccades. The fused signal stays above the instrument noise floor up to ~ 90 Hz and is the least noisy of the three simultaneously recorded modalities (Fig. S16). The smoothness of Fig. 3 is therefore not the result of a slowed or over-filtered signal.

Discussion

In this article, we present a calibration-free retinal-anterior eye-tracker, neuroFET, equipped with an internal display and suitable for clinical use. We showcased the capability of utilizing the device to accurately and precisely determine changes in gaze direction during psycho-visual tasks, applicable to both healthy, well-rested subjects (Fig. 1) and those who are unable to complete the calibration procedure necessary for video-based eye-tracking, for example, patients with neurodegenerative diseases (Fig. 2). We also showed that the synchronous recording using three distinct eye-tracking techniques—retinal, DPI, and P-CR—within a single device eliminates the requirement for patient-dependent calibration of the video-based eye-trackers, and offers more detailed insights into the movement of the entire eyeball in response to a moving stimulus (Fig. 3).

The presented device offers superior accuracy and precision compared to video-based eye-tracking systems. It eliminates the need for patient-dependent calibration prior to measurements. Moreover, it enables data acquisition from both healthy and diseased subjects using only chin and head rests, without complex head immobilization such as bite bars. The retinal imaging module can detect changes in gaze direction with a precision of $<0.09^\circ$ and an accuracy of $<0.6^\circ$ during simulated eye movements. These specifications significantly surpass those documented for the leading laboratory video-based eye-trackers, which report a precision and an accuracy of $\sim 0.14^\circ$ and 10° , respectively³⁷. High precision and accuracy are critical requirements for eye-trackers, especially in applications that measure micro-movements such as drifts and micro-saccades, whose amplitudes may be less than 1° ⁵⁵.

A characteristic feature of retinal eye-trackers, which distinguishes them from video-based eye-trackers and significantly enhances their accuracy, is the elimination of subject-dependent calibration. In typical video-based eye-trackers, the accuracy of the entire measurement is determined by calibration quality, which can be significantly reduced when patients are unable to maintain stable fixation (e.g., individuals with neurological disorders or simply fatigue, as illustrated in Figs. 2c–e)⁵⁶. Furthermore, eye-tracking using video-based techniques may be entirely unfeasible in scenarios where features such as the pupil center or Purkinje reflections are not detectable (as in the case of subjects with an implanted artificial intraocular lens or those with a droopy eyelid, as shown in Figs. 2a–b). On the other hand, retinal trackers do not require subject-dependent calibration to estimate change in gaze direction, as they calculate rotation angle directly by determining displacements between consecutive retinal images in combination with a known and constant angular extent of the scanning beam. Importantly, while the constant angular size of the imaged retinal region may correspond to different geometric extents due to factors such as different eye sizes, it is effectively constant within each individual during tracking. The calibration-free property arises from the confocal imaging geometry, which images the retina with a nearly constant angular magnification for a given subject, so that the displacement of retinal features maps directly to eye rotation without subject-specific calibration. The equivalence between rotating the eye and deflecting the imaging beam — used to validate the instrument on an artificial eye — rests on the pupil conjugation of the optical system: because every scanning element is conjugated to the eye pupil plane, physical rotation and beam deflection change the beam angle inside the eye in the same way, and the lateral pupil displacement that accompanies rotation acts on the aperture available to the beam rather than, to first order, on the angular scale of the measurement. This equivalence was verified experimentally over the tested range of -8° to $+8^\circ$: the difference between the two conditions did not exceed 2.2° , remaining within the rotation stage's specified bidirectional accuracy ($\pm 2.82^\circ$) throughout, so it is bounded by the mechanical reference rather than by the tracker, and rotations of the eye can be modelled by imaging-beam deflections at the arcminute level (Supplementary Material 1, Experiment #1). Minor deviations in apparent retinal image size are primarily attributable to the natural curvature of the retina. Across the traverse

from -8° to $+8^\circ$ tested with a single artificial eye — the nominal range being $\pm 7.5^\circ$, set by the $15^\circ \times 15^\circ$ reference frame — the measurements did not show measurable degradation in gaze estimation accuracy attributable to retinal curvature. This result applies to the artificial-eye condition; the influence of inter-individual variation in retinal curvature, axial length and refractive error on gaze estimation accuracy was not assessed in this work and would require a dedicated study in a cohort spanning these parameters. To obtain information about the absolute gaze direction, only offset calibration is required. In principle, this offset might be established by locating the fovea on the foveola SLO image. However, studies reveal that the preferred retinal location (i.e., the retinal point corresponding to the fixation point on the screen) varies among individuals, making this approach insufficient for clearly determining the line of sight⁵⁸. Consequently, on a human eye the accuracy of absolute gaze estimation cannot be attributed to the instrument alone: the distance between the reported gaze and the visual target is dominated by the subject's own oculomotor behaviour — fixation offset, drift, microsaccades and the individual preferred retinal locus — rather than by the device. Such a measurement therefore quantifies the combined subject–instrument system; the instrument's own accuracy is the value obtained on the artificial eye, where an independent ground truth is available.

It should be emphasized that our device is not solely a retinal eye-tracker, as it enables determining gaze using three eye-tracking techniques, retinal, P-CR, and DPI, synchronously. This allows us to use retinal eye-tracking signals as ground truth during calibration of both video-based eye-tracking signals, thereby enabling DPI and P-CR eye-tracking techniques that do not require subject-dependent calibration. This feature can also be used to directly compare signals from different eye-trackers. As illustrated in Fig. 3, both the amplitude and velocity of the saccade exhibit greater values when measured by P-CR and DPI eye-trackers compared to these same parameters obtained using a retinal eye-tracker for evaluation. This may enable a more comprehensive description of the movement of the entire eye during psycho-visual tasks, thereby significantly enhancing our understanding of eye movements, which presently relies predominantly on the analysis of movements of structures in the eye's anterior segment.

Like any device, our retinal-anterior eye-tracker also has specific limitations. From the perspective of eye-tracking research, the most significant are the necessity for offset correction, as that retinal gaze trajectory has an arbitrary starting point and the reference frame is not anchored to the preferred retinal location. In practice, we estimate the zero position as the average gaze position during the initial 3-second fixation, chosen based on our experimental insights and consistently applied in all our psycho-visual experiments. Because the offset is estimated from a 3-s fixation, the accuracy of the absolute gaze direction inherits the subject's fixational stability and cannot, in principle, be better than it; consequently, absolute-gaze accuracy in humans characterizes the subject–instrument system rather than the instrument alone. The second limitation is the ability to detect gaze movements only within a confined angular range using a retinal eye-tracker. The theoretical upper boundary in this instance is determined by the dimensions of the reference frame, which is $15^\circ \times 15^\circ$. In practice, the angular range over which eye movements can be tracked in a human subject is smaller, and is limited by several factors: the diameter of the undilated pupil, the width of the palpebral fissure, which may occlude the imaging beam at eccentric gaze positions, and the density of the retinal vascular pattern, which determines how many features are available for registration within the reference frame. Based on our experience of operating the device, this range is approximately $\pm 4^\circ$ in both directions. We emphasize that this figure reflects our practical operating experience rather than a systematic characterization; establishing it quantitatively would require a population study in which these factors are measured across a large number of eyes. The human recordings reported in this work extend to 5° ($\pm 2.5^\circ$). This range could be expanded by using pupil-dilating agents. Nonetheless, we opt not to incorporate this approach into our research because the measurements would no longer be completely non-invasive. A further constraint is common to all ophthalmic devices. The detection of eye movements using a retinal eye-tracker necessitates a sharp image of the retina. When the transparency of the eye structures is compromised, such as in subjects with advanced cataract, measurements cannot be conducted. In the future, addressing these issues may involve developing fusion algorithms capable of generating highly accurate and precise eye-movement trajectories, using a retinal eye-tracker when feasible, and employing video-based eye-trackers in situations when this

is not feasible. To fully exploit this future functionality, the gaze trajectory should be calculated in real-time. However, the current offline nature of the trajectory reconstruction process is merely a technical limitation that may be overcome through computational advances. Future improvements will focus on GPU-accelerated implementations of the retinal tracking algorithms, which could enable real-time eye tracking, further broadening the potential applications of the proposed device.

Conclusions

In conclusion, the retinal-anterior eye-tracker we have introduced may pave the way for new measurement opportunities. The device is capable of measuring saccadic velocities of up to 300 °/s, incorporates an integrated display for dynamic stimulus presentation, and requires only a simple headrest for subject stabilization. With an artificial eye, the retinal part of the eye-tracker recovers a commanded angular change with an accuracy below 0.6' (0.01°) and a sample-to-sample precision of 0.09' (0.0015°) for the reference-frame signal and 0.05' (0.0008°) after Kalman fusion without requiring subject-specific calibration. In human recordings the corresponding precision during fixation without microsaccades is 0.119' and 0.0409'. The absence of a subject-specific calibration step makes the approach applicable to individuals who cannot undergo conventional calibration. Nonetheless, it is crucial to acknowledge that establishing absolute gaze direction still requires at least one point of offset correction. The integration of three eye-tracking technologies into a single device enables the retinal signal to serve as ground truth for calibrating DPI and P-CR signals. This advancement could furthermore foster research aimed at improving the calibration of video-based eye-trackers, devising techniques to filter DPI and P-CR signals (e.g., reducing errors associated with the pupil-size artifacts or lens wobbling), and providing a data source for supervised training of machine-learning-based gaze estimators. However, prior to fully harnessing the potential offered by the amalgamation of three tracking technologies in one device, additional research is essential into how to best leverage the retinal trajectory for calibrating the P-CR and DPI trajectories, including the feasibility of executing continuous calibration, as all three signals are recorded synchronously throughout the entire measurement.

Methods

Optical setup

Figure 4 illustrates the optical system of the neuroFET instrument, comprising three optomechanical modules: dedicated to tracking the retina (A) and the anterior segment of the eye (C), and for presenting sharp visual stimuli to the subject (B). The system is mounted on a motorized three-axis platform, allowing the subject's head to rest freely on a chin support. No bite bars or additional head immobilization systems are used during experiments, neither for healthy nor diseased subjects.

- a. The retinal tracking module operates on the principle of a scanning laser ophthalmoscope (SLO). A 920 nm laser beam (LS) rapidly scans a portion of the eye's fundus at 620 Hz using a high-speed X-Y MEMS scanner set. The resulting tracking frames (RTF) are processed by a tracking algorithm that analyzes their spatial shifts (see below). Additionally, a pair of positioning galvo-scanners (PGS) places the RTF frame (typically 5.3° x 2.8°) on the targeted retinal area (15° x 15°). To correct for the eye's refractive errors and ensure a sharp retinal image, the module includes movable lenses (L3).
- b. Stimulus presentation display features a high-brightness LCD integrated into the pupil-viewing optical path via a DM2 dichroic mirror and connected to the retinal tracker module through a DM1 dichroic mirror. Stimuli are presented using the open-source PsychoPy software (v.2023.1.2), which provides precise temporal and spatial control of visual stimuli⁵⁷.

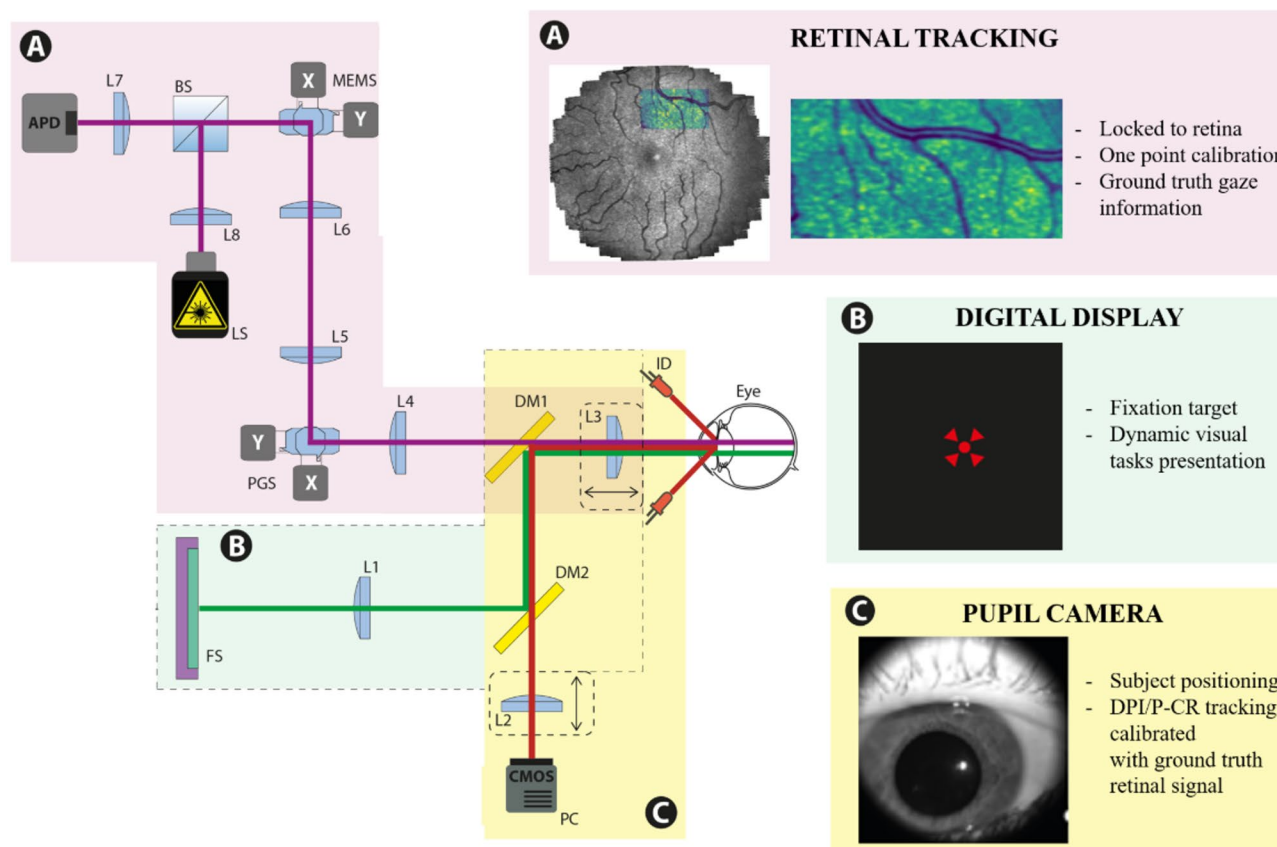


Fig. 4 Schematic illustration of the neuroFET instrument: (A) - retinal tracking module; (B) - visual stimulus display module; (C) pupil camera module; L1-L8 - lenses; L2, L3 – movable lenses; L8 – laser beam collimating lens; BS – beam splitter; DM1, DM2 – dichroic mirrors; ID – illumination diodes; PC – pupil camera; FS – fixation screen; PGS – positioning galvo-scanner module; MEMS – fast retinal scanner based on the micro-electro-mechanical systems; LS – laser source; APD – avalanche photodiode detector; RTF – retinal tracking frame.

c. The pupil camera module is designed for Purkinje reflection and subject positioning. This module employs two independently activated infrared diodes ($\lambda=810$ nm) to illuminate the anterior eye structures (only one was used for the recordings presented in this paper). Reflected light from the cornea, iris, and pupil passes through the main lens (L3) and two dichroic mirrors (DM1 and DM2) to a secondary lens (L2), which forms an image of the eye's anterior segment on a 2D CMOS camera (PC) at a scanning frequency 1.24 kHz (then reduced in post-processing to 620 Hz to align the scanning frequency of retina). A motorized sliding mechanism ensures a sharp image and accommodates variations in eye position.

The neuroFET device conforms to the applicable laser safety requirements of PN-EN 60825-1:2014, IEC 60825-1:2014, ANSI Z136.1-2014, and EN 207 regarding ocular exposure limit, which corresponds to Class 1 Laser Products. The laser operates at an optical output power of 3.155 mW, with the effective power incident on the cornea limited to 420 μ W under immobilized-beam conditions (note that the beam remains in continuous motion during actual operation and measurement).

Retinal movement tracking

The retinal tracker algorithm estimates the eye's trajectory by analyzing the relative shifts of a sequence of small retinal tracking frames (RTF). It's based on the principles presented earlier by our group⁴³. However, it has been

modified, extended, and tuned to maximize the system's usability and reliability in clinical conditions. One notable innovation is the introduction of an additional pair of galvo-scanners (PGS). The scanners allow sweeping the small-angle RTF along a predetermined, periodic trajectory, which, by overlapping, collectively cover a wider area of the retina (approximately $15^\circ \times 15^\circ$). This reference frame provides a view of the retina during the measurement and also minimizes the influence of drift and post-blink eye position changes on the success of eye movement trajectory reconstruction.

The complete retinal tracking algorithm consists of three stages. Initially, raw retinal tracking frames are pre-processed and resampled, which transforms the frames to rectangular raster form. In the second step, which is the core of the tracking algorithm, the velocities and positions of the retina are determined. The retinal velocities are estimated using an N-back algorithm, which aligns each retinal frame with $N=5$ previously acquired frames to obtain an estimate of the frame's position, based on the positions of previous frames and the displacements between the frame and the previous frames. Based on the thus obtained N-back trajectory, the retinal velocity is calculated using a two-point central difference formula. This approach results in a precise estimate of the eye's velocity but suffers from drift when integrating to recover position. The retinal position is therefore estimated using a newly developed, not previously described, reference frame tracking algorithm. The idea behind this algorithm is to create a wide-angle (444×714 pixels, $2.8^\circ \times 5.3^\circ$) reference frame of the retina by stitching together small-angle (140×140 pixels) retinal tracking frames that were collected while sweeping the galvo-scanners (PGS) over the retina as mentioned above. The estimated position of each retinal frame is obtained by registering its position to the reference frame. This new reference frame tracking algorithm does not exhibit drift, but is less precise compared to the N-back algorithm, so to reconstruct the retinal movement trajectory both accurately and precisely, in the last step, a Kalman filter⁵⁸ is used to combine the outputs obtained using these two algorithms. A detailed comparison of trajectories obtained using the N-back algorithm, reference-frame tracking algorithm, and Kalman filter, together with sample-to-sample distance distributions and RMS-S2S analysis, is included in Supplementary Material 4. In the present version of the algorithm, trajectory reconstruction is performed automatically in offline mode, with all computations executed by the processor (CPU). Processing duration depends on experiment length and data quality. Typically, the average processing speed is 20–30 frames per second.

The final output of our retinal movement tracking algorithm yields information about gaze direction, that is, where a person looks, and how this point of regard evolves over time. While the retinal tracking provides correctly scaled eye movements, it has an arbitrary starting point as the reference frame is not anchored to the preferred retinal location. We estimate the zero position as the average eye position during the initial 3-second fixation, selected based on our experimental experience and present in each of our psycho-visual experiments. This procedure is referred to as trajectory offset correction.

Validation of the retinal tracking algorithm was performed by performing the reverse operation: individual FET frames captured while the patient performed a smooth-pursuit circular movement were placed in a two-dimensional space according to the positions derived from the retinal tracking algorithm. The animation in Supplementary Video S2 demonstrates that a clear image of the retina was obtained using this method, thereby confirming the accuracy of the retinal tracking algorithm. This study was approved by the Bioethics Committee at the Nicolaus Copernicus University in Toruń, Collegium Medicum in Bydgoszcz (consent number: KB 763/2019). All experiments were performed in accordance with relevant guidelines and regulations, including the Declaration of Helsinki. Written informed consent was obtained from all subjects.

Comparison of the retinal and anterior eye-tracking signals

The device is equipped with a camera filming the front of the eye, facilitating the synchronized acquisition of images from both the retina and the outside of the eye, as used by video-based eye-trackers. The images of the anterior surface of the eye are processed to determine the parameters defining the position and shape of an ellipse fitted to the pupil (center x and y , width, height, and orientation), as well as to estimate the coordinates of the centers of the P1 (first Purkinje image) and P4 (fourth Purkinje image). A detailed description of these

algorithms is provided in Supplementary Material 7. In traditional anterior eye-tracking devices, a calibration procedure for each subject is performed prior to every experiment to enable mapping the features' relative positions to the eye's orientation. On the other hand, retinal trackers do not require subject-dependent calibration, as they deliver a retinal image of known angular extent, which enables directly determining the rotation angle by means of the displacements between consecutive retinal images, as depicted schematically in the graph included in Supplementary Material 6. In this work, therefore, instead of a calibration procedure in which participants are asked to look at a series of fixation targets, we use the retinal eye-tracker signal as the ground truth for calibrating the anterior eye-tracker signals. For this purpose, we used a general polynomial model of arbitrary rank K [59], which can be described as follows:

$$X = \sum_{k=0}^1 \sum_{l=0}^1 a_{kl} x^k y^l, \quad k, l \in [0, 2], \quad (4.1)$$

$$Y = \sum_{k=0}^1 \sum_{l=0}^1 b_{kl} x^k y^l, \quad k, l \in [0, 2], \quad (4.2)$$

where $\theta = [X, Y]$ refers to the gaze positions from the retinal tracker, and $\mathbf{P} = [x, y]$ are the raw values of the signal used by a given eye-tracking technology. For dual Purkinje image (DPI) trackers $\mathbf{P} = \mathbf{P}_{DPI} = \mathbf{p}_4 - \mathbf{p}_1$, and for the pupil to corneal reflection (P-CR) trackers $\mathbf{P} = \mathbf{P}_{PCR} = \mathbf{p}_C - \mathbf{p}_1$, where $\mathbf{p}_4$ and $\mathbf{p}_1$ are coordinates of fourth and first Purkinje reflections and $\mathbf{p}_C$ is the coordinate of pupil center. Parameters a_{kl} and b_{kl} are found using least squares regression.

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Author contributions MKS: investigation, visualisation, writing the initial draft, RK: software, methodology, PS: building the experimental setup, MNo: building the experimental setup, SzT: conceptualization, VP: software, JL: software, MM: conceptualization, MNy: validation, visualisation, ITCH: validation, visualisation, DCN: validation, visualisation, ASz: conceptualization, supervision, MSz: conceptualization, methodology, supervision. All authors discussed the results and contributed to the final manuscript.

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Data availability The datasets presented in the article are available from the corresponding author on reasonable request.

Declarations

Competing interest MNo, ASz, JL, and MSz are co-owners, and RK is an employee of Inoko Vision. DN and MNy are the full owners of eye-tracking consulting companies (DC Target AB and Emnys Consulting AB, respectively). The other authors declare no conflicts of interest.

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