Speckle interferometric sensor to measure low-amplitude high frequency Ocular Microtremor (OMT)

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ABSTRACT

Ocular microtremor (OMT) is a physiological high frequency (up to 150Hz) low amplitude (150-2500nm) involuntary tremor of the human eye. It is one of the three fixational ocular motions described by Adler and Fliegelman in 1934 as well as microsaccades and drift. Clinical OMT investigations to date have used eye-contacting piezoelectric probes or piezoelectric strain gauges. Before contact can be made, the eye must first be anaesthetised. In some cases, this induces eyelid spasms (blepharospasm) making it impossible to measure OMT. Using the contact probe method, the eye motion is mechanically damped. In addition to this, it is not possible to obtain exact information about the displacement. Results from clinical studies to date have given electrical signal amplitudes from the probe. Recent studies suggest a number of clinical applications for OMT, these include monitoring the depth of anaesthesia of a patient in surgery, prediction of outcome in coma, diagnosis of brainstem death. In addition to this, abnormal OMT frequency content is present in patients with neurological disorders such as Multiple sclerosis and Parkinson’s disease. However for ongoing clinical investigations the contact probe method falls short of a non-contact accurate measurement solution. In this paper, we design a compact non contact phase modulating optical fiber speckle interferometer to measure eye motions. Digital signal processing is then performed to extract the low amplitude high frequency displacement information.

Keywords: Ocular microtremor, Speckle interferometry, vibration analysis, inplane displacement.

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1. INTRODUCTION

Constant activity of the brainstem oculomotor units causes a fine involuntary tremor of the eye known as Ocular micro-
tremor (OMT) [1]. Along with drift and microsaccades, it is one of the three fixational eye movements described by
Alder and Fliegelman in 1934 [2]. Since then, the existence of OMT in humans has been confirmed by many authors [1,
3-10] and OMT observations have also been made on a number of mammals including cats [11], rabbits [12] and rats
[13].

The ability to accurately measure OMT could prove a useful tool to aid clinical diagnosis of a number of conditions.
Research has been carried out using contact methods to investigate OMT as a method of unambiguous brain stem death
confirmation [14], prediction of the outcome of coma [15], monitoring patients depth of anaesthesia [16-18], and atypical
OMT records have been observed in patients with Parkinson’s Disease [19] and Multiple Sclerosis [20]. A decrease in
peak OMT frequency with age, of clinically normal patients, has also been documented [21]. It has also been suggested
that OMT and the other fixational eye movements affect the visual process [22].

Eye motion amplitudes are customarily quoted in the literature in units of angular rotation of the eye. Prior to in-vivo
measurement, it has been more convenient to express displacement (obtained from the change in phase of the optical
beam) in terms of linear rather than angular units (i.e. meters as oppose to arc sec). Thus linear units have been used
during the system development, calibration and testing. We note however, that a 1 arc sec rotation corresponds to a 56
nm displacement for a typical eye of diameter 23 mm. OMT has a random, noise like appearance with sporadic
sinusoidal bursts. The peak-to-peak (pk-pk) amplitude of OMT is of the order of 1 μm (17.86 arc sec), with an estimated
range from 150 to 2500 nm (pk-pk) (2.67 to 44.64 arc sec). However, to accurately observe OMT, a minimum
resolution of 25 nm (pk-pk) (0.45 arc sec) has been suggested to be necessary [10]. OMT in clinically normal humans
has a frequency range from 40 to 100 Hz peaking around 88 Hz. However frequency observations between 10 and 150
Hz have been made [10]. As the frequency range of OMT is much greater than drift or microsaccades, OMT is
considered a physiological high frequency signal relative to the other fixational eye movements.

OMT has been observed using far field methods [7], however we are only aware of one other that has used a speckle
based technique [10, 23]. Previous non-contact based studies have observed the minute motion of OMT using contact
lenses placed on the eye. However, such methods, involving the use of a contact lens embedded with either a mirror or
an embedded search coil, are impractical and too invasive for a clinical situation. It should be noted, however that using
the resulting reflected light beam which deflected proportionally to the eye movement, investigators were able to detect
OMT [2, 3, 5]. When exposed to an AC magnetic field, the current induced in an embedded search coil can be used to
determine the orientation of the eye relative to the field. In the 1950’s, researchers were able to determine the amplitude
and frequency of OMT optically [7] using a blood vessel in the eye as a reference point, with a slit camera used to take
photographs at a constant velocity. Far-field non-contact optically based convolution methods are attractive for their
apparent simplicity and ease of use, however the cost and the relatively low sampling rates of multi-pixel digital cameras
have to date hindered their application in the measurement of OMT.

A number of studies investigating OMT using the mechanical contact probe method have been performed [4, 8, 16, 17,
19]. Much of the research has been carried out using a length of piezoelectric material, covered by a protective covering
of silicone, and makes contact with the eye by means of a sprung screw mechanism. Despite resolution as low as 10 nm
(pk-pk), a number of drawbacks exist with this method. It does not give accurate information about the amplitude of the
movement but rather a voltage is proportional to the degree of contact made between the piezoelectric material and the
eye. Also, by mechanically loading the eye, this method may significantly dampen the eye’s movement. While this is
useful in filtering out large movements such as drift, it compromises true OMT measurement. In addition, the eye must
be anaesthetised which causes the eyelid to spasm (blepharospasm) in a small number of patients. Apart from being
extremely uncomfortable to the patient, the contact method requires the use of sterile separate piezoelectric probes for
each examination increasing the difficulty and cost.

In the late 1990’s, Boyle et al. [23] designed and built a speckle interferometer using bulk optics attached to a large
optical bench to measure OMT. An optical beam originating from a Helium-Neon (HeNe) laser was optically split using
a half mirror beamsplitter with phase modulation of the reference beam in the interferometer achieved through
mechanical vibration of a mirror positioned in one arm of the interferometer. This system required subjects to focus at a
point while clinicians took readings. The physical bulk of the system inhibited the probability of its in-vivo applications. Furthermore, the minimum displacement resolution achieved using this system was 100 nm (pk-pk). This is 4 times greater than the suggested minimum sensitivity for accurate resolution of OMT [10].

Speckle interferometric techniques have advanced significantly since the work carried out by Leendertz in the late 1960’s and early 1970’s. Laboratory heterodyne speckle interferometric systems for example have been demonstrated to be capable of measuring micro-vibrations with amplitudes as low as 1 nm. However, in that case the lowest frequency measured was 5 kHz with a laser output power of 2 mW. For a device capable of measuring OMT, the system must be able to measure micro-vibrations with amplitudes from 150-2500 nm (pk-pk) and a frequency range of 10-150 Hz at low laser output power levels, i.e. at intensity values which will not cause retinal damage.

2. PRICIPLE OF OPERATION

Leendertz proposed a two beam illumination method to detect in-plane displacement in one component of the direction using the speckle effect. An area of the target surface is symmetrically illuminated by two mutually coherent beams incident at equal angles \( \alpha \), to the surface normal. The image at the observation plane, \( P(x,y) \) in fig. 1, is normal to the point of intersection of the two beams. It consists of two independent, coherent speckle patterns, arising due to each beam. A change in the relative phase of the two patterns gives rise to an intensity variation proportional to the in-plane surface distance moved.

![Figure 1. Schematic showing Leendertz’s method to measure motion using the speckle effect showing propagation vectors, and two collimating laser beams interfering at angles \( \alpha \) to the normal.](http://proceedings.spiedigitallibrary.org/)

The electromagnetic propagation vector of the illuminating beams in fig. 1 is defined as,

\[
\vec{k}_m = \frac{2\pi}{\lambda} \vec{n}_m,
\]

with \( m = \{1,2\} \) for each illuminating beam, \( \vec{n}_m \) are the unit vectors in the direction of illumination beam propagation, and \( \lambda \) is the wavelength of the source. The change of the observed phase \( \phi_0 \) at the \( P(x,y) \) is caused by a relative in-plane displacement of the surface described by the vector \( \vec{L} \),

\[
\phi_0 = \vec{k} \cdot \vec{L},
\]

\( \vec{k} = \frac{2\pi}{\lambda} \vec{n}_m \)
where \( \vec{k} \) is the propagation vector in the observed direction and the ‘·’ indicates a vector dot product. The phase change accumulated by each beam in propagating this distance is given by,
\[
\varphi_m = k_m \cdot L,
\]
with once again \( m = \{1, 2\} \) for each illuminating beam. The resultant phase shift between speckle points at the observation plane is
\[
\varphi_d = (\varphi_0 - \varphi_1) - (\varphi_0 - \varphi_2).
\]
Replacing each path phase in Eqn. (4) with the path phases from Eqn. (2) and (3), the resultant phase shift in the plane \( P(x,y) \) due to displacement vector \( \vec{L} \) of the rough surface is given by,
\[
\varphi_d = (\vec{k}_2 - \vec{k}_1) \cdot \vec{L},
\]
however, with the beams extending in the \( X\bar{Y} \) plane at equal angles ‘\( \alpha \)’ to the normal , with vectors \( \vec{i}, \vec{j} \) and \( \vec{k} \) lying in the \( X, Y \) and \( Z \) planes as shown in fig. 1, this reduces to
\[
\varphi_d = 2\pi \left\{ \frac{\sin(\alpha) \vec{j} + \cos(\alpha) \vec{k}}{-\sin(\alpha) \vec{j} + \cos(\alpha) \vec{k}} \right\} \cdot \vec{L} = \frac{2\pi}{\lambda} 2\sin(\alpha) \vec{j} \cdot \vec{L}.
\]
Thus the only component of motion for this method of illumination is in the \( \vec{j} \) direction. Thus a displacement vector of magnitude ‘\( d \)’, and direction along the vector \( \vec{j} \) results in the following phase change,
\[
\varphi_d = \frac{4\pi}{\lambda} \sin(\alpha)d.
\]
This is also true for in-plane time-varying displacement \( d(t) \). Baseband displacement information can be shifted up the frequency spectrum in order to eliminate unwanted low frequency optical signals introduced, for example, by the ambient room lighting. This is achieved by modulating one beams phase relative to the other. Assuming point detection by a photodiode, and that a sufficient number of speckles (i.e. ~100) always populate the photosensitive area of the photodiode, thus keeping the average optical power constant, the modulated interference signal will take the form,
\[
s(t) = U_0 \cos\left[\beta \cos(\omega_c t) + \varphi_d + \psi_0 \right],
\]
where \( U_0 \) and \( \psi_0 \) are random amplitude and phase components [24]. Modulation depth is denoted by \( \beta \) and \( \omega_c = 2\pi f_c \) is the angular frequency of the carrier frequency \( f_c \). For very small movement, we can assume \( U_0 = 1 \) and \( \psi_0 = 0 \).
Expanding Eqn. (8) using the Bessel [25] series we get,
\[
s(t) = \cos\left[\frac{4\pi}{\lambda} \sin(\alpha)d(t) \right] \left[ J_0(\beta) - 2J_2(\beta) \cos(2\omega_c t) + 2J_4(\beta) \cos(4\omega_c t) + \ldots \right]
\]
\[
- \sin\left[\frac{4\pi}{\lambda} \sin(\alpha)d(t) \right] \left[ 2J_1(\beta) \cos(\omega_c t) - 2J_3(\beta) \cos(3\omega_c t) + \ldots \right]
\]
where \( J_n(\beta) \) is a Bessel function of the first kind. In order to separate the movement signal from the phase modulated signal, phase demodulation is necessary. In-quadrature signals are obtained by performing synchronous demodulation at the carrier frequency \( \omega_c \) and at the first harmonic of the carrier frequency at \( 2\omega_c \), this yields,
Controlling the modulation depth $\beta$, it is possible to set $J_1(\beta) = J_2(\beta)$. In this case, taking the arctangent of the ratio of $s_s(t)$ to $s_c(t)$ gives,

$$\phi_w(t) = \arctan \left[ \frac{s_s(t)}{s_c(t)} \right],$$

where $\phi_w(t)$ is the wrapped phase information related to the in-plane movement. The arctangent function is bound to the range $[-\pi/2, \pi/2]$ radians, and as a result the phase of the motion is discontinuous and phase wrapped, where we recall that $\pi$ corresponds to one half of a phase cycle of the optical wavelength used. Phase unwrapping is the process where by integer multiples of $\pi$ are added or subtracted to the wrapped signal to achieve a continuous phase signal, $\phi_d(t)$. Applying this process, the amplitude is found by converting the unwrapped phase signal into distance using known system parameter values,

$$d(t) = \frac{1}{\sin(\alpha)} \frac{\lambda}{4\pi} \phi_d(t).$$

(12)

### 3. Design and Development of a Non-Contact OMT Sensor

Using optical techniques to measure OMT, the safety of the subjects’ eyes is of first and foremost concern. This optical non-contact system is designed so only a small area on the sclera, i.e. the white of the eye, is illuminated with laser radiation, allowing fixational eye movements, in particular OMT, to be recorded using light scattered from the sclera alone. It is not intended that any laser light enter the pupil itself.

The chief contributors to retinal damage arise due to effects from thermal, thermo-acoustic and photo-chemical interactions between internal eye tissue and the laser radiation [27]. Safety standards [28] exist to protect the human eye from retinal damage due to exposure from laser radiation. In our systems, to measure fixational eye movements, only scleral illumination with laser radiation is necessary. However, for non-intentional exposure to laser light of wavelengths in the range 400-700 nm, for time periods of between 10 seconds and 5 minutes, the Maximum Permissible Exposure limit, (MPE), in Joules per meter squared ($\text{Jm}^{-2}$), is calculated using the following equation,

$$MPE = \begin{cases} 
10^2 C_3 C_6 & \text{if } t > T_2 \\
18 t^{0.75} C_6 & \text{if } t < T_2 
\end{cases}.$$  

(13)

$C_3$ is a wavelength dependant correction factor and $C_6$ is a correction factor dependant on the visual angle with which the laser radiation enters the pupil. The exposure duration is denoted by $t$. For direct viewing of the laser radiation [23] (i.e. small visual angles), $C_6 = 1$. $T_2$, the exposure duration limit, is given by

$$T_2 = 10 \times 10^{0.02(g-550)},$$

(14)

where $g = \frac{\lambda}{d} \times 10^6$. Therefore for a 638 nm source, $T_2 = 575$ seconds. Since we are only interested in short exposures, i.e. $t < T_2$, this is the expression used to calculate the MPE later in this paper. Therefore the MPE, in Watts per meter squared ($\text{Wm}^{-2}$) is $MPE = 18 t^{0.25} C_6$. Assuming direct ‘intrabeam’ accidental direct exposure of the retina ($C_6 = 1$) and a pupil close to full dilation with a diameter of 7 mm and the light intensity is averaged over this area ($38.48 \times 10^{-6} \text{m}^2$). Thus the expression for the MPE limit in Watts (W) becomes

$$MPE = 6.927 \times 10^{-4} t^{-0.25}.$$  

(15)
This expression is derived from the International Electrotechnical Commission’s 1993 (IEC 825-1) safety standards [28]. We note however that Delori et al.’s recent paper [27] is a review of the American National Standards Institute’s (ANSI Z136.1-2000) safety standards. For accidental retinal exposures \( t < 0.7 \text{s} \), the pupil function in [27] becomes unity resulting in an equivalent expression for the MPE limit presented here. These results can be applied to our system in order to determine the permissible exposure time for the laser power used.

The principle components in our system, as seen in fig. 2 include: i) laser diode source, ii) combined Y-junction splitter and phase modulator, iii) interconnecting optical fibers, iv) collimating lenses, v) photodetector and amplifier, and vi) an analog to digital converter, post capture phase unwrapping and displacement signal extraction is then performed.

Figure 2: Schematic diagram of the fiber based speckle interferometer. Two beams exit a solid state Lithium Niobate Y-junction optical splitter and interfere at an angle \( \alpha \) with respect to the normal of vibrating sample surface. FO, phase maintaining fiber optics. SG1 and SG2, signal generators. PD, Photodiode. AMP, Amplifier. DSP, digital signal processing. \( \alpha \), angle at which the two optical beam interfere. \( L_1 \) and \( L_2 \), collimating lenses. \( L_3 \), imaging lens.

A linearly polarized adjustable power laser diode (FDL-638) is coupled by polarisation maintaining optical fiber to a combined Y-junction splitter and phase modulator (YJ-638), out of which extend two phase maintaining optical fibers terminated with Ferrule Connector (FC) optical connectors [29]. The optical wavelength used is \( \lambda = 638 \text{ nm} \). In order to collimate the spherical wave emerging from the FC connectors, visible to near infra-red FC focusing lenses (0.25 numerical aperture) were used in conjunction with thin 6 mm diameter lenses \( (f = 72 \text{ mm}) \) mounted in adjustable C-ring mount assemblies. Each beam is collimated and aligned, using lenses \( L_1 \) and \( L_2 \) to interfere on a small area at angles of ‘\( \alpha \)’ to the surface normal. For the results presented here, \( \alpha = 45 \text{ degrees} \).

The normally back scattered light intensity from the interference surface point can be collected by imaging lens \( L_3 \), however an imaging lens was at first not used as it would have introduced an aperture which would increase the speckle size [30]. In this paper, we therefore at first, while testing the device, did not include the imaging lens \( L_3 \). Later however imaging lenses were introduced when using lower illuminating intensities.

Opto-electrical conversion of the time-varying speckle signal was performed using a low capacitance Melles Griot Photodiode (13d3i001; area: \( 0.31 \text{ mm}^2 \)). Amplification of the signal was performed using a low noise Burr Brown operational amplifier (OPA637) in transimpedance gain configuration with a 20 M\( \Omega \) resistor. To avoid electrical pickup, the casings of the photodiode, as well as the operational amplifier, were caged in a grounded metal case.

A National Instruments NI-USB 6221 16-bit analog to digital (ADC) converter was employed to sample the time varying optical signal at a sampling rate of 100 kHz. To ensure adequate sampling, \( 2\omega_c + \omega_{\text{max}} < 50 \text{ kHz} \), where \( \omega_{\text{max}} \) is the highest frequency component of the motion, i.e. \( \sim 200\text{Hz} \). Post-processing of the data was performed with code written using The Mathworks Inc. MatLab (V.7) [31] environment.
A deviation to the methods employed in previous papers [10, 32, 33] was adopted to unwrap the phase signal and to process the results presented in this paper. Integer multiples of $\pi$ are added or subtracted to the phase wrapped signal $\phi(t)$ depending on both the change of signs of successive quadrature sine $\sin(t)$ and cosine terms $\cos(t)$. Discontinuities are obtained where the cosine term $\cos(t)$ is zero, i.e., goes from positive to negative or vice versa. The unwrapped phase signal is obtained when the values of discontinuities ($-1, 0$ or $1$) are convolved with $\pi$ values. The wrapped signal is then added to the result to yield the continuous phase signal. Scaling is performed to convert the phase signal into a displacement signal, using Eqn. (12).

Our method of phase unwrapping for retrieving the true phase $\phi_\text{t}(t)$ is as follows: First we determine the instances of zero values as described above. We then compare points at which zero crossing occur with the sign of the wrapped signal’s slope. The multiple number of $\pi$ values is accumulated in a feedback loop leading to integer multiples of $\pi$ added or subtracted to the wrapped phase signal $\phi_\text{w}(t)$ to determine the continuous phase signal, $\phi_\text{t}(t)$. Scaling, using Eqn. (12), yields the displacement signal, $d(t)$.

4. EXPERIMENTAL RESULTS

The total optical power from each arm extending from the combined Y-junction and phase modulator was measured using a Newport 840-C optical power meter (serial No. 2269). Fitting the resultant light power versus current curve we found the laser diode electro-optical transfer characteristic can be satisfactorily described, after reaching the threshold, using a third order polynomial of the form

$$P(i_d) = 42211 - 194.59i_d + 2.7844i_d^2 - 0.0117i_d^3,$$

where $i_d$ is the input current (mA), and $P(i_d)$ is the optical output power ($\mu$W).

When using the device in a clinical setting, each optical beam will have to be aligned so that they interfere on the sclera of the patient before measurement of OMT can take place. Allowing: (a) 270 seconds for alignment, and (b) 30 seconds to record OMT movement, the calculated MPE using Eqn. (16) for both stages is presented in Table 1. A second value for the MPE is also presented in Table 1 assuming operational safety factors of 50.68% and 17.56% of the MPE limit for the measurement and recording stages respectively.

<table>
<thead>
<tr>
<th>Experimental Stage</th>
<th>MPE (without safety factor)</th>
<th>MPE (with safety factor)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Measurement, $(t = 30 \text{ s})$</td>
<td>295.96 $\mu$W</td>
<td>150 $\mu$W (50.68% of 295.96 $\mu$W)</td>
</tr>
<tr>
<td>Alignment, $(t = 270 \text{ s})$</td>
<td>170.87 $\mu$W</td>
<td>30 $\mu$W (17.56% of 170.87 $\mu$W)</td>
</tr>
</tbody>
</table>

Table 1: Calculated MPE, Eqn (16), and proposed MPE with operational safety factor.

As stated, we assume that ~100 speckles are observed across our photodiode area. This implies that the maximum speckle diameter should be less than 60 $\mu$m. Capturing the backscattered speckle field from out test surface with a CCD camera, (Fujifilm Finepix F460 with a Fujinon zoom lens, $f = 5.8–17.4$ mm), and a lens setup to magnify the field, $(f = 60$ mm, lens diameter: 25 mm, magnification: $M = 2$), we measured the speckle diameter to be between 51.3 $\mu$m and 56.8 $\mu$m.

The algorithm to process the speckle signal and perform phase unwrapping was implemented using Matlab for offline processing. After testing our algorithm on simulated test sinusoidal displacement signals with varying levels of Gaussian white noise added, the method was then applied to experimental data. A single, undamped, length of Sensortech Ltd. piezoelectric bimorph material (SM10-2505-00), with a rated sensitivity of 400 nm per volt applied, was employed as our micro-vibration simulator. A small sample of rounded white plastic was attached to the end to mimic the sclera. Under ideal conditions, applying a sinusoidal voltage of amplitude 5 V should result in a sinusoidal displacement of amplitude 2000 nm.
This data was sampled directly for the amplified photodiode output signal, when a sinusoidal voltage was applied to a piezoelectric bimorph element to produce a 90 kHz sinusoidal signal of amplitude 1000 nm (pk-pk). The phase wrapped phase signal from the arctangent of the ratio of the two quadrature signals is shown in fig. 3 (a). The required phase additions are shown in fig 3 (b), and the resulting sinusoidal displacement is presented in fig. 3 (c).

Using this method, we have successfully demonstrated that our method can unwrap a discontinuous wrapped phase signal from two noisy quadrature signals. The resulting displacement signal, using Eqn (12) is found to have displacement amplitude of 977.06 nm (pk-pk). This is 2.3% below the expected amplitude of 1000 nm (pk-pk).

During the experiments describe above, we applied a driving current of 85 mA to the laser diode which corresponds to a measured optical power of 598.5 µW. This is above the safe limit of operation for use on a human eye. This high power level was employed in order to ensure a sufficiently high SNR of the speckle field to allow an unambiguous verification of the ability of the system to measure OMT like motion. Table 2 shows the expected displacements and the experimentally resolved values of the amplitude for a range of frequencies. We note that measurements for the smallest signal, 50 nm (pk-pk), at the three lowest frequencies were not reproducible, however measurements at 54 nm (pk-pk), were possible.

<table>
<thead>
<tr>
<th>Driving Voltage (V)</th>
<th>Expected Displacement Amplitude (nm)</th>
<th>Frequency (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>10 ±/− 2</td>
<td>30 ±/− 2</td>
</tr>
<tr>
<td>6.24</td>
<td>2500</td>
<td>2464</td>
</tr>
<tr>
<td>5</td>
<td>2000</td>
<td>1976</td>
</tr>
<tr>
<td>3.74</td>
<td>1500</td>
<td>1380</td>
</tr>
<tr>
<td>2.5</td>
<td>1000</td>
<td>920</td>
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<tr>
<td>1.24</td>
<td>500</td>
<td>583</td>
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<tr>
<td>0.74</td>
<td>300</td>
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<td>0.50</td>
<td>200</td>
<td>211</td>
</tr>
<tr>
<td>0.24</td>
<td>100</td>
<td>84</td>
</tr>
<tr>
<td>0.12</td>
<td>50</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 2: Simulator II. Measured displacement amplitude at increasing driving voltage and varying frequencies.

From Table 2, we can see our system is capable of measuring micro-vibrations over a wide range of amplitudes, from 100-2500 nm (pk-pk), and we have reproducibly measured micro-vibrations with amplitudes as low as 54 nm at 10 Hz. Using a similar optical technique [23] to measure OMT, the peak-to-peak amplitude of OMT was measured to be centered at 728 nm, over a range from 598 nm to 903 nm. A number of investigations have indicated that peak OMT frequency for clinically normal patients occurs in the range of 88 Hz [1, 4, 34]. Patients experiencing coma exhibit...
lower peak OMT frequency between 40 and 60 Hz [15]. The dashed region appearing in the centre of the table of measured results, i.e. Table 3, includes both frequency and amplitude regions. Thus, we have demonstrated, that a broad spectrum of frequencies, beyond those frequencies and amplitudes required to measure OMT, can be measured using our system.

![Figure 4](image.png)

Figure 4. (a) Trace showing 0.5 seconds of an OMT trace recorded from a clinically normal human subject using the invasive piezoelectric probe method. (b) OMT Spectrum from a clinically normal human subject with peak frequencies around 80 Hz.

Fig. 4 (a) shows a results from the contact method used to observe OMT from a clinically normal human. As stated earlier, a voltage signal proportional to the degree of contact made with the eye is obtained from this contact system. A peak in the spectrum of the data in fig. 4 (a), as shown in fig. 4 (b) occurs around 88 Hz. Our system can measure displacement signals in the range of OMT signals, as shown in Table 2, and thus is capable of observing OMT signals like the example presented in fig. 4.

However, a number of physiological aspects be taken into account when in-vivo measurement takes place, such as the contributions tear drops have on optical measurement. To test this, the optically rough surface on the simulator was wetted using a commonly available eye drop solution (Brolene Propamidine) to simulate water (tears) on the surface of the eye. In this way, the effects, if any, of tears on the recording of OMT signals using this optical system was examined.

Introduction the eye drop solution had no apparent effect on the measured result using the system. It remained possible to resolve a 684.48 nm (pk-pk) amplitude signal at 88.9 Hz signal using our simulator, see fig. 5 (a) and (b). The number of samples (N), sampling interval (SI) as well as sampling frequency (f_s) are also shown. From the amplitude of the driving signal and the responsivity of the piezoelectric simulator (400 nm/V), we expect a displacement of 698.1 nm, thus our measured result disagrees by only 1.9%.
Real tears are dynamic in the sense that the volume and density are likely to vary between patients. This simple experiment was carried out by wetting the rough surface to mimic the effects of tear drops. However, based on these results for static droplets, it appears that wetting is not likely to cause any significant operational issues in measuring OMT using this optical system.

As stated earlier, the measurements presented so far result from illuminating the rough surface with 598.5 μW of laser radiation. It is intended to illuminate the sclera of the eye, and subject safety is paramount. However, following the predictions of Eqn. (15), accidental exposure of the retina can last less than 2 seconds at this intensity level before permanent damage occurs. In order to demonstrate that our device can operate at ‘eye safe’ exposure levels and still measure OMT like micro-vibrations, we reduced the driving current in the laser so that it emits only 180 μW of light. This is equivalent to a safety factor of 60.82% of the MPE assuming a 30 seconds exposure in the recording stage, see Table 1.

An imaging lens, \( L_3 \) (\( f = 35 \text{ mm} \), lens diameter = 25 mm), was used to produce a system of magnification of \( M = 2 \), and the interference fringes were resolved. Fig. 10 (a) shows the resulting measured noisy 647 nm (pk-pk) micro-vibration signal. The measured amplitude is 5.98% below the expected peak-to-peak amplitude of 688.89 nm. Fig. 6 (b) is the corresponding spectrum which contains a peak at 88.1 Hz.
To perform an in vivo measurement using this lower power level one might proceed constrained as follows: Suppose it is found necessary to use an exposure of 180 μW for 30 second in order to record sufficient signal in order to accurately estimate OMT movements, then the device can be aligned for up to 270 seconds, prior to measurement, using a 30 μW signal with a safety factor of 17.56% of the MPE. In this case, the eye is exposed to a total time-averaged intensity of 45 μW. The requirement for eye safety is met as it is below the MPE for an exposure of 300 seconds at 166.4 μW. Thus this device is capable of measuring OMT like movements, using illumination intensities over exposure times which adhere to the safety requirements for accidental exposure of the retina.

5. DISCUSSION AND CONCLUSIONS

This paper reviewed the significance of the smallest and highest frequency of the three fixational eye movements, Ocular MicroTremor, (OMT) has in a clinical setting. Without the need to anaesthetize or make contact with the patient’s eyes, it has been shown that a non-contact measurement device for measuring eye movements is useful. In a clinical setting, it could be used to diagnose several conditions but prior to that, such a tool could help produce a better understanding of the link between several serious medical conditions and OMT. A non-contact system is desirable to allow greater convenience and patient acceptability in the clinical environment. It is also accurate to state that for vision scientists, a practical and viable method of measuring OMT non-invasively would be welcome.

In this paper, we have reviewed some of the work carried out to date in designing and developing a compact and portable fiber based interferometer to measure OMT [35]. The physical size of the system has been greatly reduced, compared to that of a bulk optical system, using optical components coupled together using optical fibres. This has resulted in a compact and easily portable system. We have tested this device, measuring in-plane motion of a rough surface using piezoelectric bimorph elements to provide calibrated test vibrations. The measurement of micro-vibrations over a range of amplitudes and frequencies, larger than those required to measure OMT, i.e. from 100-2500 nm (pk-pk), and over 10-170 Hz, has been demonstrated. In addition to this, it has been shown that the system can perform measurements while operating at safe illumination intensities for which accidental exposure of the retina will not causing permanent damage. A simple experiment was performed, the rough surface being wet to mimic the effects of static tear drops, and no operational problems using the optical method were observed. Therefore a portable device capable of measuring OMT non-invasively and safely has been implemented and tested.

For this device to be realized as a clinical tool for in-vivo use by clinicians, several practical issues must be addressed. Local ethics board approval will be required before any in-vivo test may be carried out. To measure OMT using the system described here, it will be necessary to either isolate or measure simultaneously any head movement relative to the eye.

Some suggestions for future work on this project included extending and modifying the above system to enable binocular OMT measurement and/or simultaneous vertical and horizontal displacement measurement. Isolating the recorded OMT signal from other signals such as residual head movement, environmental vibrations and cardiac pulse could be accomplished by using a second device to accurately measure large scale head movements at a stationary fixed reference point (e.g. bridge of the nose). This could be combined with an Electrocardiogram (ECG) to filter out unwanted pulse signals. In addition to this, greater flexibility may be achieved concerning exposure durations through judicious choice of illuminating wavelength with respect to the safety standards. This could increase the number of possible applications of this method as a clinical diagnostic tool.

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