mNetra: A Fundoscopy based Optometer

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Abstract: Untreated refractive error in the eye is one of the leading causes of preventable blindness. The devices necessary for this is expensive and often requires skilled technicians to operate. In this paper, a common off the shelf ophthalmoscope has been modified and integrated with a smart phone to build an affordable optometer. The device has been tested on a statistically significant population with refractive error range from $-8.00D$ to $+3.50D$. We found a reasonably good correlation with other prevalent methods of measuring refractive error in the eye.

1 INTRODUCTION

Worldwide more than 285 million people suffer from some form of visual impairment (WHO.int, 2014). Out of this 90% person live in developing countries. The WHO has published a report in 2010 that, the leading causes of visual impairment throughout the world are the untreated refractive error (43%), unoperated cataract (33%) and glaucoma (2%)(WHO.int, 2014). There are an estimated 19 million children are visually impaired and out of these, 12 million children are suffering from refractive errors, a condition that could be easily diagnosed and corrected (Pascolini and Mariotti, 2011). Worldwide myopia is leading cause of low vision in young children, especially school going children (Resnikoff et al., 2008; Saxena et al., 2015). That can hinder education, productivity, personality development and career opportunities. The impact of blindness due to refractive error (myopia) in young age is very high as compared to cataracts or glaucoma in old age because it affects the person for a longer span of time starting from childhood. It also places a greater socioeconomic burden for developing countries like India.

The human eye is a sophisticated and complex imaging system which is capable of dynamically adjusting its focal power for wide ranges of distance. The eye's refractive power is maintained by the air-cornea interface (corneal shape dependent fixed power) and crystalline lens (provides an adjustable extra power to the eye). Common refractive errors found in the eye (in Fig.1) are due to imperfections in eyeball shape, lens and cornea. The **Myopic** eye shown in Fig.1(b), has an enlarged axial length (the length from the posterior corneal surface to the retina) or an increase in the refractive power of the eye lens and focuses parallel rays to a point in front of the retina rather than on it. In contrast, the **Hyperopic** shown in Fig.1(c), has a short eye or insufficiently curved cornea and focuses parallel rays at a point behind the retina. A **Healthy eye** (In Fig.1(a)) on the other hand, can focus parallel rays to a single point on the retina.

The refractive disorder is commonly treated using appropriate corrective lenses such as eyeglasses or contact lens. For this to be possible a periodic but comprehensive examination of the eye is mandated especially in school and primary health care centers. To do this, three diagnostic methods are primarily used to evaluate the refractive error.

- **Subjective Methods:** Its screening result relies upon the response and reaction of a patient, e.g. patient's judgment of sharpness or blurriness of a test object (Snellen eye chart).
- **Objective Methods:** This method doesn't depend on the feedback from the patient. The optical instrument that can find out the refractive error automatically, e.g. refractometer and optometer.
- **Hybrid Methods:** Currently, this is the most common way to determine refractive error to prescribe corrective lens. It consists of two steps. An objective method is used in the first step followed by the subjective method to determine the accu-

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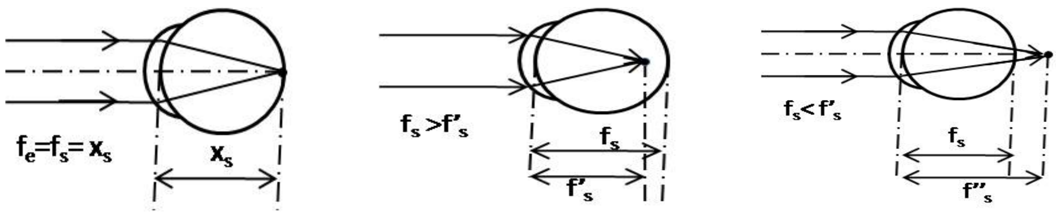


(a) Healthy eye (b) Myopia (c) Hyperopia

Figure 1: Common retinal refractive error in the human eye.

rate corrective lens.

For the above have a method of refractive error screening ophthalmologists used optical technology based diagnostic instruments such as auto refractometer, focometer, retinoscopy, and wavefront analyzer can be used for measuring the refractive power of the human eye (Ko and Lee, 2006; Berger et al., 1993; Dave, 2004) is bulky, expensive, sophisticated, highly application specific and needs a trained technician to operate these devices.

Recently, portable devices (smartphones and tablets) have emerged as a versatile mobile computing platform with high resolution display, which provides us a new opportunity for mobile health (**mHealth**). Paul et al. (Paul and Kumar, 2015) use the ophthalmoscope and mobile phone to develop an affordable fundus imaging based eye care device for glaucoma screening. *Peek Vision* provides high quality tools for professional eye exams using a mobile phone (PeekVision, 2015). It can be used to diagnose cataracts, visual acuity and eyesight, colour and contrast of the eye. In (Pamplona et al., 2010; Cameraculture.media.mit.edu, 2015), *NETRA* and *CATRA* provide affordable cell phone attachment that measure the eye refractive power and cataracts information. It creates an inverse Shack-Hartmann sensor based on a "high resolution programmable display and combines inexpensive optical elements, interactive GUI, and computational reconstruction". Currently, *Smart Vision Labs* have developed a smartphone based affordable auto refraction technology based on *Wavefront aberrometry (Shack Hartmann)* to measure the refractive error of the eye (Zhou and Kassalow, 2010).

In this paper, we introduce a novel low cost, hand held, portable, reliable, accurate and interactive instrument ***mNetra***, based on view dependent display to determine the refractive error in the human eye for fast and accurate screening. The ophthalmoscope and smart phone based device tries to get the best image/video frame of the retinal scan after proper adjustment of ophthalmoscope lens focusing wheel. This paper presents,

- A methodology to do refraction of the eye for screening using a mobile phone an ophthalmoscope.
- An Android based application has been developed, which allows collection of patient details and perform the computation for refraction. A generic Regression (R) model to demonstrate that an off the shelf ophthalmoscope can be attached to the mobile phone is also described in the paper.

We believe that our funduscopy based optometry increase the usability and application of ophthalmoscope and mobile phone. It provides a suitable solution for affordable eye care in developing countries. And this is particularly useful in prescribing corrective lenses for patients who are unable to undergo a subjective refraction that requires a judgment and response from the patient (e.g. a person with communication problems or severe intellectual disabilities).

2 mNetra

The ***mNetra*** is a monocular, funduscopy based handheld optometry which uses the principle of ophthalmoscopy to measure refractive error. The optician uses the manual rotary wheel to focus the image on the retina. This concept has been explored to do same automated refraction in the mNetra. We describe the methodology in the following subsection.

The Mobile ophthalmoscope is an optoelectronics handheld, an affordable lightweight instrument for screening the interior structure of the eye, especially the back part of the eye (**fundus**: which includes the retina, optical disk, optical cup, blood vessel and fovea etc.). This also supports computation and communication needs for processing and sharing of medical data. This optoelectronic device has been made possible by integrating the smartphone and an ophthalmoscope.

The total optical power of a multiple lens system is equal to the linear summation of individual lens power. In Fig. 4(b), ophthalmoscopy based screening system's optical unit consist of the ophthalmoscope, subject (patient) eye and a camera rather than ophthalmoscope, subject eye and observer eye (in Fig. 4(a)). This new multiple lens optical system is in equilib-

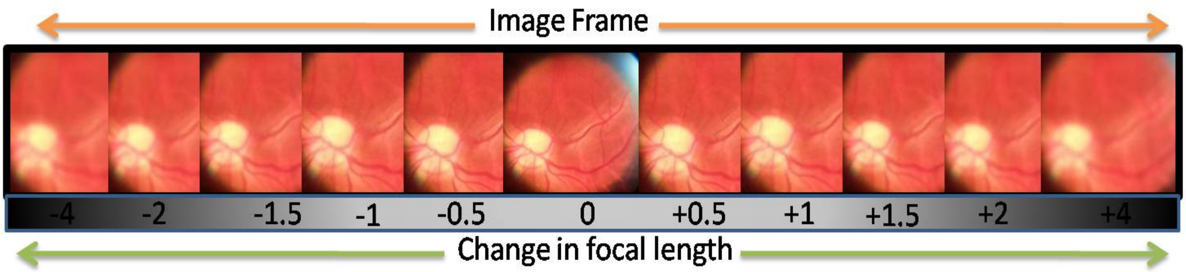


Figure 2: Basic concept of retinoscopic refractometer :In this figure, fundus image quality is detonated with increment or decrement of focusing lens focus.

rium condition (total power is zero) for the healthy eye with no refractive error. A small change in the power (δP) of any lens in this system will disturb the equilibrium of the whole optical system.To overcome and nullify this problem, we should use the other lens of opposite power ($-\delta P$) so it will again maintain the equilibrium.

We use this concept to develop a funduscopy based optometer using an ophthalmoscope and mobile phone. This has been made possible by making some changes in the basic ophthalmoscope and embedding mobile phone. In Fig. 4(b) of the new optical system, the observer's eye is replaced by CMOS camera module (Mobile phone) and focusing lens can be used to correct the ocular error i.e. due to subject/patient eye refractive alignment. The focusing lens position is controlled by the ophthalmoscope focusing lens wheel (in Fig. 5(c)). In Fig. 2, we show how the quality of the image frame depends on the focal length of the focusing lens set with the focusing wheel of the ophthalmoscope.

Mathematically, refractive error corrective power (P) is the function of focusing lens position (x). Where, x is the function of focusing wheel angle θ. So that,

$$P = F(x) = F(g(\theta)) = F(r \times \theta) \qquad (1)$$

Where, r is the radius of the focusing wheel. and in another way it can be written as $P = \frac{1}{f_s}$ and from the equation 23, $P \equiv P_s$.

2.1 Functional and Architectural Overview

In this section, we outline the design of a mobile ophthalmoscope based optometer subjected to the constraints of **Cost, Bulkiness, Portability** and **Adoptability**.The device is based on the Smartphone. This has been made possible by embedding the **Optical** (Ophthalmoscope and CMOS camera), **Computing** (Mobile phone and Microcontroller), **Sensor** (potentiometer based position sensor) functionality in a single device. As illustrated in Fig. 3, the device consists of different functional block which are described below:

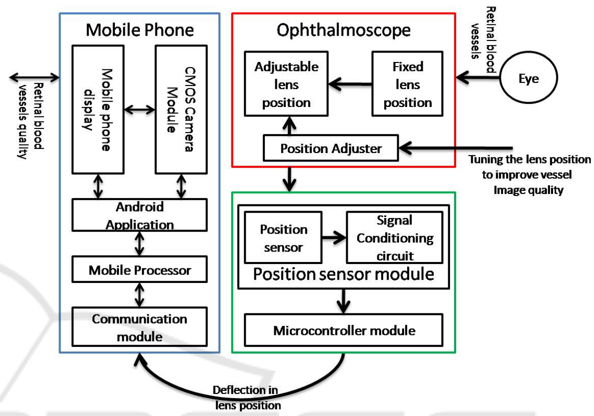


Figure 3: Working diagram of ophthalmoscope based hand held refractometer.

- **Optical System:** This block capture the magnified fundus image with a CMOS camera –typical CMOS in nature. The key innovation in the solution is to replace the eye of ophthalmologist by of a smartphone camera based CMOS sensor to capture the image of patient's fundus. We use a smart phone of a high resolution ($\geq$5MP) camera for the same. We have also designed and build a custom mobile phone holder Rapid Prototype (RP) model to attach the Smartphone camera to the ophthalmoscope.
- **Computing Unit:** This functional unit consists of a smart phone processor and the microcontroller for processing the mobile phone data (CMOS capture video/image) and controlling and processing of focusing lens sensor output respectively. Mobile phone centralized processing and control unit is able to satisfy high compute power required for image processing, result visualization and data sharing that are part of the application software.

- **Sensory System:** To measure the ophthalmoscope focusing lens position we used a sensor module that will measure the current position of focusing wheel. In Fig. 3, it consists of **Position sensor**, **Microcontroller** and **Bluetooth module** to share the current status of focusing wheel to the mobile phone for further processing and representation.

2.2 The Mathematical Basis

Ophthalmoscopy based optometer's optical setup and working rays diagram are shown in Fig. 4. It consists of the ophthalmoscope, subject eye and CMOS sensor (mobile phone camera). In this optical setup, we should consider the position of all lenses, i.e. magnifying lens ($M_L = \{M_{L1}, M_{L2}\}$), subject eye (E_L) and CMOS sensor (O_L) are fixed except focusing lens (F_L). The focusing lens is used to overcome and nullify the refractive error of the subject eye lens so that image properly focused on CMOS sensor. In this section, we explain the mathematical relationship between focusing lens (F_L) position (x) and subjective eye power ($P = 1/f_e$).

For subject lens E_L (l_s, f_s):

$$\frac{1}{x_s} + \frac{1}{q_s} = \frac{1}{f_s} \tag{2}$$

from the lens maker formula.

Similarly for lens M_L (l_2, f_2):

$$\frac{1}{x'_2} + \frac{1}{q_2} = \frac{1}{f_2} \tag{3}$$

where, $x'_2 = x_2 + q_s$. Hence 3 becomes

$$\frac{1}{x_2 + q_s} + \frac{1}{q_2} = \frac{1}{f_2} \tag{4}$$

Similarly for lens F_L (l_1, f_1):

$$\frac{1}{u} + \frac{1}{v} = \frac{1}{f} \Leftrightarrow \frac{1}{x'_1} + \frac{1}{q_1} = \frac{1}{f_1} \tag{5}$$

where, $x'_1 = X - x + q_2$. Hence 5 becomes,

$$\frac{1}{(X - x + q_2)} + \frac{1}{q_1} = \frac{1}{f_1} \tag{6}$$

Similarly for lens O_L (l_o, f_o):

$$\frac{1}{u} + \frac{1}{v} = \frac{1}{f} \Leftrightarrow \frac{1}{x'_0} + \frac{1}{q_0} = \frac{1}{f_o} \tag{7}$$

where, $x'_o = (x = x_o) + q_1$. Hence 7 becomes,

$$\frac{1}{x + q_1} + \frac{1}{q_o} = \frac{1}{f_o} \tag{8}$$

Rearranging 2 we get

$$q_s = \frac{f_s x_s}{x_s - f_s} \tag{9}$$

which implies q_s is a function of f_s.

$$q_s = F(f_s) \tag{10}$$

Similarly

$$q_2 = \frac{f_2 x'_2}{x'_2 - f_2} = \frac{f_2(x_2 + q_s)}{(x_2 + q_s) - f_2} \tag{11}$$

$$\Rightarrow q_2 = F(q_s) = F(F(f_s)) \Rightarrow q_2 = G(f_s) \tag{12}$$

where,

$$G = F(F) \tag{13}$$

From 5 and 6

$$q_1 = \frac{f_1 x'_1}{x'_1 - f_1} = \frac{f_1(X - x + q_2)}{(X - x + q_2) - f_1} \tag{14}$$

$$q_1 = H(G) = \frac{f_1(X - x + G)}{(X - x + G) - f_1} \tag{15}$$

Also from 7 and 8 :

$$q_o = \frac{f_o x'_o}{x'_o - f_o} = \frac{f_o(x + q_1)}{x + q_1 - f_o} \tag{16}$$

Similarly

$$q_o = \frac{f_o(x + H(G))}{x + H(G) - f_o} \tag{17}$$

$$q_o = I(H, x) \tag{18}$$

and q_o is constant. and from 17

$$x = \frac{f_o H(G) - q_o H(G) + f_o q_o}{(q_o - f_o)} \tag{19}$$

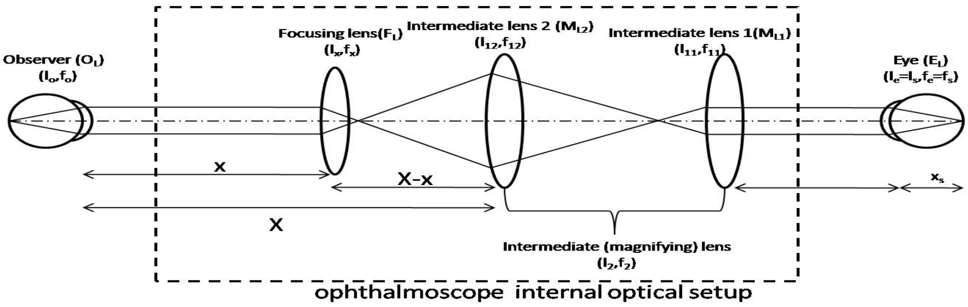
Assuming the position and focal length of all the lenses are fixed except for the focussing lens F_L. We obtain from 9 and 11

$$q_2 = \frac{f_2(x_2 + \frac{f_s x_s}{x_s - f_s})}{(x_2 + \frac{f_s x_s}{x_s - f_s}) - f_2}$$

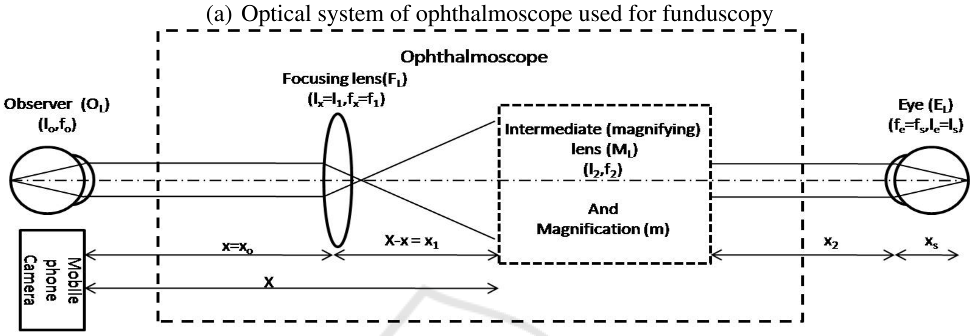
$$\Rightarrow q_2 = \frac{f_2(x_2(x_s - f_s) + f_s x_s)}{(x_2(x_s - f_s) + f_s x_s) - f_2(x_s - f_s)} \tag{20}$$

Using equations 14 and 16

$$\Rightarrow q_o = \frac{f_o(x + q_1)}{x + q_1 - f_o} = \frac{f_o(x + (\frac{f_1(X - x + q_2)}{(X - x + q_2) - f_1}))}{x + (\frac{f_1(X - x + q_2)}{(X - x + q_2) - f_1}) - f_o}$$



(a) Optical system of ophthalmoscope used for funduscopy



(b) Optical system of mNetra

Figure 4: Optical rays diagram for (a) ophthalmoscope and (b) mNetra : Retinal image formation on observer eye/CMOS sensor in ophthalmoscope from the rays coming from the subject eye.

$$\Rightarrow q_2 = \frac{\begin{array}{c} q_o(Xx - x^2 - 2f_1x + f_ox) + q_o(f_1X - f_oX \\ -f_of_1) - f_of_1X - Xxf_o + f_ox^2 - 2f_1f_ox \end{array}}{(-q_ox - q_o(f_1 - f_o) - f_ox + f_1f_o)} \tag{21}$$

Combining equation 12 and 20,

$$\Rightarrow \frac{f_2(x_2(x_s - f_s) + f_sx_s)}{(x_2(x_s - f_s) + f_sx_s) - f_2(x_s - f_s)}$$
$$= \frac{\begin{array}{c} q_o(Xx - x^2 - 2f_1x + f_ox) + q_o(f_1X - f_oX \\ -f_of_1) - f_of_1X - Xxf_o + f_ox^2 - 2f_1f_ox \end{array}}{(-q_ox - q_o(f_1 - f_o) - f_ox + f_1f_o)}$$

Rearranging the above expression and we get,

$$f_s = \frac{\begin{array}{r} (-q_ox - q_o(f_1 - f_o) - f_ox + f_1f_o)(f_2x_2 + f_2x_s) \\ -(q_o(Xx - x^2 - 2f_1x + f_ox) + q_o(f_1X - f_oX - f_of_1) \\ -f_of_1X - Xxf_o + f_ox^2 - 2f_1f_ox)(x_2x_s + f_2x_s) \end{array}}{\begin{array}{r} f_2(-q_ox - q_o(f_1 - f_o) - f_ox + f_1f_o) - (q_o(Xx \\ -x^2 - 2f_1xf_of_1) + f_ox) + q_o(f_1X - f_oX - f_of_1) \\ -f_of_1X - Xxf_o + f_ox^2 - 2f_1f_ox)(x_s - x_2 + f_2) \end{array}} \tag{22}$$

We can write, 22 as a

$$f_s = \frac{N(x)}{D(x)} = \frac{N_2x^2 + N_1x + N_o}{D_2x^2 + D_1x + D_o} \tag{23}$$

where,

$$N_2 = (q_o - f_o)(x_2x_x + f_2x_s)$$

$$N_1 = (-q_o - f_o)(f_2x_2 + f_2x_s) - (x_2x_s + f_2x_s)$$
$$(q_oX - 2q_of_1 + q_of_o - Xf_o - 2f_1f_o)$$

$$N_o = (f_1f_o - q_o(f_1 - f_o))(f_2x_2 + f_2x_s) + (f_of_1X$$
$$-q_o(f_1X - f_oX - f_of_1))(x_2x_s + f_2x_s)$$

$$D_2 = (q_o - f_o)(x_s - x_2 + f_2)f_2$$

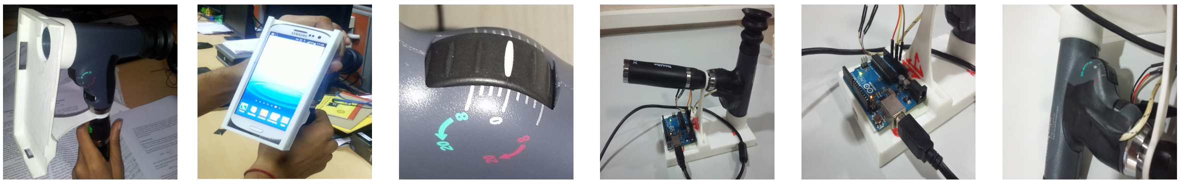
$$D_1 = -q_of_2 - f_2f_o - q_oX + (2q_of_1 - q_of_o$$
$$+Xf_o + 2f_1f_o)(x_s - x_2 + f_2)$$

$$D_o = -q_of_2(f_1 - f_o) + f_2f_1f_o + (f_of_1X$$
$$-q_o(f_1X - f_oX - f_of_1))(x_s - x_2 + f_2)$$

Now from the 22 the lens power P_s is ,

$$P_s = 1/f_s = \frac{D(x)}{N(x)} = \frac{D_2x^2 + D_1x + D_o}{N_2x^2 + N_1x + N_o} \tag{24}$$

From the above mathematical expression, we see that the patient eye refractive power (P_s is function of x. and $P_s = \dfrac{1}{f_s}$. We used this concept and mathematical relation to develop a mobile ophthalmoscope



(a) (b) (c) (d) (e) (f)

Figure 5: Ophthalmoscope based funduscope: (a) Mobile phone holder attached with ophthalmoscope (b) Mobile phone is attached with ophthalmoscope using this holder and (c) Focusing wheel of ophthalmoscope: Its is used to focussing the image onto the mobile phone camera. In funduscopy based refractometer used this wheel's rotation to find out refractive error of the eye/lens. Position Sensor module (d)to (f):Position sensor interfaced with ophthalmoscopy based refractive error screening device for communication with Android application using blue tooth.

based affordable handheld refractive error screening device.

We describe the functional aspects of the design in the next section.

3 DETAIL DESIGN

In this section, we outline the design of funduscopy based optometer using the handheld ophthalmoscope, mobile phone and its application software. This has been made possible by integrating the **optical** (imaging unit: ophthalmoscope and mobile phone camera), **computing** (micro-controller and mobile phone) and **sensor** (position sensor) unit in a single device shown in Fig. 5. The three major components are described below.

3.1 Ophthalmoscope

The ophthalmoscope is an optical instrument for examining the interior structure of the eye, especially the back part of the eye (fundus), which includes the retina, optical disk, optical cup, blood vessel and fovea etc. An Ophthalmoscope is of two kinds, direct and indirect. We use the **PanOptic ophthalmoscope** (Fig. 6(a)) which is very similar to the traditional ophthalmoscope (Welchallyn.com, 2015).

3.2 Mobile Phone

For image capturing, image processing and data sharing, we need a system that has the capability to perform all these jobs. We use a **Smart-phone**, which has these features along with communication and data sharing capability. It full fills the requirement of (a) High resolution CMOS camera (b) Centralised Control System to synchronise and control all process (c) High computation capability to perform image processing (d) high resolution display and (e) sharing result and data with remote users. We have used an Android based Samsung Galaxy S-3 smartphone.

(a) (b)

Figure 6: (a) Pan-Optic Ophthalmoscope and (b) Mobile phone holder rapid prototype model.

Mobile Phone Holder

The smartphone camera is able to capture fundus image only when if it is properly interfaced with a handheld portable ophthalmoscope. In this device, the physician's eye (Fig. 4(b)) is replaced by Smartphone Camera. For that a special phone holder as shown in Fig. 6(b) has been designed. The design of this was done so that the functionality of the basic phone was not impaired.

The material used in RP for building the case of SLS. The mobile case, when integrated with the Ophthalmoscope, is shown in Fig. 5(b).

3.3 Rotary Detail or Sensor Module

In the Fig. 7, ophthalmoscope's rotary wheel is used to focus the retinal image on CMOS sensor camera. This rotary wheel is connected to the focusing lens such that it controls its position (x). It follows the relation: $x = r \times \theta$ where, r and θ are the radius and rotational angle of the focusing wheel of an ophthalmoscope.

To measure the precise position of the rotary

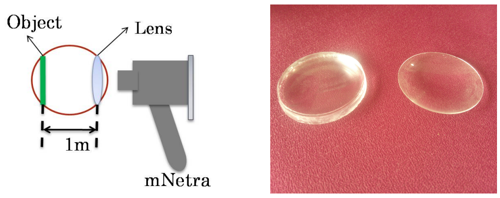
Figure 7: Housing of the position sensor inside the ophthalmoscope to measure the lens position. The sensor will measure the rotational movement (θ) of the focusing wheel.

wheel we have used a high precision position sensor module. In Fig. 5 & Fig. 7, this module consists of a **potentiometer**, a **micro-controller** (Arduino or Intel Galileo Gen 2) board and a Bluetooth module **(RN42N)** is used. This senses the current status of focusing lens position (x) as a function focusing wheel's angle (θ) and sends the processed data to the mobile phone via Bluetooth. The sensor module uses embedded software to control and communicate the wheels rotational information and focusing lens data to the mobile phone.

An Android application has been developed to provide an interactive testing environment to the device. It consists of two sections, one is to display the fundus scan (retinal scan) and another one is to display the current status of focusing lens (refractive error value of the eye).

4 RESULTS

This section discusses the experiments conducted in the laboratory and a hospital for the validation of our new ophthalmoscope based funduscopy type optometer. This has been done in two stages. In the first stage, we have performed a lab based primary concept validation (Fig. 8). Here we put different lenses in front of mobile phone based fundus imaging device and then focus the image on the CMOS sensor using the focusing wheel. This enables us to closely relate the mNetra refractive power ($P(x)$) expression 24 with potentiometer reading. For this, experimental data and the result are shown in Table. 1. It is clear that the ophthalmoscope focusing wheel reading depends on the power of the lens in the test. In the second stage, we have performed the testing and validation with real (subjective) data (**Patients**) in the



(a) (b)

Figure 8: mNetra Concept validation test: (a) Optical setup of the laboratory and (b) the optical lens.

hospital. For this experiment, we chose volunteers in the age group 20-45 year. The volunteers were selected randomly to cover a range of refractive error power.

4.1 Test Methodology

For the above two scenarios, we followed a standard predefined test procedure. In the first stage, we used an optical eye model (Fig. 8(a)) in which variable lens (+3.5D to -6.0D) (Fig. 8(b)) and the object was placed at a distance of 1 meter. And for device validation, we estimate it's lens power (shown in Table. 1) using our funduscopy based method. With the second stage of subjects, we compared our result with the hybrid method of refractive error screening. The data were collected using this protocol.

1. The lighting in the room is dim/dark: This ensures that person pupil's is dilated to the maximum.
2. The person is sitting in the rest condition.
3. To minimize the eye lens power measurement error, observer asks the patient to focus his/her eye on distinct object and infinity. This helps in accurately performing the refraction.
4. The observer/technician measuring the blur in the image using the focusing wheel.
5. Take the reading of focusing wheel when the fundus image quality is best as compared to previous

Table 1: Experimental results: Lens power measurement using mNetra.

S.No	Lens Power (D)	mNetra reading
01.	+3.5	2.0
02.	+2.0	1.25
03.	+1.0	0.75
04.	0.0	0.25
05.	-1.0	0.0
06.	-2.50	-0.25
07.	-3.5	-1.0
08.	-4.0	-1.75
09.	-5.25	-2.25
10.	-6.0	-2.75

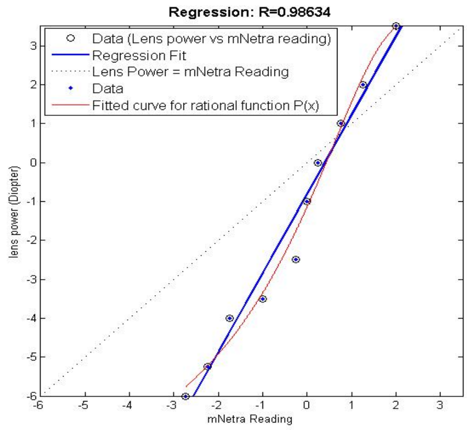


Figure 9: Regression and Curve fitting plot of the actual lens power vs mNetra measured power. R value for mNetra readout is (0.98634).

Table 2: Curve fitting parameter: Lens power $P(x)$ measurement using mNetra.

General model Rational model :	$P(x) or f(x) = (p1 * x^2 + p2 * x + p3)/(x^2 + q1 * x + q2)$
Coefficients (with 95% confidence bounds)	$p1 = -9.967(-54.24, 34.3)$ $p2 = 42.38(-11.15, 95.91)$ $p3 = -16.33(-40.82, 8.148)$ $q1 = -5.104(-27.09, 16.88)$ $q2 = 14.44(-9.752, 38.64)$
Goodness of fit	SSE: 1.276, R-square: 0.9861, Adjusted R-square: 0.9749, RMSE: 0.5052

all images shown on mobile screen.

6. Repeat 3-5 again to take a retinal scan of another eye also.

4.2 Data Analysis

The data that is collected is analyzed for accuracy and efficiency in use for practical settings. The first set of trials gives the test data for obtaining the correlation between the potentiometer based readout and the actual power of the subject lens.

Experiments for the lens of power range +3.5D to -6.0 were performed and the data are shown in Table 1. The data is subject to a curve fit as shown in Fig. 9.

For this regression analysis, we get the regression coefficient 0.98634 and curve fitting result is shown in table 2. This shows that the potentiometer readout is able to track closely the actual power of the subject lens.

Table 3: Subjective trial of device: Refractive power of eye estimated using hybrid method and using funduscopy based device.

S. No	Sex	Age	left eye power (D)	left eye mNetra reading	right eye power (D)	right eye mNetra reading
1	F	35	-1.5	-1.33	-0.75	-1.17
2	F	36	-0.25	-0.42	-0.25	-1.33
3	M	24	-2.25	NA	-1.25	NA
4	M	26	-0.5	-1.125	-0.5	-1.75
5	M	25	-1.25	-1.875	-2.25	-6.25*
6	M	24	-2.25	-2.25	-2.0	-1.25
7	M	23	-0.25	0.0	+3.5	+1.75
8	M	24	-4.0	-3.5	-4.0	-4.0
9	M	23	-8.0	-6.0	-8.0	-6.25
10	F	24	-6.5	-5.25	-6.5	NA
11	M	23	-2.25	-1.75	-2.75	-2.125
13	F	27	-4.25	- 3.0	-4.50	NA
14	M	20	-2.75	-2.75	-2.50	-2.125
*/NA Data sets are not considered for analysis						

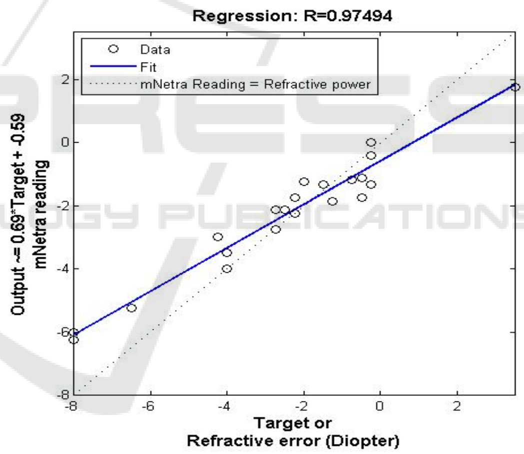


Figure 10: Regression analysis of eye lens power using an ophthalmoscope and Regression coefficient $R = 0.9749$.

Subjective

In the second stage, data are collected from volunteers using our device and hybrid method for refractive error estimation (shown in table.3) is used.

Validation

For the validation, we compared mNetra reading (D_m) with eye refractive power (i.e. true power D_T) shown in table 5. For this, we compute the estimated power using the mNetra power Equation 24 and then find out the error power ($D_e = D_T - D_P$). We got the Standard

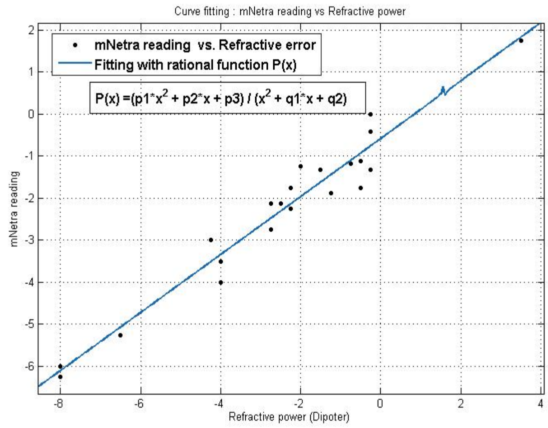


Figure 11: In this graph refractive power (D) vs mNetra potentiometer reading's relation fitted with rational function $f(x) = P(x)$.

Table 4: Curve fitting parameter's value for a rational model of expression 24 with subjective trial.

General model Rational model	$P(x) or f(x) = (p1 * x^2 + p2 * x + p3)/(x^2 + q1 * x + q2)$
Coefficients (with 95% confidence bounds)	$p1 = -2.393e + 04(-9.676e + 07, 9.672e + 07)$ $p2 = 5.816e + 04(-1.291e + 08, 1.292e + 08)$ $p3 = -3.215e + 04(-4.932e + 07, 4.925e + 07)$ $q1 = -3.472e + 04(-1.404e + 08, 1.403e + 08)$ $q2 = 5.469e + 04(-8.377e + 07, 8.388e + 07)$
Goodness of fit	SSE: 3.697, R-square: 0.9505, Adjusted R-square: 0.9381, RMSE: 0.4807

Deviation, Mean, and RMS value for refractive error power (D_e) is 0.4019, 0.4344 and 0.5686.

Table 5: Subjective refarctive data for the validation and performance (error power) analysis of mNetra.

mNetra Reading (D_m)	True power (D_T)	Estimated power $D_P = P(D_m)$	Error power $D_e = D_T - D_P$
-0.625	-0.75	-1.0191	0.2691
-0.25	-0.50	-0.7604	0.2604
-0.67	0.0	-1.0501	1.0501
-0.25	0.0	-0.7604	0.7604
-2.90	-2.25	-2.5875	0.3375
-2.67	-2.50	-2.4290	-0.0710

5 CONCLUSION AND FUTURE WORK

This paper discusses the design and use of an Ophthalmoscope based Optometer/ refractometer which can be used for affordable eye care. It has been shown that the application of this mobile phone based fundus imaging device can be used to determine retina impairment and ametropias. Primary experimental data also suggest its usability and easy handling capability. It can be used in primary health care center, OPD and Healthcare camp where fast screening is necessary. On the basis of stage one result, the device also can be used to measure the optical power of the lens.

Some of the features like image/video stabilization available in the modern phone can be integrated in the application in the future. This basic infrastructure can be used for screening of other ocular diseases.

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