Sensors and Actuators A 148 (2008) 388–394

Contents lists available at ScienceDirect

Sensors and Actuators A: Physical

journal homepage: www.elsevier.com/locate/sna

Class I infrared eye blinking detector

Fabio Lo Castro

CNR-Institute of Acoustics, Via del Fosso del Cavaliere 100, 00133 Rome, Italy

ARTICLE INFO

Article history:
Received 12 March 2008
Received in revised form 21 July 2008
Accepted 2 September 2008
Available online 10 September 2008

Keywords:
Infrared
LED
Blinking
Safety
Eye
Detector

ABSTRACT

Eyes blinking detectors are often used to monitor the drowsiness or the fatigue during any activity. Therefore, sensor has to be as safe and less intrusive as possible in order to avoid additional annoyance and hazards. The proposed detector has been developed according to the Class 1 specifications of IEC 60825-1 standard on the safety of LASER and LED products in order to work until 8 h in a safety condition for the user. The sensor used is a commercial infrared emitting diode coupled to an infrared photodiode, embedded in the same device and usually used for data link, clipped on the rod of the glasses and placed closed to the eye. The sensor uses the technique of the modulation of the infrared beam to be less sensitive to the head movements and changes of environmental light. Actually a cable connects the sensor to the control board that sends the eye blinking signal to a personal computer in order to be recorded.

1. Introduction

Eyes are a good candidate to be monitored in order to recognize the fatigue or the disease of a person. In particular the eye lid, the pupil and the gaze can be observed.

Different technique and methods can be used, as electrooculography (EOG), electromyography (EMG), infrared-oculography (IR-OG) and image based methods called video oculography (VOG) [1,2]. Each one of these has its advantages and disadvantages.

The last two methods require an infrared source as LEDs. Although, in the last years LEDs and IR-LED were considered safe on the contrary of lasers, today this is not still true considering their high non-tolerable irradiance.

Thus according to the standard IEC 60825-1, regarding the eyes safety, we have developed an infrared eye blinking detector that can be considered safe.

This paper will describe how it is possible to develop a safe eye blinking detector, the choice of the rules to make it safe and the evaluation of its level of eyes safety.

1.1. Eye movements

Eyelid movement is one of the visual activities that reflects a person's level of fatigue.

There are two principal ocular measures that characterize the eyelid movement. The first one is the percentage of eye closure over time (*PERCLOS*) [3] and the second one is the average eye-closure speed (*AECS*), i.e. the amount of time needed to fully close or open the eyes. PERCLOS is the most valid ocular parameter for monitoring fatigue and it is evaluated computing the cumulative eye-closure duration over time, excluding the time spent on normal eye blinks. In literature the eye blink is defined as a closure followed by an eye opening within a period of 1 s [4]. The degree of eye opening is characterized by the shape of the pupil. It is not always the same but changes getting more and more elliptical as soon as the eyelids start to close the eyes.

There are two types of eyes blinks: voluntary eye blink and spontaneous eye blink.

The voluntary blinking is a very rapid movement; it takes only 0.2 s, about half of the time taken by spontaneous blinking [5].

The spontaneous blinking is the most frequent movement of the eyelid. It consists in rhythmic involuntary closing and opening, at a rate of 10–30 movements per minute depending principally on his activity and health state [6]. The whole movement (opening and closing) lasts 0.3–0.4 s and it is repeated after an interval of 2–10 s. Slow-camera cinematography shows that upper lid falls like a curtain; the lower lid remains almost motionless.

Spontaneous blinking is an indicator of fatigue, increasing its rate, while we spend time watching a display [7], like a TV or a monitor, reading [8] a book or while we are driving [9].

Furthermore it gives information about the presence of disease as Parkinson syndrome [10], stroke [11] and others nervous system diseases [12]. In the first case it is commonly found a slow blinking

E-mail address: fabio.locastro@idac.rm.cnr.it.

0924-4247/$ – see front matter
doi:10.1016/j.sna.2008.09.005

rate while in the other cases there is frequently a high blinking rate during resting.

Gaze has the potential to indicate a person's level of vigilance, finding a narrow gaze in fatigued individual.

1.2. Ocular measurement techniques

Traditional methods to record the eyes blink signal are the EOG and the EMG.

EOG uses three silver/silver chloride or gold electrodes placed on the skin, up and down the eye, for measuring the voltage difference between the cornea and the fundus, that is directly related to the eyes movement and the blinking. Also the EMG uses electrodes to catch signal from muscles close to the eyes.

Those two methods present different problems regarding the electrodes. In fact they could detach from skin during session, for example if a person is sweaty, or they could have a bad contact, increasing the impedance of the electrode and causing a low signal to noise ratio. Furthermore the electrode or the plaster that hold up the electrode could cause local irritation or could disturb the person during his activity.

At last the movement of the person can cause artefacts in the signal.

To avoid these problems new methods as the IR-OG and the VOG have been developed. They are contact free and non-intrusive and for these reasons are more and more used to detect the fatigue, especially during driving.

IR-OG is based on the principle that if a fixed light source is directed against the eye, the amount of light reflected back to a fixed detector changes with the pupil position and if the eye is open or closed.

This principle has been exploited in a number of commercially available eye trackers used also on the animals [13–18].

The infrared light is used because it is invisible to the eye, and does not distract the subject (driver).

As infrared detectors, that are sensible only to the infrared wavelength, are not influenced by other light sources, as the bulb or fluorescent lamp (having them a precise working frequency) or the ambient lighting (if it changes slowly).

The IR-OG usually is used for measuring horizontal rather than vertical eye movements and it is able to achieve spatial resolutions (the size of the smallest movement that can reliably be detected) in the order of 0.1° and temporal resolutions of 1 ms or more.

The disadvantage is that during horizontal and vertical measurement blinks can be a problem, not only because the lids cover the surface of the eye, but also because the eye retracts slightly after the blink, altering the amount of light reflected for a short time. But if we are interested only on the eyelid movement this is not an inconvenient. The advantage of the IR-OG method is that it is portable, wearable and cheap compared to the previous system. The only precaution that has to be taken is on the angle between the IR emitter and detector, because it is vital for the detector in order to receive the signal reflected back from the eye.

The VOG usually uses a video camera and software for the eyes tracking and blinking evaluation [3,19] in order to evaluate the blinking rate and also the gaze direction. The direction of a person's gaze is determined by two factors: the orientation of the face (facial pose) and the orientation of the eye (eye gaze). Facial pose determines the global direction of the gaze, while eye gaze determines the local direction of the gaze. Global and local gazes together determine the final gaze of the person.

In order to find the pupil and retina on the picture the VOG systems use an infrared spot to lighten the retina so that it can be easy identified.

Table 1
Eyes pathologies related to the infrared and other exposed radiation wavelength

Spectral region	Eye	Skin
Ultra-violet C (180–280 nm)	Photokeratitis	Erythema (sunburn) Accelerated skin ageing process
Ultra-violet B (280–315 nm)		Increased pigmentation
Ultraviolet A (315–400 nm)	Photochemical cataract	Pigment darkening
Visible (400–780 nm)	Photochemical and thermal retinal injury	Photosensitive reactions
Infra-red A (780–1400 nm)	Cataract, retinal burn	Skin burn
Infra-red B (1.4–3.0 μm)	Aqueous flare, cataract corneal burn	
Infra-red C (3.0 μm–1 mm)	Corneal burn only	

The advantage of the image based method is that it leaves completely free of movement the subject, but he has to be in front of the camera, otherwise the camera has to be put on the head of the subject.

The inconvenience is that it has a low resolution, due a frame rate, and requires a heavy image recognition software and hardware.

2. Eyes hazard

Both the IR-OG and VOG considered above emit an infrared light that is pointed toward the eyes, that is not visible, hence not blinding and not giving disturbance that cause the closure of the eyes. Furthermore the emitter is located at a distance, that can change from 0.01 to 1 m (according to the type of eyes movements detector) and it could work for different hours.

These considerations point out situations that could be dangerous to the eye. In fact being the crystalline transparent to the near infrared light, it arrives directly on the retina and could cause injury or burn as shown in Table 1 [20].

This mean that the irradiance that arrives to the retina is in the order of 100k times greater than that on the cornea [21], considering 7 mm the diameter of the pupil (worst case, young eye completely dilated) and the pupil corresponding image between 10 and 20 μm in diameter. Hence an irradiance of 1 W m^{-2} at the cornea level becomes 100 kW m^{-2} on the retina. Other wavelengths that do not go through the crystalline could cause cataract or corneal irritation.

The amount of energy absorbed by the tissue changes with the wavelength of the light and the tissue itself. It could cause burns, lesions and temporary or permanent loss of vision, depending on the magnitude of the exposure. On the contrary, the skin tolerates the exposition to radiation better than the eye.

The hazard is virtually independent of the distance between the source and the eye in case of point or extended source with a well collimated beam, as a LASER.

In case of a point-type and diverging beam source, as a LED, the hazard increases with decreasing distance between the beam source and the eye. This is true until distance is greater than the shortest focal length. Thus for distance less than the shortest focal length there is a rapid growth of the retinal image and a corresponding reduction of the irradiance, even though more power may be collected.

2.1. Safety evaluation

Different studies on the hazard assessment and the reactions of the human tissues to the laser radiation have developed different standard in different country, but often they have in common the

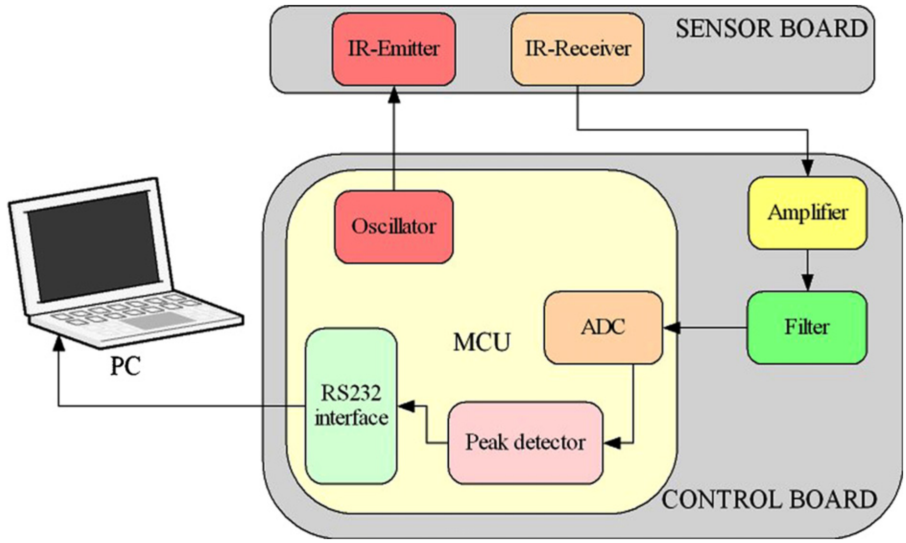

Fig. 1. Flow chart of the proposed eye blinking detector.

criteria and the formulae to evaluate the limit levels of radiation in order to classify the safety levels of the products [22,23].

Lasting time standards have considered LEDs and have treated them as Laser from the standpoint of safety. One of these standards is IEC 60825-1 and we have chosen it as reference to develop our eyes blinking detector.

It gives us some limit parameters for the exposition to both visible and invisible radiation: the maximum permissible exposure (MPE), the accessible emission level (ALE) and the nominal ocular hazard distance (NOHD):

$$\mathrm{MPE}(\lambda, T_{\mathrm{expo}}, \alpha, f) = k_1(\lambda, T_{\mathrm{expo}})t^{k_2(\lambda, T_{\mathrm{expo}})}c_4(\lambda)c_5(f)c_6(\alpha)\ [\mathrm{J\,m^{-2}}]\ \text{or}\ [\mathrm{W\,m^{-2}}] \quad (1)$$

$$\mathrm{AEL}(\lambda, T_{\mathrm{expo}}, \alpha, f) = k_1(\lambda, T_{\mathrm{expo}})t^{k_2(\lambda, T_{\mathrm{expo}})}c_4(\lambda)c_5(f)c_6(\alpha)\ [\mathrm{J}]\ \text{or}\ [\mathrm{W}] \quad (2)$$

$$\mathrm{NOHD}(\theta, d, P, \mathrm{MPE}) = \frac{1}{\theta}\left(\sqrt{\frac{4P}{\pi \mathrm{MPE}}} - d\right)\ [\mathrm{m}] \quad (3)$$

where λ is the wavelength, t is the pulse duration, α is the subtended source angle, k_1 and k_2 are the constant values tabled in the standard, T_{expo} is the time of exposure, f is the frequency of the impulse, θ is the divergence, d is the apparent source diameter and P is the radiant power.

Furthermore the functions c_4, c_5 and c_6 are the correction factors added in the last years in order to permit at the AEL and MPE values to fit more closely the actual variation of biological injury thresholds [24]. Their limit value is evaluated so that it is about one tenth of the power density that would cause permanent eye damage to 50% of the population.

$c_4(\lambda)$ permits to increase the threshold limit values at different wavelengths due to the reduced absorption of some specific tissues, $c_5(f)$ takes in consideration the presence of the pulse train and $c_6(\alpha)$ has been added in order to consider extended source.

In this study we have pointed our attention to the pulsed radiation taking into account what has been specified in the IEC 60825-1 standard as far as Class I characteristics is concerned. In this case, in order to evaluate the AEL and the MPE, the standard considers the single pulse width, the pulse train duration and the number of pulses inside the train. Hence three cases have to be studied in order to find the most restrictive value:

- The exposure from any single pulse within a pulse train shall not exceed the AEL/MPE value (for a Class 1 device) for a single pulse.
- The average power for a pulse train of duration T shall not exceed the power corresponding to an establish AEL/MPE for a single pulse of duration T.
- The exposure from any single pulse within a pulse train shall not exceed the AEL/MPE for a single pulse multiplied by the correction factor $c_5(f)$: $\mathrm{AEL_{train}} = \mathrm{AEL_{single}} \times c_5(f)$.

3. System architecture

The proposed prototype is an IR-OG device. The device circuitry is divided in two electronic boards: one is plugged on the temple containing the sensor and the other is used to drive the emitter LED, to treat the signal and to transmit the data to a personal computer (Fig. 1).

In order to realize a wearable infrared eyes blinking detector, using the smallest possible sensor, it has been chosen the couple diode/photodiode as infrared detector made by the Agilent. Its

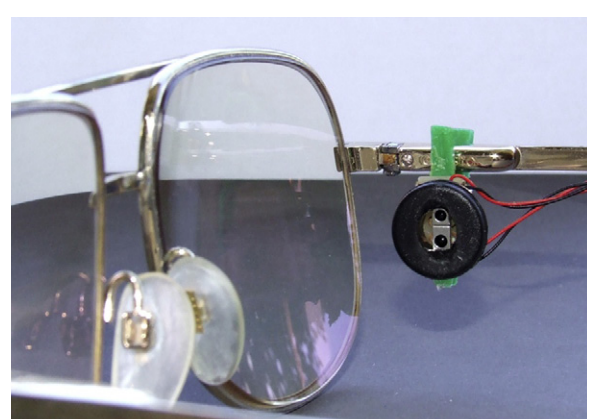

Fig. 2. Infrared sensor placed on the temple of the glasses.

Table 2
Agilent HSDL9100-21 IR-transmitter and receiver characteristics

Parameter	Value	Unit
Wavelength, λ	940	nm
Bandwidth	50	nm
Diameter, tx	1.8	mm
Diameter, rx	2	mm
Apparent source size	0.8×0.8	mm
Junction, ΔV	0.9–1.5	V
Beam divergence (°)	26[a]	degree
Radiant flux	7.6[b]	μW

[a] Measured considering an area collecting the 63% of the total power.
[b] Measured according the standard IEC 608251-1; detector model: Newport's 818 SL.

dimension allows to be plugged on the temple of the glasses as shown in Fig. 2.

The emitter/detector characteristics are shown in Table 2. The values as the apparent source size, the irradiance and the divergence that were not given by the datasheet have been evaluated.

The working principle of the control board is shown in Fig. 1. The core of the control board is a microcontroller (MCU) that it is also used as an oscillator to drive the IR-LED at the frequency around 1 kHz (value that gives us a good result considering the amount of light reflected back) in order to avoid environmental change of luminosity. The IR-LED is powered by a square wave. During the high level the IR-LED is on. The impulse is the shortest possible in order to supply the maximum admissible current. The amount of light reflected back on the photodiode by the eye produces a current proportional to the light received. In this way the signal changes if the eyelid is open or closed. As the output signal of the sensor is tiny it needs to be amplified and filtered, using an envelope detector. Then it is converted in digital form by an analogical to digital converter in order to be analyzed by a microcontroller. Here it is implemented an algorithm to simulate a peak detector able to find the blinks. The algorithm compares the acquired signal to the reference level one, used as trigger, establishing if the eye is open or closed. At last the digital signal can be sent via the RS232 interface to a data logger or a personal computer so that it can be recorded, displayed or analysed in order to give an alarm.

4. Results and discussion

Lasting time many eye blinking detectors using an infrared light source and photodiode as sensor have been developed [14–18], but the novelty of this paper is to show how it is possible to develop a safe eye blinking detector, the processing rules to follow in order to make it safe and to evaluate its safety level.

In order to choose the appropriate rules, we looked for one of the most severe standard, the IEC 60825. According to this, it is not sufficient to claim that a new device, using a LED inside, is safe because the LED itself is certified as safe by the manufacturer. But the safety of the LED product must be considered taking into account the whole system, its application and the surrounding environment.

For example the safety of the LEDs used for the data link, as the TV remote control, is evaluated considering the LED turned on for 100 s [25,26], a duration much greater than the real exposure period to the infrared beam for a person (usually less than 1 s).

The proposed device must work typically for 8 h per day and the eye is always lit by the LED (it is not developed for medical uses), so the information on the safety provided by the datasheets have to be considered with great caution. If the limit values are not available they must be evaluated, in the specific context, as described in the paper.

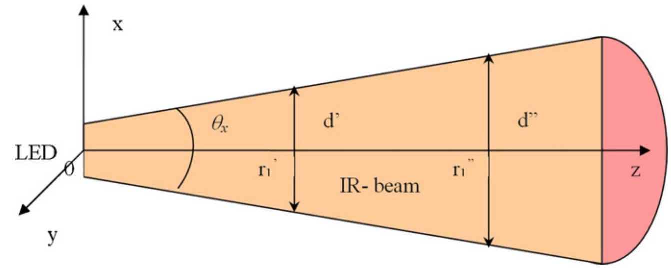

Fig. 3. IR-LED beam divergence measurement technique.

The European standard IEC 60825-1 (1998) assigns the same hazard levels to LEDs and lasers and hence it treats both as laser, even if the power density on the retina emitted by the LED is over two orders of magnitude less than that of a laser source of the same power. This is because laser light can be focused on the retina to a very small spot, several wavelengths in diameter, in contrast with LED light (that is multimode radiation), that cannot be focused on to less than the source area, typically more than 25 μm in diameter.

In this way the IEC 60825-1 overestimates the risk from LED sources [27]. For softer limit let's refer to the USA ANSI RP-27 (1996) (Recommended Practice for Photobiological Safety for Lamp Systems) [28].

In order to satisfy the IEC 60825-1 standard requirements the first parameter that has been evaluated is the divergence θ (Table 2) of the beam given by

$$\theta_i = 2\ \arctan\left(\frac{d_i - d}{2r_1}\right) = 2\ \arctan\left(\frac{d_i'' - d_i'}{2(r_1'' - r_1')}\right)\ [\mathrm{rad}] \quad (4)$$

where d_i is the diameter of the projected image area (containing, for safety, the 63% of the total power instead of 95%), on the ith axis, at r_1 distance from the source, d_i'' and d_i' are the two diameters measured, respectively, at distance r_1'' at r_1' with $r_1'' > r_1'$ (Fig. 3).

$$\theta = \frac{\theta_x + \theta_y}{2} \quad (5)$$

In order to find the solid angle Ω from the beam divergence it is used the following equation

$$\Omega = \frac{A}{r_1^2} \approx \frac{(d_1/2)^2\pi}{r_1^2} = 4\pi\ \sin^2(\theta) = 2\pi(1 - \cos(2\theta))\ [\mathrm{sr}] \quad (6)$$

where A is a portion of spherical surface at r_1 distance and d_1 is the diameter of the projection of the surface A on a flat wall.

Observe that the flat area approximation of the spherical area results in an error less than 1% when the solid angle is less than 0.03 sr.

The angle α subtended by the source at distance r = 0.1 m (the shortest human focal length) from the eye is given by

$$\alpha = 2\ \tan^{-1}\left(\frac{d/2}{r}\right)\ [\mathrm{rad}] \quad (7)$$

If the apparent source is not a circle then α will be the arithmetic mean of the two angles subtended by the source each one for each axis. Furthermore if any dimension is greater than 100 mrad it has to be chosen equal to 100 mrad and consistently the emitted power has to be reduced.

The second useful parameter is the radiant power P emitted by the source. According to the standard IEC 60825 if the light source is a in the wavelength range from 400 to 1400 nm and subtending an angle α > 1.5 mrad then the radiant power has to be measured through an aperture of diameter d = 7 mm (conventional eye diameter) place at a distance r from the source; where

$$r = 100\sqrt{\frac{\alpha + 0.46\,\text{mrad}}{\alpha_{\max}}}\ [\text{mm}] \quad \text{with}\ \alpha_{\max} = 100\,\text{mrad} \tag{8}$$

Usually the LEDs have a lens, the package itself, which shifts the position of the point source. Thus in order to find the virtual distance x of the source from the lens can be used the following equation (coming from the inverse square low),

$$E_1(x + r_1)^2 = E_2(x + r_2)^2 \tag{9}$$

where E_1 and E_2 are the irradiances (W m^{-2}) measured at distances r_1 and r_2 from the lens.

In order to know which is the NOHD, i.e. the distance over that there is not an exposure risk, Eq. (3) has to be considered. Note that Eq. (3) is an approximation because it does not take in account the atmospheric attenuation of the beam (negligible for short distance). Rearranging Eq. (3) it is possible to evaluate the maximum power emitted by a safe device at any distance

$$P < \frac{d^2}{4}\pi\text{MPE}\ [\text{W}] \tag{10}$$

Using the values in Table 2 and considering 8 h as maximum exposure time and/or an impulse duration t between 50 μs and 1000 s the AEL and the MPE given in Eqs. (1) and (2) can be written as

$$\text{AEL} = 7 \times 10^{-4} t^{-0.25} 10^{0.002(\lambda-700)} \frac{\alpha}{\alpha_{\min}} c_5\ [\text{W}]$$

$$\text{for}\ 18 \times 10^{-6} < t < 1000\ [\text{s}] \tag{11}$$

$$\text{MPE} = 18 \times t^{-0.25} 10^{0.002(\lambda-700)} \frac{\alpha}{\alpha_{\min}} c_5\ [\text{Wm}^{-2}]$$

$$\text{for}\ 18 \times 10^{-6} < t < 1000\ [\text{s}] \tag{12}$$

where $c_4 = 10^{0.002(\lambda-700)}$, $c_6 = \alpha/\alpha_{\min}$ being and $\alpha_{\min} < \alpha < \alpha_{\max}$.

The AEL and the MPE, using the curve for λ = 905 nm (the closest at 940 nm) shown in figure 10b of the IEC EN 60825-1-1998-5 can also be written as

$$\text{AEL} = 2.7 \times 10^{-4} 10^{0.002(\lambda-700)} \frac{\alpha}{\alpha_{\min}}\ [\text{W}]$$

$$\text{for}\ 1000 < t < 30,000\ [\text{s}] \tag{13}$$

$$\text{MPE} = 7 \times 10^{0.002(\lambda-700)} \frac{\alpha}{\alpha_{\min}}\ [\text{W m}^{-2}]$$

$$\text{for}\ 1000 < t < 30,000\ [\text{s}] \tag{14}$$

where $\alpha_{\min}$ = 1.5 mrad for t < 0.7 s, $\alpha_{\min} = 2 \times t^{3/4}$ mrad for 0.7 s ≤ t < 10 s, $\alpha_{\min}$ = 11 mrad for t ≥ 10 s, c_5 = 1 for single impulse calculation or for t ≥ 0.25 s and 400 nm < λ < 1 mm, $c_5 = N^{-1/4}$ for N impulse evaluation and for t < 0.25 s (400 nm < λ < 1 mm).

Considering that it must work for hours close to the eye, the base time of 30,000 s, required by the standard has limited the maximum radiant flux of the source especially in relation to the NOHD.

Using the evaluated parameters of the LED (as the radiant flux, the irradiance versus the supplied current and the angular aperture) and the parameters given by the datasheet we find that our emitter can be classified as Class 1 product and hence it is safe.

Indeed the selected LED radiant flux of 7.6 μW is less than 586 μW (AEL) required by the standard and the irradiance at the eyes is 0.20 W m^{-2} less than 15.4 W m^{-2} (MPE) too. Furthermore, the operative distance is safe because the NOHD value is zero or negative according to the radiant flux equal or less than 7.6 μW and hence any distance is safe for the eyes.

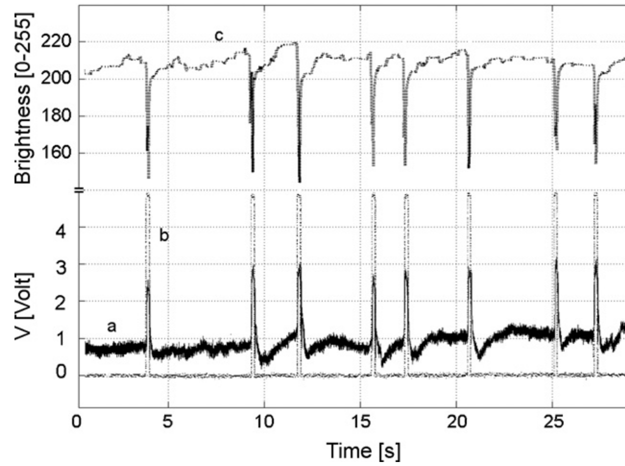

Fig. 4. Eye blinking signals. (a) Sampled data of the proposed IR-OG device; (b) peak detector digital signal; (c) VOG signal.

Finally the standard requires that for final products incorporating procured parts or modules, two sets of single fault conditions must be only considered: the procured parts single fault and the failure of the final product (exclusive of the targeted procured part which causes the procured part to exceed the AEL class limit). Any fault as a short or an open circuit inside the procured part (the LED in our case) cannot increase the emitted power by itself (part is passive, in the sense that it has only one external power that is supplied by our circuit). An electronic circuit limits the current value avoiding an overcoming of the AEL/MPE values (developer fault condition). The fault that usually could happen when the user is plugging or unplugging the sensor on the temple is to leave the LED on, because he does not take care of it being the light not visible. So that the LED could go up to the eye or in front of the pupil more than the normal condition. This is not a problem because as mentioned above it can be used at any distance normal to the pupil.

The application for which this device has been developed is to monitor the fatigue of the driver observing the duration of the blinking and the blinking rate, i.e. the number of impulses per minute detected from the device.

An example of a sequence of recorded blinks is shown in Fig. 4; the (a) track is the analog signal of the proposed device while the (b) track is the output of peak detector unit, useful to evaluate the duration and the rate of the blinking.

When the digital signal remains at the high level (closed eyes), it means that the driver is sleeping. Until now this case cannot be distinguished by a bad functioning or positioning of the sensor. In

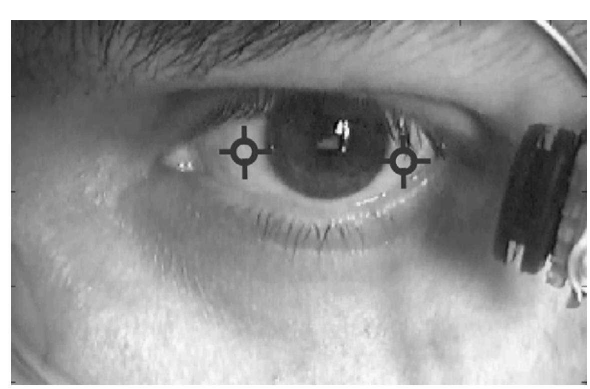

Fig. 5. Frame of a video sequence. The targets on the left and the right of the pupil show the areas used to evaluate the brightness signal. On the right of the picture is visible the infrared sensor.

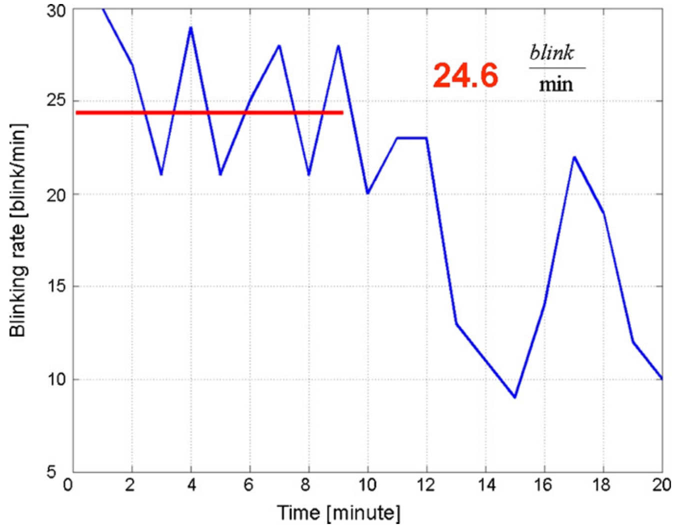

Fig. 6. Example of the blinking rate trend at the beginning of the experiment (low blinking rate).

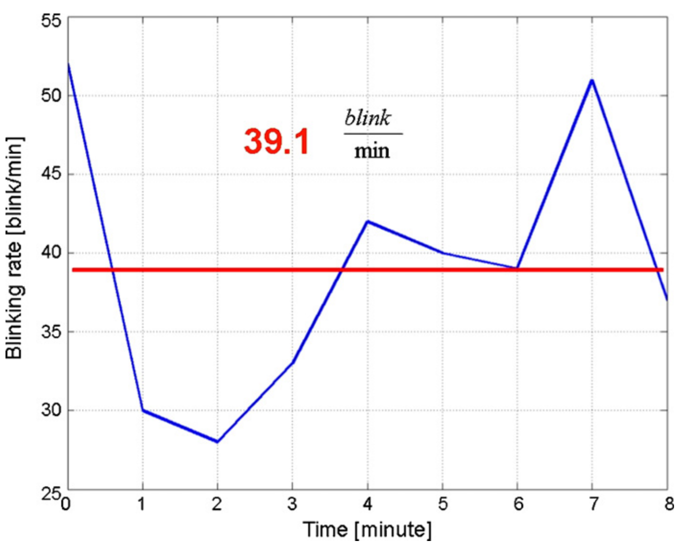

Fig. 7. Example of the blinking rate trend 2 h later the beginning of the experiment (high blinking rate).

order to avoid this type of failure it would be worthwhile to use a sensor for each eye, increasing in this way the redundancy.

In order to validate the acquired blinking signal, it was compared with that coming from another system such as a video camera. An example of the signal used to validate the Class I prototype is shown in Fig. 4 as the (c) track.

The validation signal is obtained processing a video sequence of the eye at 30 frames per second taken in the same time of the infrared signal. The signal is got evaluating the brightness of the two areas on the left and the right of the pupil, in order to avoid eyes movement, as shown in Fig. 5.

Figs. 6 and 7 show an example of the blinking rate at the beginning of a run test during driving and at the end after 2 h. Analysing the first 9 min, here was present a stable oscillation in the rang around 20–30 blinks for minute and an average of 24.6 blinks for minute, while after 2 h, considering the same time, values were in the range around 30–50 blinks for minute and its average is increased up to 39.1, showing a tiring out.

5. Conclusions

In this paper we have proposed a safe eye blinking detector used to monitor the fatigue, especially of the driver, but it has the potentiality to be used in medical field for the diagnosis of particular diseases.

As the sensor is placed closed to the eye we have focused our attention on the safety of this device. Therefore, in order to develop a detector able to work for 8 h in the maximum safety conditions it has been chosen one of the most restrictive standards as the IEC 60825-1.

According to this standard our device can be classified as Class 1, i.e. the device is safe to eye and skin under all reasonably foreseeable conditions of operation.

This have implied that the emitter LED had to be powered less then its maximum power, confirming that the safety of a product depend on its application as well as its emission irradiance, the time of exposition and the operative environment.

Furthermore the functionality of the system has been validated comparing the recording data with that coming from another system using a video camera.

Future developments of the prototype will concern the elimination of the wire to send data, allowing free movements for a person in his environments.

Acknowledgements

The author wishes to acknowledge Dr. G. Brambilla of the CNR-Institute of Acoustics and Dr. A. Reale of the Department of Electronic Engineer at the University of Rome "Tor Vergata" for their support during the development of the project.

References

[1] D.F. Dinges, M. Mallis, G. Maislin, J.W. Powell, Evaluation of techniques for ocular measurement as an index of fatigue and the basis for alertness management, Washington DC: National Highway Traffic Safety Administration, US Department of Transportation. Publication 808 762 (4) (1998).
[2] J.H.J. Allum, F. Honegger, M. Troescher, Principles underlying real-time nystagmus analysis of horizontal and vertical eye movements recorded with electro- infra-red- or video-oculographic techniques, J. Vestib. Res. 8 (6) (1998) 449–463.
[3] Z. Zhu, Q. Ji, Real time and non-intrusive driver fatigue monitoring, IEEE Trans. Vehicular Technol. 53 (4) (2004) 1052–1068.
[4] S.R. Quartz, M. Stensmo, M.S. Makeig, T.J. Sejnowski, Eye Blink Rate as a Practical Predictor for Vigilance, Society for Neuroscience, San Diego, November 1995.
[5] B.A. Houssay, Human Physiology, McGraw-Hill, 1971, pp. 1951–1955.
[6] F.H. Adler, R.A. Moses, Adler's Physiology of the eye: Clinical application, C.V. Mosby Co. (1981), 1–15.
[7] K. Kaneko, K. Sakamoto, Spontaneous blinks as a criterion of visual fatigue during prolonged work on visual display terminals, Percept. Motor Skills 92 (1) (2001) 234–250.
[8] J.A. Stern, D. Boyer, D. Schroeder, Blink rate: a possible measure of fatigue, Hum. Factors 36 (2) (1994) 285–297.
[9] F. Lo Castro, Driver Psycho-Physical Alteration Recognition System, Ph.D. Thesis, Dep. of Elect. Eng. Univ. of Rome "Tor Vergata", Italy, 2007.
[10] W.H. Chen, T.J. Chiang, M.C. Hsu, J.S. Liu, The validity of eye blink rate in Chinese adults for the diagnosis of Parkinson's disease, Clin. Neurol. Neurosurg. 105 (2003) 90–92.
[11] D. Djordjevic, A. Jovicic, D. Veljancic, S. Petkovic, Diagnostic and predictive value of long latency reflexes. Blink reflex and somatosensory evoked potentials in the patient with TIA, RIA and brain infarction, Electroencephalogr. Clin. Neurophysiol. 99 (4) (1996) 352.
[12] A.E. Smit, P. Buisseret, C. Buisseret-Delmas, C.I. De Zeeuw, F. VanderWerf, F. Zerari-Mailly, Reticulo-collicular and spino-collicular projections involved in eye and eyelid movements during the blink reflex, Neurosci. Res. 56 (2006) 363–371.
[13] M. Orlowska-Majdak, P. Kolodziejski, K. Dolecki, W.Z. Traczyk, Application of infrared detection in the recording of eyelid movements in rabbits, Acta Neurobiol. Exp. 61 (2001) 145–149.
[14] J.F. Disterhoft, H.H. Kwan, W.D. Lo, Nictitating membrane conditioning to tone in the immobilized albino rabbit, Brain Res. 137 (1) (1977) 127–143.
[15] K.H. Mannik, Eye actuated sleep prevention devices and other eye controlled devices, United States Patent, 5,682,144 (1997).

[16] M. Brannen, S. Sanderson, Sleep prevention apparatus and method, United States Patent, 5,982,287 (1999).
[17] P.D. Hiltman, Automatic reset for personal alert safety system, United States Patent, 5,909,179 (1999).
[18] T. Selker, A. Lockerd, J. Martinez, Eye-R—A Glasses-Mounted Eye Motion Detection Interface, Conference CHI, 2001.
[19] A. Amir, L. Zimetb, A. Sangiovanni-Vincentelli, S. Kao, An embedded system for an eye-detection sensor, Comput. Vision Image Understanding 98 (2005) 104–123.
[20] IEC EN 60825-1 1998-05, Safety of Laser Products- Part 1. Equipment Classification: Requirement and User's Guide.
[21] D. Slineyand, M. Wolbarsht, Safety with Lasers and Other Optical Sources, Plenum Press, New York, 1980.
[22] D.H. Sliney, Risk assessment and laser safety, Opt. Laser Technol. 27 (5.) (1995).
[23] ICNIRP, ICNIRP statement on light-emitting diode (LEDS) and laser diodes: implication for hazard assessment, Health Phys. 78 (6) (2000).
[24] D.H. Sliney, Laser effects on vision and ocular exposure limits, Appl. Occup. Environ. Hyg. 11 (4) (1996) 313–319.
[25] Agilent Technologies, LED Eye Safety, Application Note 1109, 1999.
[26] A.C. Boucouvalas, IEC-1 Eye Safety Classification of Some Consumer Electronic Products, IEE Colloquium on Free Space Communication Links, Paper No. 13, London, February 1996, pp 13/1–13/6.
[27] E. Sutter, Extended source—concepts and potential hazards, Opt. Laser Technol. 27 (1.) (1995).
[28] D.A. Kahn, Optical Wireless and Safety—A Primer, Plaintree Systems, Inc., 2000.

Biography

Fabio Lo Castro was born in Rome, Italy, in 1971. He received the Laurea and the PhD degrees in electronic engineering from the University of Tor Vergata. His research activities are concerned with the development of sensor interfaces, applications and data processing. He designed and developed the prototype of the Temperature Control System for the ST-Microelectronics DNA Chip (2002–2003). He has been co-responsible of two experiments during the Eneide Space Mission on the International Space Station (April 2005). In the year 2006 he got a research grant at the CNR Institute of Acoustics. He is author or co-author of more than 20 papers in journal and conference proceedings and more than 40 technical reports.