Electrooculography Based iOS Controller for Individuals with Quadriplegia or Neurodegenerative Disease

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***Abstract*— As the use of tablet computers and cell phones has become a standard medium of access to information, entertainment, and communication around the world, the reliance on having access to such devices has increased tremendously. For individuals with quadriplegia or neurodegenerative diseases, the access to these mobile devices is greatly hindered due to their inherent touchscreen design. Assistive technology solutions available to such patients today require families of patients to invest thousands of dollars in standalone tablet systems. There are few known options for allowing such patients to connect to their existing tablets or smartphones, which already have access to apps that can assist them in communication and daily activities. For this reason, we present in this paper a low-cost commercial off the shelf (COTS) assistive communication device to allow individuals with such conditions to access iOS based devices through electrooculography signals captured from their eye movements. Signals are captured through electrodes placed on the users face around the eyes. These signals are filtered, amplified, and processed to detect key eye movements mapped to perform control outputs sent to the iOS device. The communication capabilities are tested through the administration of a typing test to measure characters typed per minute (cpm). Testing of the device includes timed trials of directed tasks carried out by both healthy subjects and patients with ALS (PALS). It was determined that a user can type an average of 3.25 ~ 6.11 cpm using the device with an average accuracy of 89%. This could be significantly improved using a better suited keyboard application on the phone.**

Keywords—EOG; Assistive Technology; Accessibility; ALS; Bluetooth; Tablet; Smart Phone; Embedded System; Communication Device; Switch Control;

I. Introduction

A. Overview

For individuals with motor neural diseases such as Lou Gehrig's disease, also known as Amyotrophic Lateral Sclerosis (ALS), or individuals suffering from quadriplegia, many of the modern everyday activities we take for granted like checking email on a smartphone can be nearly impossible to accomplish without assistive technology (AT) devices. Because these diseases severely affect an individual's motor functions, use of any device which requires touch screen or hands on interaction to access becomes increasingly difficult as symptoms intensify. There are currently options for such individuals in terms of eye controlled communication devices, but they are quite costly and often require users to purchase an entire standalone tablet system. Depending on the capabilities of the device, the price of a communication device can range from $2,000-$15,000. Although they can be partially or fully covered by insurance, including Medicare thanks to the Steve Gleason Act (2015), some insurance plans will force patients into a choice between a communication system and crucial healthcare [22]. When combined with the additional costs of medical care, medication, a wheelchair, and caretaking, the patient and their family must prioritize where insurance coverage will be utilized. Because of this, a communication device may have to wait until the more critical areas are assessed. Poor planning in terms of disease progression can also end up making the transition to using a communication device extremely difficult and frustrating [22].

It is often the case that individuals with progressing diseases have greater difficulty using these communication devices by the time they need them because of the lack of intuitive design and high learning curve [22]. Many individuals also report being steered away from trying to learn the devices early because it is disheartening for them to imagine that they may need rely on them in the future. Due to the high costs and difficult learning curve associated with eye-tracking AT devices, we propose an eye tracking device that allows users to connect to their existing iOS based tablet or smartphone device via Bluetooth to provide communication, entertainment, and overall improvement of life with quick and easy access to the modern digital world.

B. Previous Work

A previous implementation of our EOG device was intended to control a desktop mouse pointer via eye gazes [23]. To use eye gazes to control a mouse, a calibration was used to establish a baseline reading when looking straight ahead. When the user looked a specific direction, the change in voltage compared to the baseline would be calculated and then compared to tiered threshold values. If the change in voltage passed the threshold value then the mouse would move in the direction and speed that pertains to that threshold. The desired mouse movement would be continuous by either moving up and down or left and right while eye blinks were used for mouse clicks. Another option was given where a mouse can click using a dwell timer where the mouse would have to be still for a set amount of time. This design was implemented with the intention of the user having full control of the desktop and not be limited to specific application. After testing on healthy and unhealthy patients, issues became apparent on. The accuracy of the device, learning curve, time to complete tasks, and baseline drift all attributed to a system where much could be improved. These effects were amplified due to having the cursor move at a slow speed in an attempt to augment accuracy. The time to complete a task, which was to

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move to specific locations on the screen and type a message, would take a significant amount of time [23].

With these issues being apparent, the next step taken was to use the device as a switch controller detecting eye movements based on saccades. An eye saccade is the rapid eye movement between point fixations. Using the device as a switch controller compliments eye saccades because switch control is discrete and eye saccades can be categorized into four discrete movements: up, down, left, and right. This approach should successfully eliminate all the previous issues related to the gaze control. Due to nature of discrete switch control, the magnitude of the signal will not affect the intensity with which the cursor moves. Instead it outputs a single directional signal per each input it receives.

II. Review of Existing Technology

A. Alternate Eye Tracking Solutions

One of the most attractive and commonly used implementations for hands free communication devices is that of image processing based systems. These "gaze" based systems utilize light reflections back from the eye to determine the position of the pupil in reference to center. This can be implemented through the use of infrared (IR) transmitter/receiver pairs, or stereo high resolution camera systems combined with IR filters and sophisticated processing [20]. In the case where IR is utilized, an IR transmitter aims into the user's eye where its signal is reflected back to many individual sensors dispersed strategically around the frame surrounding the eye. These sensors measure the intensity of the reflected IR signal, which allows the system to determine the position of the eye. For instance, a sensor on the right side of the frame will receive a weaker reflected signal when the user is looking left and vice versa. It is also possible to detect blinks using this sensor system when minimal light reflection is detected in each sensor (i.e. eyelid closed or open) [21].

B. Electrooculography (EOG)

An alternative method to capturing eye movement signals to imaging based systems is that of electrooculography (EOG). EOG signal magnitude is based upon the potential (voltage) difference between the front and back of the eye. The voltage measured between the cornea and retina is between .5mV ~ 4mV. Although this voltage does not directly change due to eye movement, we can detect the intensity of movement by placing measuring electrodes near the subject's cornea/retinal potentials. When the eyes change direction, the angle of the potential changes and its field generates a current through the electrodes which is can be amplified and processed.

Once the signals have been captured and processed, there are many assistive technology applications for using the eyes as a controller. For instance, one study utilized EOG to allow an individual with limited mobility to control a wheelchair. EOG signals are classified to determine which direction the user wishes to move, followed by a blink to indicate their decision to move. This design allows the user to survey their environment and decide when the wheelchair starts moving. The wheelchair is able to move forward, left, right and stop by using EOG measurements to detect if the user is looking up to go forward, left and right are by looking in the respected direction, while stopping is controlled by looking down [10].

EOG signals can also be used to assist a person in controlling a virtual keyboard. Another research group used five silver chloride electrodes to capture the signals and pass the data through a combination of low pass and high pass filters to remove noise and DC drift. Once processed they use the data to determine which direction the person wishes to move the selector in the direction they look, while using blinks to select. They report tested subjects were able to type "water" in about 24.5 seconds averaged with an accuracy of 95% [11].

III. Methodology

A. System Design

The EOG controller system consists of a set of four silver chloride (AgCl) electrodes coated with a conducting gel and an adhesive housing placed at key locations around the user's eyes (Fig 2)[23]. These electrodes are able to detect potential signals around the eyes in the microvolt range. An additional reference electrode is placed on the users arm as a skin potential reference point for the amplifier.

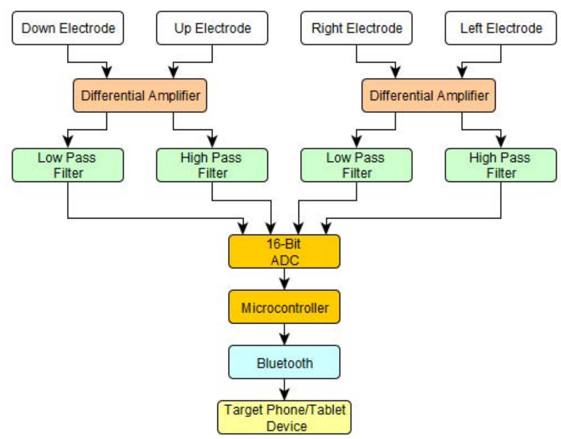

Fig.1. System Block Diagram

The wires coming from the electrodes around the eyes are fed into two Texas Instruments INA118 Precision, Low Power Instrumentation Amplifier in pairs of left/right and top/bottom at the V_{IN+} and V_{IN-} respectively. The INA118 amplifiers are set to a gain by selecting R_G that does not saturate our signal [23]. Once the signal has been amplified, it passes through two analog filters. The high pass filter consists of resistor and capacitor in series and the low pass filter consists of a resistor and capacitor in parallel to eliminate noise in the signal. From the previous implementation in [23] we have made some significant changes to the controller. The filtered signal is measured by an Adafruit ADS1115 16-Bit ADC and finally this measurement is sent to the Arduino microcontroller for processing. Once processing is complete, the device sends signals over Bluetooth connection to the iOS device through and Adafruit Bluefruit LE transmitter. The system block diagram can be seen in Figure 1.

Fig. 2. Electrode Placement [23]

B. Signal Capture/Processing

The signals captured by the Arduino microcontroller are further filtered by software in the processing stage through two filtering algorithms. The first filtering function performed determines the delta between past and present input values in order to detect significant changes in magnitude. In order to smooth the delta signal, a two tap digital filter is applied by weighting the newest raw ADC value at 0.10 weight and the latest known good value weighted at 0.90. This allows us to stay within 10% of our last known good sampled for any single new sample taken. Signal deltas greater than user specific thresholds are used to determine when eye movement events such as looking left, right, up, or down have occurred. In use, the system periodically samples the ADC and updates the values returned from each filter. These samples are checked against the threshold value each update, in order to see if the user has moved from a neutral state to looking in any direction. Once the sampled values are detected to be over the threshold for consecutive samples, it is will then stay in the "peak detect" state until the value comes down below a lower threshold for consecutive samples. Once this happens, it can be determined that the peak has passed and an output signal is sent to the Bluetooth module for transmission. To determine if the signal delta is significant enough to warrant an event trigger, a calibration process is performed before use by measuring the signal magnitude from the user looking in each direction. The calibration measures the peak ADC values when the user looks in specified directions and determines a threshold based on a percentage of those values. Control can be manually enabled or disabled during calibration to eliminate unwanted inputs during the calibration process.

C. Switch Control Accessibility Interaction

The EOG controller relies on the iOS accessibility software to gain access to action functions for the tablet or phone. For the interaction to take place between the controller and the iOS system, it is necessary to enable the "Switch Control" feature on the phone or tablet. This feature allows the user to connect their existing Bluetooth or Axillary switching device and map its buttons to different accessibility "actions" in order to control the device from the switch. To grant the user the most basic control, the actions are mapped to different "switch" inputs as follows. User action "Look Left" is mapped to the action "Move to Previous Item". User action "Look Right" is mapped to action "Move to Next Item," and user action "Look Down" is mapped to the action "Tap." The user action "Look Up" was originally mapped to begin auto-scanning of items, but seemed to cause confusion for users trying the device for the first time. For this reason, it was removed during testing. Using the move to next/previous item can be used to navigate through selection items, and the tap action can be used to make a selection. The iOS accessibility interaction works by grouping nearby selections on the screen into sub groups which can be selected in order to select an item or additional sub group (see Fig.3). Once the last sub group has been selected, a character can be chosen for output.

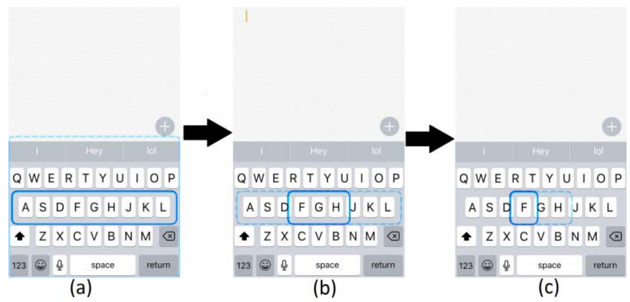

Fig. 3. User interface: (a) Shows selection of a parent group for the row containing three sub groups. (b) Shows selection of subgroup for the row containing an additional three sub selections. (c) Shows selection of the final subgroup allowing individual selection of the character.

D. Testing

The system is tested using a three step process requiring the user to type a phrase using the iOS keyboard, navigation to a web site, and a using an assistive technology speaking app to speak phrases. The set up used for testing trials can be seen in Fig. 4. The first test (Test #1) required each test subject to type the phrase "Hello World" using their eyes as the only inputs. The test was performed in the "Notes" app on the iOS 10.3.1 operating system using the switch control accessibility feature. Any input error was noted and automatically erased by the test monitor to ensure consistent completion standards. Errors were then noted for accuracy calculations. In this case, the user would be required to retry entering the current character in the test. Auto correct was not allowed and each individual character was required to be separately entered. The time from start to completion was recorded as well as the number of erroneous inputs corrected. In the second test (Test #2), users are required to navigate from the home screen to the Safari web browser app and select the input field for web sites. The keyboard appears allowing the user to enter a web site specified by the test monitor. The use of auto-correct is allowed for this test after selecting the first character.

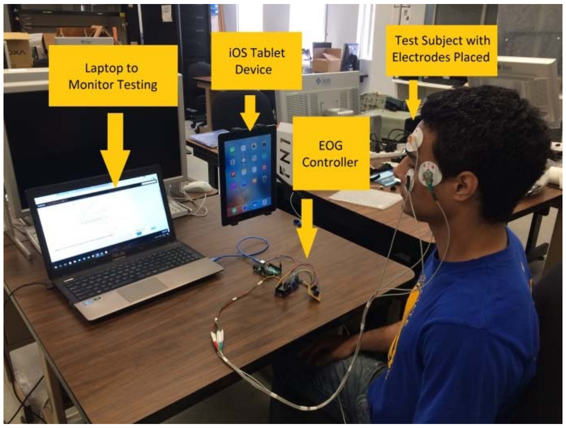

Fig. 4. Testing setup

Every test subject performed the test in the same manner by selecting the auto-finished version after selecting only one character. The time is recorded from when the subject begins the test to when they have finished selecting the web site. In the third test (Test #3), users are required to select a series of pictures from the Ablenet Sounding Board app. The user is instructed by the monitor to select a category and a button for three different phrases in the app. Time is recorded from the start instruction to the final phrase being selected.

IV. Data

TABLE I
EXPERIMENTAL DATA HEALTHY

Subj.	Test #1 (mm:ss)	Test #1 Accuracy	Test #2 (mm:ss)	Test #3 (mm:ss)
1	3:41	78%	0:58	1:18
2	3:18	84%	1:33	1:21
3*	7:10	--	3:10	2:51
4	3:22	84%	1:16	1:48
5	2:57	100%	1:30	1:38
6	2:31	91%	1:16	1:43
7	3:45	91%	2:43	1:10
8	4:25	68%	1:17	1:03
9	3:00	100%	2:41	1:35
10	3:14	100%	4:40	1:41
11	1:48	100%	1:37	1:14
12	3:41	84%	2:39	-:--
13	5:04	91%	2:58	2:05

Table I: Times of each subject for completion of tests #1-3. *We note that Subject #3 experienced technical difficulties and considered this data as an outlier for averaging.

In the first table, we obtain the results of thirteen healthy test subjects who performed each of the tests. From the data obtained, we noticed an average time of 3:23 to type 11 characters giving an output of around 3.25 characters per minute (cpm) seen in Test #1. An average accuracy of 89% was determined by considering the number of mistakes as an addition to the denominator of the averaging. For instance, a user who made zero mistakes would complete 11/11 characters correctly. If an individual makes n mistakes, their accuracy is calculated as 11/(11+n) correctly chosen characters out of the total characters chosen. Test #2 resulted in an average completion time of 1:18, demonstrating the feasibility of navigating to a web site quickly via the EOG controller. Finally, Test #3 was completed on an average time of 1:34 demonstrating the functionality of using AT communication apps with the controller. For ALS patients tested, the results were slower for the typing test with an average time of 6:10 at 1.78 cpm. The results improved as the patients became more comfortable with the device and were able to perform quickly in the accessibility tests that followed.

TABLE II
EXPERIMENTAL DATA PALS

PALS Subj.	Test #1 (mm:ss)	Test #1 Accuracy	Test #2 (mm:ss)	Test #3 (mm:ss)
1	6:07	78%	1:39	2:11
2	5:22	84%	1:42	2:27
3	7:01	78%	1:32	2:18

Table II: Times of each subject for completion of PALS tests #1-3.

TABLE III
IMPROVEMENT OVER REPEAT TESTS

Trial	Time Taken (seconds)	Mistakes	Accuracy
1	3:41	3	78%
2	2:53	2	84%
3	1:58	1	91%
4	1:49	1	91%
5	1:54	1	91

Table III: Improvement of individual subject over 5 repeated tests.

The set of data (in Table III) is based off the data from subject #1 in Table I who repeated the test over five trials. This was a test in order to determine if performance improved past the initial use. It was determined that improvement is indeed observed beyond the first use as both accuracy and the total time improved over subsequent trials until the final trial. On the fifth test no improvement was seen, showing that the device has a slight learning curve. From the first test to the fourth test where the highest improvement is seen, the time to complete the test was reduced by 1:52 nearly half the time of the original trial. This also resulted in a 5.59 cpm improved from 2.98cpm of subject #1's original test (an 87% improvement).

V. Conclusion

The EOG device was tested on healthy subjects and PALS subjects to determine its functionality as a hands free, eye controlled human computer interface device for communication. The results from the testing determined and average of 89% accuracy and 3.25 cpm with a peak of 6.11cpm. It was also determined that a learning curve of only a few trials can result in an 87% improvement for individuals using the device for the first time. PALS subjects also were able to complete the tests and seemed to improve over just a few minutes of usage when moving on to tests 2 and 3. It was also determined through the speedy and successful completion of Test #2 and Test #3 by every subject, the functionality of this device is successful in providing not only communication access, but to that of additional AT applications developed for tablet or phone use. It could be significantly improved with a custom tailored keyboard application for the phone or tablet that is optimized for minimal number of user inputs to select a letter. This would significantly reduce the number of actions required to select a letter (which can require as many as 7-8 inputs to select a single character). By implementing this keyboard, we could drastically improve the achievable characters per minute. There are also additional ways to use the signals such as "holding" a left or right glance that could be classified as an additional signal to use in the controller. By utilizing as many input classifications as possible, the user is provided more control over their target device.

VI. Future Work

In addition to these healthy subjects, tests were performed on 3 PALS patients. At the time of this paper, testing is still ongoing for PALS patients. We hope to demonstrate similar success on these subjects as we did with the healthy subjects. It has also been considered to further improve signal classification through machine learning and to implement a "universal" calibration technique allowing a user to calibrate their device quickly. Finally, the directional classifications could be expanded to include diagonal and longer duration looks to provide additional control signals. We hope to implement this in future applications.

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