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Eyeglasses Based Electrooculography Human-wheelchair Interface

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***Abstract*—This paper describes an electrooculography (EOG) based human-wheelchair interface for wheelchair users. A pair of electrodes which measures the eye-gaze direction of users is desired as wheelchair manipulation commands. In addition to EOG based wheelchair manipulations, this paper also introduces ultrasonic arrays for detecting distances to actively avoid collisions. In order to simplify setup procedures of using this system, two electrodes are mounted on two side arms of eyeglasses respectively that are possible to have tight contacts with skins around eyes. Therefore, wheelchair movements are capable of following eye-gaze directions. On the other hand, an analog biopotential signal amplifier and a laptop computer are used to develop the proposed EOG based human-wheelchair interface controller. Finally, experimental results demonstrate the operations of EOG based human-wheelchair interface.**

***Keywords*—Human-wheelchair interface, eye-gaze tracking, electrooculography, reactive navigations.**

I. INTRODUCTION

Wheelchairs are important mobile aids for elders and disabled patients. Several reports pointed out the tendency of increasing populations of elders and disabled patients [1-2]. In recent years, several literatures proposed [2-4] intelligent wheelchairs by introducing novel sensor fusion, autonomous navigation, and robotics technologies to conventional powered wheelchairs to improve the quality of life for wheelchair users. At the same time, these technologies actually extended the user spectrum from lower extremities disabled patients to high degree spinal cord injured and multiple sclerosis patients [4].

In general, wheelchairs may be manipulated via either exerting pushrims or operating joysticks to perform manual, powered, or power assisted [2] driving. Such wheelchair manipulations need at least one arm to hold the joystick or pushrims for driving. In this manner, wheelchair users may not easily hold their pets or objects when manipulating the wheelchair. Scenario photos of pushrim, joystick, and hand-free wheelchair manipulations with holding books are shown in Fig. 1. Apparently, it is not convenient to hold these books when using the pushrim and joystick based wheelchairs. It is also expected that the hand-free wheelchair manipulation is more convenient for disabled patients in their daily life.

Figure 1. Scenario photos of (a) hand-rim, (b) joystick, and (c) hand-free wheelchair manipulations when holding books.

Several works reports hand-free operations of wheelchairs by using electroencephalography (EEG) based brain computer interface (BCI) [5], electrooculography based eye-gaze tracking [1], camera based eye-gaze tracking [4], and voice control [6] technologies. Voice controlled wheelchairs [6] have to overcome noisy environments as well as to implement complicated computational programs. More specially, limited voice commands result in restricted wheelchair motion behaviors; consequently, it is very difficult to control the wheelchair in a precise manner. At the same time, users have to speak out the driving commands when the control situation changes. It is confusing to ask users to speak out the commands frequently. Therefore, the voice control based wheelchair manipulation mechanism is not feasible for practical crowded and irregular paths.

In general, the electroencephalography based BCI approach [5] is not easy to implement due to weak signal strength, complicated signal distributions, heavy computation efforts, possible wrong recognitions, and inconvenient surface electrode installations. On the other hand, eye-gaze based wheelchair manipulations are much robust and easier then the BCI approaches. Several researches proposed the computer vision solution to detect the eye-gaze directions. Nevertheless, the head mounted camera [4] based eye-gaze solution represented larger camera sizes, unstructured luminance condition, and complicated image recognition efforts. Consequently, the electrooculography based eye-gaze based approach is selected in this paper to implement the human-wheelchair interface.

EOG is a popular solution for detecting the eye-gaze directions in several assistive technologies [1]. Although the EOG signal is easy to collect, installations of surface electrodes are still challenging to most EOG based wheelchair developers. Apparent surface electrodes as well as complicated setup procedures reduce the willingness of using the EOG based wheelchair control system. As a result, most of EOG based wheelchairs were desired for users who are unable to use their upper extremities to control wheelchairs.

To simplify surface electrode setup procedure and to eliminate surface electrode shape effects on the face, the eyeglasses based configuration is used in this paper. Two surface electrodes are mounted on two side arms of the eyeglasses respectively, and they are desired possibly to have tight contacts with skins around eyes. These electrodes are used to detect the eye-gaze direction of users, and the eye-gaze direction is further generated as wheelchair driving commands using wheelchair steering coefficient function. In addition, an analog signal amplifier with multiple stages filters is implemented to extract EOG signals as well as eliminate the noise occurred from surface electrode slipping.

A laptop computer is used to capture the eye-gaze direction and to generate wheelchair driving commands in terms of the steering coefficient function. In summary, the proposed EOG based human-wheelchair interface has the advantages of:

1. Compact eyeglasses type surface electrode module design increases the willingness of using the EOG based controller.
2. Active obstacle collision avoidance improves the safety of using EOG based human-wheelchair interface controller.
3. Modular based design is easy for integration and maintenance.

Finally, this paper is organized as follows: section II presents the system architecture; section III describes the EOG based human-wheelchair interface; section IV illustrates the system integrations and experiments; and finally conclusions and future works are concluded in section V.

II. System Architecture

This paper presents an electrooculography based wheelchair control systems with active collision avoidance. To achieve the desired functions, the system architecture is designed as three major parts: EOG signal collection and wheelchair command generation, perception based active collision avoidance, and wheelchair mechatronics system. Fig. 2 shows the proposed system architecture, and these major parts are illustrated as follows.

1. EOG signal collection and wheelchair command generation: this part is responsible of collecting electrooculography of users. Due to small and noisy biopotential signals, the analog EOG signal is amplified and filtered to extract the horizontal eye-gaze direction. In addition, the calibration procedure is done to eliminate individual variations. In addition, the average speed of the wheelchair is captured from the wheelchair control panel. Consequently, the horizontal eye-gaze direction and the average wheelchair speed are aggregated to generate wheelchair driving commands. Detailed procedures are described in section III.

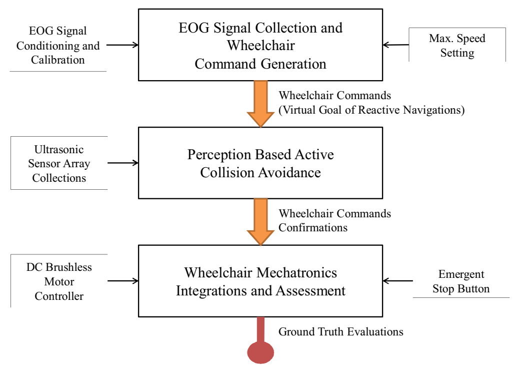

Figure 2. Electrooculography based wheelchair control system architrecture.

2. Perception based active collision avoidance: the obstacle distances are detected in terms of an ultrasonic sensor array with 9 sensors. Any sensor in the moving direction detecting the distance of obstacle smaller than the safety range will stop the operation of wheelchair.
3. Wheelchair system mechatronics integrations: this part generates the angular velocities of two driving wheels using the steering based inverse kinematics. The angular velocities are further used to control two DC brushless motors in terms of DC voltages. In addition, the start, stop, emergent stop buttons are used for practical applications. Finally, the results are evaluated based on ground truth trajectories.

III. EOG Based Human-Wheelchair Interface

In this paper, the electroencephalography (EOG) signal is used to generate the driving command of wheelchairs. The EOG is a very typical approach to measure eye movements, and it is measured based on the steady corneal-retinal potential. In general, this steady dipole is used to measure eye position in terms of placing surface electrodes to the positions around volunteer's eyes.

If the steady dipole is symmetrically placed between the left and right electrodes and the eye-gaze is straight ahead, the EOG output will be zero. When the eye-gaze shits to the left or right, the equilibrium will not be established. The EOG output depends on the relative positions of the cornea and electrodes. For example, the EOG output will be more positive when the eye-gaze shifts to the right [7].

In general, there is an approximately linear relationship between horizontal angle of EOG output and eye-gaze directions within the gaze angles of +/-30^0. The EOG output ranges from 50 to 3500 μV within the frequency range below dc-100 Hz [1,7]. However, the EOG output value is still dependent on the individual properties of users, surface electrodes, and electrode placements. It is noted that the EOG may also be affected by the activations from the EEG of the brain as well as the electromyography (EMG) of facial expressions and head movements.

A. Eye-Gaze Direction Detection

In this paper, the horizontal eye-gaze direction is captured to form the basis of wheelchair driving direction commands. Three surface electrodes are used in this work. Two surface electrodes (A-B) are placed at the right and left of the outer canthi as a pair to measure the horizontal eye-gaze direction. Another surface electrode (C) is placed at the ear lobe as a reference electrode. Note that Ag/AgCl surface electrodes are used in this paper. The placement of surface electrodes is shown at the left-hand-side of Fig. 3.

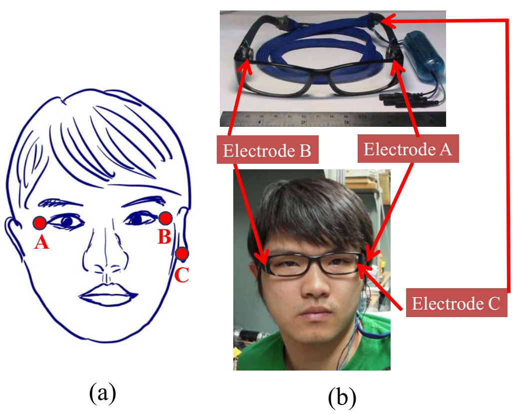

Figure 3. (a) Placement of surface electrodes; and (b) photos of EOG eyeglasses module and volunteer with this module.

Based on the geometry relationships of three electrodes, this paper uses a pair of eyeglasses to setup these surface electrodes as a compact modular. The proposed EOG eyeglasses module is convenient for setup, and it performs more compact and pleasing outlook than the cutting strips of adhesive tape holding solution. In order to maintain the tightness of holding, the proposed EOG eyeglasses module is custom made to perform tight contact. At the same time, conductive paste (with type Ten20™ [8]) is used to eliminate motion artifacts. The photo of EOG eyeglasses module is shown at the right-hand-side of Fig. 3.

In general, the EOG signal is small and noisy due to inductions of EEG, EMG signals, motion artifacts, and small EOG signals. Therefore, this paper develops an EOG signal conditioning board to extract eye-gazing corresponded EOG signals. The volume conductor of EOG signal is collected via a pair of eye-gaze surface electrodes and a reference electrode. An instrument amplifier (with type: AD 620 [9]) is used to extract biopotentials from electrodes. This signal is initially amplified with 110 times in amplitudes; then a cascade configuration with two second-order low-pass filters (50 Hz) and a high pass filter (0.5 Hz) is desired to reject noise as well as to eliminate DC offset effects. Finally, the signal is further amplified with 330 times in amplitudes.

B. Closed Loop Wheelchair Direction Control

Practically, the wheelchair is controlled in terms of the eye-gaze direction (θ). The eye-gaze direction represents a goal direction that the wheelchair would like to approach. However, the goal direction is not necessarily consistent with the wheelchair direction. In addition to the eye-gaze direction, the wheelchair also provides a wheelchair linear velocity setting switch to adjust the average wheelchair speed. Three speed categories of 10 cm/s, 30 cm/s and 50 cm/s are desired. Consequently, the eye-gaze direction and the speed category form the wheelchair motion commands.

Wheelchair directions are the most important characteristics of controlling a wheelchair. Appropriate direction controls may result in approaching destinations and avoiding collisions. A typical closed loop wheelchair direction control system of combining human's perception is shown in Fig. 4. This closed loop system is desired to maintain the direction consistence between the user and wheelchair. The volunteer may compare the difference (error) between the desired and real direction (from visual feedback).

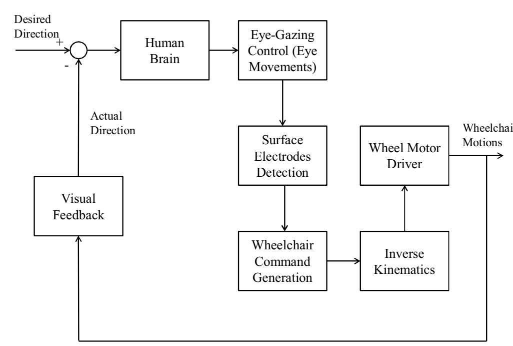

Figure 4. Closed loop wheelchair direction control system.

Differences between desired and actual direction will result in the eye-gaze control via human's brain decision system. The resulted eye-gaze direction can be further detected in terms of the eyeglasses based EOG module. The eye-gaze direction is used to generate the wheelchair steering commands. Consequently, the wheel motor control commands are generated from the steering commands.

In this system, the maximum controllable angle of the eye-gaze direction (θ_{max}) is 30^o. To reduce the sensitivity of wheelchair steering commands within central gaze directions, the wheelchair steering coefficient function is desired as a nonlinear function of horizontal eye-gaze direction, as shown in (1).

$$T_S = \begin{cases} e^{m\theta} - 1 \ (\theta \leq 0.5\,\theta_{\max}) \\ a\theta^2 + b\theta \ (\theta > 0.5\,\theta_{\max}) \\ a\theta_{\max}{}^2 + b\theta_{\max} \ (\theta \geq \theta_{\max}) \end{cases} \quad (1)$$

where T_s is the output of the wheelchair steering function; θ is the absolute value of eye-gaze direction; a, b and m are constants for various control purposes.

$$a = (2Ts_{\max} - 4Ts_{set}) / \theta_{\max}{}^2 \quad (2)$$

$$b = (4Ts_{set} - Ts_{\max}) / \theta_{\max} \quad (3)$$

$$m = 2\ell n(Ts_{set} + 1) / \theta_{\max} \quad (4)$$

where Ts_{max} is the maximum steering coefficient of the wheelchair, and it is desired as less than unity; and Ts_{set} is an adjustable parameter, and it is determined in terms of the user driving criteria.

As a result, the steering command (Tc) is generated with direction dependent signed Ts. Fig. 5 shows the curve of this hybrid velocity function. Consequently, such a manipulation approach extends conventional wheelchair commands such as "Forward", "Back", "Right", "Left", and "Stop" [1]. Furthermore, the steering coefficient can be further used to generate the wheel angular velocity commands (ω_l and ω_r) using (5) and (6), and this part indicates the inverse kinematics of Fig. 4. Noted that ω_{avg} is the control average velocity, and it is a rated value from desired average speed (10 cm/s; 20 cm/s; and 30 cm/s). The rated value is inferred proportionally to the nearest obstacle distance.

$$\omega_l = \omega_{avg} \times (1 + Tc) \quad (5)$$

$$\omega_r = \omega_{avg} \times (1 - Tc) \quad (6)$$

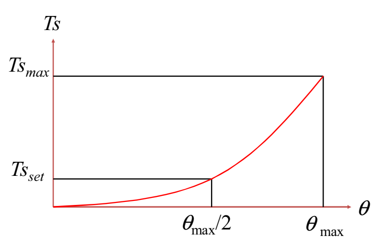

Figure 5. Curve of a hybrid velocity function.

IV. SYSTEM INTEGRATIONS AND EXPERIMENTS

To validate the proposed EOG based human-wheelchair interface, this paper uses a conventional wheelchair as a test platform. Especially, two DC brushless motors are attached on the wheels of manual wheelchair so that the wheelchair can be controlled via electronic signals. The control architecture of the proposed EOG based human-wheelchair interface is shown in Fig. 6.

In this architecture, a laptop computer is served as a supervisory controller. A program coded with Microsoft Visual C++ is used to capture the eye-gaze direction through the conditioned EOG signals (EOG board is shown in Fig. 7) and the average speed commands (ω_{avg}) via the serial communication (RS 232). The EOG signal is represented as an eye-gaze parameter (θ). Two tests of users' EOG signals are evaluated. The relationships between the eye-gaze direction and the EOG signal voltage are shown in Fig. 8. This chart is further used to calibrate and calculate the eye-gaze direction. By using the formulas of (1) to (6), the wheel velocity commands can be properly generated.

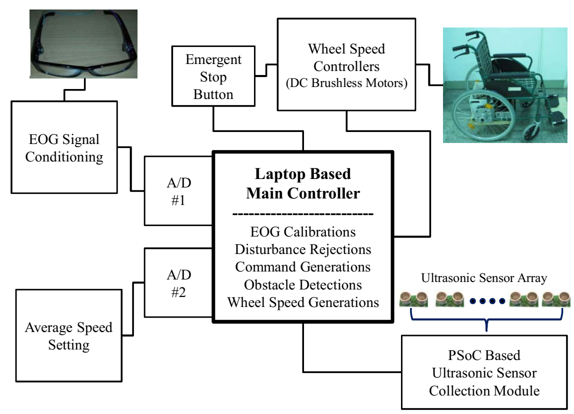

Figure 6. System architecture of EOG based human-wheelchair interface.

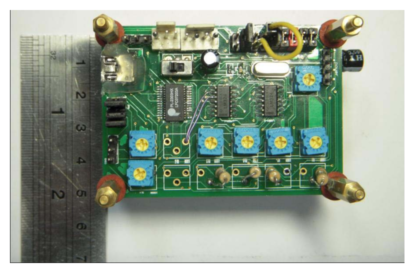

Figure 7. In-lab design EOG signal conditioning board.

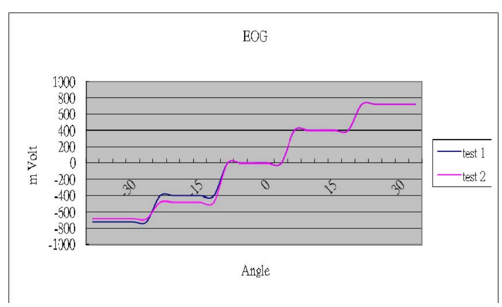

Figure 8. Chart of eye-gaze directions and EOG signal voltages.

Especially, 9 ultrasonic sensors are arranged around the wheelchair to detect obstacles. Five ultrasonic sensors are in the front of wheelchair; two ultrasonic sensors are in the right-hand-side of wheelchair; two ultrasonic sensors are in the left-hand-side of wheelchair, as shown in Fig. 9. The stop distance is adjustable, and it is set as 15 cm in this study. Noted that the proposed EOG controlled wheelchair does not support backward driving, and the detection of backward obstacles is not constructed. Therefore, these ultrasonic sensors play important roles for avoiding collisions.

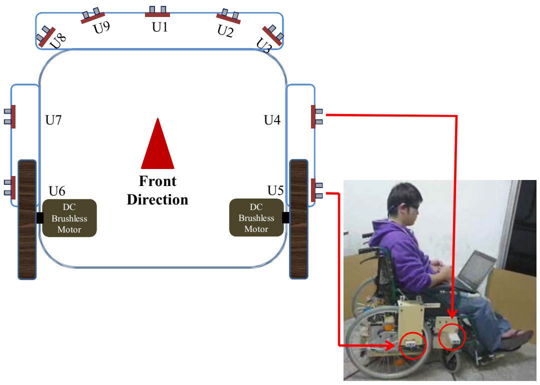

Figure 9. Sensor allocation for detecting surround obstacles.

To verify the capability of the proposed EOG based human-wheelchair interface, a small and crowded test environment is evaluated. This test environment is a 4.5m × 3.0m indoor area. A pillar is appeared in the right-upper corner to increase the driving challenge. The test photo is shown in the left-hand-side of Fig. 10. At the same time, the photo of real test for this test environment is shown in the right-hand-side of Fig. 10. Especially, the real path of the wheelchair is also drawn using a marker pen to evaluate the ground truth performance.

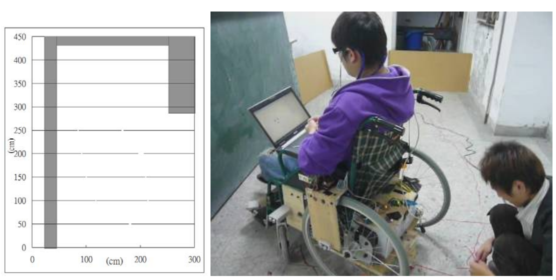

Figure 10. Test area and photo of test.

The EOG based human-wheelchair interface is a cooperation system between wheelchair controller and the human's brain. Therefore, a junior volunteer may not perform perfect driving performance. Fig 11 shows the training paths of a junior volunteer. From the paths of training, the collision avoidance activated frequently. A well trained ground truth evaluation is shown in Fig. 12. In this figure, two paths are recorded from the front and rear pens to evaluate the approximate moving path of a "wheelchair block". In this case, a trained user (around 10 times training courses) is demonstrated. Results of well-trained volunteer represented a smoother path and collision free driving performance.

In addition to the ground truth path recording, steering commands and generated wheel velocities via inverse kinematics are also collected. Fig. 13 shows the partial results of steering commands with 100 times (TC*100), rated average speed (Avg), left wheel speed (WL) and right wheel speed (WR).

Figure 11. Ground truth evaluations of junior volunteer.

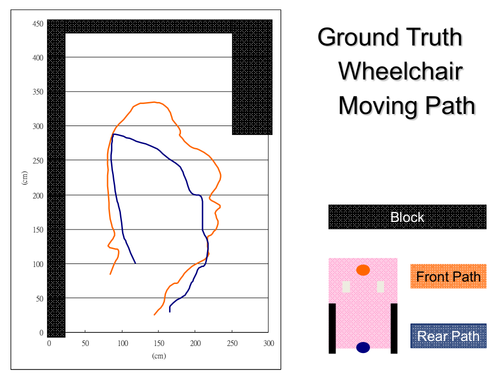

Figure 12. Ground truth evalautaions of well-trained volunteer.

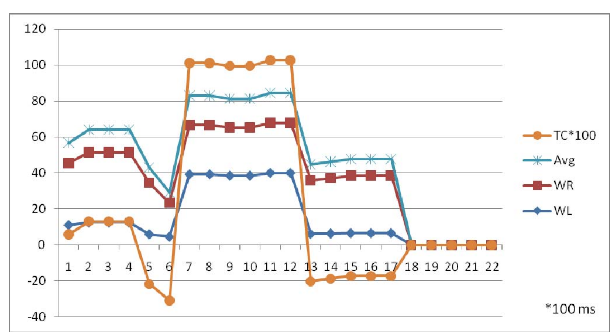

Figure 13. Partial results of steering commands with 100 times (TC*100), rated average speed (Avg), left wheel speed (WL) and right wheel speed (WR) of well-trained volunteer driving.

V. Conclusions and Future Works

This paper presents an EOG based human-wheelchair interface. In addition to the signal processing of the EOG signals, an eyeglasses based electrode devices is also developed to perform compact and nice look solutions. The eye-gaze is detected to generate the steering commands in terms of steering coefficient function. In addition, the ultrasonic sensor array is used to ensure the collision free of manipulations. The proposed EOG based human-wheelchair interface required a short training course to compensate the gap between the wheelchair controller and the user's brain. Especially, ground truth paths of front pen and rear pen are recorded for evaluations. Experiment results also demonstrate well-trained volunteers behave better performance than junior volunteers. In the future, the perception based obstacle avoidance and embedded controller will be developed to improve the driving performance and to reduce the cost of the EOG based human-wheelchair interface system, respectively.

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