Neurocognitive Assessment in Virtual Reality Through Behavioral Response Analysis

Hawkar Oagaz, Breawn Schoun, Manpreet Pooji, and Min-Hyung Choi

***Abstract*—The ability to detect and diagnose neurocognitive disorders at the earliest possible moment is key to a better prognosis for the patient. Two of the earliest indicators of potential neurocognitive problems are motor and visual dysfunction. Motor disorders and problems in visual cognition can be seen in many neurocognitive disorders, resulting in abnormal physical reactions to visual stimuli. Analyzing physical behaviors when presented with such stimuli can provide insights into the visual perception and motor abilities of an individual, yet there is currently no unbiased, objective, general-purpose tool that analyzes attention and motor behavior to assess neurocognitive function. We propose a novel method of neurocognitive function assessment that tests the patient's cognition using virtual reality with eye tracking and motion analysis. By placing the patient in a controlled virtual environment and analyzing their movements, we can evoke certain physical responses from subjects for neurocognitive assessment. We have developed a prototype system that places the subject in a virtual baseball field and captures their full body motion as they try to catch baseballs. This scenario tests the subject's ability to determine the landing time and position of the ball, as well as the test subject's balance, motor skills, attention, and memory. Preliminary tests with 20 healthy normal individuals demonstrate the ability of this tool to assess the test subject's balance, memory, attention, and reaction to visual stimuli. This platform has a twofold contribution: it is used to assess several neurocognitive constructs that affect visual and motor capability neutrally and objectively based on controlled stimuli, and it enables objective comparison between different neurocognitive disorders research in this field.**

***Index Terms*—Behavioral analysis, motor skills, neurocognitive assessment, virtual reality, visual cognition.**

I. Introduction

By 2017, 6.08 million people in the United States were diagnosed with Alzheimer's Disease (AD) or mild cognitive impairment, and this number is expected to increase to 15 million by 2060 [1]. Traumatic Brain Injury (TBI) is also prevalent in the United States, with over 2 million people affected each year [2]. Symptoms of patients with neurocognitive disorders can vary significantly depending on the type and location of damage to the brain. Neurocognitive problems such as mild Traumatic Brain Injury (mTBI) are often signified with headaches and fatigue, whereas patients with Parkinson's Disease (PD) show a progressive loss of motor control, and sufferers of AD lose their memory over time. Yet for all of the differences in how these disorders present themselves, they often share similar attributes, and can therefore be diagnosed and assessed using the same tests [3].

Patients with TBI in particular often have difficulty tracking movement and locating visual stimuli [4], [5]. Ideomotor and limb apraxia, motor disorders resulting from brain damage, are common to AD, PD, TBI and Multiple Sclerosis (MS), and affect the individual's ability to perform motor functions [6], [7]–[10]. Slow Information Processing Speed (IPS) is also common in MS and AD, often resulting in performance reduction of executive functions [11], [12]. In dementia-related disorders such as AD and PD, the patient's performance in tests of eye-hand coordination and motor tasks are often correlated with results of neurocognitive tests identifying or assessing dementia progression [8], [13], [14]. Additionally, studies have shown that TBI severity is also correlated with motor skills [15], [16]. Memory impairment is also commonly seen in individuals with neurocognitive disorders [17]–[19]. These findings suggest that motor and visual dysfunction can be a strong indicator of neurocognitive function, and assessment of this dysfunction in

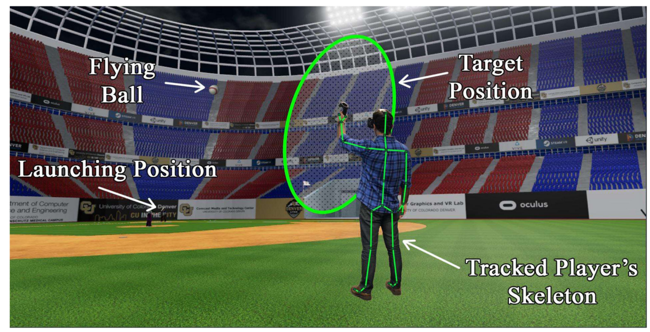

Fig. 1. Virtual baseball environment for assessment of visual cognition and behavioral responses. The system analyzes the participants' movement and responses to visual and spatial stimuli in order to assess cognitive function.

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H. Oagaz, B. Schoun, and M.-H. Choi are with the Department of Computer Science and Comcast Media and Technology Center, University of Colorado Denver, Denver, CO 80204 USA (e-mail: hawkar.oagaz@ucdenver.edu; breawn.schoun@ucdenver.edu; min.choi@ucdenver.edu).

M. Pooji is with the Department of 3D Graphics and Animation, University of Colorado Denver, Denver, CO 80204 USA (e-mail: manpreet.pooji@ucdenver.edu;).

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conjunction with memory testing could aid in the identification and assessment of neurocognitive impairment.

Neurocognitive assessment is generally performed using paper-and-pencil tests, where subjects are asked to create a drawing or respond to questions in writing [3], or computerized tests, which can provide additional objective data such as the response time of the patient [20]. These tests assess different aspects of neurocognition such as perception, executive function, and memory [3], [17]–[19], [21]. However, these tests tend to be unrelated to everyday life, and rarely take into account the behavior of the person [22]. Writing tests have the additional problem of educational and language biases, which may result in patient misclassification [17].

Neurocognitive testing can be vastly improved through the use of modern Virtual Reality (VR) and motion tracking tools. VR has previously been used to prevent, assess and treat neurocognitive ailments [21], [23]–[29], and provides a higher degree of ecological validity than other approaches [26]. Neurocognitive testing in a virtual environment limits external influences that could distract the patient, provides complete control over stimuli, and can be adapted for a variety of cases with little cost. Additionally, a VR application can provide immediate feedback to the test subject, increasing learning speed [30]. The addition of motion tracking provides accurate measures of the bodily behavior of the individual. Though VR has been previously been used to assess specific neurocognitive functions [21], [25], [26], to our knowledge, no general-purpose tool exists that assesses neurocognitive function based on physical behavior (including eye tracking) in VR.

We propose a novel VR and behavioral analysis framework that analyzes a person's physical behavior in order to assess neurocognitive function. The participant reacts to visual stimuli in a virtual environment while their eye gaze and bodily movements are tracked using an eye-tracking Head Mounted Display (HMD) [31] and a 3D depth camera [32], respectively. By placing the participant in a virtual environment, we can create a realistic and fully-controlled test setup, as well as alter the physical properties of the environment. Recording the participant's movements in response to stimuli allows for automated real-time analysis, and gives medical professionals the ability to view an instant replay of the subject's performance. Our system is intended to be a tool for comprehensive vision, memory and movement analysis that can be optimized for specific disease groups and clinical study objectives.

As a proof of concept, we have developed a prototype system set in a virtual baseball field where the participant attempts to catch baseballs. This task requires the participant to react quickly to catch the ball, accurately determine the time and location of the ball's arrival, and maintain balance and attention while performing the task. After the task, the patient can be asked questions about their experiences in the virtual environment in order to test their memory. By implementing a simple aspect of a well-known sport, we are able to reduce the educational effects between test subjects. Additionally, this gamified immersive environment provides a distraction from the awareness of being medically tested. We specifically chose the scenario of catching a flying ball because the test subject needs to continually process the scene in their visual cortex to determine the ball's position, trajectory, and estimated landing position. Through motion tracking, we can measure the physical behavior of the individual to gain insight into their attention, balance, motor skills, reaction time, memory and visuospatial understanding.

Our aim is to provide a common platform for neurocognitive -related behavioral analysis studies. This system can be used to assess a variety of neurocognitive attributes with little modification, and can be a common tool for neurocognitive research that allows for objective comparison between studies. This system is intended to be an easy-to-use behavioral assessment tool for psychologists and neurologists to use for their own research endeavors, and has been released under a GNU General Public License (GPL) license for use and modification by the wider neuroscience and VR community.

This work has two main contributions: first, it is a novel VR and behavioral analysis framework that helps to assess and analyze neurocognitive impairment related to visual and motor capability in affected populations. The novelty of this work comes from the ability of this system to record accurate, real-time, unbiased data regardless of educational level, spoken language, and dysphasia. Skeleton, eye gaze and reaction data can offer a valuable means to reach a neutral analysis of neurocognitive functions. Second, this tool offers a way to objectively compare neurocognitive disorders. Generating data for various disorders using a common platform provides another way to find relationships between them.

This paper is arranged in the following manner: Section II (Related Work) will discuss the current state of neurocognitive testing, uses for VR and motion tracking in medicine, and similar research to our own. Section III (System Design) will detail our proposed framework. Section IV (Experimental Setup) will explain our experiment, and Section V (Results) will detail the results of our experiment. Section VI (Discussion and Future Work) will discuss the future direction of this research and issues that need to be addressed. Finally, Section VII (Conclusion) will summarize our work.

II. Related Work

The standard method of assessing neurocognitive function is the paper-and-pencil test, where the subject responds to questions and prompts in writing [3]. The Mini Mental State Examination (MMSE) is one of the most widely used neurocognitive tests, and assesses neurocognitive function by requiring the patient to complete calculation and spelling tasks, as well as answer questions about orientation. Another commonly used neurocognitive test, the Clock Drawing Test (CDT), looks for problems in visuospatial abilities by asking patients to draw a clock face, and may be used in conjunction with a test called the Mini-Cog to test the patient's memory. There are also many other paper-and-pencil tests that target specific neurocognitive abilities [3], [17]–[19], [21]. Computerized versions of these tests are also frequently used, and have the advantage of being able to accurately record the subject's reaction time and accuracy [20].

While paper-and-pencil tests and their computerized counterparts work well in many cases, they also have significant problems and limitations. These methods tend to produce biased results when the patient suffers from sensory loss, dysphasia or illiteracy, or if the patient is not a native English speaker [3], [17], [21]. Translated versions of these tests exist, but they have not been validated [3]. Additionally, these tests do not take into account the physical behavior of the individual [22], and are therefore unable to assess other common attributes of neurocognitive disorders, such as difficulties in visual cognition [4], [17], [18], [33], [34], abnormal motor responses [6]–[10], slower IPS [11], [12], and difficulties maintaining balance [35], [36] and attention [12], [34].

A few tests exist that examine the motor abilities of a patient. One such test is the Test of Upper Limb Apraxia (TULIA), which rates a person's ability to perform movements and gestures [37]. As the name implies, this test only measures the upper body abilities of the individual. Additionally, a few studies have used 3D skeletal tracking for balance assessment [38], [39].

VR-based applications have been explored for use in mental and physical healthcare for more than two decades [40], [41]. VR has been shown to have applicability to Post-Traumatic Stress Disorder (PTSD) [23], [42], phobias [43], rehabilitation [44], [45], social anxiety [46], and pain management [47]. Several studies have used VR for neurocognitive assessment [21], [25], [26], [48]–[50], but these experiments only assessed specific neurocognitive functions instead of providing a general assessment. A few studies have explored using VR in combination with a balance screening device to detect and assess mild Traumatic Brain Injury (mTBI) [51], [52], and studies have also used VR for patient memory assessment [22], [49]. 3D skeletal tracking has found similar uses in healthcare applications, and has been used for physical rehabilitation [53], and to assess a participant's range of motion [54] and balance [38], [39].

III. System Design

Our system is designed to be a tool for researchers to explore the relationship between visual, motor and neurocognitive impairment. Research has shown that neurocognitive impairment is often accompanied by visual and motor deficits, and that visual and motor impairment are often early indicators of underlying neurocognitive problems. Various forms of visual agnosia, the inability to process visual information, are seen in TBI and AD, affecting the individual's visuospatial orientation and target prediction abilities [4], [17], [18], [33], [34]. Visuomotor skills, or the coordination between visual perception and motion, were found to be inhibited in AD, mTBI, and TBI, as evidenced by decreased accuracy and increased response and movement times during eye-hand coordination tests [4], [5], [13], [55].

To evaluate the participant's visual, motor and neurocognitive abilities, we designed a virtual baseball field where the participant is tasked with playing catch. We chose baseball because of the fast-paced nature of the game, as well as the visuospatial and temporal mental calculations required to determine the ball's landing time and position. The test subject plays the role of the outfielder and attempts to catch balls launched towards them at varying speeds and angles. Assessment of the participant's cognition is based on their ability to maintain balance and attention while performing the task, their ability to move their hand to the correct position to catch the ball in a timely manner, and their memory of the sequence of events during the test.

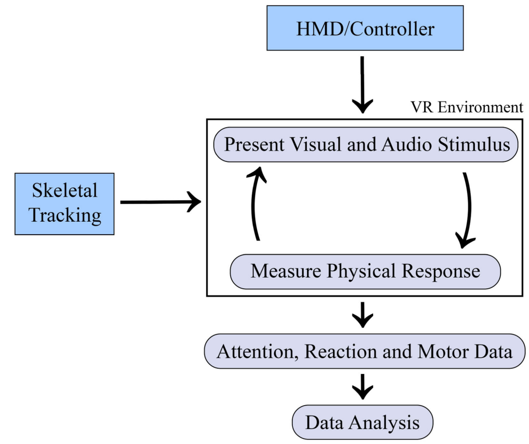

Fig. 2. The system's overall structure. The system takes Skeletal Tracking and HMD/Controller data as inputs and generates attention, reaction and motor data for analysis afterwards. The blue boxes represent hardware, while the grey rounded boxes represent processes in the system. Inside the VR environment, the processes are repeated for each trial.

A. System's Data Flow

The system is designed to record the participant's physical reactions to stimuli in order to assess their neurocognitive function. The system takes a 3D skeletal tracking device, HMD and controller as inputs, all of which are used to monitor the participant's physical movements. In the virtual environment, visual and audio stimuli are presented to the participant, and the participant's gaze and movements in reaction to the stimuli are recorded. The participant's attention is assessed by using the HMD to monitor the participant's gaze. This is done by either using an HMD that tracks the participant's eyes (such as a FOVE HMD), or by estimating gaze using the orientation of a generic HMD (such as the HTC Vive). The test subject holds a controller in one hand, which is used to measure the test subject's hand movements and reaction time, as well as provide haptic feedback to the participant when the virtual baseball hits their hand. The number of caught balls is recorded using data obtained from the controllers, while motor data is obtained using 3D skeletal tracking. This collection of information is the output of the system, and can be analyzed to find indications of any neurocognitive impairment in the test subject. The diagram in Fig. 2 illustrates the general structure of the system and its data flow.

B. Hardware and Software

The system was built using Unity 2017.4 on a conventional PC. In the initial phase of development, we used an HTC Vive HMD. However, in order to provide more accurate gaze information, we later incorporated eye tracking into the system using

Fig. 3. Virtual baseball field showing pitcher and batter animation, and overall virtual environment that replicates a baseball arena to give a realistic experience.

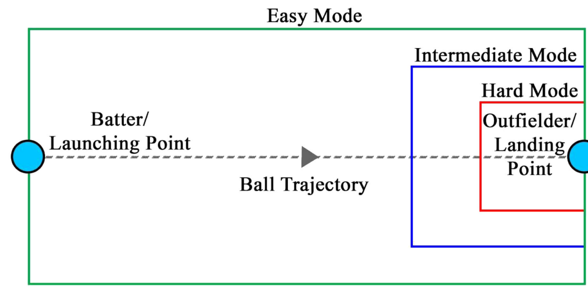

Fig. 4. Ball catching modes of easy, intermediate and hard. Each rectangle represents the time the participant needs to catch the ball.

a FOVE HMD. FOVE uses infrared sensors embedded in the HMD to track eye movements with low latency [31]. The test subject wears either the FOVE or HTC Vive HMD while holding an HTC Vive controller to track and control a virtual baseball glove. In order to better combine the FOVE HMD and HTC Vive controller, a Vive tracker was attached to the FOVE HMD so that the position and orientation of the FOVE were overwritten by the Vive tracker's position and orientation. We used a Microsoft Kinect for 3D skeleton tracking, and integrated the Microsoft Kinect SDK with Unity to perform movement analysis. The Kinect detects full-body skeletal information using an infrared RGBD sensor [32].

C. Virtual Environment

We designed the VR environment to provide an immersive experience to evoke responses that are true to the subject's condition. To that end, we simulate day and night lighting conditions in the environment and use 3D sound effects and realistic animations to further enhance the realism of the virtual experience. Fig. 3 shows our realistic virtual baseball field design.

D. Operator Variables

The system's control panel provides an easy way for the operator to modify the testing environment. The operator can modify several variables, such as the position of the participant in the field, the catching difficulty level, and the velocity, launching angle, and gravity of the ball. By modifying these variables, operators can use this system for their own research purposes or customize the experience for specific individuals.

There are three levels of catching difficulty: easy, intermediate, and hard. These modes allow the application to be adapted to the test subject's individual needs. In easy mode, the participant places their hand in the path of the ball in order to catch it, and is not required to pull the controller trigger to catch the ball. This mode can be used to evaluate the test subject's visuospatial understanding of the ball's trajectory, but does not assess the participant's reaction time. In easy mode, the participant has a significant chance of catching the ball if they are not suffering from neurocognitive impairment. Intermediate mode requires the participant to not only place their hand in the correct position to catch the ball, but also to pull the controller trigger at the exact time the ball reaches the glove. This mode is designed to evaluate the participant's visual perception of both location and time. The last catching mode is hard mode, which is similar to intermediate mode, but with a less forgiving time span to catch the ball. Fig. 4 illustrates the catching modes design with each rectangle representing the time the participant has to catch the ball from the launch position to the landing position.

E. Ball Flight

We used ballistic motion equations derived from Newton's laws of motion to simulate realistic ball flight [56]. The landing position of the ball depends on where the participant is standing; though the ball's end position should vary, it should arrive in the general vicinity of the player so that the ball can be reached without significant effort. Velocity or launching angle values can be selected by the operator, and unknown parameters are calculated by the system.

The position of the ball in flight is determined using the equation $x = x_0 + v_0 t \ + \frac{1}{2} a t^2$. In this equation, x represents the current position of the ball, x_0 is the previous position of the ball, v_0 is the ball's previous velocity, a is the ball's acceleration, and t is the delta time. The ball has a constant horizontal velocity, and the vertical acceleration of the ball is gravity, or 9.81 m/s^2. For a given launching angle, the range (x) and the highest point the ball can reach (y) depend on the initial velocity of the ball as shown in Fig. 5. The velocity components of the ball can be described as $v_x = v \ \cos \ \theta$ and $v_y = v \ sin \ \theta$. If the launching angle and the ball's position are known, velocity can be obtained using equation (1). In this equation, gravity g is used directly instead of acceleration a.

$$v = \sqrt{\frac{x^2 g}{x \ \sin 2\theta - 2y \ \cos^2 \theta}} \tag{1}$$

To obtain the launching angle with a given velocity and landing point, we use equation (2), which is a derivation of the original equation. This equation can produce zero, one or two solutions. When there are two solutions, the smaller angle results in a more direct ball flight towards the participant, whereas the larger angle results in the ball flying high into the air before arriving at the participant's location, as shown in Fig. 5. When two

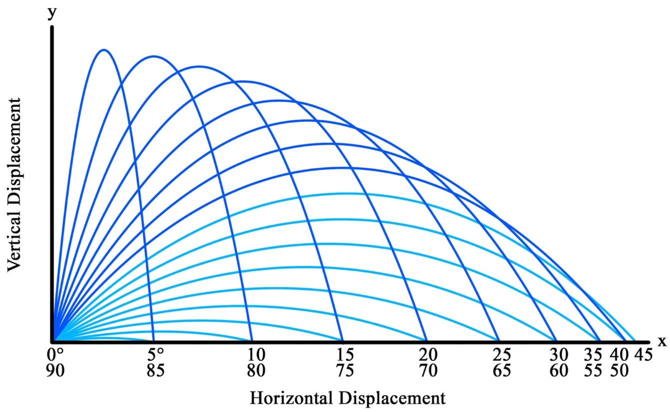

Fig. 5. Flight distance increases with velocity for the same launching angle. For certain landing positions and velocities, there are two launching angles.

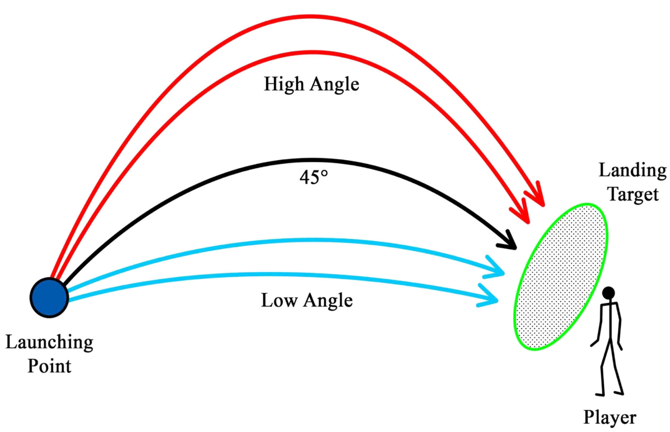

Fig. 6. Ball target position and path are randomly chosen in order to reduce educational effect and offer a more realistic experience.

solutions are generated, the system will randomly choose one.

$$\theta = \arctan\left(\frac{v^2 \pm \sqrt{v^4 - g\left(gx^2 + 2yv^2\right)}}{gx}\right) \tag{2}$$

To determine the ball's landing position, the system checks to see if the participant is within range of the chosen launching angle and velocity. The landing position is then chosen at a random location within range of the participant so that the participant doesn't have to move substantially to catch the ball. Fig. 6 shows an illustration of a target point and the ball path toward that point.

The equations used in this system provide a realistic approximation of ball flight, but do not take into consideration factors such as air drag, spin, and environmental conditions.

F. Behavioral Analysis

The system measures three main aspects of the participant's behavior while performing the virtual catching task, the first of which is the participant's attention. A healthy normal individual should be able to focus on the ball throughout the course of its flight, whereas someone with neurocognitive impairment may find it difficult to track the ball, which could be due to a deficit in target prediction or movement tracking. The second behavioral aspect monitored by the system is the participant's ability to mentally calculate the ball's landing time and position. This is indicated by whether or not they initiate proper actions with spatial and temporal coherency. Deficits in target prediction can also play a role in this behavior, as well as apraxia (a decrease in visuomotor skills), and slower IPS. The third aspect is the catch conditions, such as the participant's balance and movements while catching. Abnormal movements or imbalance while catching the ball can be indicators of neurocognitive decline, as imbalance and motor disorders are common to several neurological conditions.

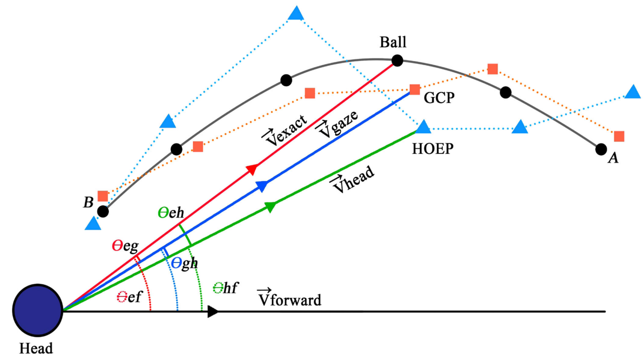

Fig. 7. Illustration of correlation among participant's gaze direction, head orientation and exact attention direction to the stimulus. $\vec{V}$ *exact*, $\vec{V}$ *gaze*, $\vec{V}$ *head* and $\vec{V}$ *forward* vectors start from test subject's head and point out to the ball, Gaze Convergence Point (GCP) and Head Orientation End Point (HOEP) respectively. The angles between $\vec{V}$ *exact*, $\vec{V}$ *gaze*, $\vec{V}$ *head* and $\vec{V}$ *forward* show the participant's behavioral responses to the flying ball. θeg, θeh and θgh are calculated when FOVE HMD is used, while θef, θhf and θeh are calculated and gaze direction is estimated when only HTC Vive HMD is used.

To assess the test subject's attention, the eye gaze direction, head orientation and a direct vector from the participant's head to the ball are obtained. Depending on the HMD used, eye gaze direction is either directly measured (using the eye-tracking FOVE HMD) or estimated by head orientation (when HTC Vive is used). While using head orientation does not provide an actual measurement of the participant's gaze, it provides a reasonable approximation. Fig. 7 illustrates how the participant tracks the movement of the ball over time. The ball, represented by black dots, moves towards the participant's head at each time step, while the participant's Gaze Convergence Point (GCP) and Head Orientation End Point (HOEP), represented by orange squares and blue triangles respectively, are within close range of the ball. GCP is gained from the FOVE eye tracking system, and HOEP is calculated using the head orientation vector extended to a point near the ball. The $\vec{V}$ *exact* vector represents the exact orientation from the participant's head to the ball, $\vec{V}$ *gaze* represents the participant's eye gaze direction, and $\vec{V}$ *head* represents the participant's head direction. $\vec{V}$ *forward* represents the z vector of the environment, which is the direction the participant faces toward the launching point of the ball. $\vec{V}$ *exact* and $\vec{V}$ *head* differ in that $\vec{V}$ *exact* is a direct vector from head to the ball regardless of head orientation, while $\vec{V}$ *head* is the vector of the head orientation. All of these vectors are normalized, but are extended for visualization purposes.

Finding the relationship between $\vec{V}$ *exact,* $\vec{V}$ *gaze and* $\vec{V}$ *head* provides a simplistic approach to analyzing the participant's attention. The difference in angle between $\vec{V}$ *exact* and $\vec{V}$ *gaze* is represented by θeg, θeh represents the angle between $\vec{V}$ *exact* and $\vec{V}$ *head*, and θgh is the angle between $\vec{V}$ *gaze* and $\vec{V}$ *head.* θeg is used to observe the relationship between where the participant needs to focus and where the participant is actually looking. The smaller this angle is, the more the participant is focused on the ball. θeh shows the head orientation with respect to the exact vector to the ball. This tells us where the head is supposed to be oriented in order to have maximum attention on the ball. The smaller the value of θeh, the better the orientation of the head towards the ball. θgh shows the relationship between eye gaze movement and head movement. Although the test subject may orient their head towards the ball, this does not mean that the participant is looking at the ball.

If a FOVE HMD is unavailable, our system is still able to offer reasonable attention analysis by estimating eye gaze using the HTC Vive HMD. In this case, we find the relationship between $\vec{V}$ *exact and* $\vec{V}$ *forward* and call the angle between them θef. The angle between $\vec{V}$ *head* and $\vec{V}$ *forward* is called θhf. Then, θeh (previously presented) represents the distraction angle between the participant's attention and the exact vector direction. Even though this estimation is not ideal, it is reasonably acceptable due to the fact that during ball flight, eye gaze is generally close to head orientation.

The values of θeg, θeh and θgh are expected to be smaller for healthy normal individuals and larger for those suffering from neurocognitive issues, as attention deficits often accompany neurocognitive impairment. Fig. 8 shows the expected changes of θeg, θeh and θgh over time for healthy individuals. While tracking the ball, it is expected that θeh should generally decrease as the ball approaches the participant, except when the participant catches the ball, at which point the error increases drastically. This is because at the last moment, the participant mentally calculates the catching position and no longer watches the ball. This drastic change can pinpoint the participant's position with respect to the ball, as well as determine whether the participant preemptively moved and waited for the ball or extended their arm. A larger value of θeh is expected when the participant extends their arm due to the larger angle between the participant's orientation and the ball. However, when the participant is in place and waiting for the ball without extending their arm, θeh should be smaller. It is expected that θeg will be similar to θeh most of the time, though sometimes the participant may get distracted and the eye direction may not have a consistent correlation with the ball distance. This is due to the fact that eye movements tend to change drastically and sporadically compared to head movements. Furthermore, when the participant blinks, the FOVE HMD loses track of eye gaze. To minimize error when the participant closes their eyes, the previous gaze direction is recorded during eye closing events. In contrast with eye movements, head movements follow the ball more smoothly, as head movements are slower than eye movements.

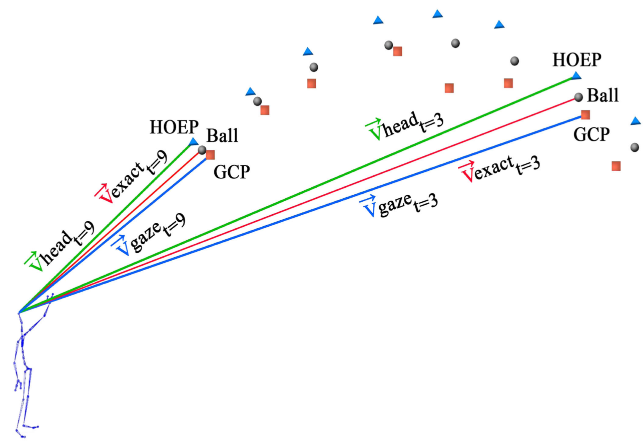

Fig. 8. θeh decreases when the object approaches the participant. θgh may change drastically at any time due to the fast and erratic nature of eye movement.

The participant's ability to mentally calculate physical phenomena is determined by recording their reaction to the ball when it lands. Accurate estimations of the arrival time and location of the ball result in catches, or near catches where the participant may have made slight miscalculations. To determine the accuracy of the participant's spatial and temporal estimations, we measure the distance between the participant's hand and the ball upon pulling the trigger, as well as the overall closest distance between the ball and glove. Smaller distances indicate better spatiotemporal estimates, whereas larger distances can indicate a problem in the participant's mental calculations. Additional information such as the target position, landing position, caught position, flight duration, gravity, ball velocity, launching angle, and whether or not the ball was caught are also recorded. This information is later used for a memory test where the participant is asked to recall the order of caught and missed balls from a certain number of trials (the number of trials is decided by the researchers).

The catch conditions are assessed by monitoring a person's balance and posture while catching. To assess the participant's balance and posture, a Microsoft Kinect records the participant's joint locations as they perform the task. This information can be replayed at a later time for assessment, and can be processed to identify any significant offsets in the participant's Center of Mass (CoM), which may indicate a loss of balance. CoM represents the mean position of the body mass, and can be measured as the sum of weighted joint vectors relative to a reference point divided by the total mass of all joints, as seen in equation (3).

$$CoM = \frac{1}{m_{total}} \sum_{i=0}^{n} m_i r_i \tag{3}$$

In this equation, m_i is the mass of the joint *i*, r_i is the vector from the reference point to the joint *i*, and m_{total} is the total mass of all joints. The skeleton recording from the Kinect does not associate any weights with joints. Therefore, in order to calculate an individual's CoM, we estimate the body's segmented weight distribution [57]. To do this, the skeleton is divided into

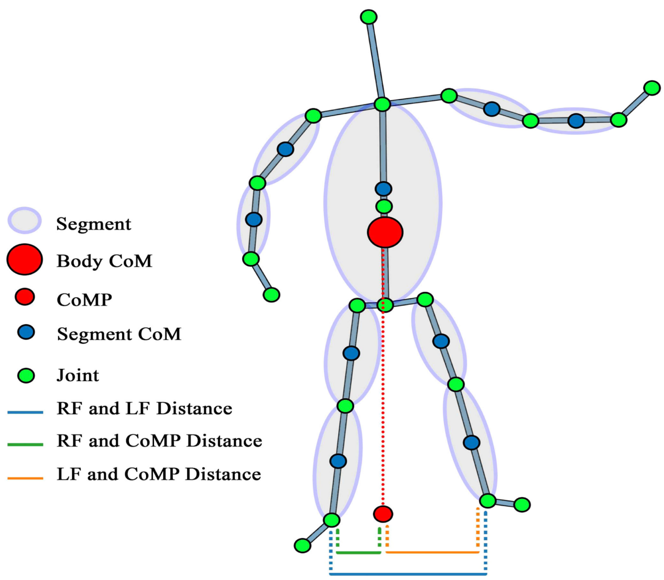

Fig. 9. Body segmentation to determine CoM and CoMP. The distance between the feet and CoMP is used for to determine balance.

9 segments as shown in Fig. 9, leaving out the feet and hands because their mass is negligible. We then calculate the CoM for each segment, multiply the output by its weight percentage, and then divide by the total body mass [57].

The participant's CoM changes depending on the way the body is postured and the individual's mass distribution. To maintain body stability while catching the ball, the CoM projection should be somewhere between the person's feet [58]. To estimate balance, we draw a vector starting from the CoM to the CoM Projection point (CoMP), which is the projection of the CoM on the ground between the feet as shown in Fig. 9. This vector is constantly updated for real-time balance assessment. When the CoMP shifts outside of either foot, this is a strong indication that the participant is not balanced. To determine the CoMP's location in relation to the feet, we calculate the distance between the CoMP and each foot separately, and then compare this measurement to the distance between both feet. If the distance between the feet is smaller than the distance between the CoMP and either foot, it means the CoMP is not between the feet, and the participant is likely unbalanced.

G. Measurements

Each of the tested neurocognitive functions were given a normalized score. The scoring is based on the mean and Standard Deviation (SD) of the participant's performance data. For instance, attention is scored by taking the standard deviation of the gaze error and giving it a weight. A smaller standard deviation indicates that the participant is less distracted and is focused on the flying ball. In a similar manner, the reaction score is based on the standard deviation of the closest distance between the player and the ball when the participant pulls the trigger. The balance score is based on the probability of being unbalanced, which is calculated based on how many times the CoMP went outside of either foot, and how far it went in relationship to the distance between both feet each time. Researchers have the freedom to choose the number of times and the distance value. The memory score is based on how accurately the participant can recall the order of caught balls or the number of the caught balls. For instance, the participant could recall which balls they caught during the first 5 trials, or how many balls they caught overall. Since all of the data taken during the experiment is stored, researchers have the freedom to change the scoring criteria to whatever is suitable for their study.

IV. Experiment Setup

We recruited 20 healthy normal test subjects to participate in a proof-of-concept experiment to demonstrate the feasibility of our system. Test subjects were between the ages of 21 and 70, with 12 male participants and 8 female participants. The average age was 36. The Kinect was placed 1 meter above the ground and 2 meters away from the subject. The test subjects signed a consent form before the experiment began. The FOVE eye tracking system was calibrated for each test subject, and a calibration profile was created for each person. Test subjects were then asked to wear the HMD and explore the virtual environment in order to become familiar with the application. After the test subjects familiarized themselves with the environment, the operator would instruct the test subject to catch as many baseballs as possible, and the operator would then start the experiment.

We conducted 30 trials per test subject with 3–5 seconds between each trial. The ball launching angles were evenly distributed between 10° and 70° (there were 5 trials between 10° and 20°, 5 trials between 20° and 30°, and so on), though their order during the trials was chosen at random. The ball velocity and target position were randomly chosen for each trial. As the participant performed the task in VR, the data needed to analyze the participant's attention, estimation of the ball's landing time and position, and catch conditions were recorded.

V. Results

To assess a person's attention over time, we plotted data for one individual over the course of one trial. Angles θeg, θeh and θgh obtained while using the FOVE HMD are plotted in Fig. 10. The figure represents the data taken from a young male participant while catching a flying ball at a speed of 24.7 m/s launched at an angle of 35.9°. The ball flight took over 2.5 seconds from the launching point to the target position. The test subject successfully caught the ball at less than half a meter away from the target position. The recorded data shows that θeh changes smoothly over time, indicating that the participant only changed their head position and orientation slightly. θgh has some occasional dramatic changes due to the fast and sometimes erratic nature of human eye movement. A strong correlation between θeh and θgh can be seen in the data, which is due to the fact that the participant will generally not move their head and eyes in a colinear manner, but will tend to shift their eye gaze towards stimuli more than their head. This is why θeg is smaller than θeh and θgh. The figure shows that when the ball approaches the participant, both θeh and θgh exhibit the same behavior, indicating that the participant is focused on the ball.

Fig. 10. θeh and θgh behave similarly as the participant tries to focus on the ball, while θeg has smaller values because it is physically impossible to look at some points without turning the head. Additionally, θeg is less smooth because the participant moves their eyes more freely compared to their head.

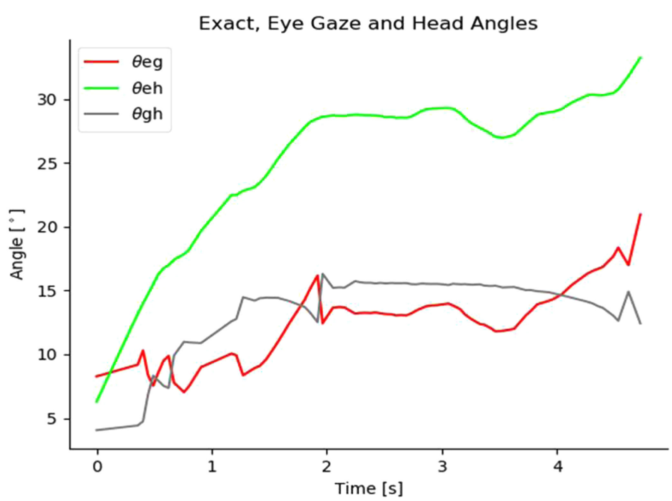

Fig. 11. This figure shows the data from a trial when the test subject is distracted and is not focusing on the ball. θeg continually increases, indicating that the participant's eye gaze was not following the ball. θeh increases over time, which shows that the subject wasn't orienting their head as the ball approached.

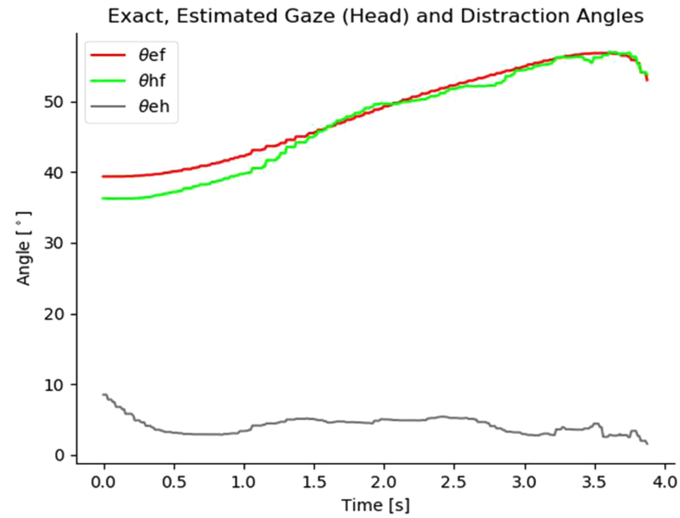

Fig. 12. θef and θhf change smoothly over the time due to the fact the participant has minimum head movement and changes in orientation, which leads to a decreased distraction angle θef.

θeg has a smaller value because the participant's gaze is focused on the ball, which makes the eye gaze direction closer to the exact vector, or the direction the participant is supposed to look. This causes the gaze direction and exact direction to be closer to each other, and both of them farther from the head orientation.

When the test subject is distracted, the data shows very different behavior. Fig. 11 illustrates data from a trial where the ball speed was launched at an angle of 66.08° and at a speed of 27.9 m/s. The ball took 5.26 seconds to reach the target position, and the test subject failed to catch the ball. The recorded data shows that θeh gradually increased during the ball flight, which indicates that the head and exact vectors start close to each other, but diverge over time. This happened due to the fact the participant didn't change their head orientation as the ball approached, and because this ball was launched with a high angle, the exact vector continually diverged from the head vector. Meanwhile, θeg and θgh showed different behaviors with respect to the exact and head vectors. Even though the eye gaze vector was closer to the exact vector, the test subject wasn't focused on the ball, which is why θeg increased over time. This figure also shows that θeg and θgh intersect a few times, which means that the subject's head and eye gaze were pointed in the same direction, and were both equally distant from the exact vector.

Fig. 10 and 11 accurately reflect the participant's head and eye gaze behaviors. This information can help researchers to better understand the participant's attention while performing a task. As mentioned previously, in the absence of a FOVE HMD, our system is able to approximate gaze direction using an HTC Vive HMD. Fig. 12 shows a plot of a trial using an HTC Vive HMD when the participant caught the ball. In this instance, the ball was launched at an angle of 69.25° and a speed of 30.60 m/s, and the ball flight lasted 4 seconds. As shown, the distraction angle was large when the ball was launched and decreased as the ball approached the participant. The distraction angle was near zero when the ball reached the catching position, meaning the participant looked directly at the ball. The figure shows θef changing smoothly over time, whereas θhf changes less smoothly over time, indicating that the participant was most likely still and didn't change their head position, but rather changed their head orientation.

The participant's ability to make accurate visual spatiotemporal calculations is assessed by monitoring the distances between the participant's hand and the ball during trials. Fig. 13 shows the distances between the ball and the glove when the participant pulled the trigger for 10 trials, as well as the overall minimum distance between the ball and glove. Fig. 13 also shows whether the ball was caught, missed, or if the participant never pulled the trigger. In the three instances when the participant caught the ball, the trigger distance and closest distance are nearly identical, meaning the participant pulled the trigger at the perfect time and location. When the participant missed, the distance between

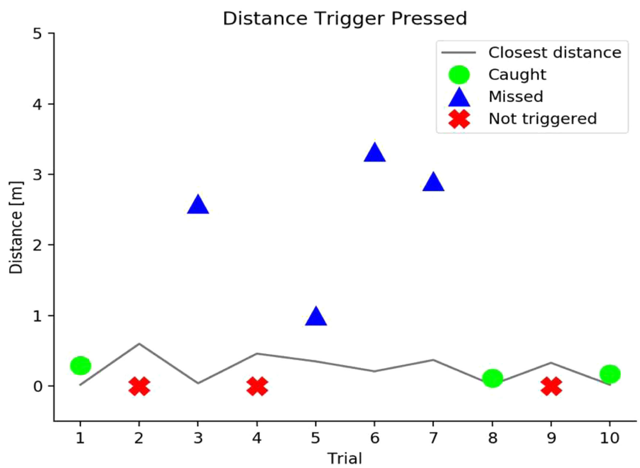

Fig. 13. Participant's response when the ball reaches the landing position. Three balls were caught, four missed and three not triggered.

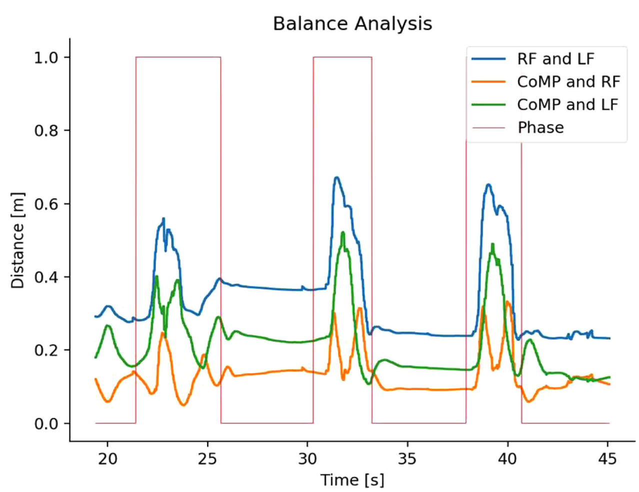

Fig. 14. Distance between the feet, and distances between CoMP and each foot as a function of time over consecutive trials.

the ball and the glove was between 1 and 3.3 meters, meaning that the participant triggered the controller at the wrong time. The data from Fig. 13 also shows three instances where the participant did not pull the trigger. After the experiment, the recorded data from the trials is used for memory assessment, where the participant could be asked to recall the order in which they caught the first 5 balls or the number of total caught ball (or any other information since all of reaction behavior is recorded).

To analyze balance, we examined data from the same subject test plotted in Fig. 13. For each trial, we measured the distance between the CoMP and each foot, as well as the distance between both feet. The data was divided into two phases: the anticipation phase, when the participant is waiting for the ball and tracking it visually, and the reaction phase, when the participant attempts to catch the ball. Fig. 14 shows the measured distance between the right foot (RF) and left foot (LF), as well as the distance between the CoMP and each foot as a function of time. The red line indicates the phase, where zero represents the anticipation phase and one indicates the reaction phase.

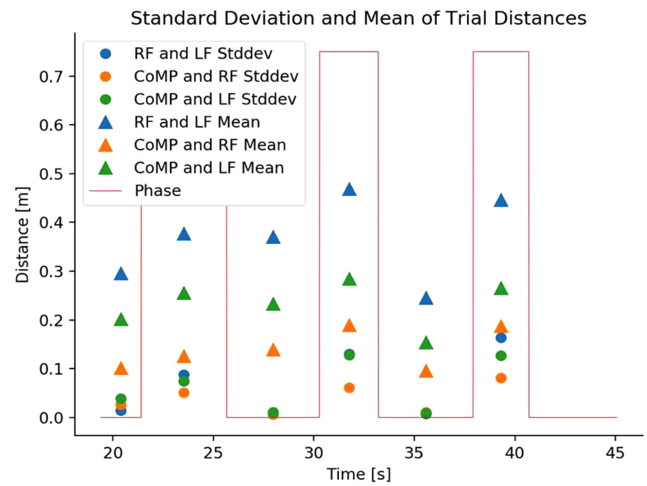

Fig. 15. Standard deviation and mean of the distance between the feet, and distances between CoMP and each foot.

Fig. 14 presents the data for three consecutive trials. As shown, the distance between both feet is always greater than the distance between the CoMP and either foot, showing that the participant was balanced during the trials. At one point during the first trial, the distance between the CoMP and LF is similar to the distance between both feet, meaning the CoM and left foot were nearly on the same vertical line. If the participant was unbalanced at any point, the orange line or the green line would have a higher value than the blue line. All three lines have a consistent horizontal value over the anticipation phase, indicating that the participant was mostly standing still. In contrast, during the reaction phase, the values indicate an increase in motion as the participant attempts to catch the ball. During the anticipation phase of the second trial, the overall values were greater than during the anticipation phase of the third trial, meaning the participant's feet were a greater distance apart. During the reaction phase of each trial, the distance varies depending on the participant's reaction to the flying ball. If the ball is launched at a high angle, the values are generally greater than for balls launched at a lower angle due to the larger movements required to catch the ball. Using this information, we can determine when the participant is unbalanced by examining the relationship between recorded distances.

In order to gain further understanding of the participant's balance, we can perform statistical analysis of the data presented in Fig. 14. Fig. 15 shows the standard deviation and mean of the distances plotted in Fig. 14. From the data in Fig. 15, we can see that the mean of the distance between both feet is always higher than the distance between the CoMP and RF or CoMP and LF, indicating that the participant is generally balanced throughout the phase. Additionally, during the first trial, we can see in the anticipation phase that the standard deviation of both CoMP and RF and CoMP and LF are bigger than the standard deviation of RF and LF. At the same time, the mean distance of RF and LF is larger than the other distance means. From this information, we can infer that the participant was balanced and moving their upper body, but keeping their feet stationary. Swinging the upper body causes the CoMP to move, generating varying distance

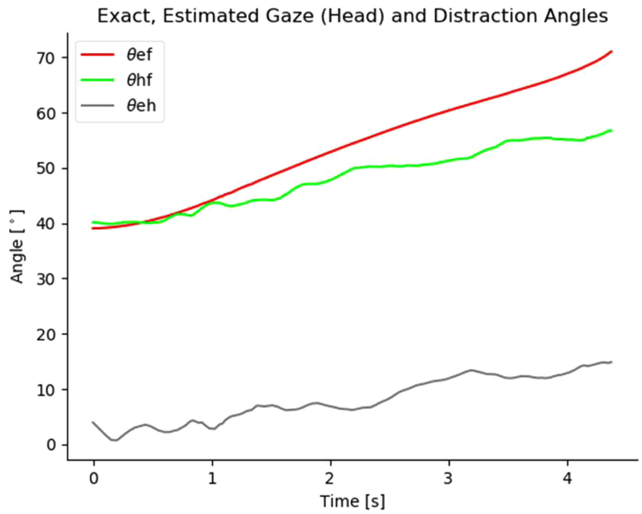

Fig. 16. θef and θhf change smoothly over the time due to the fact the participant has minimum head movement and changes in orientation. However, this changed shows that the participant was not focusing on the ball. This is why θef increases when ball approaches the participant.

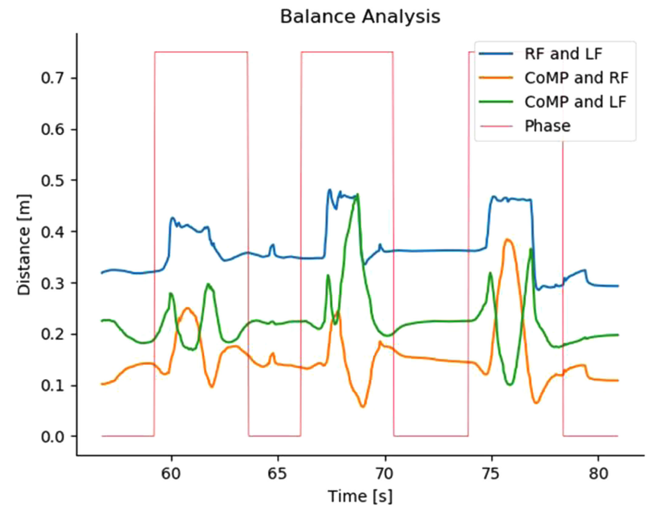

Fig. 17. Distance between the feet, and distances between CoMP and each foot as a function of time over consecutive trials. This data shows that the participant has minimum movement and the distance between RF and LF almost stayed unchanged.

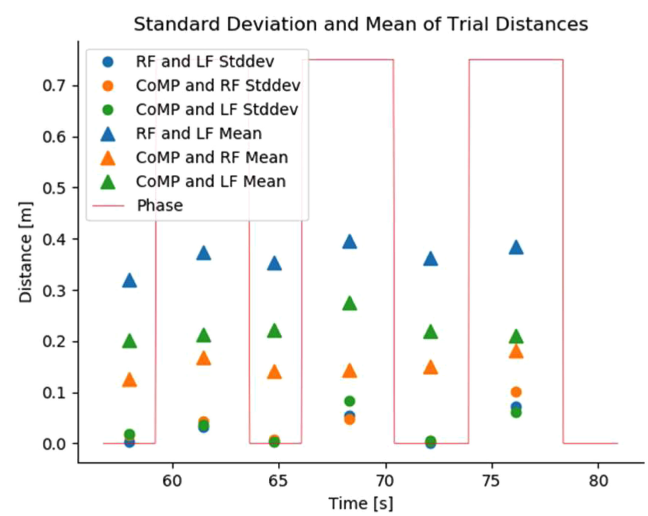

Fig. 18. SD and mean of the distance between the feet, and distances between CoMP and each foot. The mean of the RF and LF has stayed similar over the time which indicates the minimum movement.

values between the CoMP and each foot, whereas keeping the feet stationary generates smaller distance values, and therefore smaller standard deviation values. In the anticipation phase of the second and third trials, the standard deviation values were nearly the same, indicating that the participant's feet and body were mostly still. Further statistical analysis could reveal additional clues about the participant's behavior.

As mentioned previously, one of the main contributions of this work is to show the physical behavior of the test subjects. Therefore, we took data from a 70-year-old female participant and compared it with data taken from younger test subjects. Fig. 16 illustrates attention data where the ball is launched at a 62.4° angle and a speed of 26.5 m/s. The data shows that the test subject consistently pays attention to the ball at the beginning of the trial, but the attention error increases over time. The error reaches close to 20° when the ball reaches the target, and the participant missed the ball. If we compare this data to Fig. 12, we can see that the subject in Fig. 12 pays consistent attention to the visual stimulus, and the attention error decreases when the ball reaches the target, in contrast to the subject in Fig. 16. This may indicate that the subject in Fig. 16 has slower IPS and cannot follow the ball trajectory at the right time, or may have limited motor abilities and are not able to move their neck as quickly as necessary.

Body movement data for the same subject reveals an important physical aspect of the participant. Fig. 17 depicts that in the three consecutive trials, the subject had minimal movement, and sometimes was less balanced than the younger subject. The distance between RF and LF remained unchanged, which means the subject didn't try to move at all. In addition, during the second trial, the distance between CoMP and LF overlaps with the distance between RF and LF for a short time, which indicates that the participant's CoMP was on a vertical line with the LF. The data shows that in 7 frames the CoMP wasn't between the feet, which may indicate imbalance. Fig. 14 shows that the younger test subject moved more than the older test subject, and with more stability. This doesn't necessarily mean that the elderly subject suffers from a cognition impairment, but it distinctively shows the difference between both subjects' physical behaviors and motor capabilities.

The standard deviation and mean of the data shown in Fig. 17 gives a more direct interpretation of what has been mentioned previously about the subject's movement. Fig. 18 clearly illustrates that the standard deviation and mean values of all three distances (RF and LF, CoMP and RF, and CoMP and LF) are similar for both anticipation and reaction phases because of the minimum movement by the subject. In contrast to this, Fig. 15 shows significant changes between the anticipation and reaction phases. It's also noticable in Fig. 18 that the mean of the distance between CoMP and LF is almost always bigger than the mean of the distance between CoMP and RF. This happens when the subject is always leaning to the right side. Most of our test subjects are right-handed, and that causes them to put more weight on their right side when they hold the controller. We had one left-handed test subject and the data shows that he leaned more to the left.

Our results show that our system is able to record many physical behaviors and attributes that may allude to neurocognitive function. We believe that providing these properties in our system will make this a very effective tool for researchers to explore the relationship between neurocognitive disorders and physical functions.

VI. Discussion and Future Work

Trials with healthy normal test subjects showed data that reflects the physical behavior of the subjects, as well as their attention and balance. One unexpected result was that in some instances, the participant may not attempt to catch the ball. We considered these data points to be outliers, and disregarded this data when performing reaction analysis. One participant reported that their reasoning for not pulling the trigger was the ball was too fast to attempt to catch it, which agreed with the recorded landing position and velocity information.

Data produced by this system should be analyzed in collaboration with neurocognitive disorder specialists to avoid misleading and biased conclusions. For example, if the data shows that the participant was balanced and attentive but failed to catch the ball, this may not indicate neurocognitive impairment. Factors such as the speed and position of the ball may contribute to a failed catch. In contrast, if the participant could not follow the ball's trajectory, was unbalanced, or couldn't catch a ball that could have been easily caught by a healthy individual, this may indicate underlying neurocognitive issues. If this scenario repeats several times, we can consider that a strong differentiator.

Data quality from the HMD and Kinect also greatly impacts analysis. For instance, when using the HTC Vive HMD and a Kinect, the participant's gaze is estimated using the HMD orientation, which will not always indicate the participant's actual gaze direction. In contrast, using the FOVE HMD with embedded eye tracking provides accurate gaze analysis, and will improve the quality of the data. However, FOVE's current design has some limitations; it doesn't support sound, and it has some difficulties with orientation and position when combined with HTC Vive hardware. Although the Vive tracker can solve this problem, it increases system overhead. Using the Kinect for balance assessment without measuring the Center of Pressure (COP) has some limitations as well, especially in detecting balance during forward and backward body movements.

The proposed system is intended to be easy to adopt and deploy to clinical experiments with minimal modification. However, in order to provide additional biophysical information, our open architecture can be extended for use with additional sensors to measure vital signs and brain waves. In addition to monitoring the physical behaviors of test subjects, we plan to incorporate other measures to the test procedure to assess neurocognitive function. One immediate goal of this project is to experiment with physically improbable movement of the ball to see if the test subject notices the abnormal movement. This may involve having the ball ricochet off of a surface at an improbable angle, or artificially changing the velocity of the ball in a way that would be noticeably wrong to healthy normal individuals. We hypothesize that physically improbable movement will be easily detected by those who do not suffer from neurocognitive issues, and will not be as easily detected by test subjects that have neurocognitive impairment.

In developing this work, we do not claim that our system can definitively diagnose neurocognitive disorder, but we claim that it offers an insight into the probability of impairment. We consider this tool to be a preliminary prototype that aims to inspire further research in this direction, and recognize that further study and additional subject testing is needed. At our current stage, we are not focusing on diagnosing actual patients, and therefore a comparative analysis was not conducted. In the near future, more in-depth patient diagnosis and cross examinations will be conducted. We plan to recruit participants with neurological impairments so that we can compare their results to results from healthy participants.

VII. Conclusion

In this paper, we have presented a prototype VR and motion analysis framework that can be used by psychologists and neurologists for general purpose neurocognitive assessment and movement analysis. This work was motivated by the absence of a system that could objectively and neutrally assesses neurocognitive disorders. This work provides a general framework for behavioral assessment for several types of neurocognitive disorders, including AD, PD, and mTBI, and provides a common platform for comparisons between studies. By attempting to catch a baseball in a virtual environment, the neurocognitive functions of the test subject can be quantitatively assessed based on the accuracy with which the test subject performs the task, as well as the attention, balance, memory, and motor skills of the test subject while they perform the task. The results from our experiment show that our system is able to provide informative data related to a participant's ability to track visual stimuli, as well as their balance and body posture while performing a task. This system has been released under a GPL license so that it can be downloaded, modified and used by researchers. In the future, this framework can be easily adapted to suit a variety of research needs by interfacing with various biophysical sensors, and by implementing additional cognitive tests.

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