TECHNOLOGY-ASSISTED SCREENING AND BALANCE TRAINING SYSTEMS FOR STROKE PATIENTS

A thesis submitted in partial fulfillment of the requirements
for the degree of Doctor of Philosophy in
Electrical Engineering

By

DEEPESH KUMAR
Roll Number: 13310019

Under the guidance of:

DR. UTTAMA LAHIRI
Associate Professor, Electrical Engineering

INDIAN INSTITUTE OF TECHNOLOGY GANDHINAGAR
April, 2018

DEDICATION

To my source of inspiration, My parents

DECLARATION

I declare that this written submission represents my ideas in my own words and I have adequately cited and referenced the original sources, where others' ideas or words have been included, I have adequately cited and referenced the original sources. I also declare that I have adhered to all principles of academic integrity and have not misrepresented or falsified any idea/data/fact/source in my submission. I am fully aware that any violation of the above can cause disciplinary action by the Institute and can also evoke penal action from the sources which either have not been properly cited or from whom proper permission has not been sought for before being used.

Signature

Name: Deepesh Kumar

Roll No. 13310019

Date: April 30, 2018

CERTIFICATE

It is certified that the work contained in the thesis titled "**Technology-Assisted Screening and Balance Training Systems for Stroke Patients"** by Deepesh Kumar (13310019), has been carried out under my supervision and that this work has not been submitted elsewhere for a degree.

Dr. Uttama Lahiri

Associate Professor

Indian Institute of Technology Gandhinagar

Palaj, Gandhinagar-382355, India.

ABSTRACT

Stroke is the second most common cause of death and fourth leading cause of disability worldwide. The morbidity and mortality rate due to stroke can be addressed at least to some extent if the treatments are given in short time window post the onset of stroke symptoms. This makes early screening of stroke critical. Some of the common symptoms of stroke are sudden trouble seeing in one or both eyes and visual neglect. These symptoms can be identified as related to stroke by expert clinicians. Unfortunately, lack of adequately trained neurologists and healthcare workers has limited its use only in major urban centers in countries like India. Faced with such challenge, a cost-effective, simple-to-use, a clinically valid device that can pick up potential biomarkers representative of neurological dysfunction can be critical and useful in both urban and rural healthcare settings. One such biomarker can be captured by picking up one's oculomotor signature for screening stroke cases. This is because one’s oculomotor system connected with a vast network of brain areas become vulnerable to various neurological disorders such as stroke resulting in unique clinical patterns. Thus, the oculomotor examination can serve as a sensitive and also early indicator of neurological dysfunction. In this research, my **first objective was to develop a cost-effective, simple-to-use and clinically-valid device (SmartEye) for screening one’s probable neurological disorder from oculomotor signature.** To achieve this objective, I designed a gaze-sensitive computer-based system called SmartEye. Results of the SmartEye-based study indicate that one's gaze-related indices such as gaze fixation, smooth pursuit, and blinking might serve as potential quantitative biomarkers for screening of stroke cases.

Once screened for the possible neurological disorder, it is also critical to address at least some of the accompanying deficits such as those related to balance and mobility. As a result of stroke,

individuals often succumb to hemiplegia causing complete or partial paralysis on one side of the body. These patients show impaired balance due to asymmetric body posture which causes them to fall while performing activities of daily living. Such deficits are usually addressed by rehabilitation exercises done under the supervision of trained therapists spanning over repeated exposures. These conventional techniques, though powerful often suffer from subjectivity, restricted availability of trained therapists particularly in primary health centers, monotonicity with repetitive training/exercises stealing away patients' motivation. Thus, investigators have been exploring technology-assisted training.

Among the technology-assisted balance rehabilitation systems use of robot-assisted, computer-based and Virtual Reality (VR) based techniques have started gaining popularity. Among these techniques, I chose VR augmented with peripheral device such as Balance Board and Kinect (to determine center of pressure (CoP) and center of mass (CoM)) for developing a balance training system that offers the flexibility of design, controllability, and individualized approach to balance rehabilitation. My system offers individualization along with variations in tasks used for balance training that is not there in the currently existing VR-based systems that use off-the-shelf games (designed with an entertainment perspective) along with limited individualization thereby restricting the applicability for rehabilitation. Thus, the **second objective of my research was to develop and study the implication of intelligent adaptive VR-based balance training platforms on one's task performance in a balance training task**. To achieve this objective, I have explored the applicability of different VR-based training systems through three studies.

In the first study, I developed a VR-based CoP-assisted Balance Training (**VBaT**) platform, where VR-augmented user-interface using a single wireless balance board (WiiBB) was used. The VBaT offered tasks of varying challenges to the participants and was adaptive to one's performance quantified through weight-shifting capability during balance training. During the weight-shifting task, the position of a virtual object in the VR environment was controlled with CoP excursion measured by the WiiBB. Results of a usability study indicate the potential of the VBaT system to cause improvement in overall average task performance over the course of the training.

In my second study, I explored another modality such as use of CoM instead of CoP. For this, I developed a VR-based CoM-assisted Balance Training (**Virtual CoMBaT**) system interfaced with WiiBB and Kinect instead of marker-based motion capture systems as used by other researchers. In this system, I have used one's personalized CoM, estimated using Kinect sensor while offering individualized VR-based balance exercises. Here, the participant was allowed to interact with the VR-based tasks by shifting weight in different directions while standing on the ground. During the weight-shifting task, the position of a virtual object in the VR environment was controlled with CoM excursion. Results of a usability study indicate the potential of the Virtual CoMBaT system to (i) provide one's quantitative estimates of direction-specific residual balance capability and (ii) contribute to improvement in one's weight-shifting capability through an increase in performance in balance-related tasks of different challenge levels. Though both the VBaT and Virtual CoMBaT systems were able to contribute to the improvement in one's balance (weight-shifting ability) in the course of the training, yet these systems did not give the contribution of each of the two legs (Affected leg and comparatively healthy (Unaffected) leg of a hemiplegic patient) of an individual towards the improvement of his/her balance. This information on the relative contribution of each leg to one's weight-shifting ability is important to the clinician since it can assist a clinician in modifying the training paradigm so as to condition the rehabilitation effort in a way that the Affected leg gains greater usage by the participant. To achieve this, clinicians use targeted weight-shifting training for the patients in which the clinicians instruct the patients to increase the usage of Affected leg. However, these techniques operate as in open loop without any real-time visual feedback.

To achieve this, in my third study, I have designed a VR-based Balance Training platform interfaced with two WiiBB (**V2BaT**) augmented with a distributed weight paradigm (conditioning) coupled with a closed loop visual feedback. Here, during the weight-shifting task, the position of the virtual object in the VR environment was proportionally controlled with CoP excursions measured by two (one for each leg) WiiBB. Results of a usability study indicate the potential of the V2BaT system to contribute to (i) the improvement in one's balance (weight-shifting ability) in the course of the training and also (ii) increased usage of the Affected leg along with the Unaffected leg.

Having seen the use of VR-based systems augmented with peripheral devices in balance rehabilitation and the use of oculomotor signature as a screening biomarker of stroke, I wanted to understand the connectivity between one's eye movement and balance while performing a goal-directed balance task offered by the V2BaT system. Thus, the **third objective of my research was to extend the previous study using V2BaT augmented with operant conditioning to examine the implication of such a paradigm on one's gaze fixation behavior during a goal-directed balance task**. Similar to the previous study, here an individual was expected to maneuver a virtual object from a start location to a pre-defined static target location through CoP excursion while using distributed weight paradigm. At the same time, I monitored the individual's fixation pattern. The results of my study indicate that after the participant was exposed to the balance task augmented with operant conditioning, the participants demonstrated improved task performance coupled with increased fixation towards the static target location and decreased fixation towards the dynamic virtual object.

To summarize, through my research, I have developed (i) technology-assisted, easy-to-use and clinically valid screening device (SmartEye) that can identify one's oculomotor-based biomarkers as predictors of stroke, (ii) VR-based balance training platforms (VBaT, Virtual CoMBaT and V2BaT) augmented with Balance Board and Kinect that can offer flexibility to the balance training task regime, motivational exercise platform for patients, individualization and complementary tool for clinicians and (iii) an understanding into the connectivity of one's gaze fixation pattern with task performance in a VR-based balance task setting.

ACKNOWLDGEMENT

I would like to express my special appreciation and thanks to my advisor Dr. Uttama Lahiri. She has been supportive since the day I started my Ph.D. journey. I appreciate all her contribution of time, ideas and her patience to make my Ph.D. experience productive and stimulating. Many thanks are extended to my doctoral studies committee for their valuable feedback and encouragement. I thank Dr. Malavika Subramanyam and Dr. Shanmuganathan Raman for their insightful comments which incented me to widen my research from various perspectives.

Many thanks to Dr. Anirban Dutta, Dr. Mitsuhiro Hayashibe and Dr. Alejandro Ganzalez (formerly at DEMAR team of INRIA, Montpellier, France) for their valuable suggestions that substantially improved this research. Also, I thank them for all the help they have provided me during my visit to their laboratory. I would like to thank Dr. Abhijit Das for his valuable inputs in my research that helped me to improve the protocols of various clinical trials of usability studies.

I am grateful to Department of Science and Technology, India and Institut National de Recherche en Informatique et en Automatique, France for financially supporting this research work. Also my sincere thanks to Dr. Biman Kanti Ray from Bangur Institute of Neuroscience Kolkata, Dr. Anjali Bhise from Spine Institute at Civil Medical Hospital Ahmedabad, Dr. Vaittianadane Kesavan from CMP college of Nursing Gandhinagar and Mr. Sourabh from Sadbhavna Charitable Trust Ahmedabad for supporting this work by providing the participant pool and also by providing valuable clinical feedback. Also, my sincere gratitude to all the participants and their family who made this study possible.

Working with my fellow lab members of the Intelligent Rehabilitation and Affective Computing Systems Laboratory (IRACS) has been both enjoyable and educational. I was fortunate to get some of the bright students as my lab colleagues. I want to extend my gratitude to the following wonderful lab members who were helpful at various instants - Dr. Selvia Kuriakose, Pradeep Raj, Dhaval Solanki, Adyasha Dash, Sunny Verma, Taruna Yadav, Ritika Jain, Megh Patel, Ashish Dhiman, Valay Patel, Poojan Oza, Sutapa Bhattacharya, and Nirvik Sinha. I take this opportunity to express gratitude to all of the electrical lab staff members for their help and support, particularly, Mr. Pragnesh Parekh for his invaluable help.

Last but not the least, I thank my family for their love, encouragement and patient support without which this goal could not have been accomplished. I dedicate this dissertation to my mother and my late father who has been my constant source of inspiration.

TABLE OF CONTENT

LIST OF TABLES

LIST OF FIGURES

LIST OF ABBREVIATIONS

BBS	Berg Balance Scale
BPM	Blinks Per Minute
CoM	Center of Mass
CoP	Center of Pressure
CT	Catch Trial
FP	Fixation Point
Hemo-GaSiD	Head Mounted Gaze Sensitive Display
HLD	Heel Lift Detection
NT	Normal Trial
PSD	Power Spectral Density
ROI	Region of Interest
SDS	Stimulus Display Screen
SESC	Statistically Equivalent Serial Chain
SPL	Smooth Pursuit Length
TTL	Transistor-Transistor Logic
US	Ultrasonic Sensor
USB	Universal Serial Bus
VBaT	Virtual Reality based Balance Training
Virtual CoMBaT	Virtual Reality based Center of Mass assisted Balance training
VR	Virtual Reality
WiiBB	Wii Balance Board

LIST OF PUBLICATIONS

D. Kumar, A. Gonzalez, A. Das, A. Dutta, P. Fraisse, M. Hayashibe, and U. Lahiri, “Virtual Reality based Center of Mass Assisted Personalized Balance Training System,” *Front. Bioeng. Biotechnol.*, vol. 5, p. 85, 2017.

S. Verma, **D. Kumar**, A. Kumawat, A. Dutta, and U. Lahiri, “A low-cost adaptive balance training platform for stroke patients: A usability study,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 25, no. 7, pp. 935–944, 2017.

D. Kumar, A. Dutta, A. Das, and U. Lahiri, “SmartEye: Developing a Novel Eye Tracking System for Quantitative Assessment of Oculomotor Abnormalities,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 4320, no. c, pp. 1–1, 2016.

D. Kumar, A. Das, U. Lahiri, and A. Dutta, “A Human-machine-interface Integrating Low-cost Sensors with a Neuromuscular Electrical Stimulation System for Post-stroke Balance Rehabilitation.,” *J. Vis. Exp.*, no. 110, p. 2016, 2016.

D. Kumar, A. Dutta, A. Das and U. Lahiri, " A step towards developing a gaze-based screening tool for neurological isorders," *International Journal of Stroke* Vol. 11, no. 3S , pp. 35, 2016 (Abstr).

D. Kumar, A. Dutta, A. Das, and U. Lahiri, “Engagement sensitive visual stimulation,” *Eur. J. Transl. Myol.*, vol. 26, no. 2, 2016.

S. Verma, A. Kumawat, **D. Kumar**, A. Dutta and U. Lahiri, “A Step Towards an Adaptive Human Computer Interaction System for Balance Rehabilitation,” *2nd International IEEE conference on Human Computer Interaction*, 10th-11th March, Chennai, India, 2016.

D. Kumar, G. Aggarwal, R. Sehgal, A. Das, U. Lahiri, and A. Dutta, “Engagement-sensitive interactive neuromuscular electrical therapy system for post-stroke balance rehabilitation - a concept study,” *2015 7th Int. IEEE/EMBS Conf. Neural Eng.*, vol. 2015–July, pp. 190–193, 2015.

A. Dutta, **D. Kumar,** U. Lahiri, A. Das, and M. V. Padma. "Post-stroke engagement-sensitive balance rehabilitation under an adaptive multi-level electrotherapy: clinical hypothesis and computational framework." *Neuroscience and Biomedical Engineering* 2, no. 2 (2014): 68-80.

CHAPTER 1

INTRODUCTION

1.1 Introduction

This dissertation presents the design and development of technology-assisted screening device and balance training systems for individuals with stroke. In my research, I have developed a user-friendly, easily accessible, cost-effective and clinically valid gaze-based screening device (SmartEye) (described in Section 1.2.1). Additionally, I have designed intelligent adaptive Virtual Reality (VR)-based balance training platforms to address the post-stroke balance deficit in hemiplegic post-stroke survivors (described in Section 1.2.2). Finally, I have designed study to understand the gaze behavior of post-stroke hemiplegic participants in VR-based goal-directed balance tasks (described in Section 1.2.3). In the forthcoming sections, I briefly present the motivation behind such design and development.

1.1.1 The Necessity of Early Screening of Stroke Cases and Role of Oculomotor Signature

A neurological disorder such as stroke is a medical emergency situation [1] whose aftereffects can often be minimized if the treatment is given within a golden hour, that is, 60 minutes post onset of stroke symptoms [2]. The importance of early treatment in the case of stroke can be realized through the common saying “time is brain” [3]. This phrase often emphasizes that human brain is adversely affected at a fast rate as the post-stroke period progresses thereby laying stress on rapid emergent evaluation and therapy. This makes early screening of stroke cases critical. While using conventional techniques for stroke screening, clinicians deploy manual methods such as physical test [4], Head Impulse Nystagmus and Test of Skewness (HINTS) [5], Optokinetic Drum [6] for identifying possible symptoms of stroke, namely, face drop, arm weakness, speech irregularity and vision-related problems [7]. Though the conventional techniques are useful yet they suffer from lack of availability of trained clinicians and subjectivity while deciphering the symptoms of stroke based on observation alone [8],[9]. There are high-end technological solutions for screening the stroke cases such as Magnetic Resonance Imaging (MRI) [10] and Computed Tomography (CT) Scans [11]. However, these technological solutions are costly and limited to specialized healthcare settings only [8]. This often causes these equipments to be inaccessible to sections of the society residing in the rural areas.

This necessitates the development of a clinically-valid screening device that is cost-effective, accessible, easy-to-use and able to provide a quantitative measure of probable stroke condition. Such a screening device may play a critical role in improving identification and management of neurological dysfunction caused by a stroke in both rural as well as in urban settings. To achieve this, it is critical to identify and quantify potential biomarkers of stroke in a cost-effective and objective manner. Among the potential biomarkers such as cerebrospinal fluid, blood plasma, Matrix metalloproteinase9 [12], [13] and oculomotor signatures [14], [15], I chose to pick up the oculomotor signature since a vast network of brain areas (vulnerable to neurological disorders) are engaged in oculomotor functionality, from low-level visual processing to motor control of gaze orientation [16]. Again, literature indicates that stroke patients can demonstrate gaze abnormalities marked by unique clinical patterns [17] even within less than six hour after the onset of stroke symptoms [6], [18] making oculomotor signature an early stage biomarker. Apart from being an early predictor of stroke, the oculomotor disturbance has been reported to be prevalent among stroke patients. Evidence from literature reports that up to 86% of stroke patients have some form of oculomotor disturbance [15]. Thus, the oculomotor examination can serve as a sensitive and also early indicator of stroke condition. Given the potential of oculomotor-related features to serve as biomarkers of one's neurological disorder, in my present research, I have developed a low-cost, easy-to-use gaze-based screening device for probable stroke condition (Objective 1; Section 1.2.1 below).

1.1.2 Role of Technology-assisted Balance Rehabilitation Systems

Post-stroke vision problem can be one of the major contributors of impaired balance in stroke participants. This is because one's vision along with vestibular and somatosensory systems play important role in maintaining balance [19]. For example, Uchiyama et al. [20] and Thomas et al. [21] reported that vision (that can be adversely affected due to stroke condition) contributes to postural control by providing afferent feedback to cerebellum through gaze fixation, smooth pursuit and saccades. Literature indicates that visual neglect (one of the post-stroke symptoms) that is a disorder of attention whereby patients characteristically fail to orientate or to respond to stimuli located on the contralesional side [22] leads to balance disorder [23], [24]. Deficits in balance need to be addressed since this can lead to falls [25] that in turn can adversely affect one's mobility. This deficit comes due to partial paralysis. In the case of partial paralysis, one suffers from muscle weakness in both the upper and lower limbs of one side (right or left) of the body. This condition is called as hemiplegia [26]. Due to hemiplegia, the patients exhibit asymmetric body weight distribution leading to impaired balance [27].

To address such balance deficits, conventional balance rehabilitation techniques are often used. Though these techniques have been reported to show promise as far as improving one's balance [28] is concerned yet it suffers from certain limitations. For example, these rely heavily on physical presence of therapists during rehabilitation. As a result, the conventional techniques suffer from requirement of one-to-one therapist's supervision and restricted availability of trained therapists in primary health centers on account of low doctor:patient ratio [29], [30].

Again, these techniques need the patients to undergo repetitive training/exercises in similar settings devoid of variations that often turn out to be monotonous while stealing away patients' motivation [31]. Given these disadvantages, investigators have been exploring the applicability of technology-assisted training platforms that can play an important role in providing more accessible, quantitative, intensive, motivating and individualized training platforms. For example, many studies have investigated the application of various technology-assisted platforms, such as robot-assisted, computer-based and Virtual Reality (VR) based rehabilitation systems to improve one's balance [32], [33], [34], [35]. Among these techniques, I chose VR for developing a balance training system that offers the flexibility of design, controllability, and individualized approach to balance rehabilitation [36], [37].

The technology-assisted balance rehabilitation platforms are grounded on different approaches such as Center of Pressure (CoP) and Center of Mass (CoM). This is because, balance is a generic term that describes the dynamics of body posture needed to prevent falls [38]. Specifically, balance is related with the inertial characteristics of body segments [38] quantified through CoP and CoM-based indices. The CoP position can be measured using force platform data [39] and the CoM can be measured by using motion capture systems [40]. This necessitates the integration of peripheral devices such as force platform, motion capture systems and other devices with rehabilitation platforms.

In recent years, VR-based balance training platforms coupled with peripheral devices such as force platform and motion capture device have started to gain widespread usage [35], [38], [41], [42]. However, most of the existing VR-based systems addressing balance issues have used off-the-shelf games (designed with an entertainment perspective) for rehabilitation. Also, the balance tasks offered to the participants are not individualized. In other words, the systems are not adaptive to one's performance and residual balance capability that are critical requirements for effective rehabilitation. Also, for effective postural stability in hemiplegic stroke patients, it is important to quantify the extent of symmetry in weight distribution as far as both the lower limbs are concerned. This is because, there is evidence from literature that in extreme cases, the weight bearing ability of stroke survivors can be reduced by up to 43% on the paretic side of the lower extremity [42], [43]. This implies that both the legs of hemiplegic post-stroke survivors are not effectively working symmetrically during a balance rehabilitation task that involves weight-shifting. Thus, balance rehabilitation coupled with weight-shifting efforts should be directed towards regaining the symmetry in one's body weight distribution to promote effective use of both the Affected and Unaffected legs (that is legs on the Affected side and Unaffected sides of the body, respectively). This is possible if the patient's contribution of both the legs individually towards his / her overall balance can be measured. Such information can allow a clinician to condition the balance rehabilitation effort in such a way that the patient is encouraged to increase his/her usage of both the Affected and Unaffected legs as equally as possible. However, the existing VR-based balance rehabilitation systems [35], [38], [41], [42] do not offer such facility. In this research, I have developed intelligent VR-based balance training systems coupled with peripheral devices such as force platform and motion capture devices (Objective 2; Section 1.2.2 below) aimed towards offering individualized balance rehabilitation. All of my VR-based systems have been designed with an aim to improve one's balance. Among these, one of the VR-based systems have been designed to investigate the implication of conditioning the contribution of each of the two legs towards one's overall balance.

1.1.3 Role of Gaze Behavior in Goal-directed Balance Task

Since one's vision-based network is an integral part of human balance system [44], post-stroke abnormal eye movements can have implications on one's coordinated movement and balance. For example, literature review indicates that following a stroke, hemiplegic patients often suffer from slower, less accurate and less-coordinated visually-guided reaching movements than their healthy counterparts [45]. Also, studies on gaze behavior in stroke patients during visually-guided stepping task [46], [47] and locomotion task [48] highlight that deficits in one's eye movement control can have a detrimental effect on one's locomotion and dynamic balance leading to falls [49]. Again, investigators have reported their observations on static balance (that is, maintenance of posture on stable ground or Base of Support) [50] while standing. These researchers have indicated the importance of gaze during standing balance. However, none (to my knowledge) have reported their observations on gaze behavior during standing balance tasks. Since, one's visual input is an important ingredient for maintainance of balance [19], [50], studying of one's gaze behavior during standing balance task is critical. Thus, in this research, I also aimed to understand the gaze behavior of hemiplegic participants while they were exposed to VR-based standing balance tasks (Objective 3, Section 1.2.3).

1.2 Research Objectives

The objectives of my research were three fold, as follows:

1.2.1 Objective 1: Develop a cost-effective, simple-to-use, clinically-valid device (SmartEye) for screening one's probable neurological disorder from oculomotor signature.

I plan to develop a cost-effective, easy-to-use, easily-accessible and clinically-valid device (SmartEye) for screening oculomotor dysfunction that can occur as a result of stroke. The SmartEye system will present static and dynamic visual stimuli on the screen of a Task

Computer. The visual stimuli will be programmed to elicit participant's gaze fixation and smooth pursuit of eye. A cost-effective remote Eye Tracker will be used to monitor participant's eye movements while he/she follows the visual stimuli appearing on the computer screen. While the participant follows the visual stimulus, his/her gaze data will be recorded on the Task Computer at the backend. The gaze data corresponding to the static visual stimulus will be used to analyze participant's gaze fixation capability and that corresponding to the dynamic visual stimulus will be used to analyze his/her smooth pursuit. Apart from these two gaze-related indices, I will also look at the blinking action.

I plan to conduct a usability study with SmartEye system while involving stroke survivors and age-matched healthy participants. The plan will be to investigate whether SmartEye can (i) quantitatively identify gaze-related indices that can be used as potential biomarkers for probable neurological disorder? and (ii) can serve as a user-friendly and easily-accessible screening device for chronic stroke patients?

1.2.2 Objective 2: Develop and study the implication of Intelligent Adaptive VR-based Balance Training systems on one's task performance in a Balance Training task.

To achieve this objective, I will explore the applicability of different VR-based training systems through three studies.

In the first study, I intend to develop a VR-based Balance Training (**VBaT**) platform, where VR-augmented user-interface using Wireless Balance Board (WiiBB) from Nintendo will be tested in a laboratory setting for its feasibility. I plan to provide the participants with real-time feedback on their CoP excursion while they maneuver virtual objects (VR_{Obj}) in the VR environment through directional weight-shifting. The VR-based tasks will be designed so that these will necessitate effective use of both the lower limbs to perform a task. Specifically, the task will require one to stand on the WiiBB and shift his weight in different directions to maneuver the VR_{Obj} from its initial position to a fixed Target position. The VR-based tasks will be of different difficulty levels and the difficulty of the task presented to the participants will be adaptive to one's individualized performance. Also, in-house fabricated Heel Lift Detection (HLD) unit will be used to help the participants to adhere to the Ankle Strategy [51] during weight-shifting task. I plan to conduct a usability study of VBaT system with hemiplegic stroke patients to address the research question on understanding the implication of intelligent adaptive VR-based Balance Training platform augmented with WiiBB on one's weight-shifting capability.

In the second study, I plan to explore the applicability of VR-based Balance Training platform by developing a VR-based CoM-assisted Balance Training (**Virtual CoMBaT**) system. Here, in contrast to the first study, instead of the CoP, I will use one's CoM. This is because one's balance is strongly related to the position and velocity of CoM [28]. In Virtual CoMBaT system, I will use one's personalized CoM, estimated using Kinect sensor while offering individualized VR-based balance exercises. The VR-based tasks will also have varying difficulty levels and the difficulty of the tasks will be decided based on one's individualized performance quantifying one's weight-shifting ability. During a weight-shifting task, the position of a virtual object (VR_{Obj}) in the VR environment will be controlled with CoM excursion. Similar to that in the first study, here, I will also use an HLD unit to monitor whether the Ankle Strategy is being followed or not followed. I plan to conduct a usability study with Virtual CoMBaT system while enrolling hemiplegic participants to address the research questions, such as (i) what is the implication of Virtual CoMBaT system on the balance of post-stroke hemiplegic participants?

and (ii) is it possible to quantify one's residual directional weight-shifting capability based on the performance measures in the tasks offered by Virtual CoMBaT?

In the third study, I plan to design a VR-based Balance Training platform interfaced with two (2) WiiBB (**V2BaT**) augmented with an operant conditioning paradigm. The first two studies (involving VBaT and Virtual CoMBaT) will help us gain insights into the potential of VR-based Balance Training platforms augmented with peripheral devices into improving one's balance. Then following this, using the knowledge acquired in the first two studies, I plan to extend the VR-based system with an operant conditioning paradigm where I will allocate different weightage to the contribution of each leg (Affected and the comparatively healthy that is, Unaffected) of hemiplegic participant interacting with VR-based balance tasks. The idea behind this is to understand whether such a Balance Training system can encourage post-stroke hemiplegic participants to use both their Affected and Unaffected legs as equally as possible while performing a weight-shifting task. For this, I plan to augment the VR-based Balance Training platform with two WiiBB (one for each leg). A usability study with hemiplegic participants is planned to understand (i) whether the system can quantify the contribution of one's individual legs during a weight-shifting task? and (ii) what is the implication of operant conditioning on one's balance?

1.2.3 Objective 3: Investigate the Gaze behavior of hemiplegic post-stroke participants during VR-based Balance Training augmented with operant conditioning paradigm.

During standing balance task, one's gaze behavior can offer important information (as explained in Section 1.1.3). Also, it might be interesting to study one's gaze behavior before and after being exposed to the VR-based Balance Training augmented with operant conditioning. Thus, to achieve the third objective of the research, I plan to extend the V2BaT system augmented with operant conditioning by integrating an Eye Tracker-based gaze monitoring unit with the system. While doing this, I plan to address the issues of Eye Tracker calibration during a standing balance task. For this, I plan to design a head-mounted gaze-sensitive assembly that can record participants' gaze data while they perform weight-shifting tasks offered by the V2BaT system. Through this study, I intend to address two research questions, namely, (i) what is the gaze behavior of hemiplegic post-stroke participants when they perform goal-directed VR-based weight-shifting tasks? and (ii) what are the implications of operant conditioning on the gaze behavior of post-stroke hemiplegic participants?

There is a rich body of literature available on fundamentals of screening and rehabilitation for stroke patients. These fundamentals have been applied in conventional techniques for a long time and have shown promises. Yet, there are some limitations of conventional techniques that can be addressed through the use of technology-assisted systems. Thus, in the next chapter, I will describe in details the motivation and background for developing technology-assisted screening and rehabilitation systems for stroke patients.

This dissertation is organized as follows: In **Chapter 2**, I will be briefly discussing the significance and the background for the present research work. In **chapter 3**, I will present the design of a usability study of SmartEye system to understand the possibility of using one's gaze-related indices as potential biomarkers for screening of probable neurological disorder. The usability study will also focus on validating SmartEye as a user-friendly and easily-accessible gaze monitoring device for patients with stroke. In **Chapter 4,** I will offer the designed VR-based Balance Training (VBaT) system augmented with WiiBB. In this chapter, the focus will be on designing a variety of VR-based tasks of different difficulty levels and also making the VBaT system individualized based on one's performance capability in a task. Additionally, this chapter will present the implication of VBaT system on weight-shifting capability in terms of improvement in one's performance score in the VR-based tasks. In **Chapter 5**, I will present the development of CoM-assisted VR-based Balance Training (Virtual CoMBaT) system where VR-based tasks augmented with Kinect will be studied for its feasibility. Additionally, I will investigate the possibility of using Virtual CoMBaT system to quantify one's direction-specific weight-shifting ability and also the implication of the Virtual CoMBaT system on one's balance. **Chapter 6** will be focused on developing a VR-based Balance Training platform using two WiiBB and augmented with operant conditioning paradigm (V2BaT). This chapter will discuss the steps involved in quantifying the contribution of one's individual legs and also the process of implementing the operant conditioning paradigm by varying the contribution of each leg towards one's balance. The focus will be to conduct usability study with hemiplegic participants to understand the implication of operant conditioning on the usage of both the Affected and Unaffected limbs while maneuvering the virtual object during VR-based weight-shifting tasks. In **Chapter 7**, I will extend this study to investigate gaze behavior of a small sample of hemiplegic stroke participants while they interact with the VR-based tasks offered by the V2BaT (Chapter 6) system. In this chapter, I will also describe the use of in-house fabricated head-mounted gaze-sensitive assembly to monitor the participants' gaze behavior while they perform the VR-based weight-shifting tasks. Finally, **Chapter 8** summarizes the contributions of the present work and describes the scope of future work.

Reference

[1] C. Lott, H. J. Hennes, and W. Dick, "Stroke--a medical emergency.," J. Accid. Emerg. Med., vol. 16, no. 1, pp. 2–7, 1999.

[2] R. Advani, H. Naess, and M. W. Kurz, "The golden hour of acute ischemic stroke,"

Scand. J. Trauma. Resusc. Emerg. Med., vol. 25, no. 1, 2017.

[3] J. L. Saver, “Time is brain - Quantified,” Stroke, vol. 37, no. 1. pp. 263–266, 2006.

[4] G. Gulli, HS. Markus, "The use of FAST and ABCD2 scores in posterior circulation, compared with anterior circulation, stroke and transient ischemic attack," J Neurol Neurosurg Psychiatry, vol. 83, pp. 228-229, 2012.

[5] J. C. Kattah, A. V. Talkad, D. Z. Wang, Y. H. Hsieh, and D. E. Newman-Toker, “HINTS to diagnose stroke in the acute vestibular syndrome: Three-step bedside oculomotor examination more sensitive than early MRI diffusion-weighted imaging,” Stroke, vol. 40, no. 11, pp. 3504–3510, 2009.

[6] T. Brandt, M. Dieterich, and M. Strupp, Vertigo and dizziness: Common complaints, second edition. 2013.

[7] H. P. Adams et al., “Guidelines for the early management of adults with ischemic stroke,” Circulation, vol. 115, no. 20, pp. e478-534, 2007.

[8] J. Joubert et al., “Stroke in rural areas and small communities,” Stroke, vol. 39, no. 6. pp. 1920–1928, 2008.

[9] J. D. Pandian et al., “Public awareness of warning symptoms, risk factors, and treatment of stroke in Northwest India,” Stroke, vol. 36, no. 3, pp. 644–648, 2005.

[10] G. W. Albers et al., “Magnetic resonance imaging profiles predict clinical response to early reperfusion: The diffusion and perfusion imaging evaluation for understanding stroke evolution (DEFUSE) study,” Ann. Neurol., vol. 60, no. 5, pp. 508–517, 2006.

[11] P. A. Barber, A. M. Demchuk, J. Zhang, and A. M. Buchan, “Validity and reliability of a quantitative computed tomography score in predicting outcome of hyperacute stroke before thrombolytic therapy. ASPECTS Study Group. Alberta Stroke Programme Early

CT Score,” Lancet, vol. 355, no. 9216, pp. 1670–1674, 2000.

[12] M. Ward and E. L. Schofield, “Biomarkers for brain disorders,” Therapy, vol. 7, no. 4, pp. 321–336, 2010.

[13] M. Castellanos, R. Leira, J. Serena, J. M. Pumar, I. Lizasoain, J. Castillo, and A. Davalos, “Plasma metalloproteinase-9 concentration predicts hemorrhagic transformation in acute ischemic stroke,” Stroke, vol. 34, no. 1, pp. 40–45, 2003.

[14] V. Kumar, "Eye is the Window to the Brain Pathology." Curr Adv Ophthalmol, vol. 2017, no. 1, pp. 3-4, 2017.

[15] A. Pollock et al., “Interventions for disorders of eye movement in patients with stroke,” in Cochrane Database of Systematic Reviews, 2011.

[16] J. Leigh and D. S. Zee, The neurology of eye movements, vol. 53, no. 9. 2013.

[17] E. Purves D, Augustine GJ, Fitzpatrick D, et al., “Neuroscience,” Sunderl. Sinauer Assoc., 2001.

[18] O. C. Singer, M. C. Humpich, H. Laufs, H. Lanfermann, H. Steinmetz, and T. Neumann-Haefelin, “Conjugate eye deviation in acute stroke: Incidence, hemispheric asymmetry, and lesion pattern,” Stroke, vol. 37, no. 11, pp. 2726–2732, 2006.

[19] C. L. Horak, F. B., and Shupert, “Role of the vestibular system in postural control,” Vestib. Rehabil., vol. 2, pp. 98–113, 1994.

[20] M. Uchiyama and S. Demura, “The role of eye movement in upright postural control,” Sport Sci. Health, vol. 5, no. 1, pp. 21–27, 2009.

[21] N. M. Thomas, T. M. Bampouras, T. Donovan, and S. Dewhurst, “Eye movements affect postural control in young and older females,” Front. Aging Neurosci., vol. 8, no. SEP, 2016.

[22] K. Li and P. A. Malhotra, “Spatial neglect,” Practical Neurology, vol. 15, no. 5. pp. 333–339, 2015.

[23] Wee, Joy YM, and Wilma M. Hopman. "Stroke impairment predictors of discharge function, length of stay, and discharge destination in stroke rehabilitation." American journal of physical medicine & rehabilitation 84.8 (2005): 604-612.

[24] Malhotra, Paresh, Elizabeth Coulthard, and Masud Husain. "Hemispatial neglect, balance and eye-movement control." Current opinion in neurology 19.1 (2006): 14-20.

[25] V. Weerdesteyn, M. de Niet, H. J. R. van Duijnhoven, and A. C. H. Geurts, “Falls in individuals with stroke,” J. Rehabil. Res. Dev., vol. 45, no. 8, pp. 1195–213, 2008.

[26] B. Bobath, "Adult hemiplegia: evaluation and treatment", Elsevier Health Sciences, 1990.

[27] C. B. de Oliveira et al., “Balance control in hemiparetic stroke patients: Main tools for evaluation,” J. Rehabil. Res. Dev., vol. 45, no. 8, pp. 1215–1226, 2008.

[28] L. C. Pereira, A. C. Botelho, and E. F. Martins, “Relationships between body symmetry during weight-bearing and functional reach among chronic hemiparetic patients.,” Rev. Bras. Fisioter., vol. 14, no. 3, pp. 229–266, 2010.

[29] P. Langhorne, F. Coupar, and A. Pollock, “Motor recovery after stroke: a systematic review,” The Lancet Neurology, vol. 8, no. 8. pp. 741–754, 2009.

[30] R. Teasell et al., “Stroke Rehabilitation: An International Perspective,” Top. Stroke Rehabil., vol. 16, no. 1, pp. 44–56, 2009.

[31] K. R. Lohse, C. G. E. Hilderman, K. L. Cheung, S. Tatla, and H. F. M. Van Der Loos, “Virtual reality therapy for adults post-stroke: A systematic review and meta-analysis exploring virtual environments and commercial games in therapy,” PLoS One, vol. 9, no. 3, 2014.

[32] G. Saposnik and M. Levin, “Virtual reality in stroke rehabilitation: A meta-analysis and implications for clinicians,” Stroke, vol. 42, no. 5, pp. 1380–1386, 2011.

[33] B. S. Rajaratnam et al., “Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?,” Rehabil. Res. Pract., vol. 2013, pp. 1–6, 2013.

[34] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer, “Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: A pilot randomized clinical trial in patients with acquired brain injury,” J. Neuroeng. Rehabil., vol. 8, no. 1, 2011.

[35] S. E. Fasoli, H. I. Krebs, and N. Hogan, “Robotic Technology and Stroke Rehabilitation: Translating Research into Practice,” Top. Stroke Rehabil., vol. 11, no. 4, pp. 11–19, 2004.

[36] A. Darekar, B. J. McFadyen, A. Lamontagne, and J. Fung, “Efficacy of virtual reality-based intervention on balance and mobility disorders post-stroke: A scoping review,” J. Neuroeng. Rehabil., vol. 12, no. 1, 2015.

[37] H. Y. Lee, Y. L. Kim, and S. M. Lee, “Effects of virtual reality-based training and task-oriented training on balance performance in stroke patients.,” J. Phys. Ther. Sci., vol. 27, no. 6, pp. 1883–8, 2015.

[38] D. A. Winter, “Human balance and posture control during standing and walking,” Gait Posture, vol. 3, no. 4, pp. 193–214, 1995.

[39] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, “Stiffness Control of Balance in Quiet Standing,” J. Neurophysiol., vol. 80, no. 3, pp. 1211–1221, 1998.

[40] A. Gonzalez, M. Hayashibe, and P. Fraisse, “Estimation of the center of mass with Kinect and Wii balance board,” in IEEE International Conference on Intelligent Robots and Systems, 2012, pp. 1023–1028.

[41] G. I. Yatar and S. A. Yildirim, “Wii Fit balance training or progressive balance training in patients with chronic stroke: a randomised controlled trial,” J. Phys. Ther. Sci., vol. 27, no. 4, pp. 1145–1151, 2015.

[42] A. Da Gama, P. Fallavollita, V. Teichrieb, and N. Navab, “Motor Rehabilitation Using Kinect: A Systematic Review,” Games Health J., vol. 4, no. 2, pp. 123–135, 2015.

[43] G. U. Jung, T. H. Moon, G. W. Park, J. Y. Lee, and B. H. Lee, “Use of augmented reality-based training with EMG-triggered functional electric stimulation in stroke rehabilitation,” J. Phys. Ther. Sci., vol. 25, no. 2, pp. 147–151, 2013.

[44] M. Land, N. Mennie, and J. Rusted, “The roles of vision and eye movements in the control of activities of daily living,” Perception, vol. 28, no. 11, pp. 1311–1328, 1999.

[45] A. K. Rao and A. M. Gordon, “Contribution of tactile information to accuracy in pointing movements,” Exp. Brain Res., vol. 138, no. 4, pp. 438–445, 2001.

[46] M. a. Hollands, D. E. Marple-Horvat, S. Henkes, and a. K. Rowan, “Human Eye Movements During Visually Guided Stepping.,” J. Mot. Behav., vol. 27, no. 2, pp. 155–163, 1995.

[47] M. A. Hollands and D. E. Marple-Horvat, “Coordination of Eye and Leg Movements During Visually Guided Stepping,” J. Mot. Behav., vol. 33, no. 2, pp. 205–216, 2001.

[48] K. A. Crowdy, M. A. Hollands, I. T. Ferguson, and D. E. Marple-Horvat, “Evidence for interactive locomotor and oculomotor deficits cerebellar patients during visually guided stepping,” Exp. Brain Res., vol. 135, no. 4, pp. 437–454, 2000.

[49] T. Paillard, C. Lafont, M. C. Costes-Salon, D. Rivière, and P. Dupui, “Effects of brisk walking on static and dynamic balance, locomotion, body composition, and aerobic capacity in ageing healthy active men,” Int. J. Sports Med., vol. 25, no. 7, pp. 539–546,

2004.

[50] R. Hammami, D. G. Behm, M. Chtara, A. Ben Othman, and A. Chaouachi, “Comparison of static balance and the role of vision in elite athletes,” J. Hum. Kinet., vol. 41, no. 1, pp. 33–41, 2014.

[51] Y. Lee, J. G. Her, Y. Choi, and H. Kim, “Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors’ Names,” J. Phys. Ther. Sci., vol. 26, no. 2, pp. 179–182, 2014.

CHAPTER 2

SIGNIFICANCE AND BACKGROUND

2.1 Significance

Stroke, also known as a cerebrovascular accident, occurs when one's brain is deprived of blood supply either by a blockage of blood vessels due to clot (ischemic stroke) or when the blood vessel rupturs (hemorrhagic stroke) due to high blood pressure, aging blood vessels and other reasons [1]. Stroke is the second most common cause of death and fourth leading cause of disability worldwide [2]. India suffers from an epidemic of neurological disability [3] with a prevalence rate of stroke being 84-262/100,000 in rural and 334-424/100,000 in urban areas [4], [5]. Further with aging population, the risk of stroke increases with reports on prevalence rates being 27-34/100,000 in the 35-44 age group to 822-1116/100,000 in the 75+ age group [6], [7].

Researchers have shown that acute stroke management can be facilitated if treatments are given within a short time window post onset of stroke symptom [8]. Therefore, early screening of stroke cases is critical. Given the criticality of stroke screening, expert clinicians in well-established urban healthcare centers often use conventional and technology-assisted platforms. In conventional stroke screening, expert clinicians often employ manual techniques such as Face Arm Speech Test (FAST) [9], Opto-kinetic drum [10], Head Impulse test, Nystagmus and Test of Skew deviation (HINTS) [11] to name a few. In technology-assisted screening, use of Magnetic Resonance Imaging (MRI) and computed tomography (CT) are common. The conventional techniques though powerful, are often subjective and depend on the clinician's expertise [4], [12]. Again, the above-mentioned technology-assisted screening platforms, though provide quantitative estimates of one's probable neurological disorder, are very costly and often inaccessible to the common man. Given these limitations, it is critical to have simple-to-use, cost-effective and user-friendly device that can be used in home-based, community-based and hospital-based settings to screen for possible neurological disorder.

Individuals with neurological disorder often succumb to deficits in motor control. Literature review indicates that more than 50% of stroke survivors suffer from neurological impairments that can manifest itself as deficits in motor control [13]. About 80% of these patients lack control over movement and coordination skill required for balance that can adversely affect one's mobility [2], [13]. This is often accompanied with loss of muscle strength, endurance and coordination causing post-stroke patients to be either bed-ridden or with restricted mobility due to impaired balance. About 85% of stroke survivors are often reported to suffer from total or partial loss of function of body parts, usually a limb or limbs causing quadriplegia or hemiplegia [14], [15].

The hemiplegic patients often show impaired balance due to asymmetric body posture thereby increasing their risk of falls [16], [17]. The incidence of fall has been reported to be up to 73% in the first year of post-stroke [18]. Falls may have severe consequences both physically and psychologically that may lead to decreased physical activity, social deprivation with eventual loss of independence [16]. Literature review indicates that the after-effects of falls can be lacerations, hip fractures and head traumas often seen in around 20 to 30% of such individuals [19], [20]. Although the cause of falls is multifactorial, post-stroke balance-related impairment represents one of the largest contributing factors to falls in stroke survivors [21], [22], [23].

To address balance deficits, patients are exposed to rehabilitation. The conventional balance rehabilitation techniques being repetitive and without variations often leads to poor engagement and reduced interest in stroke patients while performing balance rehabilitation exercises [24], [25]. This technique also suffer from the need to have an always-present therapist that calls for one-to-one services which is often problematic given the scarcity of adequately trained clinicians [26], high cost for specialized healthcare settings [27] and other factors.

Given these limitations, researchers have started to explore the possibility of using alternate technology-assisted solutions. Many studies have investigated the application of various technology-assisted balance rehabilitation systems, such as robot-assisted, computer-based and Virtual Reality (VR) based rehabilitation systems to improve one's balance [25], [28], [29]. Researchers have been using computer-based tasks augmented with peripheral devices such as force platform and motion capture device that provide the inertial characteristic of body segments related to balance (such as Center of Pressure (CoP) [26], [30] and Center of Mass (CoM)) [31], [32]. However, most of the existing systems used to address balance issues have off-the-shelf games that are designed with an entertainment perspective rather than from rehabilitation perspective. Also, the balance tasks offered to the participants are not adaptive to one's performance quantifying one’s balance which is a critical requirement for individualized rehabilitation. This individualization is important since rehabilitative interventions must take into account the spectrum of residual abilities of different patients due to the heterogeneity of mechanisms underlying motor recovery [33]. Thus, it is critical to develop a technology-assisted balance rehabilitation system that can understand the patient's residual balance ability and accordingly plan out the balance training tasks for the post-stroke patients in an individualized manner.

2.2 Background

2.2.1 Gaze-based Screening Techniques for Neurological Disorder

In conventional stroke screening, clinicians often use observation-based methods for picking up different symptoms of stroke that also includes one’s gaze-related anomalies. The gaze-related indices can be potential indicators of stroke with reports showing that about 86% of stroke patients have some form of oculomotor disorder [34]. Given the potential of possible connectivity between stroke and oculomotor signatures, clinicians often use simple gadgets and follow clinical steps to identify one's oculomotor abnormalities. For example, Opto-kinetic drum [10] is used to observe the patient's eye movement for visual examination of one's optokinetic response that is a combination of a slow-phase and fast-phase eye movement. The outer surface of Opto-kinetic drum is striped with alternate black and white colored bars. The participants are asked to seat while facing the drum. As the drum is rotated by the clinician, the participants' eyes are subject to a moving visual field while the participants remain stationary. During this process, a clinician manually looks out for any probable abnormality in eye movements. Also, clinicians use a three-component bedside oculomotor examination, namely, Head Impulse test, Nystagmus and Test of Skew deviation (HINTS) that has been shown to diagnose stroke in the case of acute vestibular syndrome with more accuracy than diffusion-weighted magnetic resonance imaging (MRI) [35]. Unfortunately, administration of such tests requires expert clinicians who are usually not readily available, particularly in rural healthcare centers. Also, the observation-based evaluations depend on the clinician's expertise and can be subjective in nature. Hence, a quantitative, autonomous, user-friendly oculomotor screening device that needs limited specialized knowledge to operate can play an important role in the screening of one's probable neurological disorder.

Initiation of accurate and coordinate eye movements is triggered and controlled by a large network of brain areas. There are four types of eye movements namely, saccades, smooth pursuit, vergence, and vestibulo-ocular movement [36]. These eye movements are controlled by different extraocular muscles that in turn, are innervated by various cranial nerves. For example, superior oblique is innervated by cranial nerve IV; lateral rectus extraocular muscle is innervated by cranial nerve VI. Other extraocular muscles, such as inferior oblique superior rectus, inferior rectus and medial rectus muscles are innervated by cranial nerve III [36]. These muscles are responsible for horizontal, vertical and torsional movements of one's eyes. These ocular movements lead to voluntary conjugate horizontal gaze (looking side-to-side), voluntary conjugate vertical gaze (looking up and down), smooth tracking of objects, convergence and eye movements associated with head movement [37], [38]. All the ocular movements that are produced by the central nervous system are conjugate (that is, both eyes moving in the same direction in order to keep the eyes focused on a target) except for convergence, which adducts the eyes to focus on near objects [37], [38]. Voluntary horizontal gaze in one direction originates in the contralateral frontal eye fields (located in the premotor cortex of the frontal lobe). This region has upper motor neurons that project to the contralateral paramedian pontine reticular formation (PPRF), which is the organizing center for lateral gaze in the brain stem. The PPRF projects to the ipsilateral abducens nucleus (causing eye abduction on the ipsilateral side). Such distributed nature of oculomotor system often makes it vulnerable to various neurological disorders marked by unique clinical patterns [37], [38].

For example, if a person has a lesion in his frontal lobe, his eye fixation has been reported to be more likely shifted to the ipsilateral side of the lesion. In contrast, if the lesion occurs in the PPRF, then the eye fixation has been reported to shift towards the contralateral side of the lesion [36]. In turn, one gaze fixation pattern can get adversely affected due to stroke. Also, one's smooth pursuit of the eye while following a target can be adversely affected by lesion in the higher cortical centers of one's brain [36]. Hence, any neurological disorder associated with brain-related abnormality can have implications on one's eye movement causing oculomotor signatures to serve as important biomarkers for screening cases with probable neurological disorders.

2.2.2 Balance Rehabilitation Techniques

Added to post-stroke vision problems, post-stroke patients often report balance deficits. This is because, one's vision system provides important sensory information to the human balance system via different eye movements [39]. So, any disruption to the vision system can lead to balance deficits. For assessment of one's residual balance, therapists use different clinical measures, such as Berg Balance Score (BBS) scale, Brunel Balance Assessment (BBA), Fugl-Meyer Assessment of sensorimotor function [40], [41], [42] and other types of assessment. Based on the clinical measures, clinicians recommend different rehabilitation exercises for the patients. The rehabilitation involved in conventional methods require regular visit of patients to health centers followed by one-to-one repetitive therapy sessions lasting over several days. The conventional rehabilitation exercises, though powerful often suffer from certain limitations on account of low doctor:patient ratio, monotonicity with repetitive training/exercises stealing away patients' motivation [43]. Thus, technology-assisted balance training platforms can serve as an alternative while providing quantitative, intensive, motivating and individualized balance training platform with variations.

Literature have reported various technology-assisted balance rehabilitation systems such as robot-assisted therapy, computer-based balance rehabilitation system, virtual reality (VR) based rehabilitation system [25], [28], [29]. The outcomes of these studies suggest that for people with long-term ill-health conditions such as those with stroke, technology-assisted rehabilitation systems can provide a feasible method for monitoring their condition as well as offer therapeutic guidance to alter maladaptive behavior. Among the different technology-assisted platforms used for rehabilitation, VR-based rehabilitation has been widely used by researchers [24], [25], [26], [29].

This is because VR-based systems offer the flexibility of design, controllability, safety [44]. Using VR for rehabilitation allows one to create a synthetic environment with precise control over a large number of task parameters that can influence one's behavior in an individualized manner. Additionally, VR-based platform can be easily integrated with peripheral devices such as data acquisition boards, balance boards, Kinect and other devices thereby offering an avenue to simultaneously record one's real-time physiological (such as one's muscle activation, pulse rate, skin temperature and others) and kinematic responses (such as body CoP and CoM) [45]. Also, the VR-based engine can be programmed to offer quantitative feedback while one performs a task set in VR environment. Further, VR-based environment has been shown to offer an interactive and enjoyable medium with promise to improve the motor function in adults with stroke [46]. The flexibility provided by the VR-based programming platform can provide a designer with an option of manipulating the number, speed or order of stimulus presentation effortlessly [47]. The inherent flexibility of VR-based system allows a designer to incorporate incremental variations in task difficulty so as to challenge a participant and scaffold the development of participant's skill such as balance in a precise, objective and quantitative manner for individualized treatment.

In literature, the existing technology-assisted balance rehabilitation platforms use different approaches, such as CoP and CoM [26], [30], [31], [32]. This is because, one's balance is a multidimensional concept with no universally accepted way of defining or measuring it [48]. A general approach to describe the dimension of balance can be through one's ability to maintain position, incorporate postural adjustment to voluntary movement and react to external disturbance [48]. Specifically, balance is related with the inertial forces acting on one's body mass and the inertial characteristics of body segments [49] quantified through CoP and CoM-based indices. The CoP is a point location on the base of support on which the vertical ground reaction force acts. The CoM is a point where the total body mass is concentrated in the global reference system and is a weighted average of the CoM of each body segment in 3D space. The CoP position can be measured using force platform data [50], such as WiiBB and the CoM position can be estimated by using motion capture systems such as Vicon, Kinect and other devices [51]. Therefore, the VR-based systems augmented with force platform and/or motion capture device can be used to provide real-time visual feedback of one's CoP/CoM excursion while a participant performs weight-shifting tasks while standing on the force platform and in the field of view of the motion capture device. Various researchers have used one's CoP and/or CoM to interact with the computer-based games for balance rehabilitation. For example, Rajaratnam et al. [24] evaluated the positive contribution of VR-based games on balance rehabilitation. In this study, they offered off-the-shelf games available with either a Nintendo Wii-Fit or Microsoft Kinect game console. Results of this study indicate that the inclusion of interactive VR-based balance-related games can lead to improved functional mobility and balance. Researchers (Cho et al. [52]) have used VR-based balance training within a video-game environment with an aim to improve balance in stroke patients who were in chronic phase. They have used a Wii Balance Board to interact with off-the-shelf games such as balance bubble, ski slalom and others. Though the offered games were of varying difficulty levels, yet the order in which the games were offered (that is game switching) was not individualized. Again, Gil-Gómez et al. [25] designed a VR-based system integrated with WiiBB (eBaVir) for balance rehabilitation. Though the eBaVir system offered three different games to the stroke participants in each session, yet their system was not individualized to the participant’s ability.

In short, the existing VR-based balance rehabilitation systems [25], [26], [30], [31] (i) used off-the-shelf games that were designed for entertainment purpose instead of balance rehabilitation and (ii) did not have any rule engine that could offer balance training tasks of varying difficulty in a systematic, controlled and adaptive manner depending on the user's performance, essential for effective balance rehabilitation. Additionally, none of the existing balance rehabilitation systems have used mechanisms that can monitor whether a patient performing balance tasks was following any strategy, critical for postural stability. Specifically, none have used any of the three postural strategies, namely, Ankle, Hip and Step strategies [53] essential for maintaining stable posture during balance training. There is evidence from literature that among the three strategies, the Ankle strategy that use the contraction of the muscle associated with one's ankle joint, is most commonly used to improve one's standing balance [54].

Given the advantages offered by VR-based platform for rehabilitation and the limitations of the existing balance rehabilitation systems, it is critical to develop an intelligent VR-based balance training platform that can be individualized and adaptive to the balance capabilities of post-stroke hemiplegic patients. This platform augmented with cost-effective peripheral devices such as WiiBB (force platform) and/or Kinect device (motion capture device) should be able to quantify one's residual balance capability and accordingly adjust the challenge level of VR-based balance tasks for individualized balance rehabilitation. Also, such a system should have a mechanism to encourage the patients to follow Ankle strategy while performing the standing balance tasks so as to improve the quality of weight-shifting. Such a VR-based balance training system must provide quantitative estimates of one's improvement in balance without necessitating any specialized knowledge for it to be operated, thereby enabling it to be used in home-based, community-based and hospital-based settings.

Added to balance, understanding one's gaze behavior in a goal-oriented task is important. This is because, for maintaining balance, one needs to decipher sensory information obtained from vision in addition to that from vestibular and somatosensory systems [39], [55]. The vision system is an integral part of human balance and provides important information on one’s position, posture while operating in an environment [56]. One's vision is aided via various eye movements. Patients with stroke often exhibit deviations from normative eye movement patterns that can be captured by measuring the anomalies in oculomotor signatures (Section 2.2.1). Thus, I wanted to understand the gaze behavior of post-stroke participants performing VR-based goal-directed balance tasks.

2.2.3 Gaze Behavior of Participants while interacting with Computer based Tasks

Since this study involved use of limbs while participating in VR-based tasks, I was interested to understand the gaze behavior of post-stroke participants suffering from balance deficits. One way to achieve this can be to monitor one's gaze pattern while he/she interacted with VR-based balance tasks. This is because, researchers have reported that individuals move their eyes in different ways followed by fixation and one’s gaze behavior can be understood in the context of a particular task [57], [58], [59]. Various researchers have studied gaze behavior of individuals with neurological disorders during goal-directed visual tasks. For example, Rizzo et al. [60] have studied the gaze behavior in patients with chronic cerebral injury and reported that the spatial and temporal relationships between the eye and hand are disrupted in stroke survivors. Additionally, there are research studies that have looked into the gaze behavior of both healthy [61], [62] and post-stroke [63], [64] participants while interacting with computer-based tasks that require the use of one’s upper limbs.

Though there are a body of literature on one's gaze behavior in relation to the activities requiring upper limbs, yet, studies of one's gaze behavior during standing balance tasks are limited and therefore require further exploration. There are few studies to my knowledge that looks into one's gaze behavior during standing balance tasks. For example, Uchiyama et al. [65] have studied the role of fixation on postural stabilization while individuals were asked to maintain upright standing posture in a dark room. These authors reported that in contrast to the individuals not provided with any visual cue, the individuals provided with the visual cue in terms of a target position to be fixated, exhibited reduced body sway velocity (inferring greater control over balance) compared to the former. Dutta et al. [66] have explored the gaze behavior of young healthy participants in a visuomotor balance task. The task required a participant to stand on a WiiBB and maneuver a cursor on a computer screen from an initial central position to fixed peripheral target positions presented on the screen. The position of the cursor on the screen was controlled by the CoP excursion of the participant performing weight-shifting while standing on the WiiBB. The result obtained from this study suggests that the ratio of fixation duration (that is total time spent in fixating on a region of interest) towards the static target to that on the moving cursor increased with increase in performance score. These researchers have investigated one's gaze behavior during postural control in quiet standing task [65] and visuomotor balance task [67]. However, these observations have been reported only for healthy individuals. Given the fact that patients with neurological disorder exhibit deficits in oculomotor signatures [36] and also suffer from balance deficits [21], detailed investigation into their gaze behavior when subjected to VR-based balance tasks is warranted. Motivated by this, I plan to investigate the gaze behavior of post-stroke participants while they performed VR-based balance training tasks through weight-shifting.

The primary goal of my research was to carry out gaze-based stroke screening and post-stroke balance rehabilitation by utilizing the rapidly growing technology. While addressing the issue of stroke screening, I designed a cost-effective, easy-to-use, quantitative gaze-based screening device (SmartEye) that can objectively measure one's gaze-related indices when exposed to a computer-based visual task. The idea was to identify potential gaze-based biomarkers of one's probable neurological disorder. While addressing the post-stroke balance deficits, I developed intelligent adaptive and individualized VR-based balance training systems augmented with peripheral devices such as Kinect and WiiBB. The idea was to understand the applicability of my VR-based balance training platforms to improve balance of post-stroke hemiplegic patients by offering them VR-based balance tasks of varying challenges in an adaptive and individualized manner.

In the next chapter, I will present the design and development of a cost-effective, easy-to-use, easily-accessible, quantitative gaze-based screening device (SmartEye) for identifying one's probable neurological disorder. I will evaluate the possibility of using gaze-based features as potential biomarkers of the stroke condition by designing a usability study with SmartEye in which stroke survivors and age-matched healthy counterparts participated.

Reference

[1] "National Stroke Association. The Complete Guide to Stroke.," 2003. [Online]. Available: http://www.stroke.org/site/DocServer/NSA_complete_guide.pdf?docID=341. [Accessed: 13-Jan-2018].

[2] C. J. L. Murray and A. D. Lopez, "Alternative projections of mortality and disability by cause 1990-2020: Global Burden of Disease Study," *Lancet*, vol. 349, no. 9064, pp. 1498–1504, 1997.

[3] A. Das, A. L. Botticello, G. R. Wylie, and K. Radhakrishnan, "Neurologic Disability: A Hidden Epidemic for India," *Neurology*, vol. 79, no. 21, pp. 2146–2147, 2012.

[4] J. D. Pandian, A. Jaison, S. S. Deepak, G. Kalra, S. Shamsher, D. J. Lincoln, and G.

Abraham, “Public awareness of warning symptoms, risk factors, and treatment of stroke in Northwest India,” *Stroke*, vol. 36, no. 3, pp. 644–648, 2005.

[5] V. L. Feigin, C. M. Lawes, D. A. Bennett, S. L. Barker-Collo, and V. Parag, “Worldwide stroke incidence and early case fatality reported in 56 population-based studies: a systematic review,” *The Lancet Neurology*, vol. 8, no. 4. pp. 355–369, 2009.

[6] P. M. Dalal et al., “Population-based stroke survey in Mumbai, India: Incidence and 28-day case fatality,” *Neuroepidemiology*, vol. 31, no. 4, pp. 254–261, 2008.

[7] S. E. Sridharan et al., “Incidence, types, risk factors, and outcome of stroke in a developing country the trivandrum stroke registry,” *Stroke*, vol. 40, no. 4, pp. 1212–1218, 2009.

[8] R. Advani, H. Naess, and M. W. Kurz, “The golden hour of acute ischemic stroke,” *Scand. J. Trauma. Resusc. Emerg. Med.*, vol. 25, no. 1, 2017.

[9] T. G. Robinson, A. Reid, V. J. Haunton, A. Wilson, and A. R. Naylor, “The face arm speech test: Does it encourage rapid recognition of important stroke warning symptoms?,” Emerg. Med. J., vol. 30, no. 6, pp. 467–471, 2013.

[10] T. Brandt, M. Dieterich, and M. Strupp, *Vertigo and dizziness: Common complaints, second edition*. 2013.

[11] J. C. Kattah, A. V. Talkad, D. Z. Wang, Y. H. Hsieh, and D. E. Newman-Toker, “HINTS to diagnose stroke in the acute vestibular syndrome: Three-step bedside oculomotor examination more sensitive than early MRI diffusion-weighted imaging,” *Stroke*, vol. 40, no. 11, pp. 3504–3510, 2009.

[12] J. Joubert et al., “Stroke in rural areas and small communities,” *Stroke*, vol. 39, no. 6. pp. 1920–1928, 2008.

[13] K. Strong, C. Mathers, and R. Bonita, "Preventing stroke: saving lives around the world," *Lancet Neurology*, vol. 6, no. 2. pp. 182–187, 2007.

[14] S. B. O'Sullivan and T. J. Schmitz, *Physical Rehabilitation*. 2007.

[15] D. S. Nichols-Larsen, P. C. Clark, A. Zeringue, A. Greenspan, and S. Blanton, "Factors influencing stroke survivors' quality of life during subacute recovery," *Stroke*, vol. 36, no. 7, pp. 1480–1484, 2005.

[16] V. Weerdesteyn, M. de Niet, H. J. R. van Duijnhoven, and A. C. H. Geurts, "Falls in individuals with stroke," *J. Rehabil. Res. Dev.*, vol. 45, no. 8, pp. 1195–213, 2008.

[17] L. C. Pereira, A. C. Botelho, and E. F. Martins, "Relationships between body symmetry during weight-bearing and functional reach among chronic hemiparetic patients.," *Rev. Bras. Fisioter.*, vol. 14, no. 3, pp. 229–266, 2010.

[18] G. S. A. F. Verheyden et al., "Interventions for preventing falls in people after stroke.," *Cochrane database Syst. Rev.*, no. 5, p. CD008728, 2013.

[19] B. H. Alexander, F. P. Rivara, and M. E. Wolf, "The cost and frequency of hospitalization for fall-related injuries in older adults," *Am. J. Public Health*, vol. 82, no. 7, pp. 1020–1023, 1992.

[20] A. Forster and J. Young, "Incidence and consequences of falls due to stroke: a systematic inquiry.," *BMJ*, vol. 311, no. 6997, pp. 83–6, 1995.

[21] S. F. Mackintosh, K. D. Hill, K. J. Dodd, P. A. Goldie, and E. G. Culham, "Balance Score and a History of Falls in Hospital Predict Recurrent Falls in the 6 Months Following Stroke Rehabilitation," *Arch. Phys. Med. Rehabil.*, vol. 87, no. 12, pp. 1583–1589, 2006.

[22] K. Hill, P. Ellis, J. Bernhardt, P. Maggs, and S. Hull, "Balance and mobility outcomes for stroke patients: A comprehensive audit," *Aust. J. Physiother.*, vol. 43, no. 3, pp. 173–180,

1997.

[23] P. T. Cheng, M. Y. Liaw, M. K. Wong, F. T. Tang, M. Y. Lee, and P. S. Lin, “The sit-to-stand movement in stroke patients and its correlation with falling,” *Arch. Phys. Med. Rehabil.*, vol. 79, no. 9, pp. 1043–1046, 1998.

[24] B. S. Rajaratnam et al, “Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?,” *Rehabil. Res. Pract.*, vol. 2013, pp. 1–6, 2013.

[25] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer, “Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: A pilot randomized clinical trial in patients with acquired brain injury,” *J. Neuroeng. Rehabil.*, vol. 8, no. 1, 2011.

[26] G. I. Yatar and S. A. Yildirim, “Wii Fit balance training or progressive balance training in patients with chronic stroke: a randomised controlled trial,” *J. Phys. Ther. Sci.*, vol. 27, no. 4, pp. 1145–1151, 2015.

[27] H. Bateni, “Changes in balance in older adults based on use of physical therapy vs the Wii Fit gaming system: A preliminary study,” *Physiother. (United Kingdom)*, vol. 98, no. 3, pp. 211–216, 2012.

[28] S. E. Fasoli, H. I. Krebs, and N. Hogan, “Robotic Technology and Stroke Rehabilitation: Translating Research into Practice,” *Top. Stroke Rehabil.*, vol. 11, no. 4, pp. 11–19, 2004.

[29] G. Saposnik and M. Levin, “Virtual reality in stroke rehabilitation: A meta-analysis and implications for clinicians,” *Stroke*, vol. 42, no. 5, pp. 1380–1386, 2011.

[30] B. Lange, S. Flynn, R. Proffitt, C.-Y. Chang, and A. “Skip” Rizzo, “Development of an Interactive Game-Based Rehabilitation Tool for Dynamic Balance Training,” *Top. Stroke Rehabil.*, vol. 17, no. 5, pp. 345–352, 2010.

[31] Y. C. Pai and J. Patton, “Center of mass velocity-position predictions for balance control,” *J. Biomech.*, vol. 30, no. 4, pp. 347–354, 1997.

[32] B. Lange, C. Y. Chang, E. Suma, B. Newman, A. S. Rizzo, and M. Bolas, “Development and evaluation of low cost game-based balance rehabilitation tool using the Microsoft Kinect sensor,” in *IEEE Engineering in Medicine and Biology Conference*, 2011, vol. 2011, pp. 1831–1834.

[33] N. Takeuchi and S. I. Izumi, “Rehabilitation with poststroke motor recovery: A review with a focus on neural plasticity,” *Stroke Research and Treatment*. 2013.

[34] A. Pollock et al., “Interventions for disorders of eye movement in patients with stroke,” in *Cochrane Database of Systematic Reviews*, 2011.

[35] J. C. Kattah, A. V Talkad, D. Z. Wang, Y.-H. Hsieh, and D. E. Newman-Toker, “HINTS to Diagnose Stroke in the Acute Vestibular Syndrome,” *Str*, vol. 40, no. 11, pp. 3504–3510, 2009.

[36] J. Leigh and D. S. Zee, *The neurology of eye movements*, vol. 53, no. 9. 2013.

[37] P. Dale, G. J. Augustine, D. Fitzpatric, L. Katz, A.-S. LaMantia, J. O. McNamara, and S. A. Williams, *Types of eye movements and their functions*. 2001.

[38] R. S. Reeves, Alexander G. Swenson, “Chapter 4 - Extraocular movement,” *Dartmounth Medical School*. [Online]. Available: https://www.dartmouth.edu/~dons/part_1/chapter_4.html. [Accessed: 13-Jan-2018].

[39] R. J. Peterka and M. S. Benolken, “Role of somatosensory and vestibular cues in attenuating visually induced human postural sway,” *Exp. Brain Res.*, vol. 105, no. 1, pp. 101–110, 1995.

[40] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, “Measuring balance in the

elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[41] S. F. Tyson and L. H. DeSouza, "Development of the Brunel Balance Assessment: A new measure of balance disability post stroke," *Clin. Rehabil.*, vol. 18, no. 7, pp. 801–810, 2004.

[42] D. J. Gladstone, C. J. Danells, and S. E. Black, "The Fugl-Meyer Assessment of Motor Recovery after Stroke: A Critical Review of Its Measurement Properties," *Neurorehabilitation and Neural Repair*, vol. 16, no. 3. pp. 232–240, 2002.

[43] K. R. Lohse, C. G. E. Hilderman, K. L. Cheung, S. Tatla, and H. F. M. Van Der Loos, "Virtual reality therapy for adults post-stroke: A systematic review and meta-analysis exploring virtual environments and commercial games in therapy," *PLoS One*, vol. 9, no. 3, 2014.

[44] A. Darekar, B. J. McFadyen, A. Lamontagne, and J. Fung, "Efficacy of virtual reality-based intervention on balance and mobility disorders post-stroke: A scoping review," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, 2015.

[45] R. V. Kenyon, J. Leigh, and E. A. Keshner, "Considerations for the future development of virtual technology as a rehabilitation tool," *J. Neuroeng. Rehabil.*, vol. 1, 2004.

[46] J. H. Crosbie, S. Lennon, M. C. Mcgoldrick, M. D. J. Mcneill, J. W. Burke, and S. M. Mcdonough, "Virtual reality in the rehabilitation of the upper limb after hemiplegic stroke: a randomised pilot study," *Screen*, pp. 229–235, 2008.

[47] M. T. Schultheis and A. A. Rizzo, "The application of virtual reality technology in rehabilitation," *Rehabilitation Psychology*, vol. 46, no. 3. pp. 296–311, 2001.

[48] K. Berg, "Balance and its measure in the elderly: A review," Physiother. Canada, vol. 41, no. 5, pp. 240–246, 1989.

[49] D. A. Winter, "Human balance and posture control during standing and walking," *Gait Posture*, vol. 3, no. 4, pp. 193–214, 1995.

[50] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, "Stiffness Control of Balance in Quiet Standing," *J. Neurophysiol.*, vol. 80, no. 3, pp. 1211–1221, 1998.

[51] A. Gonzalez, M. Hayashibe, and P. Fraisse, "Estimation of the center of mass with Kinect and Wii balance board," in *IEEE International Conference on Intelligent Robots and Systems*, 2012, pp. 1023–1028.

[52] K. H. Cho, K. J. Lee, and C. H. Song, "Virtual-Reality Balance Training with a Video-Game System Improves Dynamic Balance in Chronic Stroke Patients," *Tohoku J. Exp. Med.*, vol. 228, no. 1, pp. 69–74, 2012.

[53] H. Y. Lee, Y. L. Kim, and S. M. Lee, "Effects of virtual reality-based training and task-oriented training on balance performance in stroke patients.," *J. Phys. Ther. Sci.*, vol. 27, no. 6, pp. 1883–8, 2015.

[54] Y. Lee, J. G. Her, Y. Choi, and H. Kim, "Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors' Names," *J. Phys. Ther. Sci.*, vol. 26, no. 2, pp. 179–182, 2014.

[55] "The human balance system." [Online]. Available: http://vestibular.org/understanding-vestibular-disorder/human-balance-system. [Accessed: 13-Jan-2018].

[56] C. L. Horak, F. B., and Shupert, "Role of the vestibular system in postural control," *Vestib. Rehabil.*, vol. 2, pp. 98–113, 1994.

[57] M. F. Land and M. Hayhoe, "In what ways do eye movements contribute to everyday activities?," in *Vision Research*, 2001, vol. 41, no. 25–26, pp. 3559–3565.

[58] M. Land, N. Mennie, and J. Rusted, “The roles of vision and eye movements in the control of activities of daily living,” *Perception*, vol. 28, no. 11, pp. 1311–1328, 1999.

[59] M. M. Hayhoe, A. Shrivastava, R. Mruczek, and J. B. Pelz, “Visual memory and motor planning in a natural task,” *J. Vis.*, vol. 3, no. 1, p. 6, 2003.

[60] J. R. Rizzo et al., “Eye control deficits coupled to hand control deficits: Eye-hand incoordination in chronic cerebral injury,” *Front. Neurol.*, vol. 8, no. JUL, 2017.

[61] U. Sailer, “Eye-Hand Coordination during Learning of a Novel Visuomotor Task,” *J. Neurosci.*, vol. 25, no. 39, pp. 8833–8842, 2005.

[62] R. S. Johansson, G. Westling, A. Bäckström, and J. R. Flanagan, “Eye–Hand Coordination in Object Manipulation,” *J. Neurosci.*, vol. 21, no. 17, pp. 6917–6932, 2001.

[63] K. L. Gao, S. S. M. Ng, J. W. Y. Kwok, R. T. K. Chow, and W. W. N. Tsang, “Eye-hand coordination and its relationship with sensorimotor impairments in stroke survivors,” *J. Rehabil. Med.*, vol. 42, no. 4, pp. 368–373, 2010.

[64] K. Caeyenberghs et al., “Static and dynamic visuomotor task performance in children with acquired brain injury: Predictive control deficits under increased temporal pressure,” *J. Head Trauma Rehabil.*, vol. 24, no. 5, pp. 363–373, 2009.

[65] M. Uchiyama and S. Demura, “Low visual acuity is associated with the decrease in postural sway.,” *Tohoku J. Exp. Med.*, vol. 216, no. 3, pp. 277–85, 2008.

[66] A. Dutta, U. Lahiri, A. Das, M. A. Nitsche, and D. Guiraud, “Post-stroke balance rehabilitation under multi-level electrotherapy: A conceptual review,” *Front. Neurosci.*, vol. 8, no. DEC, pp. 1–18, 2014.

[67] A. Dutta, D. Kumar, U. Lahiri, A. Das, and M. V Padma, “Post-stroke Engagement-sensitive Balance Rehabilitation Under An Adaptive Multi-level Electrotherapy : Clinical

Hypothesis and Computational Framework," Neuroscience and Biomedical Engineering, vol. 2, no. 2, pp. 68–80, 2014.

CHAPTER 3

EYE TRACKING SYSTEM FOR QUANTITATIVE ASSESSMENT OF OCULOMOTOR ABNORMALITIES

3.1 Introduction

This chapter presents the design and development of a technology-assisted device for screening cases with neurological disorder using oculomotor signature. Accurate and precise eye movements are continuous and vital part of our sensory perception. To initiate accurate and co-ordinate ocular movements, a vast network of brain areas from low-level visual processing to motor control of gaze orientation are involved. Eye movement can be of four different types, namely, saccades, smooth pursuit, vergence, and vestibulo-ocular movement [1]. The different eye movements are controlled by different extraocular muscles that are innervated by various cranial nerves. For example, the extraocular muscles innervated by Cranial oculomotor (CNIII), Trochlear (CNIV) and Abducens (CNVI) nerves control one's eye movement. For normal viewing, one needs to have synchronized conjugate eye movement. Conjugate eye movement is effected by the third, fourth and sixth brainstem nuclei controlling the CNIII, CNIV and CNVI. The smooth pursuit and saccadic eye movements are controlled by cerebral cortex, mid brain and vestibular nuclei of brain [1], [2]. The distributed nature of oculomotor system often makes it vulnerable to various neurological disorders marked by unique clinical patterns [2]. For example, if a person has lesion in his frontal lobe, his eye fixation is more likely to be shifted to the ipsilateral side of lesion. On the other hand, if the lesion occurs in Paramedian pontine reticular formation (PPRF) then the eye fixation is shifted towards the contralateral side of the lesion [1]. Also, lesion in the higher cortical centers of one's brain may affect smooth pursuit of the eye while following a target [1]. Hence, detection of abnormality in the eye movement can serve as an important biomarker in screening cases with neurological disorder [3]. Neurological disorders can be of varying types, namely, stroke, multiple sclerosis and spinal cord injuries to name a few [4] that can have varying manifestations on one's gaze. In my research, I have studied the gaze-related indices as screening biomarkers for patients with stroke.

The after-effects of stroke condition can be often addressed through timely screening followed by medication [5]. Unfortunately, in a country like India having scarcity of adequately trained neurologists and healthcare workers, specialized stroke care resources are often limited in major urban areas [6] leaving vast areas of rural healthcare unattended. Consequently, India suffers from epidemic of neurological disability [7], [8]. Thus, developing a cost-effective, easy-to-use, clinically-valid device for screening cases with probable neurological disorder and suitable for both urban and rural healthcare settings, can be critical. In my research, I have used one's oculomotor signature for identification of neurological dysfunction caused by stroke. This is because research shows that up to 86% of stroke patients have some form of oculomotor disturbance [9].

This chapter describes the design of a screening device called SmartEye that can quantify one's probable neurological disorder by measuring one's gaze-related indices in response to static as well as dynamic visual stimuli. In this, I have designed a computer-based visual task and captured one's gaze data in a time-synched manner while one attended to the visual task. Subsequently, the gaze data was processed and analyzed to understand the participant's (i) eye fixation, (ii) smooth pursuit and (iii) blinking activity, with an aim to map the gaze-related indices to one's probable neurological dysfunction. I hypothesize that SmartEye will help in screening stroke condition, a type of neurological disorder through quantitative assessment of gaze-related indices.

3.2 Design of SmartEye System

The eye tracking system (SmartEye) used for quantitative assessment of gaze abnormalities comprised of four units, such as (a) Eye Tracker (b) Graphical user interface (GUI) for data collection (c) Gaze data analysis and (d) Feedback unit.

3.2.1 Eye Tracker Unit

In this study, I recorded the eye movement of stroke survivors (S) and healthy (H) individuals by using an Eye Tracker from EyeTribe Llc [10]. This Eye Tracker is one of the least expensive (priced at around $100) among the off-the-shelf available commercial eye trackers, such as EyeLink, Tobii [11] and other trackers priced at around $100. Technically, the Eye Tracker from the EyeTribe has been shown to be comparable with the available standard EyeLink 1000 in terms of spatial precision and accuracy [11]. The EyeTribe Eye Tracker used in this study is a non-intrusive eye tracking unit which houses a camera and a high resolution infrared source for tracking one’s eye movement. It offers varying sampling rates such as 30Hz and 60Hz with accuracy varying from 0.5^0 to 1^0. Execution in 30Hz allows for a larger tracking area, whereas that in the 60 Hz mode is faster but tolerates smaller head movement [10]. It is recommended to use a chin rest so as to restrict one’s unwanted head movement during gaze data acquisition since the Eye Tracker calibration setting is sensitive to one's head movement. The Eye Tracker can be powered using USB3.0 port that makes it easy to be integrated with computers and tablets. This Eye Tracker is capable of recording one's gaze co-ordinates, pupil size and pupil center.

3.2.2 Graphical User Interface (GUI) for Data Collection

A Graphical User Interface (GUI) was designed to present a visual stimulus on the 2-D monitor of a Task Computer. A participant was asked to follow the visual stimulus. Simultaneously, the EyeTribe Eye Tracker connected to the Task Computer observed the participant's eyes and recorded the participant's gaze data in a time-synchronized manner.

3.2.2.1 System Requirement

In the current study, I used a Task Computer having a monitor of screen size 20.6 inches and display resolution of 1680 pixels×1050 pixels. The Task Computer has Intel(R) Core(TM) i7-3770 CPU @ 3.40GHz as the processor and 8.00 GB RAM. The SmartEye system was developed using a 64-bit Windows 7 operating system. The Eye Tracker connected to the Task Computer was used to track the participant's eyes while the participant followed the visual stimulus presented on the 2-D computer monitor. The Eye Tracker was used in high precision mode with a sampling frequency of 30 Hz. Using EyeTribe in high precision mode provides wider tracking area as well as permits larger head movement compared to the lower precision mode (that is 60Hz) [10].

3.2.2.2 Design of the Graphical User Interface (GUI)

The GUI was developed in-house using Virtual Reality programming software (Vizard) from Worldviz Llc. The GUI comprised of two components (a) background screen and (b) visual stimulus. The background screen was white and the visual stimulus was a black colored circle with a tiny white dot at the center. While looking from a distance of 50 cm, the size of black colored circle was 3 degree and size of tiny while dot was 0.6 degree. Please note that these dimensions were chosen as an initial approximation based on the visibility of the stimulus from a distance of 50 cm. While designing the GUI, I considered a distance of 50 cm since this distance has been used in other eye tracking related studies [12]. The black and white shades of the target stimulus and the background, respectively were chosen to accommodate all participants irrespective of any possible color blindness if any. The GUI was used to display the visual stimulus (dynamic with intermediate static positions) that moved slowly in the horizontal and vertical directions on the Task Computer monitor spanning over 0 to 1680 pixel along the horizontal axis and 0 to 1050 pixels along the vertical axis. My aim was to trigger participant's eye fixation and smooth pursuit eye movement along the horizontal and vertical directions. The slowest possible smooth pursuit speed has been reported to be about 1 to 2 degrees/second [13]. In this study, I chose the speed of the dynamic visual stimulus presented on the computer monitor to be 1.15 degrees/sec. Also, the low speed of the dynamic visual stimulus served to dampen the possibility of one's probable overshoot in the gaze data when the visual (target) stimulus suddenly stopped at the end of the path to offer the static stimulus.

The visual stimulus first appeared at the center of the Task Computer monitor ('position C' at 840×510 pixels of 2-D monitor) (Fig. 3.1) for 5 seconds. Then the visual stimulus slowly moved along a horizontal straight line towards the right side of the 2-D monitor with a speed of 1.15

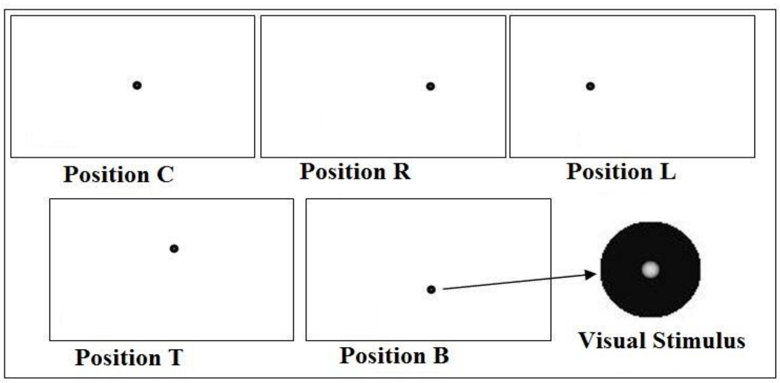


Figure 3.1. presentation of visual stimulus by SmartEye system

degrees/sec (looking towards the monitor from a distance of 50 cm) to trigger a smooth pursuit of the participant's gaze until the stop position (static) at the right side of the screen ('position R' at 1500×510 pixels of 2-D monitor) (Fig. 3.1). At the ‘position R’, the visual stimulus was static for 5 seconds before moving back smoothly at the same speed towards the 'position C' where it again waited for 5 seconds. This half-cycle was then repeated towards the left hand side of the 2-D monitor (‘position L’ at 180×510 pixels of 2-D monitor) (Fig. 3.1). In other words, the visual stimulus was programmed to move horizontally that is positions C-R-C-L-C on the Task Computer monitor to trigger one’s horizontal eye movement towards the right side and then the left side of the 2-D monitor. Subsequently, to trigger a participant’s vertical eye movement, the same half-cycle was repeated first towards the top of the screen (‘position T’ at 840×340 pixels of 2-D monitor) (Fig. 3.1) and then towards the bottom of the computer screen (‘position B’ at 840×680 pixels of 2-D monitor) (Fig. 3.1). In short, for the vertical movement of the visual stimulus, it was programmed to move in the positions C-T-C-B-C on the Task Computer monitor.

During the visual task, the participant was asked to follow the stimulus presented on the Task Computer. The SmartEye system recorded the participant’s time-synchronized gaze data along with the 2-D screen coordinates of the visual stimulus.

3.2.3 Gaze Data Analysis Unit

Fig. 3.2 shows the block schematic of the SmartEye algorithm. While the participant followed the visual stimulus presented on the 2-D monitor, the Eye Tracker connected to the Task Computer recorded the participant’s time-synchronized gaze data such as 2-D gaze co-ordinates and pupil size. Subsequently, the gaze data along with the 2-D screen coordinates of the visual stimulus were processed offline to analyze the gaze-based indices for quantitative assessment.

3.2.3.1 Computation of Deviation of one’s Fixation from Target

To understand participant's eye fixation capability, the gaze data corresponding to static visual stimulus presented at five different positions (such as positions C, R, L, T and B) on the monitor was processed to extract the centroid of the eye fixation coordinates for both the eyes. A valid

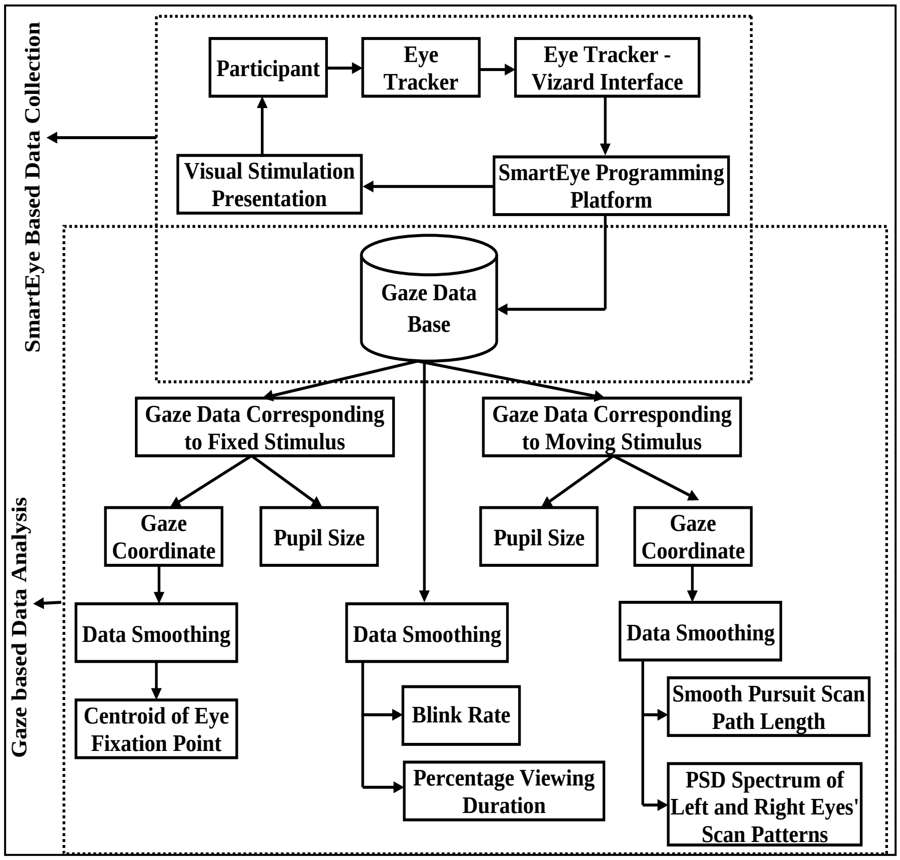


Figure 3.2. Block schematic of SmartEye algorithm
Note: PSD - Power Spectral Density

fixation was the one that lasted for more than 200 msec. The eye fixation coordinates were considered only for the valid fixations. The idea was to avoid noise due to blinks which can be upto 200 msec [14]. Subsequently, the percentage deviation of Fixation Point (%FP_{DEV}) was computed by using eq. (1).

$$FP_{DEV}(\%) = \frac{Target_{Coordinates} - Fixation_{Coordinate}}{Target_{Coordinate}} \times 100 \tag{1}$$

Here, $Target_{Coordinates}$ indicate screen coordinates of the static visual stimulus (that is the Target, in this case); $Fixation_{Coordinates}$ indicate the 2-D gaze coordinates of the centroid of the participant's fixation points corresponding to the presented target stimulus position. Since the Target stimulus was moving either along horizontal axis or vertical axis at a time, I considered either the x-coordinate of one's gaze (for C-R-C-L-C movement of visual stimulus) or the y-coordinate of one's gaze (for C-T-C-B-C movement of visual stimulus) that is, 1-D gaze-coordinates while calculating FP_{DEV}(%). For a healthy individual, one can expect that both the eyes triangulate on the presented visual stimulus with no deviation between the gaze point and the Target position while fixating on the Target. Any deviation (other than the inherent error of 0.5 degree of the EyeTribe tracker) in one's eye fixation point from the Target (actual) position can be an indicator of pathological condition. This deviation may be due to participant's gaze palsy or oscillatory eye movement (nystagmus) [15] resulting on account of from neurological disorder.

3.2.3.2 Computation of Smooth Pursuit Length and Percent Overshoot

I analyzed participant's gaze data to understand smooth pursuit (of eyes) corresponding to dynamic visual stimulus that was programmed to move either along horizontal or vertical direction on the Task Computer monitor. Subsequently, the gaze data was used for understanding participant's ability to scan the visual environment while making horizontal and vertical eye movement. Here the time domain feature of gaze, namely, smooth pursuit length was computed when the stimulus moved from (i) C to R (ii) R to C (iii) C to L (iv) L to C (v) C to T (vi) T to C (vii) C to B and (viii) B to C positions (where, C stands for Center, R stands for Right, L stands for Left, T stands for Top and B stands for Bottom) of the Task Computer monitor screen.

The smooth pursuit length (SPL) was computed from the gaze data corresponding to all the above-mentioned eight ((i) - (viii) above) movements of the visual stimulus. The SPL is the distance between the 2-D gaze coordinates corresponding to the two extreme points of participant's scan path on the stimulus screen, as recorded by the Eye Tracker. Let us consider a typical case in which the visual stimulus (dynamic Target) moved from the 'position C' to 'position R' of the screen. The distance between center and right positions of the Target visual stimulus on the Task Computer monitor was considered as the actual distance (SPL_{Actual}). In this study, when the visual stimulus was programmed to move along the horizontal direction, such as from center to right positions, only the x-coordinate of the Target visual stimulus changed while the value of y – coordinate was maintained as constant. So, the SPL_{Actual} was measured in terms of 1-D coordinates (similar to that in Section 3.2.3.1). Now while following the dynamic visual stimulus (moving along the horizontal direction, say from center to the right), the participant formed a scan path trajectory on the Task Computer monitor between the two extreme points (that is the center and right positions of the monitor in this case). The distance between these two extreme points in terms of 1-D coordinates is $SPL_{Experimental}$. I was interested to study the undershoot or overshoot that might occur while one performs smooth pursuit task. By undershoot, I mean that the participant was not able to follow the dynamic visual stimulus till the end location of the trajectory of the dynamic Target (that is $SPL_{Experimental} < SPL_{Actual}$). In contrast, overshoot was considered to occur when the participant looked beyond the end location of the trajectory of the dynamic visual stimulus (that is $SPL_{Experimental} > SPL_{Actual}$). I was interested to understand whether the SmartEye could pick up participant's undershoot or overshoot, since, literature review indicates that individuals with neurological disorders often demonstrate undershoot [16].

Actually, neurological disorder like stroke often affects one's neural networks that can have implications on one's SPL making the measured SPL as a valuable indicator of neurological disorder. The neural correlates can be used to connect one's abnormality in smooth pursuit length to undershoot. Literature review indicates that one's smooth pursuit eye movement is controlled by the upper motor neuron that controls the eye muscle using the signal from the cerebral cortex. Any damage to the upper motor neuron due to neurological disorder can adversely affect one's smooth pursuit eye movement capability [17].

Using the participant's smooth pursuit length ($SPL_{Experimental}$) and the actual length of the trajectory of the presented dynamic visual stimulus (SPL_{Actual}), I computed the percent deviation in SPL ($SPL_{Overshoot}$%) using Eq. 2.

$$SPL_{Overshoot}(\%) = \frac{SPL_{Experimental} - SPL_{Actual}}{SPL_{Actual}} \times 100 \tag{2}$$

Thus, if the $SPL_{Experimental}$ was greater than the SPL_{Actual}, then the $SPL_{Overshoot}$% was positive and vice-versa in the case of undershoot.

3.2.3.3 Computation of Power Spectral Density

There are evidences in literature that movement of human eyes have characteristic frequency while the eyes move during viewing of a stimulus [18]. Related research indicates that neurological disorder can cause variation in the frequency [19]. This can disturb the coordination between and triangulation of the eyes causing double vision (diplopia) [20]. Thus, I calculated a frequency domain feature, namely, Power Spectral Density (PSD) of the participant's gaze coordinates while the participant followed the dynamic visual stimulus. The PSD was computed separately for the left and right eyes to identify any differences in frequency of movement of both the eyes while fixating on the dynamic visual stimulus. The results obtained for stroke participants were compared with that for the healthy participants. Since the dynamic visual stimulus was programmed to move along a single direction (along X-axis for horizontal and along Y-axis for vertical movement) at a time, I computed the PSD of the participant's 1-D gaze coordinate while one was asked to follow the dynamic visual stimulus. Therefore, I computed the

PSD of the participant's 1-D gaze data along the direction of movement of the dynamic visual stimulus between the different positions of the Task Computer monitor. Once the PSD values for both the eyes were calculated, the difference between the frequencies corresponding to the peak amplitude of the PSD spectrum for left and right eyes were calculated. This gaze-related feature was used to study the participant's scan behavior in horizontal (that is, C-R-C-L-C positions) and vertical (that is, C-T-C-B-C positions) directions during smooth pursuit tasks.

3.2.3.4 Computation of Blink Rate and Percentage Viewing Duration

Stroke survivors often suffer from the problem of dry eyes which not only affects their normal blinking pattern but also affects their control over eye lids [21]. One's blinking can be adversely affected by neurological disorder since the control of eye lid activity in the cortical regions is distributed bilaterally among multiple motor areas that make it vulnerable to neurological disorder [1]. However, there is a study that did not show such effect in the cases of acute stroke [22]. Given such conflicting views, I wanted to understand whether the SmartEye can pick up information on one's blinking that can be linked with dry eye effect. Thus, to study the effect of stroke on one's blinking pattern, I computed average blink rate (blinks/minute) in each eye of the participant while he/she was performing the visual task in a synchronized manner. The Eye Tracker used in this study did not come with blink detection algorithm. For this study, I used in-house developed blink detection algorithm to measure one's blink rate while attending to the visual task. The blink detection algorithm was based on the detection of pupillary occlusion. Specifically, when one blinks, the pupil will be occluded by the eye lids, at least partially by the eye lids. This partial occlusion was recorded by the Eye Tracker in the form of either sudden reduction in pupil size (say by, 60%) or pupil size being recorded as zero. Taking inputs from literature, I considered a blink to be valid if duration varied upto 200 msec [23]. Again, blinking also effects the duration for which one's eyes are open. Therefore, it is also important to have information on the duration for which the participants had their eyes opened while attending to the visual stimulus. This information was obtained by calculating the percentage (%) viewing duration out of total task duration. I also measured the percentage of time for which the eyes were closed during the task.

3.2.4. Feedback Unit

Once the participant finished the visual task, I provided visual feedback on the gaze-related features associated with participant's eye movement to both the participant and the accompanying clinician. This unit was used to provide quantitative estimate of the participant's gaze data in terms of (a) percentage deviation in participant's gaze fixation coordinates from the position of the static visual stimulus (b) length of scan path trajectory of both the eyes of a participant corresponding to the dynamic visual stimulus (c) average blink rate during the task and (d) comparative measure of the percentage of duration during which the participant's eyes were open or closed. Apart from informing the participant on his gaze-related features, this feedback also served as complementary information to the clinician regarding the participant's pathological profile.

3.3 Experiment and Methods

3.3.1 Participants

In my current study, eight healthy (H) participants (mean (SD)= 59.87y (9.07) years) volunteered from my institute and also from the neighborhood. I first conducted this study with healthy participants to make sure that the SmartEye system was functioning as desired. Subsequently, I recruited eight stroke survivors (S) (mean (SD)=55.87 years (8.09) years) for the study. The stroke survivors were recruited from the local civil medical hospital where they were undergoing treatment. My present study had 8 stroke survivors based on their availability at a local civil hospital. This being a proof-of-concept study and the availability of a small participant pool with heterogeneous post-stroke symptoms, I tried to carry out in depth analysis of their

Table 3.1. Participants' metadata for usability study with SmartEye

	Age (y)	Gender	Post-stroke Period	Side of Lesion
S1	55	Male	7.5 years	Left
S2	54	Male	2.5 years	Left
S3	50	Male	6 Months	Bilateral
S4	59	Male	1 year	Left
S5	72	Male	5 Months	Left
S6	45	Male	4 Months	Right
S7	60	Female	7 Years	Left
S8	52	Male	5 Months	Bilateral
H1	67	Male	-	-
H2	64	Male	-	-
H3	49	Female	-	-
H4	70	Male	-	-
H5	55	Male	-	-
H6	45	Female	-	-
H7	65	Male	-	-
H8	64	Female	-	-

H1-8= Healthy, S1-8= Stroke-survivor

gaze-related measures. No compensation was provided to the participants for taking part in this study. Table 3.1 shows the metadata of healthy as well as stroke participants. With a reduced participant pool, I used non-parametric hypothesis test namely Wilcoxon Signed Rank Test. An independent sample test was performed on the participants' age which showed that the stroke and healthy participants were age-matched with p-value > 0.05. The participants did not have any prior exposure to computer-based task. The inclusion criteria were (1) ability to follow the instructions (2) should be able to see the visual stimulus appearing on 2-D monitor from the distance of 50 cm (3) should not have gone through any ocular surgery in the recent past. The ethics for the study was approved by the institutional ethics committee of IIT Gandhinagar, India. Also, before the start of the study the participants were asked to sign a consent form in which the details of the study, such as purpose of the study, type of task to be performed, possible benefit and risk were mentioned.

3.3.2 Experimental Setup

Fig. 3.3 shows the experimental setup of the SmartEye device. The setup consisted of a (i) Task Computer (PC) (ii) Eye Tracker and (ii) chin rest. A GUI was designed in the VR to present

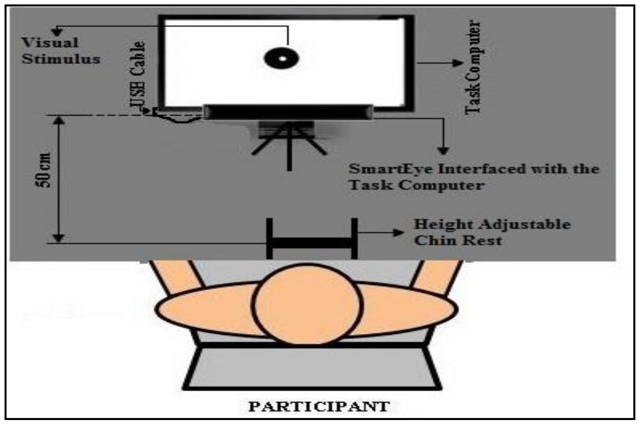


Figure 3.3. Experimental setup for SmartEye system

the visual stimuli to the participants on a 20.6" 2-D monitor of the Task Computer. Each participant was asked to sit on a chair kept in front of the Task Computer. An Eye Tracker (Section 3.2.1) was connected to Task Computer to capture participant's eye movement. The Eye Tracker was connected to the PC through USB 3.0 cable which was used to power up the Eye Tracker and also to transmit the acquired gaze data to the PC. Since the Eye Tracker calibration was sensitive to the user's head movement, a height-adjustable chin rest was used to avoid any unwanted head movement by the participants. The chin rest also helped to position the participant's eyes so that these were collinear with the center of Task Computer monitor. The chin rest was placed at a distance of 50 cm from the Task Computer monitor whereas the Eye Tracker was kept at approximately 45 cm from the chin rest. The distance of 50 cm was maintained similar to that used in other studies [23]. This distance was ensured through proper set-up related measurements. The distance of 45 cm between the Eye Tracker and the chin rest was maintained so as to satisfy the Eye Tracker mounting specifications that state that participant's eyes should be placed at a minimum of 45cm in front of the Eye Tracker [10]. The Task Computer was used to collect the time-synchronized gaze data while the participant was asked to follow the visual stimulus appearing on the 2D computer monitor. While the participant performed the visual task, my system processed the acquired raw gaze data at the backend of the Task Computer in real-time to extract the relevant gaze features.

3.3.3. Procedure

My study required a commitment of approximately 20 minutes from each participant. Once the participant arrived in the experiment room, he/she was asked to sit on a chair and relax for 5 minutes. Then, a clinician in the team ensured that the participant satisfied the inclusion criteria for the study. This step took around 10 minutes for each participant. If the participant fulfilled the inclusion criteria of the study, the experimenter (myself) briefed the participant about the study protocol and demonstrated the experimental setup. Then the experimenter ensured that the participant understood the task followed by signing of the consent form. Before the start of the study, the participant was told that he/she was free to quit from the study at any point if he/she felt uncomfortable during the experimental session. Once the participant understood the task and expressed his/her willingness to participate in the task, the experimenter performed the Eye Tracker calibration. To perform the Eye Tracker calibration, the experimenter sat on the chair kept in front of the Task Computer and placed his chin on the adjustable chin rest (the height being adjusted as per the participant’s convenience). Then the tilt angle of the Eye Tracker and the height of the chin rest were adjusted so that the eyes could be tracked by the Eye Tracker. Subsequently, the experimenter performed a 9-point calibration routine of the Eye Tracker that took approximately 20 seconds. The same experimenter performed Eye Tracker calibration for all the participants so as to reduce the calibration-related variations and also to provide a visual task environment in which the Eye Tracker calibration was not adversely affected due to probable oculomotor abnormalities of the stroke survivor. Once the experimenter performed Eye Tracker calibration successfully, the participant was asked to replace the experimenter while keeping the experimental setup unchanged. Then the experimenter started the study by asking the participant to look and follow the visual stimuli (Section 3.2.2.2) appearing on the monitor of the Task Computer. While the participant was performing the visual task, the Eye Tracker monitored his/her eye movement. The real-time gaze data of the participant was recorded and processed by the Task Computer at the backend. At the end of the task, my system provided visual feedback to the participant through a graphical representation of the participant's $\%FP_{DEV}$ (Section 3.2.3.1), the trajectory of the smooth pursuit (Section 3.2.3.2), blink rate and measure of percentage viewing and non-viewing time duration (Section 3.2.3.3).

3.4 Results

In the following sections, I present the results of the study in which 8 stroke survivors and 8 age-matched healthy participants participated. These results are also reported in one of my published articles [24].

3.4.1 Gaze Stabilization during Gaze Fixation Task

There is evidence in literature that neurological disorder can affect a participant's gaze stabilization system [25]. For example, postero-inferior cerebellar strokes can cause nystagmus problem in which a patient's eyes make repetitive, uncontrolled movements [25] while trying to fixate at a particular point. Given the evidence of such oculomotor disorders accompanying stroke cases, in this study, I wanted to understand a stroke survivor's ability to maintain ocular position at a fixed visual target presented on the 2-D monitor of Task Computer. Given a Target stimulus, I measured participant's fixation by calculating the centroid of participant's gaze

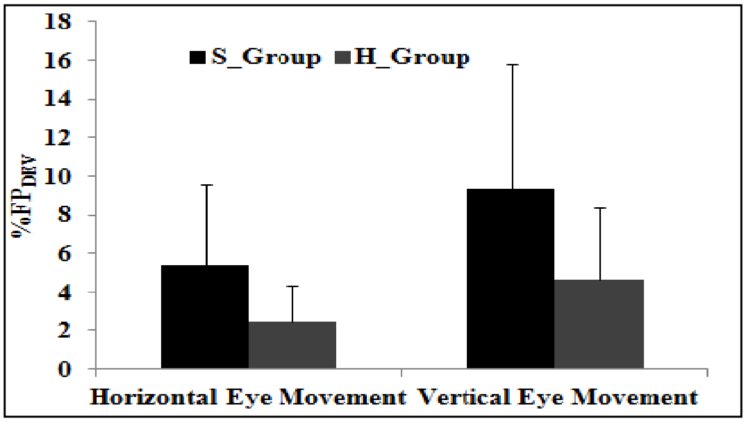


Figure 3.4. Comparative group analysis of %FP$_{DEV}$ for healthy and stroke group

coordinates in response to static visual stimuli appearing at five different positions (positions L, R, C, T, and B; Section 3.2.2.2) on the 2-D monitor. Subsequently, I studied the deviation in the centroid of participant's gaze fixation points from the positions of static visual stimulus. The idea was to understand whether a participant was able to properly fixate on the presented visual stimulus. Specifically, this gaze-related index can be used to predict abnormal gaze stabilization, probably resulting from gaze palsy, nystagmus, and/or strabismus possibly due to lesion in the cortical regions. Here, I hypothesized that the stroke participants will exhibit higher deviation in gaze fixation than their healthy counterparts.

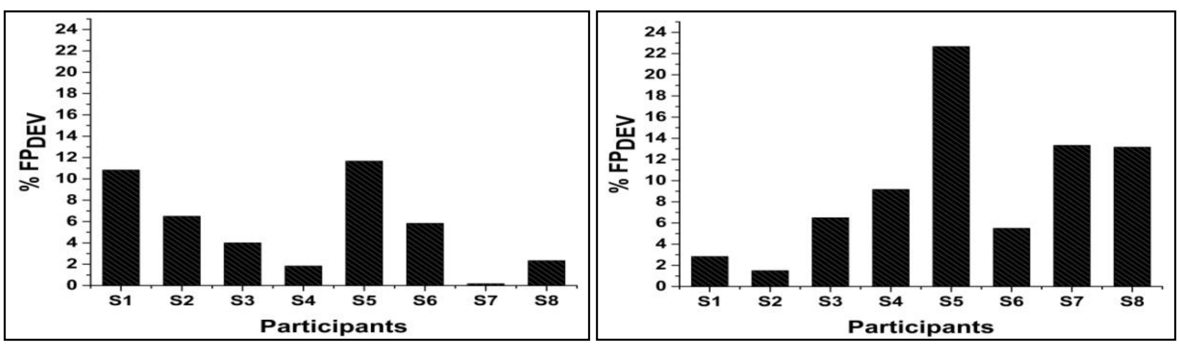


Figure 3.5. Percent FP$_{DEV}$ of individual Stroke participants for (a) C-R-C-L-C and (b) C-T-C-B-C movement of Visual Stimulus

For this, I calculated percentage deviation of participant's centroid of Fixation Point from Target Position (FP$_{DEV}$) corresponding to five different positions on the 2-D screen (Section

3.2.2.2) where the visual stimulus was static for a short duration (5 secs). Fig. 3.4 shows the group average of the percentage FP_{DEV} for the Healthy (H) and Stroke (S) groups. It can be seen from Fig. 3.4 that the S group showed higher $\%FP_{DEV}$ corresponding to the static visual stimulus appearing at different locations in the horizontal direction (that is, C, R and L positions) as well as in the vertical direction (that is, C, T and B positions) than that for the H group. The group average $\%FP_{DEV}$ for S group was beyond one standard deviation compared to that for the H group.

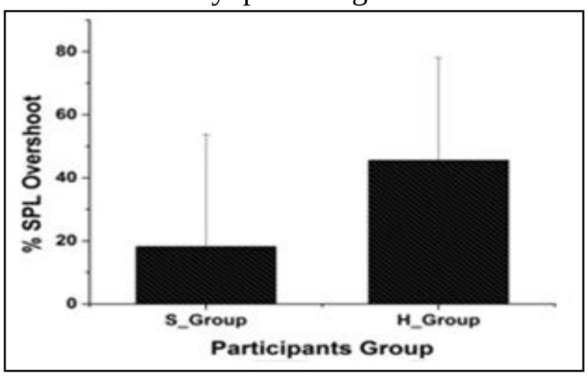
The Figs. 3.5 (a) and (b) show the $\%FP_{DEV}$ corresponding to the static visual stimulus appearing along the horizontal and vertical directions, respectively for the stroke participants at an individual level. I found that almost all the stroke patients (except, S3) demonstrated $\%FP_{DEV}$ which is beyond 1 std. dev. of the group average of $\%FP_{DEV}$ for healthy participants in either or both of the horizontal and vertical directions. Here, participant S3 showed normal fixation ability similar to healthy participants. However, S3 showed anomaly in other gaze-related indices, as explained in the following sections.

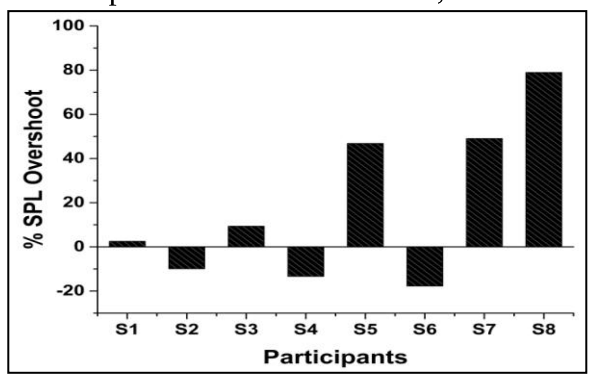
Also, to understand the possible relationship between the region of lesion and the abnormality in stroke participants' gaze fixation pattern, I further analyzed the $\%FP_{DEV}$ data. Literature indicates that lesion in one's frontal lobe can cause deviation in the gaze fixation towards the ipsilateral side of lesion. In contrast, lesion in the Pre-Pontine Reticular Formation (PPRF) area can cause gaze fixation to be deviated in the contralateral side of the lesion [1], [8]. To see the effect of hemiplegia, I computed the $\% FP_{DEV}$ separately for each eye that is, left eye and right eye for the stroke patients. I found that six (S1-S5 and S8) out of eight participants showed higher $\% FP_{DEV}$ for the eye which was ipsilateral to the side of the lesion while they fixated their gaze on the static visual stimulus appearing at positions C, R, and L position along the horizontal direction. In contrast, participants S6 and S7 showed higher % FP_{DEV} for the eye which was contralateral to the side of the lesion. These findings suggest that participants S1-S5 and S8 might have a lesion affecting their frontal lobe neural networks. In contrast, the oculomotor disturbance in participants S6 and S7 might be due to lesion affecting their PPRF area of the brain. To verify this possibility of brain lesion in either frontal lobe and/or PPRF area, I looked at the neuroimaging (MRI) report of the stroke patients. In the current study, I could get the neuroimaging (MRI) result of only two participants, S1 and S8 that indicated that both of them had lesion in the frontal lobe area of their brain. Again, I performed similar analysis of gaze data corresponding to static visual stimulus appearing in vertical direction (that is at positions C, T and B). I found that for all the stroke participants (except S5) there was variation in % FP_{DEV} for both left and right eyes. Specifically, participants S2, S3 and S7 had larger %FP_{DEV} in their eyes that were contralateral to the side of the lesion that that for the eyes in their respective ipsilateral sides. A possible explanation to such an observation can be that these participants might have lesions in their midbrain or pontine regions. In the case of participant S5, I found equal %FP_{DEV} for both the eyes (that is, left and right eyes) corresponding to the static visual stimulus appearing in the vertical direction. This may suggest that he might have lesion in pontine region of both left and right hemispheres of the brain. However, I did not get MRI report for S5. Also, the result of %FP_{Dev} for S and H groups was not normally distributed and therefore I performed non-parametric dependent sample paired wilcoxon signed-ranked test. However, I did not get any statistical significance %FP_{Dev} between S and H groups. A possible reason can be that the post-stroke participants possessed diffused nature of stroke conditions, as reported by the clinicians.

To summarize, the comparative group analysis of $\%FP_{DEV}$ shows that the S and H groups demonstrated variations in the fixation pattern while looking towards a static stimulus. However, given the spectrum nature of the S-group and the small sample size, no statistical difference was observed. A large-scale study is required before drawing inferences.

3.4.2 Time Domain Features of Gaze Shifting during Smooth Pursuit Task in Stroke Participants

Hart et. al. [16] have suggested that the smooth-pursuit eye-tracking test is useful in screening neurological disorder such as posterior or middle cerebral artery thrombosis [16]. Thus, in this study, I investigated the implication of stroke on participant's ability to track the slowly moving visual stimulus by providing him/her with a visual pursuit task. To do this, I studied the





(a) (b)

Figure 3.6: (a) Group average of percentage Overshoot in SPL for Stroke and Healthy Groups, (b) Percentage overshoot in SPL for SS participants

participant's gaze scan path in response to the presented dynamic visual stimulus which was programmed to move smoothly between different positions (C, R, L, T and B positions) on the 2-D monitor. I wanted to understand the implication of stroke on participant's gaze shifting behavior when the participant was asked to follow the dynamic visual stimulus with his/her eyes that is perform the smooth pursuit task. The smooth pursuit eye movement is partially controlled by upper motor neuron in the cortical region among others [25]. The decision-making associated with generation of normal smooth pursuit eye movement depends on the health of the cortical regions such as parietal occipital temporal junction, the frontal lobe and frontal eye field [26], [27]. Thus, any lesion in the above-mentioned cortical regions can cause abnormal smooth pursuit in the eye movement. In order to understand the implications of stroke on participant's smooth pursuit, both time domain and frequency domain indices were extracted from the smooth pursuit gaze data of the participants. Specifically, I processed the scan path trajectory to calculate the length of the trajectory (that is, smooth pursuit length (SPL)) (time domain feature) and the frequency corresponding to the peak power spectrum density (frequency domain feature). Again literature indicates that stroke patients often suffer from undershoot when asked to follow a dynamically moving visual stimulus [1], [9]. Undershoot refers to an individual’s smooth pursuit length ($SPL_{Experimental}$) of scan path trajectory in response to a dynamic visual stimulus being less than the actual length traversed by the visual stimulus (SPL_{Actual}) (Section 3.2.3.2). Here I hypothesized that S group participants in my study will exhibit Undershoot similar to that in other studies [1], [9]. In contrast, the healthy counterparts might not exhibit Undershoot.

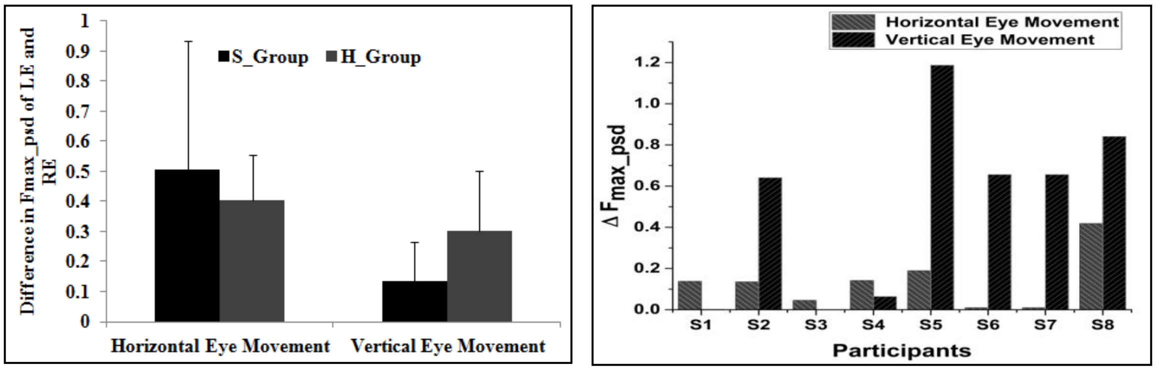
As explained in the section 3.2.2.2, in my current study, the visual stimulus was programmed to move between four horizontal paths (that is, C-to-R, R-to-C, C-to-L and L-to-C) and four vertical paths (that is, C-to-T, T-to-C, C-to-B and B-to-C) positions on the 2-D computer monitor slowly to trigger participant's smooth pursuit eye movement (Section 3.2.2.). Fig. 3.6 (a) shows the comparative group (S and H groups) analysis of participants' gaze fixation pattern while tracking the slowly moving visual stimulus in terms of percentage overshoot of SPL ($\%SPL_{Overshoot}$; Section 3.2.3.2). The group average $\%SPL_{Overshoot}$ for H group was higher than that for the S group. However, the S group showed greater average standard deviation of $\%SPL_{Overshoot}$. The higher standard deviation of $\%SPL_{Overshoot}$ for the S group was because, the stroke patients (except S5, S7 and S8) show either negative or a small positive percentage overshoot (-17.72% to 9.34% of overshoot in SPL) of SPL (Fig. 3.6 (b)) compared to the healthy participants (12.59% to 72.5%). Literature indicates that healthy individuals often exhibit rapid micro-saccades that form an integral part of active gaze strategy to help to focus the fovea onto the visual stimulus [28]. Thus, while following slowly moving stimulus, participant's $SPL_{Overshoot}$ might be positive for healthy participants. However, stroke survivors often exhibit slow micro-saccades [29], which may therefore cause lower $SPL_{Overshoot}$ than that of the healthy participants.

Again, to understand the clinical validity of the smooth pursuit gaze behavior, I analyzed the neuroimaging (MRI) report of the stroke patients. Since I had MRI report of S8 and he showed higher $\%SPL_{Overshoot}$ like healthy participants, the MRI report of S8 revealed that he had had no lesion in the frontal or occipital regions. Again, the lateral rectus and the medial rectus muscles of both eyes are controlled by frontal and occipital lobes of one's brain [1], [9]. Thus, the high value of $SPL_{Overshoot}$ (%) in S8 might be due to his frontal and occipital regions being healthy.

In short, $SPL_{Overshoot}$ (%) can be used to capture the difference in smooth pursuit gaze behavior of the H and S groups. With the small sample size and heterogeneous nature of stroke condition, no statistical difference was observed between the $SPL_{Overshoot}$ (%) of the H and S groups.

3.4.3 Frequency Domain Features of Gaze Shifting during Smooth Pursuit Task in Stroke Participants

Lesion in the frontal lobe that controls one's smooth pursuit of eyes can affect the frequency of the eye movement [1], [9]. To assess this, I computed the gaze-related frequency domain index corresponding to the participants' smooth pursuit data (that is, in response to dynamic visual stimulus). Specifically, I calculated the power spectral density (PSD) of the gaze data for both the eyes of the participant while they followed the slowly moving visual stimulus. From the PSD of the gaze data, I extracted the frequency (F_{max_psd}) corresponding to the peak amplitude of the PSD spectrum. The F_{max_psd} was extracted for the gaze data from both the eyes. Fig. 3.7 (a) shows the comparative analysis of the absolute average difference between F_{max_psd} (ΔF_{max_psd}) for left and right eyes for both the S and H groups. This gaze-related index (ΔF_{max_psd}) was calculated in response to horizontal (C-R-C-L-C) and vertical (C-T-C-B-C) movements of the visual stimulus. Here I hypothesized that H group having healthy gaze will demonstrate nearly



(a) (b)

Figure 3.7. (a) Comparative group average ΔF_{max_psd} in response to C-R-C-L-C (Horizontal) and C-T-C-B-C (Vertical) movement of stimulus, (b) ΔF_{max_psd} for stroke participants in response to dynamic visual stimulus

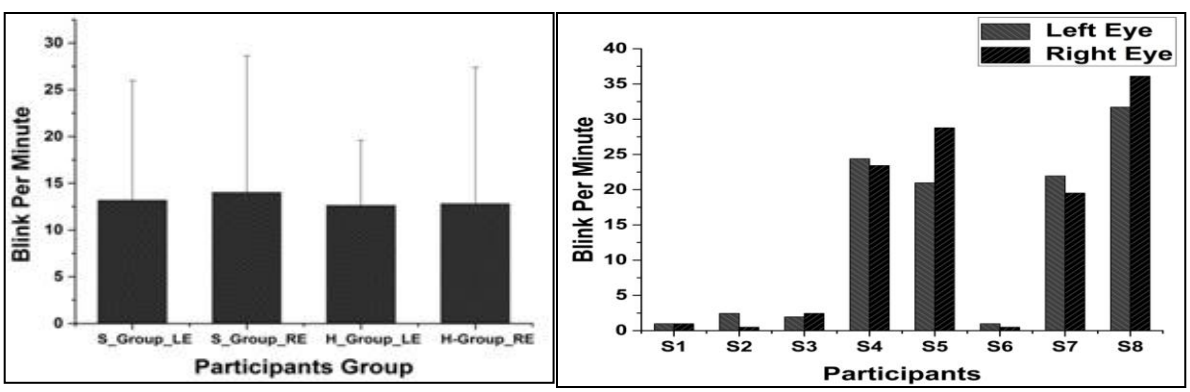
identical ΔF_{max_psd} between the left and right eyes along both the horizontal and vertical directions. In contrast, S group participants might demonstrate variations in ΔF_{max_psd} when considered along horizontal and vertical directions.

From Fig. 3.7 (a) it can be seen that the mean difference in peak frequency (that is, ΔF_{max_psd}) between left and right eyes along horizontal and vertical directions are quite different for the S group (ΔF_{max_psd} = 0.14 Hz for C-R-C-L-C and ΔF_{max_psd} = 0.50 Hz for C-T-C-B-C). On the other hand, for the H group, the mean ΔF_{max_psd} was closely similar for both the horizontal and vertical directions (ΔF_{max_psd} = 0.30 Hz for C-R-C-L-C and ΔF_{max_psd} = 0.40 Hz for C-T-C-B-C). At an individual level, I find that (from Fig 3.7 (b)) there were variations in ΔF_{max_psd} for all the stroke participants (S1-S8).

I analyzed the data of the participants S1 and S8 (whose MRI reports were available with me) in detail. As per the MRI report of S1, he had lesion in the left frontal lobe of the brain which controls one's saccadic eye movement. It can be seen from Fig. 3.7 (b), that S1 exhibited higher ΔF_{max_psd} (ΔF_{max_psd} = 0.14 Hz) for the horizontal direction as compared to that for the vertical direction (ΔF_{max_psd} = 0.001 Hz). The relatively larger ΔF_{max_psd} for the horizontal direction compared to that for the vertical direction for S1 can be possibly attributed to his lesion in the frontal lobe [1], [8] as mentioned in the MRI report. Participant S8 showed comparatively high ΔF_{max_psd} along both the horizontal and vertical directions (Fig. 3.7(b)). Such an observation on the frequency domain gaze-related index derived from the smooth pursuit data of S8 can be possibly attributed to the reported lesion in the bilateral basal ganglia and multiple chronic lacunar infarcts, as can be seen from literature [30].

3.4.4 Blink Activity in Participants with Stroke

Stroke survivors often report problem of dry eyes that is related to the reduction in the blink rate or excessive blinking to deal with the dry eye effect [15] thereby causing the normal blinking pattern to get adversely affected. From literature review I find that healthy individuals



(a) (b)

Figure 3.8. (a) Group average for BPM for both eyes of stroke and healthy Groups, (b) Blink per minute for both eyes of individual stroke participant

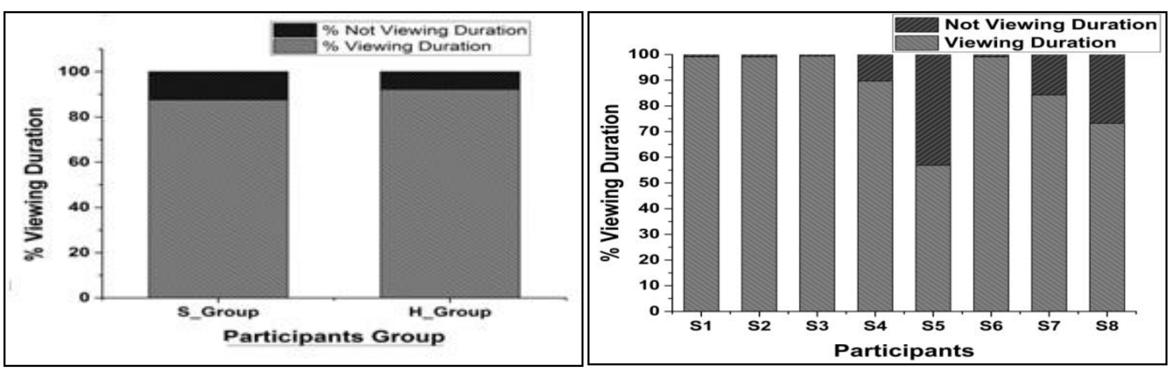
often show variation in blink rate (expressed as blinks per minute (BPM)) depending on the task in which they are involved. Literature indicates that BPM of healthy participants have been reported to vary between 4.5 BPM for visual task or reading to 17 BPM during rest condition, and 26 BPM during a conversation with others [31]. Here I hypothesized that the participants in the S group will exhibit either too reduced or too high BPM as compared to normative blinking.

In my current study, the participants were asked to look at the visual stimulus presented on the Task Computer monitor. Also, while doing the task, they were not involved in any conversation with anyone. Thus, in this study case, I considered 4.5 BPM to 17 BPM as normative blink pattern. From Fig. 3.8 (a), it can be seen that the group average BPM for both the S and H groups were closely similar. Statistical independent sample test showed that there was no significant difference in the BPM between the S and H groups. A probable reason for this can be the limited sample size and the heterogeneous nature of disability of stroke participants. Specifically, the stroke participants in my study came with different types of brain lesion and varying post-stroke periods (Table 3.1). Therefore, the group analysis of the BPM across all the stroke participants might have hidden some of the key results. This was evident when I carried out individual participant analysis for the S group (Fig 3.8 (b)) that showed abnormality in terms of either too reduced or too large blink rate compared with the normative range of 4.5 BPM to 17 BPM. From Fig. 3.8 (b), it can be seen that four (S1 - S3 and S6) out of eight stroke participants showed very low (<4.5 BPM) blink rate while the rest showed very high (> 17 BPM) blink rate.

To understand the clinical validity of the blink rate data for the participants of the S group, I tried to find the relation of the MRI reports with the blink rate data. As mentioned before, I had the MRI data available for only S8 and S1. The high blink rate (31.7 BPM for Left eye and 36.09 BPM for Right eye) of S8 can be possibly attributed to age-related cerebral atrophic changes, as mentioned in his MRI report. Literature states that the age-related cerebral atrophy may cause high blink rate in an individual [32]. Participant S1 demonstrated extremely low blink rate for both the eyes ((1 BPM for both Left and Right eyes) as can be seen from Fig. 3.8 (b). Literature indicates that the insular cortex is responsible for controlling one's eye movement [31] and plays an important role in blink suppression [33]. The extremely low BPM of S1 can be possibly attributed to the lesion in the frontal lobe as well as the insular cortex, as evident from his MRI report.

3.4.5 Gaze Pattern of Stroke Patients in terms of Percentage of Viewing Duration

Here, the participants' gaze data corresponding to the complete visual task was analyzed to find the implication of stroke on the participants' gaze pattern in terms of the viewing duration. Given the abnormal blink pattern of the participants of S group, it can be expected that their percentage Viewing Duration towards the presented stimulus will not be normative. Though the comparative group (S and H groups) analysis on the percentage of time the participants spent viewing the presented visual stimulus and had their eyes closed while taking part in the visual



(a) (b)

Figure 3.9. (a) Group average for percentage viewing duration of stroke and healthy groups, (b) Percentage viewing duration of individual stroke participant

task were quite similar (Fig. 3.9 (a)), on an individual level for the S group participants, a different picture can be seen. It can be seen from Fig. 3.9 (a) that the average % looking duration for the S group (87.17%) was not considerably different from that of the H group (92.17%). The

% viewing duration of S and H groups were not normally distributed. A non-parametric independent sample paired Wilcoxon signed-ranked test revealed that the % viewing duration of S group and H group was not statistically different (p-value > 0.05). However, at an individual level (Fig. 3.9 (b)), four (S1, S2, S3 and S6) out of eight stroke participants showed an extremely high (mean=99.13%) average % viewing duration that can be related with very low blink rate (mean=1.34 BPM) (Fig. 3.8 (b)). Again, the low average % viewing duration (56.89%) of S5 can be partly due to his high blink rate (24.87 BPM) as can be seen from Fig. 3.8 (b). For other stroke participants (S4, S7 and S8), the average % viewing duration (82.4%) was fairly close to that for the H group (92.17%). This finding can be attributed to a cumulative effect of high blink rate and reduced blink duration. The blink duration was considered as the duration for which the participant's eyes were occluded from the view of the Eye Tracker due to blinking. Specifically, the participants S4, S7 and S8 exhibited a high average blink rate (26.17 BPM) coupled with reduced blink duration of 388.97 msec than that of the other stroke participants having a mean blink duration of 575.67 msec.

3.5 Discussion and Limitation

The prevalence of neurological disorders such as stroke condition is causing it to be one of the major contributors to the global health burden that is likely to increase as a result of ageing and population growth [34]. Particularly developing countries like India are in midst of neurological epidemic due to limited specialized neurology units that are restricted to few urban centers and also are quite expensive to be accessed. Clinicians use conventional observation-based screening techniques. These are heavily dependent on clinician’s expertise to pick up the signatures to facilitate screening of the stroke condition. Given the low doctor:patient ratio and the socio-economic status in India, getting access to such expert clinicians is often difficult to the common man. Additionally, though such observation-based screening techniques are promising, yet these are not quantitative and often suffer from issues of subjectivity during prediction. Given the evidence of oculomotor signatures being used by clinicians and the limitations of observation-based techniques, developing an oculomotor-based screening technique that can offer quantitative measures and does not necessitate specialized technical knowledge to be operated is critical. Being motivated by this, I have designed SmartEye system that is an easily-accessible and cost-effective system which can provide quantitative estimates of gaze-related indices as biomarkers of one's neurological dysfunction. In this chapter I have presented the design of the SmartEye system along with administering a study with SmartEye in which both healthy and post-stroke participants volunteered.

In this chapter, I have presented two sets of goals, namely, (i) understanding the applicability of SmartEye as an easily accessible and user-friendly oculomotor monitoring device for stroke participants and (ii) identifying gaze-related indices, such as gaze fixation, smooth pursuit length, BPM and % viewing duration that can provide quantitative estimates in terms of deviation from normative pattern for the stroke participants compared to their healthy counterparts.

Though, the result of the study was promising, yet, the study had some limitations. In this study with a limited sample size and the heterogeneity of clinical profiles of the stroke participants, I did not report results of statistical tests of significance. In fact, I could not find statistically significant variation in the gaze-related indices between the two groups of participants (that is the S and H groups). In future, I plan to extend this study with larger participant pool. Also, I did not have access to the neuroimaging reports of all the stroke participants that restricted in-depth analysis, particularly for specific types of stroke. For example, it might be interesting to study the oculomotor anomalies in cases with posterior circulation stroke (PCS). This is because, the conventional face arm speech test (FAST) screening instrument is less sensitive for PCS as compared with that for anterior circulation stroke [35]. Again, for the PCS patients, dizziness is one of the common symptoms [36] that is often confused with inner ear problems [37]. However, in the cases of PCS, patients often report a visual field defect affecting either the two right or the two left halves of the visual fields of both eyes [38]. Thus, SmartEye can be applicable in such cases as well. In spite of the applicability of SmartEye even to specific types of stroke, we could not look into such details due to limited access to patients' neuroimaging reports as stated earlier. In future, we plan to conduct such studies in association with hospital settings that maintains connectivity with a cohort of stroke patients along with MRI reports.

Though I acquired participant's 2-D gaze coordinates corresponding to fixation and smooth pursuit tasks, I have considered only 1-D gaze-coordinates to compute different gaze-related indices such as $FP_{DEV}\%$, $SPL_{Overshoot}$ and ΔF_{max_psd}. This was because, in this preliminary study, my aim was to understand the anomaly in the participant's eye movement either in the horizontal or vertical directions. In future, I plan to study more complex gaze shifting patterns where visual stimulus appearing in different regions of the Task Computer monitor can be offered. Another limitation of the study was the low sampling rate (30 Hz) of the Eye Tracker due to which I did not measure the participants' rapid eye movements such as saccadic movement. In the current study, I wanted to study the participant's gaze fixation, smooth pursuit and blink rate and these can be measured with the low-cost 30 Hz Eye Tracker making it feasible for this current study. In future, I plan to incorporate Eye Trackers with higher sampling rate as an integral part of the SmartEye setup, if saccadic eye movement data is required. Though I have applied SmartEye device for post-stroke individuals, yet, SmartEye can be used to assess oculomotor abnormalities in patients with various neurological disorders, of course with varying manifestations based on the type and intensity of neurological disorder. Literature reports that oculomotor abnormality is evident in various neurological disorders such as Spinocerebellar ataxia [39] and atypical Parkinsonism [40], and others besides stroke. In my research, I wanted to understand the feasibility of SmartEye to be used as a screening device for post-stroke cases to begin with. However, in future, I have plans to extend the applicability of SmartEye to other neurological disorders and in turn try to understand differences in the oculomotor manifestations.

Nevertheless, the results obtained in this study are promising and thereby the SmartEye system can be a stepping stone towards developing a quantitative, inexpensive and easily-accessible screening device for neurological disorder. The idea behind the SmartEye is to provide a complementary tool in the hands of the clinician so that probable cases of neurological disorder can be easily screened before deciding to go ahead with the costly neuroimaging techniques that is often inaccessible to the common man.

Post-stroke vision problems accompany many other deficits such as those related with balance that in turn can adversely affect one's mobility and community living. Specifically, one's vision system provides important sensory information to the human balance system via different eye movements [41], [42]. Thus, once screened for probable neurological disorders, it is important to rehabilitate the patients on balance-related issues. In the forthcoming chapters I am going to present my work aimed to address balance disorders of post-stroke participants by exposing them to Virtual Reality based motivating serious games. Additionally, I will present my findings on the gaze behavior of post-stroke participants in response to goal-directed balance task. In short, the forthcoming chapters of this dissertation looks into settings that can be used for balance rehabilitation along with understanding the subtle nuances of variation in one's gaze behavior during a balance task.

Reference

[1] J. Leigh and D. S. Zee, *The neurology of eye movements*, vol. 53, no. 9. 2013.

[2] P. Dale et al., *Types of eye movements and their functions*. 2001.

[3] P. H. Tseng, I. G. M. Cameron, G. Pari, J. N. Reynolds, D. P. Munoz, and L. Itti, "High-throughput classification of clinical populations from natural viewing eye movements," *J. Neurol.*, vol. 260, no. 1, pp. 275–284, 2013.

[4] World Health Organization. Neurological disorders: public health challenges. World Health Organization, 2006.

[5] L. L. Yan et al., "Prevention, management, and rehabilitation of stroke in low- and middle-income countries," *eNeurologicalSci*, vol. 2. pp. 21–30, 2016.

[6] J. Joubert et al., "Stroke in rural areas and small communities," *Stroke*, vol. 39, no. 6. pp. 1920–1928, 2008.

[7] A. Das, A. L. Botticello, G. R. Wylie, and K. Radhakrishnan, "Neurologic Disability: A Hidden Epidemic for India," *Neurology*, vol. 79, no. 21, pp. 2146–2147, 2012.

[8] J. D. Pandian et al., "Public awareness of warning symptoms, risk factors, and treatment of stroke in Northwest India," *Stroke*, vol. 36, no. 3, pp. 644–648, 2005.

[9] A. Pollock et al., "Interventions for disorders of eye movement in patients with stroke," in *Cochrane Database of Systematic Reviews*, 2011.

[10] "THEEYETRIBE." [Online]. Available: http://theeyetribe.com/theeyetribe.com/about/index.html. [Accessed: 01-May-2015].

[11] E. S. Dalmaijer, "Is the low-cost EyeTribe eye tracker any good for research?," *PeerJ*

Prepr., vol. 4, no. 606901, pp. 1–35, 2014.

[12] M. J. Wieser, P. Pauli, G. W. Alpers, and A. Mühlberger, "Is eye to eye contact really threatening and avoided in social anxiety?-An eye-tracking and psychophysiology study," *J. Anxiety Disord.*, vol. 23, no. 1, pp. 93–103, 2009.

[13] J. M. Groh, *Making Space - How the brain knows where things are*. 2014.

[14] A. Poole and L. J. Ball, "Eye Tracking in Human-Computer Interaction and Usability Research: Current Status and Future Prospects," *Encycl. Human-Computer Interact.*, pp. 211–219, 2005.

[15] M. Strupp et al., "Central ocular motor disorders, including gaze palsy and nystagmus," *Journal of Neurology*, vol. 261, no. SUPPL. 2. pp. 542–558, 2014.

[16] K. Pawlak-Osińska, H. Kaźmierczak, and W. Kaźmierczak, "Saccadic and smooth-pursuit eye movement in neurootological diagnostic procedures," *Int. Tinnitus J.*, vol. 11, no. 1, pp. 52–53, 2005.

[17] R. H. Wurtz, H. Komatsu, D. S. Yamasaki, and M. R. Dursteler., "Cortical visual motion processing for oculomotor control," *Vis. brain*, pp. 211–231, 1990.

[18] M. K. Ohlbaum, "Mechanical resonant frequency of the human eye in vivo," Aerospace Medical Research Laboratory-TR-75-113 Diss., 1976.

[19] M. Panouillères et al., "Effects of structural and functional cerebellar lesions on sensorimotor adaptation of saccades," *Exp. Brain Res.*, vol. 231, no. 1, pp. 1–11, 2013.

[20] R. L. Sacco, P. a Wolf, W. B. Kannel, and P. M. McNamara, "Survival and recurrence following stroke. The Framingham study.," *Stroke*, vol. 13, no. 3, pp. 290–295, 1982.

[21] "Stroke related eye conditions." [Online]. Available: http://www.rnib.org.uk/eye-health-eye-conditions-z-eye-conditions/stroke-related-eye-conditions. [Accessed: 14-Jan-2018].

[22] E. Anagnostou, I. Kouzi, S. Vassilopoulou, G. P. Paraskevas, and K. Spengos, “Spontaneous eyeblink rate in focal cerebrovascular lesions,” *Eur. Neurol.*, vol. 67, no. 1, pp. 39–44, 2012.

[23] A. R. Bentivoglio, S. B. Bressman, E. Cassetta, D. Carretta, P. Tonali, and A. Albanese, “Analysis of blink rate patterns in normal subjects.,” *Mov. Disord.*, vol. 12, no. 6, pp. 1028–34, 1997.

[24] D. Kumar, A. Dutta, A. Das, and U. Lahiri, “SmartEye: Developing a Novel Eye Tracking System for Quantitative Assessment of Oculomotor Abnormalities,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 4320, no. c, pp. 1–1, 2016.

[25] J. Bogousslavsky and O. Meienberg, “Eye-Movement Disorders in Brain-Stem and Cerebellar Stroke,” *Arch. Neurol.*, vol. 44, no. 2, pp. 141–148, 1987.

[26] K. Fukushima, J. Fukushima, T. Warabi, and G. R. Barnes, “Cognitive processes involved in smooth pursuit eye movements: behavioral evidence, neural substrate and clinical correlation,” *Front. Syst. Neurosci.*, vol. 7, 2013.

[27] P. Pouget, “The cortex is in overall control of ‘voluntary’ eye movement,” *Eye (London, England)*, vol. 29, no. 2. pp. 241–245, 2015.

[28] M. Rolfs, “Microsaccades: Small steps on a long way,” *Vision Research*, vol. 49, no. 20. pp. 2415–2441, 2009.

[29] P. De Weerd, *Visual Perception - Fundamentals of Vision: Low and Mid-Level Processes in Perception*, vol. 154. 2006.

[30] G. Tirelli, S. Rigo, F. Bullo, C. Meneguzzi, D. Gregori, and A. Gatto, “Saccades and smooth pursuit eye movements in central vertigo.,” *Acta Otorhinolaryngol. Ital.*, vol. 31, no. 2, pp. 96–102, 2011.

[31] A. Shumway-Cook and M. H. Woollacott, *Motor control: Translating research into clinical practice: Fourth edition*. 2014.

[32] B. D. Berman, S. G. Horovitz, and M. Hallett, “Modulation of functionally localized right insular cortex activity using real-time fMRI-based neurofeedback,” *Front. Hum. Neurosci.*, vol. 7, 2013.

[33] A. Lerner, A. Bagic, T. Hanakawa, E. A. Boudreau, F. Pagan, Z. Mari, W. Bara-Jimenez, M. Aksu, S. Sato, D. L. Murphy, and M. Hallett, “Involvement of insula and cingulate cortices in control and suppression of natural urges,” *Cereb. Cortex*, vol. 19, no. 1, pp. 218–223, 2009.

[34] R. L. Sacco et al., “An updated definition of stroke for the 21st century,” *Stroke*, vol. 44, pp. 1–26, 2013.

[35] G. Gulli, HS. Markus, "The use of FAST and ABCD2 scores in posterior circulation, compared with anterior circulation, stroke and transient ischemic attack", J Neurol Neurosurg Psychiatry, vol. 83, pp. 228-9, 2012.

[36] A. A. Tarnutzer, A. L. Berkowitz, K. A. Robinson, Y. H. Hsieh, and D. E. Newman-Toker, “Does my dizzy patient have a stroke? A systematic review of bedside diagnosis in acute vestibular syndrome,” *CMAJ*, vol. 183, no. 9. 2011.

[37] T. C. Hain, "Brainstem strokes associated with vertigo or hearing symptoms. [Available online: www. dizziness-and-balance. com/disorders/central/strokes/brainstem% 20strokes. htm, Accessed May 16 (2014)].

[38] Á. Merwick and D. Werring, “Posterior circulation ischaemic stroke.,” *BMJ*, vol. 348, no. May, p. g3175, 2014.

[39] C. Wu et al., “Oculomotor deficits in spinocerebellar ataxia type 3: Potential biomarkers

of preclinical detection and disease progression," *CNS Neurosci. Ther.*, vol. 23, no. 4, pp. 321–328, 2017.

[40] E. H. Pinkhardt and J. Kassubek, "Ocular motor abnormalities in Parkinsonian syndromes," *Parkinsonism and Related Disorders*, vol. 17, no. 4. pp. 223–230, 2011.

[41] "The human balance system." [Online]. Available: http://vestibular.org/understanding-vestibular-disorder/human-balance-system. [Accessed: 13-Jan-2018].

[42] R. J. Peterka and M. S. Benolken, "Role of somatosensory and vestibular cues in attenuating visually induced human postural sway," *Exp. Brain Res.*, vol. 105, no. 1, pp. 101–110, 1995.

CHAPTER 4

VIRTUAL REALITY-BASED CENTER OF PRESSURE ASSISTED BALANCE TRAINING SYSTEM

4.1 Introduction

Human balance is a complex set of sensorimotor control systems that require sensory input from vision (sight), proprioception (touch), and the vestibular system (motion, equilibrium, spatial orientation) [1], [2]. Subsequently, based on these sensory inputs, the central nervous system activates the muscles that in turn cause eyes and other body appendages to move so as to maintain the balance [1], [2]. There is evidence from literature that states the role of one's vision, vestibular and somatosensory systems to maintain balance [3], [4]. In the current study, I wanted to address the issue of post-stroke balance disorder by designing technology-assisted balance training system that can expose the post-stroke hemiplegic participants to Virtual Reality (VR)-based weight-shifting tasks. In this chapter, I address the research question on understanding the implication of intelligent adaptive VR-based Balance training platform on one's weight-shifting capability.

To answer this research question, I have developed different VR-based balance training platforms while utilizing one's Center of Pressure (CoP) and Center of Mass (CoM) during weight-shifting tasks. In this chapter, I will discuss the design of balance training platform that used the CoP. The CoP allows easy assessment of one's balance, particularly important in the context of geriatric balance as used in my present study. Here, I present (i) the design of CoP-assisted **V**irtual Reality based **Ba**lance **T**raining (**VBaT**) platform and (ii) the results of a usability study of the VBaT system with hemiplegic stroke participants. The target population for this study was post-stroke individuals who were suffering from balance disorders.

I chose this target population, since, stroke is the fourth leading cause of disability worldwide [5] and is often responsible for impaired motor function. The incidence of stroke has been an epidemic in the developing countries like India with a prevalence rate of stroke being 84-262/100,000 in rural and 334-424/100,000 in urban areas [6]**.** Over the last four decades, developed countries have shown 42% decrement in stroke cases whereas, developing countries have seen greater than 100% increase in stroke incidence [7]. Literature indicates that up to 85% of stroke patients end up with hemiparesis and approximately 55%-75% of stroke patients continue to suffer from motor deficits along with impaired balance [8]. Post-stroke balance deficits can occur due to several factors such as loss of muscle strength, reduced joint movement, impaired sensory information and motor control [9]. Balance-related deficit often causes asymmetric weight distribution in post-stroke hemiplegic patients. Researchers have shown that the weight bearing ability of hemiplegic patients can be reduced by up to 43% on the paretic (henceforth Affected) side of the lower limb thereby making them vulnerable to falls during ambulation [10], [11]. The incidence of falls has been reported to be up to 73% in the first year post-stroke [12]. The after-effects of falls can be lacerations, hip fractures and head traumas that are seen in around 20 to 30% of such individuals [13]. As a result, stroke patients with impaired balance often suffer from fear of falling, reduction in self-confidence, and reduced functional independence which in turn adversely affect their social wellbeing [11].

During the early stages after stroke incidence, balance rehabilitation programs have shown to be effective in addressing balance impairment and mobility [14]. Post-stroke functional recovery of an individual depends on the neuronal plasticity of his/her brain that allows different areas of the brain to take over the functions controlled by the affected zone, where the chances of this neural reorganization strongly depend on the severity of the lesion [15]. Therefore, rehabilitative systems should be designed with an aim to induce cortical reorganization that helps in functional recovery [8]. Though, conventional techniques of balance rehabilitation have shown promising outcomes [11], yet, these techniques often suffer from the requirement of one-to-one supervision tolling on the therapist's time and monotony of rehabilitation tasks due to repetitive movement exercise without any variations. As a result, stroke patients show poor engagement and reduced interest while performing balance rehabilitation exercises [16], [17]. Also, there exist problems related to scarcity of adequately trained clinicians [11], high cost of one-to-one services in specialized healthcare settings [18] among others.

Given these limitations, researchers have started to explore the possibility of using alternate technology-assisted solutions, such as robot-based, Virtual Reality (VR) based [19], [20], [16] and others. In the current study, I have chosen VR-based system due to its inherent advantage of offering an individualized, inexpensive, safe, flexible, interactive practice environment with variations to avoid monotony [21] and making it motivational [17] for the users. Specifically, variations can be introduced in the VR environment through mimicking real-world scenarios by developing realistic and interactive tasks. The flexibility in the design of VR-based exercise platforms allow us to incorporate physiotherapist's inputs while designing the balance training tasks. Also, the VR-based autonomous exercise platform can be flexibly tuned with physiotherapist's input based on one's individualized residual balance capability, thereby allowing a physiotherapist to administer exercise for multiple patients at the same time. Another advantage is that such VR-based cost-effective training platforms can be used in home-based settings for the patients to exercise without the need to commute to health centers for availing rehabilitation sessions [11]. With rapid technological progress, the VR-based rehabilitation exercise units do not work in standalone modes, but instead couple with peripheral devices to offer augmented stroke rehabilitation platforms [21]. For example, VR allows the simulated world to be interfaced with peripheral devices, such as Kinect, Balance Board [22] while designing exercise platforms to address balance disorders [16].

While standing, one's balance is mainly addressed by three postural strategies, namely, Ankle, Hip and Step strategies [22]. All of these strategies are useful in maintaining one's stable posture depending on the type of activity one performs. Among these strategies, the Ankle strategy that uses muscle contraction of one's ankle joint is most commonly used to improve the standing balance [23]. While following Ankle strategy, an individual's CoP is controlled by ankle joint when his/her head and hip move in the same direction within the limits of stability (LOS) [6]. The LoS is defined as the amount of maximum excursion an individual can intentionally cover in a particular direction without losing his/her balance or taking a step [9]. Researchers have been using one's CoP as an index of postural stability while standing [24]. The CoP is the point of application of the ground reaction force vector while one stands on a base. It represents a weighted average of all the pressures applied by the individual over the surface of the base in contact with the ground [25]. Studies on upright posture use one's ''body sway'' to characterize the performance. One way of estimating body sway is from the trajectory of one's CoP that can be obtained by using force platform data [26]. Researchers have shown that the low-cost force plates, such as Nintendo Wii balance board (WiiBB) ca be reliably used for assessing an individual's balance in clinical settings [27]. Therefore, in the current study, I integrated Nintendo WiiBB (a low-cost portable force platform) with a VR-based platform to provide real-time visual feedback of the individual’s CoP (measured by WiiBB) that was integrated with various context-relevant virtual objects in the VR environments.

From literature review, I find that there are various studies supporting the use of computer-based games integrated with WiiBB to address one's balance impairment [8], [11], [16], [17], [22], [28], [29]. For example, Gil-Gómez et al. [17] designed a VR-based system integrated with WiiBB (eBaVir) for balance rehabilitation. In the eBaVir system, they offered three different games to the stroke participants in each session. However, their system was not individualized to be tuned to the participant’s ability to perform the tasks. In other words, the games did not offer different challenge levels and so the task difficulty was not adaptive to individual's performance during balance training. Also, most of the research studies [11], [16], [16], [29] used off-the-shelf games designed from an entertainment perspective to be used by healthy individuals, and not for individuals with balance disorders. Additionally, they did not have any rule engine that could present balance training tasks of varying difficulty in a systematic, controlled and adaptive manner depending on the user’s performance, essential for effective balance rehabilitation.

In my current study, the objectives were two fold, namely, (i) develop a VR-based Balance Training (VBaT) platform in which a VR-based system augmented with a WiiBB can provide balance training exercises in an individualized and adaptive manner based on an individual’s performance capabilities and (ii) conduct a usability study with VBaT system to understand the implications of VBaT system on the performance score of individuals having balance impairment. The VBaT system facilitated the participants by alarming them if Ankle strategy was not followed during weight-shifting task. For this, I fabricated a Heel Lift Detection (HLD) unit to detect an indivdual’s incorrect posture (that is, lifting of the heel from the surface of WiiBB) during weight-shifting based balance training task. To follow the Ankle strategy during weight-shifting, one's foot should remain in contact with the Base of Support (BoS) [9]. The VBaT system was programmed to offer users a variety of VR-based balance tasks with various levels of challenge in a controlled and systematic manner. The tasks required a participant to shift his weight in different directions (North, East, West, North-east, and North-west) while following the Ankle strategy.

4.2 System Design

The VBaT system consisted of five units, namely, (a) VR-based task (b) Heel Lift Detection (c) WiiBB-VR handshake (d) Performance evaluation and (e) Task switching units.

4.2.1 VR-based Task unit

In this study, I have developed VR-based balance training tasks to quantitatively estimate an individual's balance ability during weight-shifting. The VR-based balance tasks required participants to shift their weight in different directions. The idea was to simulate real-life directional weight-shifting similar to our daily life activities that often require weight-shifting in different directions to be able to perform tasks such as reaching tasks while standing at a particular position. Execution of such tasks might be difficult for hemiplegic stroke patients who often show tendencies to fall [30] possibly due to asymmetric biped balance. Practicing bipedal weight-shift exercise is critical for improved balance. Thus, in this study, I have designed VR-based tasks that needed effective use of both the lower limbs to perform a task. Physiotherapists often recommend balance training exercises that last for about 20 minutes [16] consisting of passive and active range of motion, static and dynamic balance, stretching, muscle strengthening, gait training and activities of daily living exercises. Here, I timed the VR-based balance training system to offer tasks that ran for a duration of about 20 minutes. I used Vizard software toolkit (from Worldviz Llc.) to design a variety of realistic VR environments to make the weight-shifting exercise session interesting for the participants.

The VR-based balance tasks needed a participant to maneuver virtual objects (VR_{Obj} henceforth) in the VR environment (shown on the Task Computer monitor) by the excursion of their CoP through shifting of weight while standing on the WiiBB. The raw CoP values recorded from WiiBB at the rate of 30 Hz were processed by a five-point moving average filter. The position of the virtual object was controlled from filtered CoP data by using equation (4.1).

$$\begin{bmatrix} x \\ y \end{bmatrix}_{VR_{obj}} = \begin{bmatrix} CoP_x & 0 \\ 0 & CoP_y \end{bmatrix} \begin{bmatrix} \epsilon_1 \\ \epsilon_2 \end{bmatrix} \tag{4.1}$$

Where, $\boldsymbol{\epsilon_1}$ and $\boldsymbol{\epsilon_2}$ are constants used for real-time mapping of the CoP position obtained from WiiBB to the position of VR_{Obj} on the monitor of the Task Computer. There was no perceptible visual lag between the CoP position acquired from WiiBB and corresponding movement of VR_{obj} on the Task Computer monitor during the VBaT task. Additionally, I programmed VBaT system to

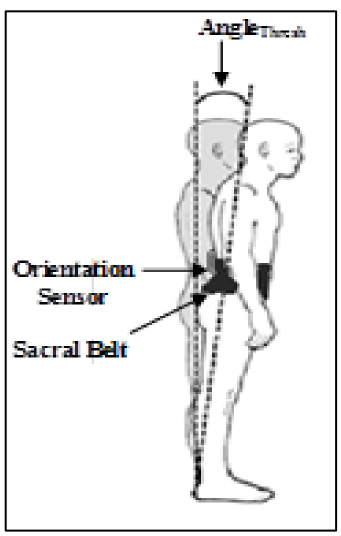


Figure 4.1. Weight-shifting using Ankle strategy.

provide audio-visual feedback to the participants based on their performance in a task. The VR-based balance training tasks were of four difficulty levels (DL1-DL4) with the tasks of DL1 and

DL4 being least and most difficult respectively. This was because, taking inputs from Flow Theory of game design [31], a task should not be too easy to bring in a feeling of boredom in the participant neither it should be too hard for the participant to be frustrated. Thus, the idea was to offer the VR-based tasks with varying challenges while being adaptive to the individualized weight-shifting capability. However, we designed tasks of varying challenges so that the participants felt motivated (devoid of any boredom) and at the same time be offered with tasks of higher challenges only when they were capable thereby avoiding any element of frustration. The difficulty of the tasks was decided based on the threshold angle of shifted weight from the vertical direction (that is, $Angle_{Thresh}$ for North as shown in Fig. 4.1). Also, tasks belonging to each difficulty level had an associated threshold task completion time ($Time_{Thresh}$) (Table 4.1). These $Angle_{Thresh}$ and $Time_{Thresh}$ values were obtained from a pilot study carried out with age-matched healthy participants (age-matched with the stroke participant pool). The VR-based

Table 4.1. Threshold angle and time of weight-shifting

	Direction	$Angle_{Thresh}$ (degrees)	$Time_{Thresh}$ (sec)
DL1	North	3.38	12
	East	5.11	12
	West	5.11	12
	North-East	4.245	12
	North-West	4.245	12
DL2	North	5.29	12
	East	8.09	12
	West	8.09	12
	North-East	6.69	12
	North-West	6.69	12
DL3 and DL4	North	5.29	60/Template
	East	8.09	
	West	8.09	

balance training tasks (Fig. 4.2) were designed using Google Sketchup and imported into VR environment (Vizard platform). For maneuvering the VR_{Obj} in the VR environment, I have used scaling factors of $\in_1$and $\in_2$ (Eq. 4.1) (to map the CoP data to VR-based world coordinates) that were (2.45, 4.51), (1.72, 3.16), (0.26, 0.48), and (1.52, 2.72) for DL1, DL2, DL3 and DL4 respectively.

In the pilot study, the healthy participants were asked to perform VR-based tasks in each difficulty level by shifting their weight in different directions while standing on the WiiBB and following the Ankle strategy. During the study, each participant was asked to wear a Sacral Belt fitted with an android phone (with the orientation sensor application program) placed on the Sacrum. When a participant shifted his weight to perform a task, an orientation sensor application program running on the android phone provided the orientation angles (roll, yaw, and pitch) in degrees. The $Angle_{Thresh}$ with the vertical was estimated from the pitch angle for 'North' direction, roll for 'East' and 'West' directions and the combination of roll and pitch angle for 'North-East' and 'North-West' directions.

4.2.1.1 Design of VR-based Tasks of Difficulty Level 1 (DL1)

A typical Graphical User Interface (GUI) developed for the VR-based task of DL1 is shown in Fig. 4.2 (a). I created a database of 30 unique combinations of such VR-based environments and VR_{Obj} related to tasks of daily activity so that the tasks can also be entertaining. The tasks needed the participants to shift their weight in different directions from the central position ($Central_{Hold}$, henceforth) that corresponded to their standing straight (upright) posture without shifting weight while balancing on both legs. To move the VR_{Obj} in the VR environment from $Central_{Hold}$ state to a predefined target position, the participants were asked to shift their weight in five different directions, namely, 'North,' 'East,' 'West,' 'North-east,' and 'North-west'. The VR_{Obj}, Target object and VR environment were chosen randomly from the developed database. Also, the tasks required the participants to hold the VR_{Obj} at the Target position for one second. The idea was to

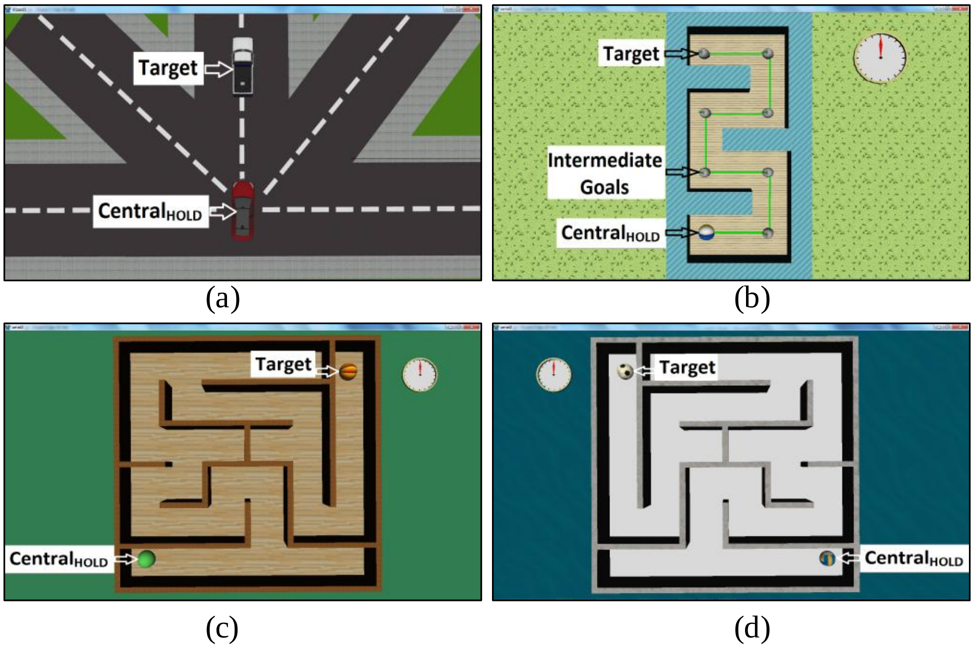


Figure 4.2. (a) Graphical User Interface (GUI) applicable for for DL1and DL2 tasks, (b) GUI of DL3 task, (c) GUI of DL4 task (Template$_{Right}$), (d) GUI of DL4 task (Template$_{Left}$)

understand whether the participants were able to maintain their shifted weight at each angular position. The tasks of DL1 were kept fairly easy, in which a participant had to shift his/her weight by a minimum displacement quantified by the smallest value of $Angle_{Thresh}$ (Table 4.1). The task in DL1 began with the VR_{Obj} at the $Central_{Hold}$ position and the Target object presented at the end (Target position) of one of the virtual paths pointing towards one of the five directions (Fig. 4.2 (a) shows Target position towards 'North') along with directional cue (in the form of an arrow pointer). The participant had to maneuver VR_{Obj} by shifting his/her weight in the direction of the Target object. Once the position of VR_{Obj} went beyond certain threshold distance (corresponding to $Angle_{Thresh}$ (Table 4.1)) in the VR environment, the VR_{Obj} was programmed to automatically shoot towards the Target position. Also, once the participant was able to take the VR_{Obj} to Target position, he/she was expected to hold his/her posture with shifted weight for 1 sec with the VR_{Obj} remaining at the Target position for 1 sec. If the participant was able to do that, then the VBaT system provided an audio feedback with a 'ting' sound to let him/her know that the task was complete. The maximum time allowed to complete the task was 12 sec. ($Time_{Thresh}$ (Table 4.1)). If a participant could not complete the task within that time, the VR_{Obj} was relocated back to the $Central_{Hold}$ position, and a performance score of zero was awarded to the participant for that specific direction. Likewise, the participant was asked to execute the tasks for the remaining four directions that were randomly presented. At the end of each task trial, the VBaT system provided audio-visual feedback, such as "Well done! You are doing great" if the task completion was successful or "Keep trying, you can do better" otherwise.

4.2.1.2 Design of VR-based Tasks of Difficulty Level 2 (DL2)

The task template and the directional cues for DL2 were similar to those for DL1. However, to make the task more difficult, tasks of the DL2 level required a greater $Angle_{Thresh}$ (Table 4.1) thereby requiring the participants to shift their weight by greater amount.

4.2.1.3 Design of VR-based Tasks of Difficulty Level 3 (DL3)

Fig. 4.2 (b) shows a typical GUI for a task of DL3 which was a maze-like structure. The maze structure consisted of 7 segments with intermediate goal position at the end of each segment. The task was to maneuver the VR_{Obj} from the $Central_{Hold}$ position to the final goal position kept at the end of the seventh segment in the maze-like path while traversing through the intermediate goal positions (as shown in Fig. 4.2 (b)) within a specified duration (Table 4.1). To perform this task, the participants were required to shift their weight starting from $Central_{Hold}$ position towards ‘North’, ‘East’ and ‘West’ directions to maneuver the VR_{Obj} in the task environment. Also, once the participant maneuvered the VR_{Obj} (that was integrated with the CoP) to an intermediate goal position, the VR_{Obj} was kept stationary at that position (achieved by disintegrating the VR_{obj} from the CoP) until the participant returned back to the $Central_{Hold}$ position. As soon as the participant returned back to the $Central_{Hold}$ position, he/she was allowed to maneuver the VR_{Obj} towards the next intermediate goal in the maze-like path. Thus, this task required a participant to (i) plan and (ii) dynamically shift weight to maneuver the VR_{obj} from a $Central_{Hold}$ position to the Target position while traversing through intermediate goal positions in the maze-like path. The $Angle_{Thresh}$ used for DL3 was same as that for DL2 (Table 4.1). Unlike DL1 and DL2, in DL3 tasks, the participants were not presented with any directional cue by the VBaT system. Also, a timer was added to the environment (Fig. 4.2 (b)) for the participants to keep track of their time. Here the participant was expected to look at the maze-like path with intermediate goal positions presented on the Task Computer screen and decide his/her direction and speed of weight-shifting. Specifically, the participant was expected to decide the direction in which he/she would need to shift his weight and also the speed of his weight-shifting action so as to maneuver the VR_{obj} from the $Central_{Hold}$ to the final Target position within the allotted $Time_{Thresh}$ (Table 4.1).

4.2.1.4 Design of VR-based Tasks of Difficulty Level 4 (DL4)

Figs. 4.2 (c) and 4.2 (d) show examples of GUI designed for balance training tasks belonging to DL4. Similar to DL3, the task of DL4 also required a participant to plan and shift his/her weight so as to maneuver a VR_{Obj} along the complex maze-like path. However, compared to DL3 task, in DL4 task, (i) the complexity of maze-like path was increased and (ii) there were no intermediate goal positions to be reached. Here, the participant's CoP position was integrated to the VR_{Obj} throughout the task, unlike that in DL3. If the participant returned back to the $Central_{Hold}$ position then the VR_{Obj} would also trace back to the $Central_{Hold}$ position which would delay the task completion. Instead, the participant was expected to maneuver the VR_{Obj} from $Central_{Hold}$ position to Target position in the same go. Here, I designed two templates for each task, namely, (a) $Template_{Right}$: one in which the start position was at the bottom left corner (corresponding to $Central_{Hold}$) and Target position at top right corner and (b) $Template_{Left}$: one in which the start position was at the bottom right corner (corresponding to $Central_{Hold}$) and Target position at the top left corner. For $Template_{Right}$, the participant needed to maneuver the VR_{Obj} in a path having combination of ‘North’, ‘East’ and ‘North-East’ directions with the Target position being fixed towards ‘North-East’ with respect to the $Central_{Hold}$ position. Whereas for $Template_{Left}$, the participant needed to maneuver the VR_{Obj} in a path having combination of ‘North’, ‘West’ and ‘North-West’ directions with the Target position being fixed towards ‘North-West’ with respect to the $Central_{Hold}$ position. The order in which the two task templates of DL4 were presented to the participants was random. Similar to DL3, a timer was added in DL4 (Figs. 4.2 (c) and 4.2 (d)) so that the participants can keep track of their time while taking part in the task. For DL4, the $Angle_{Thresh}$ was similar to DL2 and DL3 (Table 4.1) and the width of the complex maze-like path was lesser than DL3 to add to the task difficulty.

4.2.2 Heel Lift Detection unit

In this study, I wanted to monitor one’s ability to shift weight in different directions by displacing the CoP while standing on the WiiBB and maneuvering the VR_{Obj} by following Ankle strategy. To follow the Ankle strategy during weight-shifting task, the heel should not be lifted from the base of support (in this case, the surface of the WiiBB). Therefore, I developed a Heel

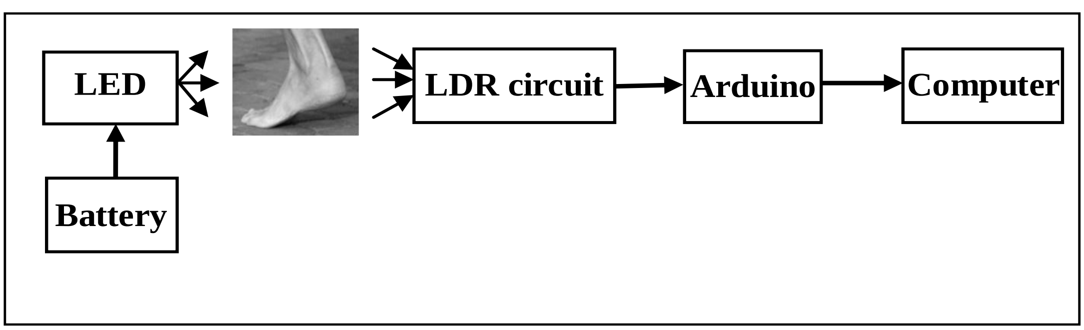


Figure 4.3. Block schematic of Heel Lift Detection unit for VBaT system

Lift Detection (HLD) unit to monitor the participant’s heel position during tasks so as to help him/her in following the Ankle strategy while shifting weight. Though many studies refer to use of Ankle strategy [22], [23], yet none have come up with an HLD unit that can be integrated with the VR-based environment. In my present study, I designed an HLD unit that was novel. The HLD unit consisted of a 5-watt light emitting diode (LED) source, a light dependent resistor (LDR) and an Arduino board. The HLD unit was setup in such a way that when one stood upright on the WiiBB, with the heels in contact with the surface of the board, the heels obstructed the light coming from the light source (that is, LED) to the detector (that is, LDR) and caused a high resistance of the LDR that sent a higher voltage output to the analog input pin of the Arduino board. On the other hand, if the participant's heel was lifted from the surface of the WiiBB, then the light from the source was incident on the detector that caused the resistance of the LDR to fall with subsequent reduction in the voltage input to the analog pin. Then, the output from the LDR was processed by the microcontroller of the Arduino board (Fig. 4.3) and transferred as a flag to the task execution routine running on the Task Computer.

4.2.3 Handshaking Unit Integrating WiiBB with VR platform

Nintendo WiiBB was connected to the Task Computer executing the VR-based tasks via Bluetooth connectivity by using licensed version of BlueSoleil 10.0.483.0 software. A

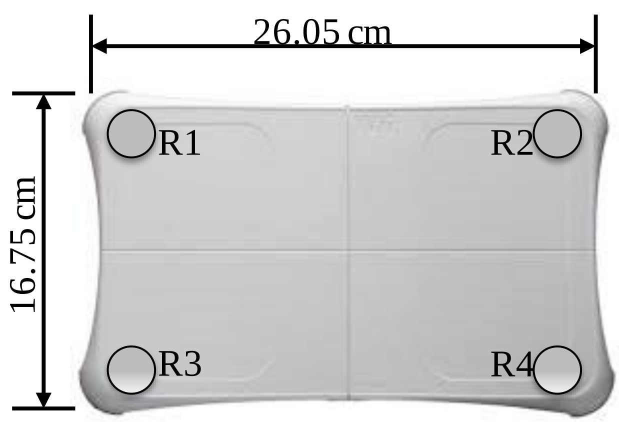


Figure 4.4. Wii Balance Board

customized Matlab-based script [28] was used to interface the WiiBB with the VR environment that accessed data from the WiiBB at a sampling rate of 30 Hz. The data provided by the WiiBB consisted of the force values from the Back Left (R_1), Back Right (R_2), Front Left (R_3) and Front Right (R_4) pressure sensors of the WiiBB (Fig. 4.4). Using these values, I computed the *x* and *y* coordinates of the CoP measure (CoP_x, CoP_y) (Eqs. 4.2 and 4.3).

$$CoP_x(\text{in cm}) = 26.05*(R_2 + R_4 – R_1 – R_3)/(R_1 + R_2 + R_3 + R_4) \tag{4.2}$$

$$CoP_Y\ (\text{in cm}) = 16.75*(R_1 - R_3 + R_2 - R_4)/(R_1 + R_2 + R_3 + R_4) \tag{4.3}$$

The constants '26.05' and '16.75' in Eqs. (4.2) and (4.3) represent the physical dimension of the WiiBB.

4.2.4 Performance Evaluation Unit

While the participants performed the VR-based weight-shifting tasks, the VBaT system computed their performance scores for tasks of each difficulty level using different performance metrics. In literature, researchers have used CoP related metrics such as CoP sway distance, sway area, sway velocity and sway angle to assess one's balance [21], [32]. Here I chose one's sway distance quantified in terms of the length of CoP trajectory, and sway angle quantified as a deviation in the direction of shifted weight from the direction of Target for tasks belonging to different difficulty levels. Additionally, I also used three other performance metrics namely, ability to hold shifted weight (a measure of postural stability), the total time taken in a task, and ability to shift weight by following Ankle strategy. The threshold values for different performance metrics were decided based on a pilot study and the feedback from a physiotherapist in the team.

4.2.4.1 Performance Evaluation Criteria for Tasks of DL1

The design of the tasks of DL1 is presented in Section 4.2.1.1. For the tasks of DL1, I evaluated the score in each of the five directions separately. The score was based on four performance metrics, namely, (i) P_{S1}: length of trajectory (T_L) of participant's CoP while standing on the WiiBB before reaching to the $Angle_{Thresh}$ of weight shifting, (ii) P_{S2}: deviation (D_A) of participant's shifted weight from the instructed straight path between the $Central_{Hold}$ and the Target positions, (iii) P_{S3}: one's ability to hold his/her shifted weight at the Target position for 1 second (Hold time (H_T)), and (iv) P_{S4}: penalizing factor if the participant did not follow Ankle strategy while shifting weight.

The first metric (P_{S1}) was a measure of an individual's body sway in terms of his/her CoP trajectory before reaching the $Angle_{Thresh}$. While the participant maneuvered the VR_{Obj} in the VR environment, based on his/her body sway in random directions, he/she was penalized as follows,

$$P_{S1} = \begin{cases} 100 & ;\ if\ T_L \leq D_{TH} \\ 100 - \propto * \left(\frac{T_L - D_{TH}}{D_{TH}}\right) * 100 & ;if\ D_{TH} < T_L < 3 * D_{TH} \\ 0 & ;\ T_L \geq 3 * D_{TH} \end{cases} \tag{4.4}$$

where, D_{TH} (Threshold distance) = 1.8 * the length of the straight line path between the $Central_{Hold}$ and the Target positions in each direction. The value of D_{TH} was decided based on the pilot study with age-matched healthy participants. In this pilot study, while an individual performed the tasks of DL1, I computed the length of the CoP trajectory before reaching the Target position for each of the five directions. Subsequently, the value of D_{TH} was computed by taking average of the distance travelled by the individual's CoP in different directions. It was found to be 1.8 * the length of the straight-line path between the start and Target positions. The factor α was chosen as ½. This was because in Eq. (4.4), the parameter P_{S1} can have three possible values depending on the value of T_L. For example, if $T_L < D_{TH}$ and $T_L \geq 3*D_{TH}$, the P_{S1} score would be 100 and 0, respectively. Whereas, for the intermediate values of T_L, I used a multiplication factor of α=½ so that the penalty factor due to increase in the value of T_L between

D_{TH} and $3{*}D_{TH}$ can be linearized. The range of the values of T_L (as function of D_{TH}) was chosen as an initial approximation. This can be changed based on the study design.

The second metric (P_{S2}) was used to evaluate the quality of participant's weight-shifting. This was measured in terms of deviation (D_A) of the position of VR_{Obj} from the instructed direction (defined by $\theta_x = 0^0$ for East; 45^0 for North-East; 90^0 for North; 135^0 for North-West; and 180^0 for West) with a tolerance range (θ_{RANGE}) of $\pm 22.5^0$ around the instructed direction.

$$P_{S2} = 100 - \propto * \left(\frac{abs(\theta_x - D_A)}{\theta_{RANGE}}\right) * 100 \qquad (4.5)$$

Here, α =½.

The third metric (P_{S3}) was used to encourage a participant to hold the shifted weight for 1 second that can help the participant to make a stable weight-shifting.

$$P_{S3} = \begin{cases} 100 & ;\ if\ H_T > 1s \\ 0 & ;\ if H_T < 1s \end{cases} \qquad (4.6)$$

The fourth metric (P_{S4}) was used to encourage the participants to follow the Ankle strategy while performing weight-shifting tasks.

$$P_{S4} = \begin{cases} -100 & ;\ for\ heel\ lifting \\ 0 & ;\ otherwise \end{cases} \qquad (4.7)$$

After computing the different performance metric scores for each direction, the weighted performance score (P_S^x) for each of the five directions (x=East; North-East; North; North-West; West) was calculated as

$$P_S^x = 0.5P_{S1}^x + 0.25P_{S2}^x + 0.25P_{S3}^x + 0.2P_{S4}^x \qquad (4.8)$$

The range of P_S^x score was kept as 0-100 and if P_S^x was negative, then it was rounded off to zero. Here, I aimed to quantify the participant's performance in terms of parameters such as (a) reduced swaying, (b) reduced deviation from the instructed direction, (c) increased ability to hold the posture with the shifted-weight and (d) Ankle strategy being followed/not followed. As advised by the therapist, I gave more weightage (0.5) to parameter (a), and less to parameters (b) and (c). I wanted to understand the implication of VBaT system in improving one's balance. An important indicator to assess such an improvement at the initial stages of balance training exercise, is the ability to shift weight with reduced body sway that is, controlled profile of CoP trajectory (captured through P_{S1}) while shifting weight. Once an individual is able to do controlled weight shifting, then only he/she can work on the finer aspects, such as reaching as close as possible to the Target location (captured through P_{S2}) and hold the posture of the shifted weight (captured through P_{S3}). Thus, as suggested by the therapist, I assigned more weightage to P_{S1} than P_{S2} and P_{S3}. Additionally, I wanted that the participants should follow the Ankle strategy during weight-shifting tasks. A penalty factor (0.2) was used if the participant lifted his heel during the tasks. The factor of 0.2 (chosen as an initial approximation) was used to discourage the participant from lifting his/her heel during shifting weight. Here we chose a lesser penalty factor compared to that in DL3 and DL4 (discussed below) since, DL1 and DL2 were of lower difficulty levels. Also, since there is no previous literature on the amount of penalty that can considered in the participant's performance due to not following Ankle strategy, we started with a comparatively lower penalty factor of 0.2. It can be changed based on the requirements of the study.

The final performance score (P_s) for DL1 was average of the performance scores for all the directions as follows:

$$P_S = \frac{1}{5}\sum_{x} P_S^x \tag{4.9}$$

4.2.4.2 Performance Evaluation Criteria for Tasks of DL2

The design of the tasks of DL2 is presented in Section 4.2.1.2. For tasks of DL2, I used the same metrics (described in Section 4.2.4.1) for computing the participant's performance score.

4.2.4.3 Performance Evaluation Criteria for Tasks of DL3

The design of the tasks of DL3 is presented in Section 4.2.1.3. As shown in the Fig. 4.2 (b), the tasks of DL3 used a maze-like path consisting of a combination of 7 different segments (3 segments pointing upwards from 'South' to 'North' and two each towards the right and left directions that is, 'East' and 'West' directions). The task required a participant to maneuver the VR_{Obj} along the maze-like path while traversing through intermediate goal position placed at the end of each segment. For this task, I evaluated the performance score using three performance metrics, namely (i) P_{S5}: length of trajectory (T_L) of participant's CoP (integrated with VR_{Obj}) between $Central_{Hold}$ and the intermediate goal positions (that corresponded to the participant's shifted weight to reach the $Angle_{Thresh}$) for each segment (ii) P_{S6}: total time taken to complete the task, and (iii) P_{S7}: penalizing factor in case the participant did not follow Ankle strategy while shifting weight towards the intermediate goal positions.

The first metric (P_{S5}) was evaluated separately for each of the seven path segments. The other two metrics were calculated for the entire path consisting of a total of 7 segments. Thus, P_{S5_i} (i=1 to 7) was calculated similar to P_{S1} (Section 4.2.4.1) and subsequently, the mean value of P_{S5} was calculated, by using

$$P_{S5} = \frac{1}{7}\sum_{i=1}^{7} P_{S5_i} \qquad (4.10)$$

The second metric (P_{S6}) was computed as

$$P_{S6} = \begin{cases} 100 & ; if T_{CT} \leq T_{TH} \\ 100 - \alpha * \left(\frac{T_{CT}-T_{TH}}{T_{TH}}\right) * 100 & ; if\ T_{TH} < T_{CT} < 3 * T_{TH} \\ 0 & ; if\ T_{CT} \geq 3 * T_{TH} \end{cases} \quad (4.11)$$

Where, T_{TH} (=20 seconds) was the minimum threshold time to complete a task to achieve maximum possible score; T_{CT} was the total time taken by the participant to complete the task. The value of T_{TH} was decided based on the pilot trial with age-matched healthy participants. The value of α was chosen as ½ so that the penalty factor due to increase in the value of T_{CT} between T_{TH} and $3*T_{TH}$ can be linearized.

To make the DL3 tasks more challenging than the previous tasks of DL1 and DL2, the penalty factor due to a participant's lifting was made sensitive to the duration of heel lift from the surface of the WiiBB. This was because, once a participant reached DL3 after having exposure to tasks of DL1 and DL2, we wanted to get an idea on the proportion of interaction time for which he/she has lifted his/her heel unlike only considering whether he/she has lifted his/her heel during the task (that was true in the case of DL1 and DL2). Here the penalty factor for lifting the heel from the surface of WiiBB during weight-shifting was made proportional to the amount of time a participant lifted his heel (T_{Lift}) during the task. Thus, P_{S7} was calculated as

$$P_{S7} = \frac{T_{Lift}}{T_{CT}} \times 100 \quad (4.12)$$

The final weighted performance score (P_s) for a task of DL3 was calculated as

$$P_S = 0.7P_{S5} + 0.3P_{S6} - P_{S7} \quad (4.13)$$

If P_S was negative, then it was rounded off to zero. As suggested by the therapist, while computing a participant's performance score for the task of DL3, I provided increased weightage (0.7) to the maneuvering capability (P_{S5}) and reduced weightage (0.3) to the maneuvering speed (P_{S6}). This was because, in the early stage of balance rehabilitation, the ability to complete a task is often more critical than the speed with which one can perform the task. Additionally, an increased penalty factor (P_{S7}) was introduced to discourage the participants from lifting heel during the weight-shifting tasks, since a participant was expected to have already practiced weight-shifting in different directions by following Ankle strategy while interacting with previous tasks of lower difficulty such as DL1 and DL2.

4.2.4.4 Performance Evaluation Criteria for Tasks of DL4

The design of the tasks of DL4 is presented in Section 4.2.1.4. As discussed in Section 4.2.1.4, each task of DL4 consisted of two templates ($Template_{Left}$ and $Template_{Right}$). These tasks were designed to understand how effectively a hemiplegic participant can use both of his lower limbs while executing tasks. The participant was required to complete both the templates ($Template_{Left}$ and $Template_{Right}$) within a fixed duration (Table 4.1). The performance score was calculated as the difference between the time taken by the participant to complete both the templates. For example, if T1 and T2 were the times taken by the participant to complete $Template_{Left}$ and $Template_{Right}$ respectively, then the first performance metric, P_{S8} was computed as

$$P_{S8} = \begin{cases} 100 - \alpha * \left(\frac{T2-T1}{T1}\right) * 100 & ; if\ T1 < T2 \\ 100 - \alpha * \left(\frac{T1-T2}{T2}\right) * 100 & ;\ if\ T1 > T2 \end{cases} \tag{4.14}$$

Here, α=½. I have also used performance metric P_{S7} (see Section 4.2.4.3) for Ankle strategy. Thus, the overall performance score (P_s) for a task of DL4 was

$$P_S = P_{S8} - P_{S7} \tag{4.15}$$

4.2.5 Task Switching Unit

In this present study, I designed a usability study with the VBaT system. In this, the participants were exposed to tasks of different difficulty levels (DL1-DL4) with task switching being adaptive to their performance scores that was labeled as ‘Adequate’ or ‘Inadequate’. A participant’s performance in a task was considered as ‘Adequate’ (condition: C1) if he/she was

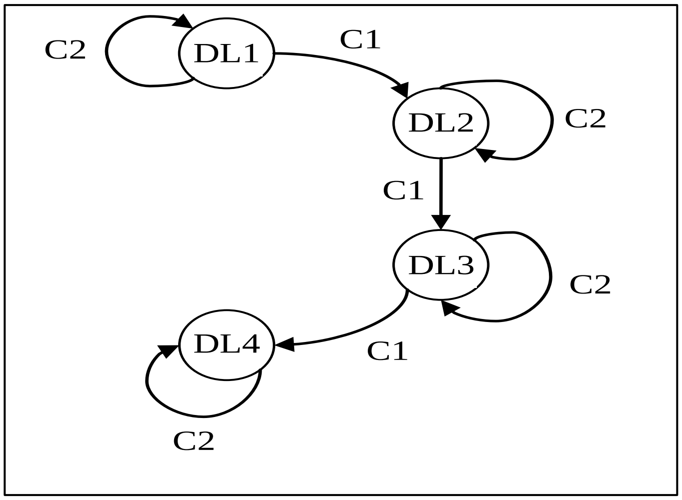


Figure 4.5. State machine based task switching rationale for VBaT system
Note: C1- Adequate Performance; C2 – Inadequate Performance

able to score ≥ 70%, otherwise his performance was considered ‘Inadequate’ (condition: C2). Here, the threshold for 'Adequate' performance score was taken as 70% as an initial approximation. The idea of choosing the threshold of 70% was borrowed from literature that indicates 70% as the average initial exercise performance for robotic rehabilitation tasks [29], for outpatient clinical rehabilitation [33], and technology-based skill learning [34]. This threshold can be changed based on the requirement of the study. Fig. 4.5 shows the state machine representation of task switching (that is order in which tasks of different difficulty levels were offered by VBaT based on the participant’s task performance) used by the VBaT system to offer tasks of different difficulty levels based on the individualized performance score. According to this state machine representation, once a participant achieved ‘Adequate’ performance in a task trial, the VBaT system offered him a task of higher difficulty level. In contrast, if his performance score was ‘Inadequate,’ the VBaT system offered tasks of the same difficulty level so that the participant got more practice to improve his performance with repetitive exercise.

4.3 Experiment and Methods

4.3.1 Participants

The usability study was carried out at Spine institute, civil medical hospital, Ahmedabad where the stroke participants were undergoing treatment. All subjects consented to participate in

Table 4.2. Participants' meta data for usability study with VBaT system

Participant's ID	Age (Years)	Gender	Hemiplegic Side	Post-stroke period	BBS Score
S1	33	Female	Left	1.5 years	46
S2	20	Male	Left	11 months	49
S3	43	Male	Right	9 years	50
S4	70	Male	Right	3 years	50
S5	21	Male	Right	13 months	53
S6	45	Female	Left	4 years	51
S7	52	Male	Right	2 years	51

Note: BBS =Berg Balance Scale

the study, which was approved by the Institutional Ethics Review Committee of Indian Institute of Technology Gandhinagar. In the current usability study, seven hemiplegic stroke participants (S1-S7) (mean (SD)=40.57years(17.73)) with varying residual balance and post-stroke period participated. The participants were enrolled based on their availability. Table 4.2 shows the participants' meta data. The inclusion criteria to participate in this study were (1) post-stroke period>6 months (2) ability to follow instructions (3) ability to stand for at least 20 minutes without orthopedic aid (4) Berg Balance Scale (BBS) score > 40 (measured by a physiotherapist in the team) and (5) should not have gone through any surgery in recent past that may interfere with their capability to do the weight-shifting tasks.

4.3.2 Experimental Setup

The experimental setup shown in Fig. 4.6 (a) consisted of a (i) WiiBB (ii) pair of slippers (iii) HLD unit and (iv) Task Computer (PC) with a 2-D computer monitor executing VR-based tasks (Fig. 4.6 (b)). A participant was asked to stand on the WiiBB fitted with slippers (kept in front of the Task Computer monitor (Fig. 4.6 (a))) and HLD unit set on the surface of the WiiBB to capture any possible heel lift (Fig. 4.6 (b)). Though the slippers attached to the surface of the WiiBB restricted the movement of the participants on the WiiBB, yet, I used the slippers to restrict the unwanted movement of participants' feet over the WiiBB. This was necessary as,

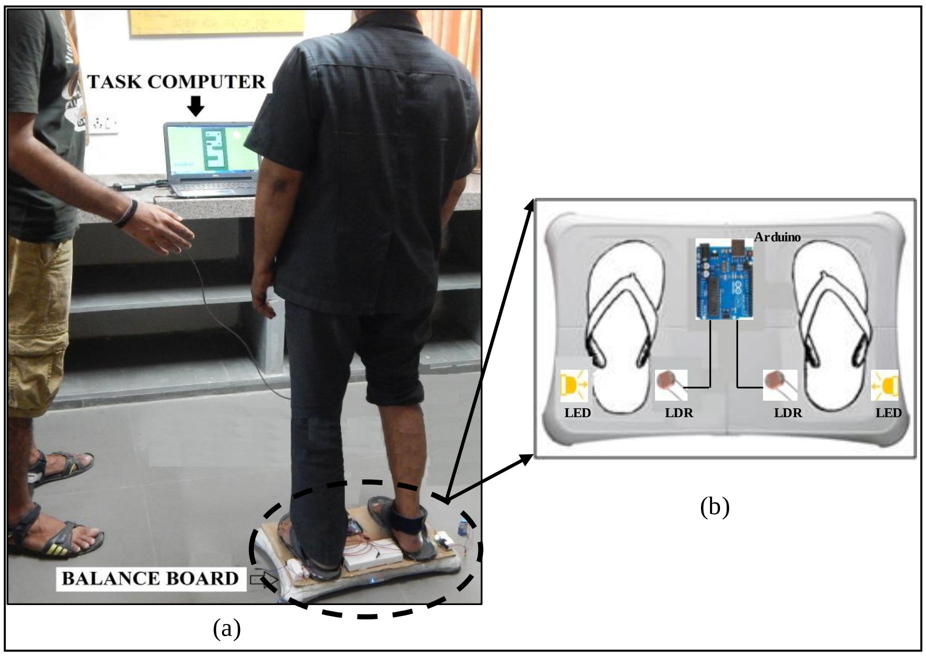


Figure 4.6. (a), Experimental setup for VBaT system, (b) Balance board with HLD unit for monitoring lifting of heel (Affected side)

without the slippers, the stroke participants may change their position in between the balance training sessions which can lead to unwanted fluctuation in the CoP values.

4.3.3 Procedure

The study required a commitment of approximately forty minutes from each participant. Once a participant arrived in the experiment room, he/she was asked to sit down on a chair and relax for approximately 5 minutes. Then the physiotherapist in the team assessed the participant's balance using BBS [35] measure and also ensured that the inclusion criteria were satisfied. This process took around 10 minutes for each participant. If the participant fulfilled the inclusion criteria of the study, the experimenter (myself) explained the experimental setup and also demonstrated four VR-based tasks (one in each difficulty level) to the participant while standing on the WiiBB. After that, the experimenter ensured that the participant understood the task to be performed followed by administering the signing of consent form. The participant was told that he/she was free to quit the balance training session at any time if he/she felt uncomfortable. Before starting the study, the experimenter asked the participant for his/her verbal consent.

Once the participant was ready, the experimenter asked him/her to stand on the WiiBB. Subsequently, the experimenter started the VR-based tasks that lasted for approximately 20 minutes. After the participant finished interacting with the VBaT system, he/she was given a questionnaire to get his/her views on the usability of the VBaT system.

4.3.4 Statistical Analysis

While the participants interacted with the VBaT system, I measured their performance score along with the time taken (in seconds) by them to complete the tasks. I was interested to understand whether the VBaT system contributed to any statistical improvement in (i) one's performance score (%) and (ii) reduction in time taken to complete the task in the *Last* attempt (that is, the last task in a particular difficulty level before being switched over to a task of higher difficulty level for DL1 – DL3 or the last task in DL4 before finishing the interaction with the VBaT system) from that in their *First* attempt (that is, the first task in a particular difficulty level) for tasks of each difficulty level. A Shapiro-Wilk test of normality was performed on the participants' performance data which suggested that the data was not normally distributed. Therefore, I opted for non-parametric dependent sample paired Wilcoxon signed-rank test [36] to check the statistical significance. While performing the Wilcoxon signed-rank test, I kept the % performance scores in the *First* and *Last* attempts as between-subject factor whereas difficulty level (DL1, DL2, DL3 and DL4) was kept as within-subject factor. Same was followed for the task completion times.

4.3.5 Post-study Survey

When interacting with a system or tool that helps users to achieve desired goals for which the system is designed, it is critical to measure user's experience to know the usability of the system [33], [34]. Researchers have used different questionnaires such as, System Usability Scale (SUS) [32], [33], VR usability (VRUSE) diagnostic tool [35], and others to evaluate the usability of various systems. Gil Gomej et al. [37] have designed User Suitability Evaluation Questionnaires (USEQ) to test the usability of VR-based rehabilitation platform. In this research, I have framed usability questions (relevant to my study) taking idea from USEQ proposed by Gil Gomej et al. [37] to understand the participants' experience with the VBaT system. The experimenter administered a post-study survey with questionnaire at the end of the study. The questionnaires asked to the participants were (i) Did you face any difficulty in understanding the tasks? (ii) Did you find the tasks interesting? (iii) Do you want to play again? and (iv) Do you think that you can benefit by using such a system. The participants' responses to these questions were recorded using binary scale ('Yes' or 'No'). I had asked questions (i) and (ii) to know the participants' thoughts on the type of tasks they were offered by the VBaT (useful from the perspective of task design). The question (iii) was asked to understand the participants' motivation to interact with the VBaT system again. The question (iv) was asked to know the participants' views on the potential of the VBaT system to help them in their rehabilitation program. The overall aim of conducting this survey was to understand the qualitative aspect of the VBaT system without going for participant-specific quantitative measures.

4.4 Result and Discussion

I conducted a usability study with VBaT system in which seven post-stroke hemiplegic participants took part. I wanted to understand the implication of VBaT system on participants' balance in terms of their performance score in the VR-based balance training tasks offered by the VBaT system. Also, I wanted to examine whether the VBaT system was acceptable to the stroke participants. In the following sections I present the results of the usability study. These results are also reported in one of my published articles [38].

4.4.1 Acceptability of the VBaT System

At the end of the study, the experimenter conducted a post-study survey (Section 4.3.5). From the participants' responses to the survey questionnaires, I found that the participants did not face any difficulty in understanding the tasks and they enjoyed interacting with the system. In response to the question on the possibilities of their future participation with the VBaT system, all the participants expressed that they were interested in participating again. Also, all of them were very positive about the usage of such a system since they felt that this system can offer potential benefit in balance rehabilitation. Thus, this qualitative finding shows that the VBaT system has a potential to be accepted by the target group of participants.

4.4.2 Effects of the VBaT System on Participants' Performance

While the participant interacted with tasks offered by the VBaT system, I recorded various task performance measures, like percentage performance score, task completion time, and number of tasks played in each difficulty level. During the weight-shifting task, aid in the form of therapist's support (as and when required) was allowed to prevent the participant from falling. Most of the participants needed physiotherapist's assistance while stepping on the WiiBB. However, only participants S1 and S4 took therapist's assistance in shifting their weight towards their Affected side during the first few initial tasks offered by the VBaT. As the session progressed, they were able to perform the tasks independently without any external assistance.

4.4.2.1 Effects of the VBaT system on participants' Performance Score (%)

The Fig. 4.7 shows percentage performance scores ($\% P_{f\_Score}$ henceforth) achieved by the participants in all of their attempts in the tasks of each difficulty level. Here the tasks of different difficulty levels were presented serially by the Task Switching Unit. Specifically, each participant was first offered tasks of DL1. Once the participant scored Adequately in DL1, the Task Switching Unit presented him/her with tasks of DL2. Once, in DL2, the participant was not switched back to DL1 in case of 'Inadequate' performance in DL2. The idea behind offering task in DL2 serially to the participant was that we did not intend to cause boredom to the participant by offering him/her a task of lower challenge (DL1 in this case). Instead, we wanted to maintain the challenge by offering him a task of DL2 serially and looking out if he/she acquired Adequate performance in DL2 before switching to a task of next higher challenge level (DL3 in this case). This progressed till the total task training session was over. Fig. 4.7 shows that the overall average $\%P_{f\_Score}$ achieved by all the participants was 51.52%. It can be seen that the number of VR-based tasks offered in each difficulty level was not evenly distributed. This was because the stroke participants had varying residual balance abilities and the VBaT system offered VR-based tasks of different difficulty levels with task switching being adaptive to one's balance capability. From Fig. 4.7, it can be seen that during the 20-minute exercise duration, the VBaT system offered more number of tasks in DL1 and DL2 as compared to that for DL3 and DL4. This was because the participants took more number of trials to achieve 'Adequate' performance for DL1 and DL2 possibly due to their poor residual balance with which they came in for the study. However, after interacting with a number of trials in each of DL1 and DL2, it can be expected that their balance has improved with repeated exposure helping at least some of the participants to move to higher difficulty level (that is DL3). For DL4, the number of tasks offered to the participants was the least (Fig. 4.7). This was because, all the participants (except S3, S5-S7) did not reach DL4 (Table 4.3). For the other participants, most of their time (20 minutes) was spent interacting with tasks of DL1 - DL3 before reaching DL4.

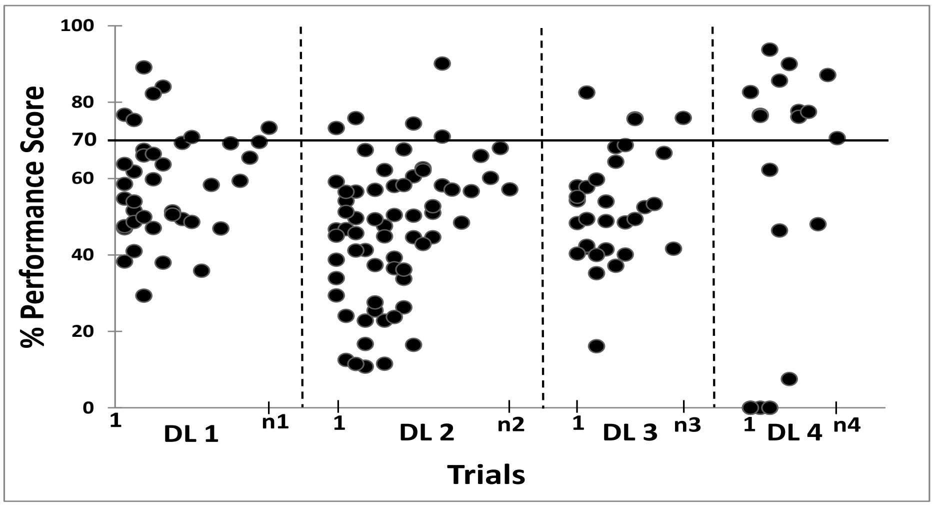


Figure 4.7. Participants' %performance score in all trials for each difficulty level

The Fig. 4.8 shows average *%* P_{f_Score} of stroke participants in their *First* and *Last* attempts in tasks of each difficulty level. The overall improvement in the average $\%P_{f\_Score}$ from *First* to *Last* attempts was 59.72%. The group average $\%P_{f\_Score}$ in the *First*, as well as *Last* attempts, was 'Adequate' only for the tasks belonging to DL1 (easiest difficulty level). Though the group average of $\%P_{f\_Score}$ was 'Inadequate' for the tasks of DL2 to DL4, yet, there was an

Table 4.3. Task of highest difficulty level played by the participants

Participant	S1	S2	S3	S4	S5	S6	S7
Difficulty Level	DL2	DL2	DL4	DL3	DL4	DL4	DL4

improvement in participants' performance score from their *First* to *Last* attempts in tasks of all difficulty levels. Again, though, DL3 was of higher difficulty level than DL2, yet, there was greater $\%P_{f\_Score}$ in the *First* attempt of DL3 compared to that of DL2. A possible reason behind this can be that before the participants started interacting with the tasks of DL3, they were offered balance practice while executing the tasks belonging to DL1 and DL2 offered by the VBaT that were also challenging to them towards the beginning (as evident from the number of tasks before reaching 'Adequate' performance (Fig. 4.7)). However, tasks of DL4 (Fig. 4.2 (c) and Fig. 4.2 (d)) were the most challenging tasks for the participants that may have caused least performance score while interacting with the *First* attempt of the DL4 task. But, still an improvement in the $\%P_{f\_Score}$ in DL4 tasks can be seen during the *Last* attempt from that in the *First* attempt that can be attributed to the practice with the VBaT system. Again, in DL4, a larger spread in $\%P_{f\_Score}$ achieved by the participants was observed compared to that for the other difficulty levels. This large spread in data at DL4 can be due to both the number of tasks played and the participants reaching DL4 were being the least among all the difficulty levels.

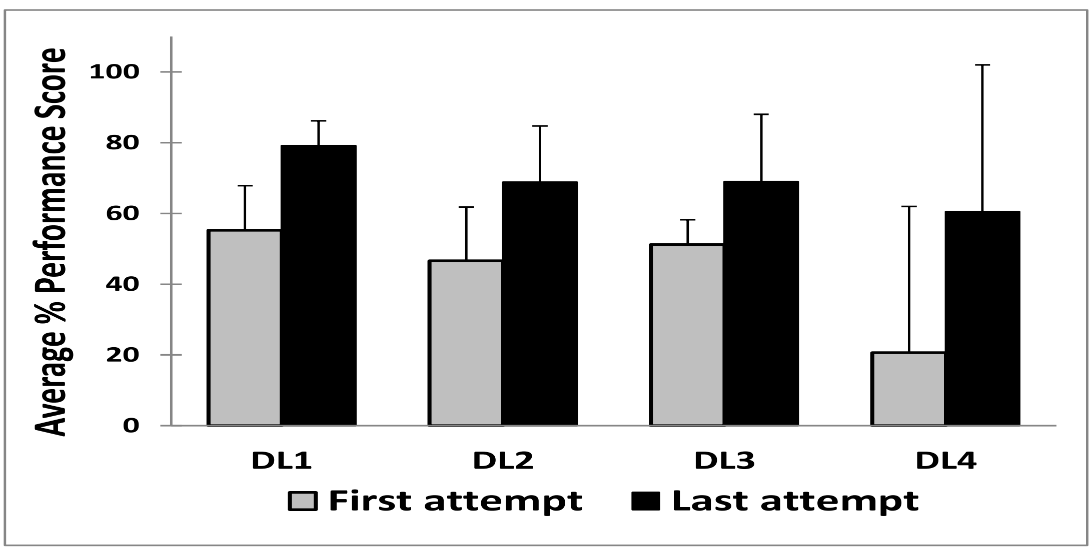


Figure 4.8. Average performance scores of participants in their *First* and *Last* attempts for each difficulty level

In short, I can say that the VBaT system was able to contribute to the improvement in $\%P_{f\_Score}$ of the participants possibly due to practice effect through interaction with a number of task trials. I performed a dependent sample Wilcoxon signed-rank test on the average $\%P_{f\_Score}$ of the participants, and I found that the improvement in the performance score from the *First* to the *Last* attempts was statistically significant with p-value<0.05.

4.4.2.2 Effects of the VBaT system on the Average Interaction Time

In the previous section, it was seen that for all the difficulty levels, the group average of participants' $\%P_{f\_Score}$ improved in the *Last* attempt of the VR-based tasks compared to that in the *First* attempt. I was interested to see whether this improvement in performance was coupled with improvement in the speed with which the participants' completed the tasks. The Fig. 4.9 shows a

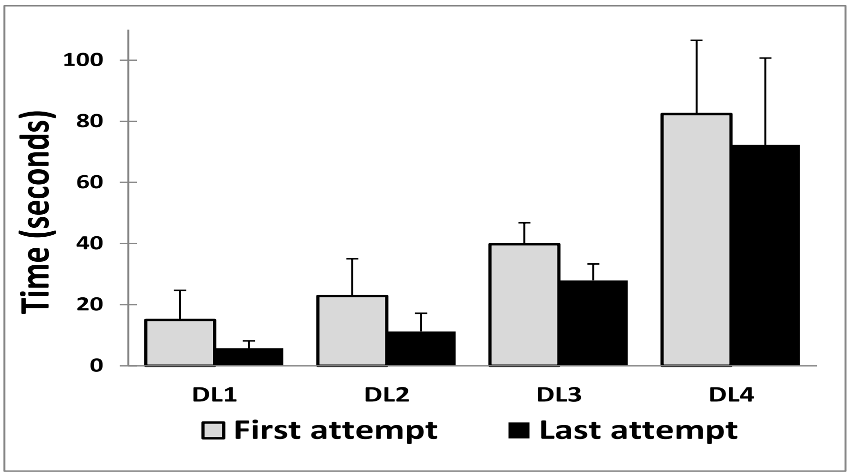


Figure 4.9. Average interaction time of the participants in their *First* and *Last* attempts for each difficulty level

comparative analysis of participants' average interaction time during their *First* and *Last* attempts in tasks of each difficulty level. I can see from Fig. 4.9 that for all the difficulty levels, there was a reduction in the group average interaction time in the *Last* attempt as compared to that in the *First* attempt. From this, it can be inferred that the participants showed improvement in terms of lesser interaction time. Also, this improvement in terms of lesser interaction time reduced as the difficulty level of the task increased. Here, I also performed a dependent sample Wilcoxon signed-rank test on average interaction time of the participants in their *First* and *Last* attempts for the tasks of all the difficulty levels. The improvement in terms of lesser interaction time in the *Last* attempt compared to that in the *First* attempt was statistically significant with p-value<0.05.

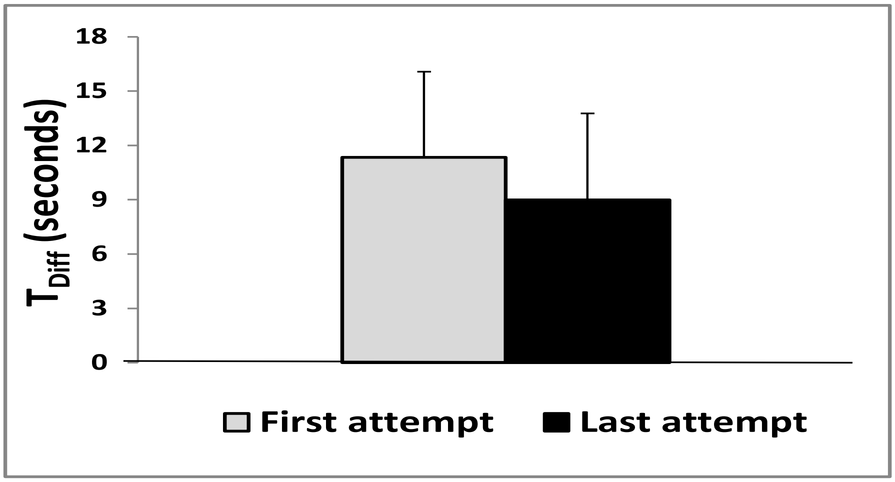


Figure 4.10. Average T_{Diff} on *First* and *Last* attempts for DL4 tasks

Also, while performing daily living activities, an individual often indulges in tasks related to biped balance. To maintain balance during such tasks, an individual need to use both the lower limbs effectively to shift his/her body weight in a controlled manner. Since tasks of DL4 (Fig. 4.2 (c) and (d)) were specifically designed to leverage participant's ability to use both the lower limbs effectively and as equally as possible, I analyzed the difference in time taken (T_{Diff}) between $Template_{Left}$ and $Template_{Right}$ for the participants. Here, smaller (or more equal task completion times for the two templates) the value of T_{Diff}, better was the usage of both the limbs for the task. To understand the implication of practice with the tasks of DL4 offered by VBaT on T_{Diff}, I analyzed the data to find the variation in T_{Diff} from *First* to *Last* attempts (Fig. 4.10). It can be seen from Fig. 4.10, that the T_{Diff} was reduced in the *Last* attempt compared to that in the *First* attempt in the tasks of DL4. Specifically, the improvement was 20.76% in terms of reduction in T_{Diff} from the *First* to the *Last* attempts. Thus, even though the participants were exposed to lesser number of tasks in DL4, the participants had shown some improvement in the average % performance score along with reduced T_{Diff} from the *First* to the *Last* attempts.

4.5 Conclusion

The main contribution in the presented work was the design of an adaptive VBaT system that could understand an individual's ability of weight-shifting in a balance task and accordingly offered tasks of varying difficulty levels to the participant. Also, the VBaT system was added with an in-house built Heel Lift Detection unit to encourage the participant to follow the Ankle strategy during shifting of weight, an important strategy for balancing the posture during standing balance tasks. The VBaT system being adaptive to one's ability of weight-shifting, offered tasks of different difficulty levels to the participants based on their performance. Results indicate that the participants needed more number of tasks in the lower difficulty levels before reaching the higher difficulty levels. The VBaT system being adaptive to individualized performance, intelligently offered more tasks at lower difficulty levels that in turn provided them opportunity to practice before achieving 'Adequate' performance and move to higher difficulty levels. The VBaT system not only helped the participants to improve their $\%P_{f\_Score}$ but also helped them to decrease their average interaction time (by varying amounts) that implies an improvement in the speed of task execution.

In this study, performance measure in balance tasks being decided based on different balance parameters related to weight-shifting, such as quality of CoP trajectory, ability to hold shifted weight while using Ankle strategy, is novel as compared to the existing off-the-shelf games [8], [11], [16], [28], [29]. Also, the rule engine used by the VBaT system to present VR-based tasks ensures that the participant can repeatedly practice specific weight-shifting tasks in a variety of environments and in a controlled manner until he/she achieves 'Adequate' performance and be ready for interacting with tasks of higher challenge, a feature not available with the presently existing off-the-shelf game environments.

Though the results obtained in this study are promising yet this study had some limitations. This study had a limited sample size. Also, during the post-study survey, instead of using a 1-10 or 1-5 Likert scale, I considered the binary scale for recording participants' responses to my survey questions. This study being the first VR-based exercise task platform that I have designed, I wanted to get at least some views of the participants on usage of my system. Again, this being a proof-of-concept study, the participants were exposed to the VR-based balance training environment for a limited duration and for one session. Such a limited exposure may not be sufficient to prove the rehabilitation efficacy of the system. To see a significant improvement in an individual's balance, a longitudinal study with more number of stroke participants is warranted. Additionally, this must be associated with the clinical assessment of balance ability by measuring BBS score before and post the study. Another limitation of the study was the wide spectrum of participants' clinical characteristics particularly with regard to different residual balance capabilities on account of participants being recruited based on their availability. This might have affected the group average of the participants' performance scores. This study can be extended to carry out a more in-depth longitudinal study by enrolling a larger patient population categorized based on residual balance capability before exposing them to the VBaT system. Also, in future, I plan to design more structured post-study survey along with a Likert scale instead of the binary scale as used in my present study.

In the study presented in this chapter, I have used an individual's CoP information to interact with VR-based balance training tasks. Specifically, I provided the feedback of an individual's CoP excursion on the Task Computer in the form of displacement of the VR_{Obj} in the VR environment. However, an individual’s balance is not only quantified based on CoP displacement, but it can also be quantified using body Center of Mass (CoM) displacement. Literature review indicates that human balance is strongly related to the position and velocity of the CoM [35] and to ensure that the balance task is sufficiently challenging, the CoM-based approach can be preferred over the CoP-based approach [39]. However, measurement of an individual’s CoM is limited to laboratory settings due to its requirement of marker-based motion capture systems that are not only costly but also space intensive [40]. As an alternative, I plan to utilize recent advancements in technology to estimate one's CoM using marker less, low-cost motion capture system along with a WiiBB to estimate one's personalized CoM [41]. Thus, in my next chapter, I will be presenting the design of CoM-assisted VR-based balance training system and a usability study framed to understand the implication of CoM-assisted weight-shifting tasks on an individual’s balance.

Reference

[1] “The human balance system.” [Online]. Available: http://vestibular.org/understanding-vestibular-disorder/human-balance-system. [Accessed: 13-Jan-2018].

[2] M. Shumway, Cook A Woollacott, *Motor Control: Theory and Practical Applications*. Philadelphia: Williams and Wilkins, 2001.

[3] C. L. Horak, F. B., and Shupert, “Role of the vestibular system in postural control,” *Vestib. Rehabil.*, vol. 2, pp. 98–113, 1994.

[4] R. J. Peterka and M. S. Benolken, “Role of somatosensory and vestibular cues in attenuating visually induced human postural sway,” *Exp. Brain Res.*, vol. 105, no. 1, pp. 101–110, 1995.

[5] R. L. Sacco et al., “An updated definition of stroke for the 21st century,” *Stroke*, vol. 44,

pp. 1–26, 2013.

[6] J. D. Pandian and P. Sudhan, “Stroke Epidemiology and Stroke Care Services in India,” *J. Stroke*, vol. 15, no. 3, p. 128, 2013.

[7] V. L. Feigin, C. M. Lawes, D. A. Bennett, S. L. Barker-Collo, and V. Parag, “Worldwide stroke incidence and early case fatality reported in 56 population-based studies: a systematic review,” *The Lancet Neurology*, vol. 8, no. 4. pp. 355–369, 2009.

[8] G. Saposnik et al., “Effectiveness of virtual reality using wii gaming technology in stroke rehabilitation: A pilot randomized clinical trial and proof of principle,” *Stroke*, vol. 41, no. 7, pp. 1477–1484, 2010.

[9] S.-N. Jeon and J.-H. Choi, “The effects of ankle joint strategy exercises with and without visual feedback on the dynamic balance of stroke patients,” *J. Phys. Ther. Sci.*, vol. 27, no. 8, pp. 2515–2518, 2015.

[10] G. U. Jung, T. H. Moon, G. W. Park, J. Y. Lee, and B. H. Lee, “Use of augmented reality-based training with EMG-triggered functional electric stimulation in stroke rehabilitation,” *J. Phys. Ther. Sci.*, vol. 25, no. 2, pp. 147–151, 2013.

[11] G. I. Yatar and S. A. Yildirim, “Wii Fit balance training or progressive balance training in patients with chronic stroke: a randomised controlled trial,” *J. Phys. Ther. Sci.*, vol. 27, no. 4, pp. 1145–1151, 2015.

[12] G. S. A. F. Verheyden et al., “Interventions for preventing falls in people after stroke.,” *Cochrane database Syst. Rev.*, no. 5, p. CD008728, 2013.

[13] B. H. Alexander, F. P. Rivara, and M. E. Wolf, “The cost and frequency of hospitalization for fall-related injuries in older adults,” *Am. J. Public Health*, vol. 82, no. 7, pp. 1020–1023, 1992.

[14] D. S. Marigold, J. J. Eng, A. S. Dawson, J. T. Inglis, J. E. Harris, and S. Gylfadóttir, "Exercise leads to faster postural reflexes, improved balance and mobility, and fewer falls in older persons with chronic stroke," *J. Am. Geriatr. Soc.*, vol. 53, no. 3, pp. 416–423, 2005.

[15] R. J. Seitz, C. M. Bütefisch, R. Kleiser, and V. Hömberg, "Reorganisation of cerebral circuits in human ischemic brain disease.," *Restor. Neurol. Neurosci.*, vol. 22, no. 3–5, pp. 207–29, 2004.

[16] B. S. Rajaratnam et al., "Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?," *Rehabil. Res. Pract.*, vol. 2013, pp. 1–6, 2013.

[17] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer, "Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: A pilot randomized clinical trial in patients with acquired brain injury," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, 2011.

[18] H. Bateni, "Changes in balance in older adults based on use of physical therapy vs the Wii Fit gaming system: A preliminary study," *Physiother. (United Kingdom)*, vol. 98, no. 3, pp. 211–216, 2012.

[19] G. Saposnik and M. Levin, "Virtual reality in stroke rehabilitation: A meta-analysis and implications for clinicians," *Stroke*, vol. 42, no. 5, pp. 1380–1386, 2011.

[20] S. E. Fasoli, H. I. Krebs, and N. Hogan, "Robotic Technology and Stroke Rehabilitation: Translating Research into Practice," *Top. Stroke Rehabil.*, vol. 11, no. 4, pp. 11–19, 2004.

[21] A. Darekar, B. J. McFadyen, A. Lamontagne, and J. Fung, "Efficacy of virtual reality-based intervention on balance and mobility disorders post-stroke: A scoping review," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, 2015.

[22] H. Y. Lee, Y. L. Kim, and S. M. Lee, "Effects of virtual reality-based training and task-oriented training on balance performance in stroke patients.," *J. Phys. Ther. Sci.*, vol. 27, no. 6, pp. 1883–8, 2015.

[23] Y. Lee, J. G. Her, Y. Choi, and H. Kim, "Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors' Names," *J. Phys. Ther. Sci.*, vol. 26, no. 2, pp. 179–182, 2014.

[24] A. Ruhe, R. Fejer, and B. Walker, "Center of pressure excursion as a measure of balance performance in patients with non-specific low back pain compared to healthy controls: A systematic review of the literature," *European Spine Journal*, vol. 20, no. 3. pp. 358–368, 2011.

[25] D. A. Winter, "Human balance and posture control during standing and walking," *Gait Posture*, vol. 3, no. 4, pp. 193–214, 1995.

[26] D. A. Winter, A. E. Patla, F. Prince, M. Ishac, and K. Gielo-Perczak, "Stiffness Control of Balance in Quiet Standing," *J. Neurophysiol.*, vol. 80, no. 3, pp. 1211–1221, 1998.

[27] D. J. Goble, B. L. Cone, and B. W. Fling, "Using the Wii Fit as a tool for balance assessment and neurorehabilitation: The first half decade of 'wii-search,'" *Journal of NeuroEngineering and Rehabilitation*, vol. 11, no. 1. 2014.

[28] D. K. A. Singh, B. S. Rajaratnam, V. Palaniswamy, H. Pearson, V. P. Raman, and P. S. Bong, "Participating in a virtual reality balance exercise program can reduce risk and fear of falls," *Maturitas*, vol. 73, no. 3, pp. 239–243, 2012.

[29] K. H. Cho, K. J. Lee, and C. H. Song, "Virtual-Reality Balance Training with a Video-Game System Improves Dynamic Balance in Chronic Stroke Patients," *Tohoku J. Exp. Med.*, vol. 228, no. 1, pp. 69–74, 2012.

[30] V. Weerdesteyn, M. de Niet, H. J. R. van Duijnhoven, and A. C. H. Geurts, "Falls in individuals with stroke," *J. Rehabil. Res. Dev.*, vol. 45, no. 8, pp. 1195–213, 2008.

[31] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[32] A. Ruhe, R. Fejer, and B. Walker, "The test-retest reliability of centre of pressure measures in bipedal static task conditions - A systematic review of the literature," *Gait and Posture*, vol. 32, no. 4. pp. 436–445, 2010.

[33] K. Jack, S. M. McLean, J. K. Moffett, and E. Gardiner, "Barriers to treatment adherence in physiotherapy outpatient clinics: A systematic review," *Manual Therapy*, vol. 15, no. 3. pp. 220–228, 2010.

[34] B. M. Young et al., "Changes in functional connectivity correlate with behavioral gains in stroke patients after therapy using a brain-computer interface device," *Front. Neuroeng.*, vol. 7, 2014.

[35] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[36] J. D. Gibbons and S. Chakraborti, *Nonparametric Statistical Inference*, vol. 15, no. 2. 2003.

[37] Gil-Gómez et al., "USEQ: A short questionnaire for satisfaction evaluation of virtual rehabilitation systems," *Sensors (Switzerland)*, vol. 17, no. 7, 2017.

[38] S. Verma, D. Kumar, A. Kumawat, A. Dutta, and U. Lahiri, "A low-cost adaptive balance training platform for stroke patients: A usability study," *IEEE Trans. Neural Syst. Rehabil.*

Eng., vol. 25, no. 7, pp. 935–944, 2017.

[39] L. E. Cofré Lizama, M. Pijnappels, N. P. Reeves, S. M. Verschueren, and J. H. van Dieën, "Centre of pressure or centre of mass feedback in mediolateral balance assessment," *J. Biomech.*, vol. 48, no. 3, pp. 539–543, 2015.

[40] L. E. Cofré Lizama, M. Pijnappels, G. H. Faber, P. N. Reeves, S. M. Verschueren, and J. H. van Dieën, "Age effects on mediolateral balance control," *PLoS One*, vol. 9, no. 10, p. e110757, 2014.

[41] M. Hayashibe, A. González, and P. Fraisse, "Personalized Modeling for Home-Based Postural Balance Rehabilitation," in *Human Modeling for Bio-Inspired Robotics: Mechanical Engineering in Assistive Technologies*, 2016, pp. 111–137.

CHAPTER 5

VIRTUAL-REALITY BASED CENTER OF MASS ASSISTED PERSONALIZED BALANCE TRAINING SYSTEM

5.1 Introduction

In the previous chapter, I have studied the implications of VR-based balance training (VBaT) system on the balance of post-stroke hemiplegic participants. The VBaT system used participant's Center of Pressure (CoP) measured by a Wii Balance Board (WiiBB) to interact with VR-based balance tasks. An individual's balance can also be quantified using information on Center of Mass (CoM) position. The CoM is a point at which the total body mass is concentrated within the global reference system. For a human body, the net CoM can be decided from the weighted average CoM of each body segment in 3D space [1]. Literature review indicates that human balance is strongly related to the position and velocity of the CoM [1].

To make the balance task sufficiently challenging, the CoM-based approach is preferred over the CoP-based approach [2]. In this chapter, I wanted to address two research questions, namely, (i) is it feasible to quantify participant's residual directional weight-shifting capability based on the performance measures in the CoM-assisted VR-based balance task? And if so, (ii) what is the implication of CoM-assisted VR-based balance training system on the balance of post-stroke hemiplegic participants? To address these research questions, I developed a **Virtual** Reality-based **CoM**-assisted **Ba**lance **T**raining (Virtual CoMBaT) system that uses participant's CoM to offer real-time feedback on his/her weight-shifting while participating in various VR-based balance tasks.

The balance can be quantified in terms of CoP as well as CoM-related measures. Most of the existing VR-based research studies addressing balance disorders have used body CoP to interact with off-the-shelf games and offering real-time feedback during a balance task possibly due to the ease of access to cost-effective force platform such as WiiBB [3], [4], [5], [6], [7], [8]. The CoP is often preferred over CoM in the context of geriatric assessment or clinical settings due to easier estimation of CoP [9] along with lesser challenge offered by CoP-based tasks. However, research studies suggest that CoM-based approach is preferable over the CoP-based approach to make the balance task sufficiently challenging [2].

Given the importance of CoM-based approach, measurement of an individual's CoM during a balance task requires use of marker-based motion capture systems that might be cumbersome, space intensive and costly [10]. This warrants for alternative approaches to estimate body CoM. With recent technological progress, researchers have been exploring techniques to estimate one's personalized CoM using portable and marker-less motion capture systems, namely, Kinect and force platform (such as WiiBB) [11], [12]. With this, estimation of an individual's personalized CoM has become a reality even outside laboratory settings.

In the current study, I have used the approach developed by Gonzalez et al. [11] to estimate a participant's CoM while using a VR-based platform coupled with peripheral devices such as Kinect and WiiBB. The Kinect is a motion capture device consisting of color sensor, IR depth sensor, IR emitter and Microphone Array. It has a field of view (43° vertical x 57° horizontal) and it works at 30 frames per second [13]. In this study, I have designed a VR-based balance training system that uses participant's CoM for offering real-time feedback during a VR-based weight-shifting task. To estimate personalized CoM, I have used the Statistically Equivalent Serial Chain (SESC) method augmented with Kinect device [11], [12]. In the present study, I have estimated participant's CoM. For this estimation, I used inexpensive off-the-shelf motion capture sensor namely, Kinect and a low-cost force platform namely, WiiBB to identify SESC parameters that were used to estimate participant's body mass distribution. Subsequently, the SESC parameters were used to compute personalized CoM using only the Kinect sensor while a participant performed VR-based standing balance tasks. While the stroke participants performed VR-based balance tasks, they were asked to follow Ankle strategy. This was because, among the three main postural control strategies, namely, Ankle, Hip and Step strategies [14], the Ankle strategy enabling muscle contraction of the ankle joint is most commonly used for addressing standing balance-related issues [15]. To encourage the participants to follow the Ankle strategy during the weight-shifting task, I designed a Heel Lift Detection (HLD) unit that was interfaced wirelessly with the VR-based task platform. This HLD unit continuously monitored the position of participant's heel during the balance task and alerted the participant in case the Ankle strategy was not followed.

The primary objectives of the study presented in this chapter were two-fold, namely, (i) develop a Virtual CoMBaT system using WiiBB and Kinect and (ii) conduct a usability study with Virtual CoMBaT system involving post-stroke patients. Additionally, my aim was to explore the feasibility of quantifying a participant's initial (residual) weight-shifting ability based on his performance indicators derived from the usability study with the Virtual CoMBaT system.

5.2 System Design

The Virtual CoMBaT system consisted of five units, namely, (A) Personalized CoM Estimation, (B) Design of VR-based Balance Training Tasks, (C) Estimation of Individualized Threshold for VR-based Task, (D) CoM-VR_{obj} Integration, (E) Monitoring of Ankle strategy during Task Execution, (F) Performance Evaluation, (G) Task Switching and (H) System Usability related Questionnaire.

5.2.1 Personalized Center of Mass (CoM) Estimation Unit

One's personalized CoM can be estimated by using a Statistically Equivalent Serial Chain (SESC) method initially proposed by Espiau and Boulic [16]. The position of the CoM of any multi-body structure can be expressed as the end effector position of an open-ended serial chain, known as the SESC. The structure of the SESC can be defined by the static and geometric parameters of the whole body structure [17]. The SESC method can effectively translate one's body mass distribution to the link-length of a linked chain. These links that match with body segment can be subsequently used to compute CoM position for any posture, even when constrained to a plane. Personalized CoM estimation using SESC procedure has been validated for both young [12], [18] and elderly healthy population [19].

In this study, in order to estimate the CoM position for a human subject, a skeleton model composed of three links (two leg segments and one torso segment) is used (Fig. 5.1). A frame (R_i) is attached to each link which is used to compute the CoM of an articulated structure using Eq. (5.1) [11]

$$\mathrm{c} = [\mathrm{E}\ \mathrm{A}_1\ \mathrm{A}_2 \ldots \mathrm{A}_\mathrm{n}][\mathrm{p}_0\ \mathrm{r}_1\ \mathrm{r}_2\ \ldots\ \mathrm{r}_\mathrm{n}]^\mathrm{T} \tag{5.1}$$

Where, c is CoM position of an articulated structure with respect to global reference system (GRS), E is an identity matrix, A_i is a 3x3 orientation matrix of a link, and p_0 is the position of the origin of R_1 (Fig. 5.1; a floating frame attached to the torso of skeleton as a base for the SESC) with respect to the GRS. The values of r_i are function of the linked masses and geometries. In order to provide a personalized CoM estimation, the subject-specific r_i values can be identified experimentally once a number of measurements of body segment orientations and ground projections of the CoM are obtained. The number of links/segments considered for the model should be sufficient to accurately describe the performed motion. Previously, Gonzalez et al. [11], [12] have used a nine-link SESC model which required 40 static postures consisting of

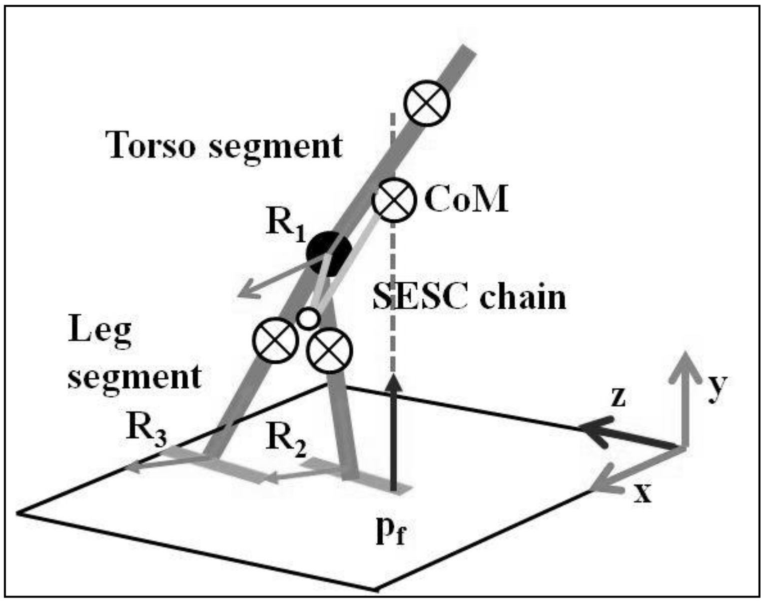


Figure 5.1. Multi-segment skeleton model

squatting, standing on one leg and doing different static postures and others to identify the SESC parameters. In that study, the participants were young and healthy individuals with no balance impairment. However, holding these 40 static postures while standing on a WiiBB is inconvenient for individuals with balance disorders. Since in the current study, the participants were post-stroke hemiplegic individuals, I simplified the SESC parameter estimation that required fewer static postures (For details, please see section 5.3.3 below). For this, I optimized the personalized CoM estimation algorithm by reducing the SESC model from nine-link to three-link model for mapping the participant’s motion (one torso and two legs) as shown in Fig. 5.1. With the optimized SESC model, that is, three link SESC model, the identification of SESC participants did not necessitate static postures (as mentioned in [11]) that would be difficult for the stroke survivors. Instead the participants needed to make only three static postures, namely, lean forward and lift each leg away from the body without stepping off the WiiBB.

In order to achieve this, I measured an individual's CoP (from WiiBB) to estimate the ground projection of CoM for three different static postures out of the ones as described by Gonzalez et al. [11]. Then by using the optimized SESC algorithm, I estimated the CoM. The estimated CoM and the measured CoP differed by approximately 5% in the medio-lateral direction and 12% in the antero-posterior direction on an average. These deviations were comparable with that reported in a previous study [11]. Again, to minimize the effect of such deviation, the system used relative change in participant's personalized CoM position during Task Execution Stage (Section 5.3.3).

5.2.2 Design of VR-based Balance Training Tasks

In the current study, I have designed the VR-based Balance Training tasks with an aim to (i) leverage a participant's weight-shifting capability while standing on the ground with fixed base of support (BoS) in different directions by following Ankle strategy and (ii) quantify the participant's balance capability during the VR-based weight-shifting tasks. I have used Vizard software toolkit (from Worldviz Llc.) to design various VR environments and VR objects (VR_{Obj} henceforth). In order to make the balance training session interesting and interactive, I designed 30 unique combinations of VR environments (such as road, playground, river and others) and virtual objects (such as car, ball, fish, carom board items and other objects) that one often encounters in daily living and entertainment. The VR-based tasks needed participants to shift their weight (and therby vary CoM position) in pre-defined directions, namely, North (*N*), East (*E*), West (*W*), North East (*NE*) and North West (*NW*) while maintaining their balance while moving within their Limit of Stability (LoS) by following Ankle strategy (with heel being in contact with the ground surface). The current study did not consider the South (*S*) direction in the weight-shifting tasks, since while estimating the CoM from the CoP measure, it has been observed that the CoP of the participants during standing upright on the WiiBB were shifted towards the *S* direction due to their over-weight physique. The Virtual CoMBaT system showed

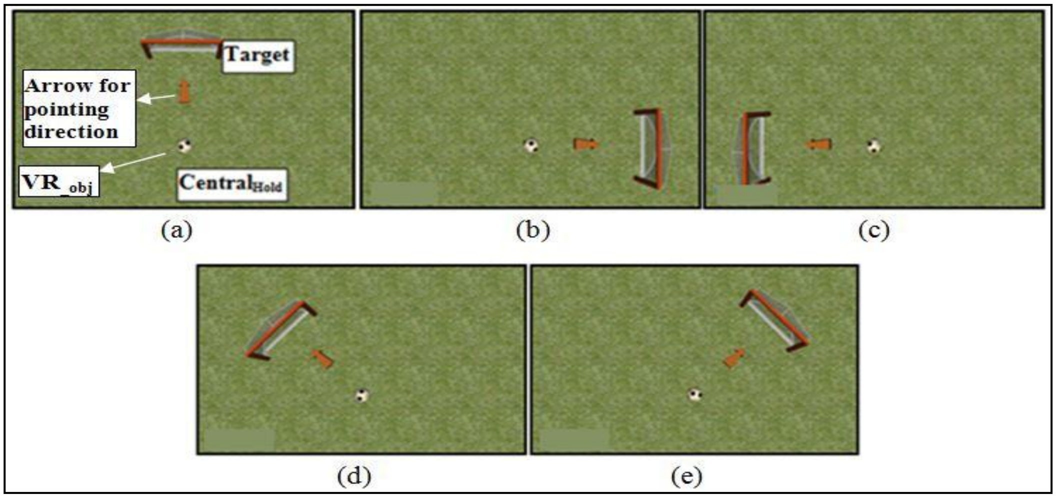


Figure 5.2. Graphical User Interface (GUI) of a VR-based balance training task in (a) North, (b) East, (c) West, (d) North-West and (e) North-East directions.
Note: $VR_{\_obj}$ = Virtual object

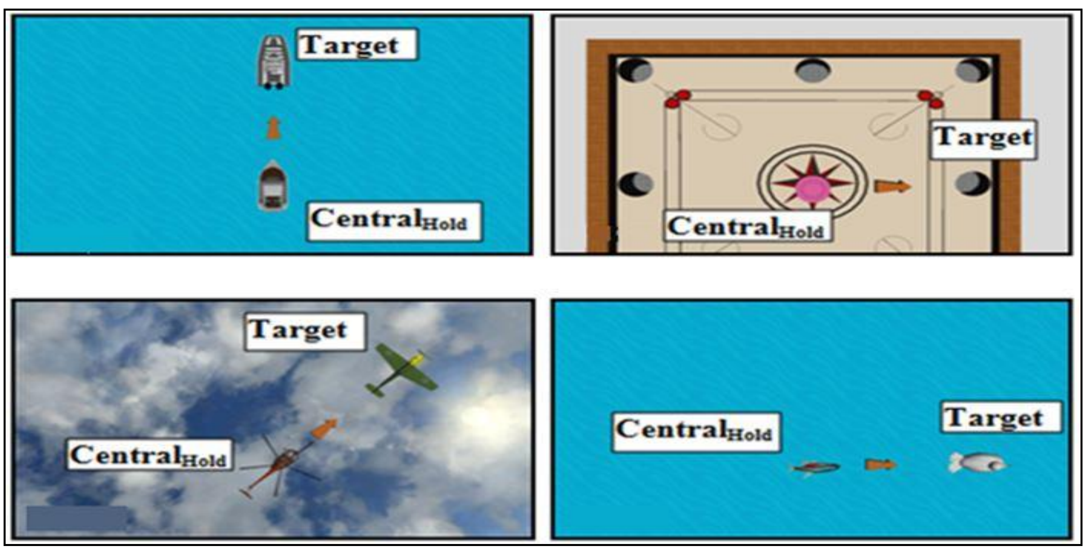


Figure 5.3. GUI of few VR-based balance training tasks

real-time feedback of participant's estimated CoM by moving the virtual object (VR_{Obj}) inside the VR environment in the direction of participant's weight-shifting. The participants were required to shift their weight to displace the CoM position to maneuver the VR_{Obj} in the VR environment from their initial (*$Central_{Hold}$*) to target (*Target*) positions (Fig. 5.2 and Fig. 5.3). The Fig. 5.2 shows an example of Graphical User Interface (GUI) of VR-based task in different directions (*N, E, W, NE, NW*) and Fig. 5.3 shows few other templates of VR-based tasks used in this study. Here, I designed the VR-based tasks having two different difficulty levels (DL1 and DL2) with the difficulty being individualized based on participant's weight-shifting capability. For this, I measured individual participant's initial ability to shift the CoM position in different directions from their *$Central_{Hold}$* position to decide Threshold CoM displacement required to complete the task. Details of the individualized threshold estimation is explained in the following section (Section 5.2.3).

5.2.3 Estimation of Individualized Threshold for VR-based Task

In the present work, the aim was to develop Virtual CoMBaT system which can estimate participant's weight-shifting ability in different directions so as to provide individualized balance training to the participants. For this, I estimated the individualized thresholds of weight-shifting ability for each stroke participant. Therefore, before starting the study, the participants were asked to stand on the ground and shift their weight in the forward (North (*N*)), left (West (*W*)) and right (East (*E*)) directions as much as possible three times while following the Ankle strategy. Simultaneously, the system recorded the participant's corresponding CoM position using SESC method that was used to estimate the individualized threshold of weight-shifting. The thresholds of weight-shifting were decided from the participant's maximum CoM displacement (ΔCoM_{max}) along the *N, E* and *W* directions out of the three trials while maintaining Ankle strategy. As far as the *NE* direction was concerned, the CoM displacement (ΔCoM_{max}) was calculated from the average of the ΔCoM_{max} for the *N* and *E* directions. For the *NW* direction, the threshold ΔCoM_{max} was calculated from the average of the ΔCoM_{max} for the *N* and *W* directions. The direction-specific ΔCoM_{max} was chosen as the threshold of participant's CoM displacement required to reach the *Target* (Figs. 5.2 and 5.3) position for the tasks of DL2. For the tasks of DL1, the threshold was kept at 80% of the corresponding ΔCoM_{max} for each direction. Thus, the tasks of DL2 required more weight-shifting than that for tasks of DL1. Please note that the weight-shifting threshold for DL1 tasks was chosen to be 80% of ΔCoM_{max} as an initial approximation and this value can be changed based on the study design.

5.2.4 CoM-VR_{Obj} Integration Unit

While the participants performed VR-based weight-shifting tasks, the Virtual CoMBaT system recorded their CoM position along different directions (that is *N*, *E*, *W*, *NE* and *NW*). Subsequently, the CoM positions were mapped to the position of VR_{obj} inside the VR environment in real-time using Eq. (5.2).

$$\begin{bmatrix} x \\ y \end{bmatrix}_{VR_{obj}} = \begin{bmatrix} CoM_x & 0 \\ 0 & CoM_y \end{bmatrix} \begin{bmatrix} \epsilon_1 \\ \epsilon_2 \end{bmatrix} \tag{5.2}$$

Where, ϵ_1 and ϵ_2 are scaling factors for the VR_{Obj} coordinates (x, y) corresponding to the CoM position (CoM_x, CoM_y) in the VR environment presented on the Task Computer monitor. There was no perceptible visual lag between the change in CoM position (acquired from the Virtual CoMBaT system) and the corresponding change in the position of the VR_{obj} on the Task Computer monitor.

5.2.5 Monitoring of Ankle Strategy during Task Execution

Similar to the previous system, that is, VBaT, here also a Heel Lift Detection (HLD) unit was used to monitor whether the participants followed the Ankle Strategy or not. However, in contrast to the previous study (that needed the participant's feet to be fixed on the WiiBB through slippers), here, a participant was independent to change his relative position while standing on the Base Location (Fig. 5.4). So, instead of a static HLD unit, here the HLD unit needed to be fixed onto the participant's shoes. Also, the HLD unit used in the previous study (with VBaT) was meant for only sensing an individual's heel lift. However, in the present study,

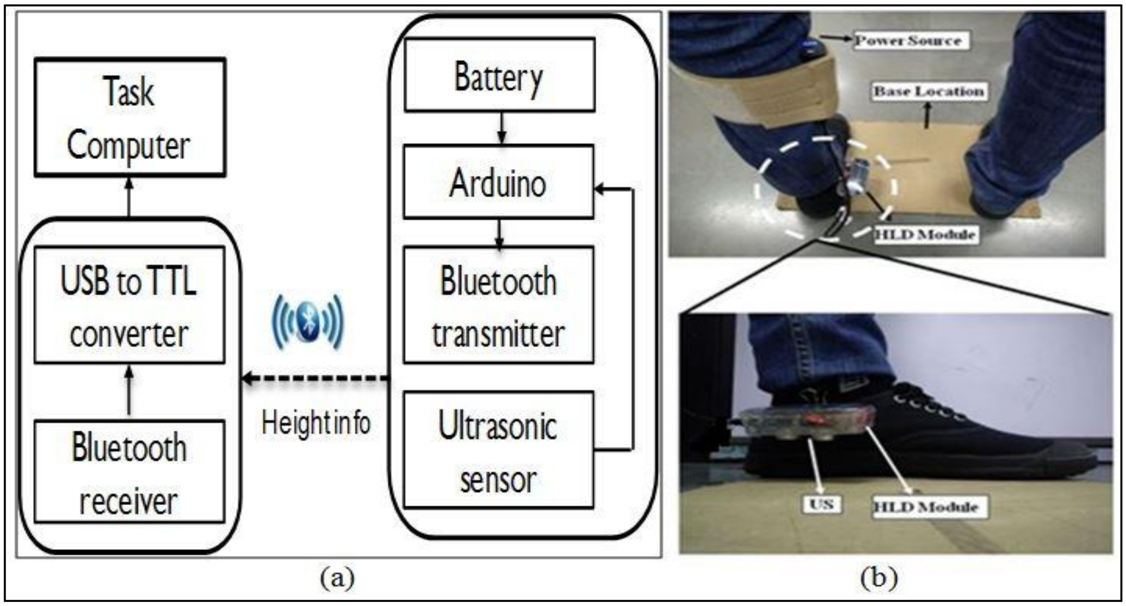


Figure 5.4. (a) Block diagram and (b) Placement of HLD unit on the shoe

since the participants were free to move on the Base Location while standing, I wanted to consider the heel lift as a valid one provided the amount of heel lift from the surface of the ground was beyond a particular threshold (described below). The threshold distance (determined from observations) was considered beyond that which can generally occur while a participant repositions himself/herself on the ground surface during the study. Thus, for this study, I needed a different HLD unit. For this, I designed Ultrasonic Sensor (US)-based HLD unit (Fig. 5.4). This HLD unit was attached to the participant's shoe and wirelessly communicated with Virtual CoMBaT to (i) alert the participant in case of heel lift by providing an audio alarm as a feedback and (ii) penalize the participant by adding a penalty factor to the performance score in case the Ankle strategy was not followed. The HLD unit consisted of an US, an Arduino board, a Bluetooth HC-05 transmitter and receiver pair and a USB (universal serial bus) to TTL (transistor-transistor logic) converter. Fig. 5.4 (a) shows the block schematic of the HLD unit.

The US sensor used in the HLD unit could measure the distance between 2 cm to 400 cm with resolution of 0.2 cm [20]. The Fig. 5.4 (b) shows the placement of HLD unit on the participant's shoe worn on the leg of the Affected side (Affected leg *henceforth*). The HLD unit was positioned on the medial side of the shoe closer to the inner ankle bone of the Affected leg. The US sensor of the HLD unit was arranged in such a manner that it faced downwards towards the ground so that it could determine the distance between the surface of the ground and the US sensor position. Initially, the participant was asked to stand upright with his heels in contact with the ground surface while wearing the HLD unit. The US sensor mounted on the participant's shoe measured the initial height (d_{ini} in mm) between the US sensor and the ground surface. Thereafter, the HLD unit was used to continuously monitor the distance between the US sensor and ground surface while the participant took part in the weight-shifting tasks. The output from the US sensor was processed by the microcontroller of the Arduino board that in turn provided the instantaneous distance between the US sensor and ground surface (d_{ins}). This data was transmitted wirelessly to the Task Computer to identify participant's heel lift by using Equation (5.3)

$$Ankle\ strategy = \begin{cases} Followed;\ \ if\ d_{ins} < d_{Limit} \\ Not\ Followed; if\ d_{ins} \geq d_{Limit} \end{cases} \tag{5.3}$$

$$d_{Limit} = d_{ini} + \ d_{th} \tag{5.4}$$

Where, d_{ins} = instantaneous distance between the ground surface and US sensor of the HLD unit and d_{th} = 20 mm = height tolerance for heel lift detection. In this study, the value of d_{th} was chosen as 20 mm (as a typical case) while considering the participant pool. The hemiplegic stroke participants involved in the study often demonstrated small movement in different directions during quite standing to stabilize their posture within the tasks. Such small movements resulted in variation in the value of d_{ins} that could cause false alarm while detecting whether the

Ankle strategy was followed or not. Also, the post-stroke participants often demonstrated the problem of foot inversion while standing. The purpose of using the d_{th} was to minimize the chances of false alarm due to such minor movements while standing. Also, there was no tolerance or margin between the two states of Ankle strategy, namely, ***Followed*** or ***Not Followed***, as far as this initial study was concerned, as can be seen from Equation (5.3). While the participant performed VR-based tasks, the HLD unit transmitted d_{ins} (for the heel lift information) to the serial port of the Task Computer presenting the VR-based tasks in real-time. In this study, the frame rate of transmission of the HLD information was chosen to be 60Hz.

Also, based on the d_{ins} value, if the participants did not follow Ankle strategy, then a penalty factor was added in their performance score (described below).

5.2.6 Performance Evaluation Unit

I wanted to understand the participants' ability of weight-shifting in different directions while they performed the VR-based tasks presented by the Virtual CoMBaT system. To do that, it was required to evaluate their performance in the direction-specific VR-based tasks. Therefore, the Virtual CoMBaT system computed participant's performance scores for the tasks of DL1 and DL2 in each of the five directions. The performance score was evaluated based on (i) P_{S1}: length of CoM trajectory (T_L) before reaching the *Target* position, (ii) P_{S2}: deviation of CoM from the instructed straight path between *Central*$_{Hold}$ and *Target* positions, (iii) P_{S3}: ability to hold his shifted weight at the *Target* position for Hold time (H_T) of 1 second, and (iv) P_{S4}: penalizing factor to discourage heel lifting. This unit was similar to that used in evaluating performance for the task of DL1 and DL2 in the previous Chapter 4. For details, please see Section 4.2.4 of Chapter 4.

The first metric (P_{S1}) was used to evaluate the participant's CoM trajectory for body sway while shifting weight by using Eq. (5.5).

$$P_{S1} = \begin{cases} 100 & ;\ \text{if } T_L \leq D_{TH} \\ 100 - \propto * \left(\frac{T_L - D_{TH}}{D_{TH}}\right) * 100 & ;\ \text{if } D_{TH} < T_L < 3 * D_{TH} \\ 0 & ;\ T_L \geq 3 * D_{TH} \end{cases} \tag{5.5}$$

$$D_{TH} = 1.8 * T_L \tag{5.6}$$

Where, T_L is the length of CoM trajectory. In Equation (5.5), P_{S1} can have three possible values based on the value of T_L. The value of P_{S1} was programmed to be 100 and 0 for $T_L < D_{TH}$ and $T_L \geq 3*D_{TH}$, respectively. Again, for the intermediate values of T_L, the value of P_{S1} was linearly reduced from 100 to 0 with the help of a multiplication factor of α=0.5 in Equation (5.5) due to increasing value of T_L between D_{TH} and $3*D_{TH}$ (Fig. 5.5). The range of the values of T_L (as

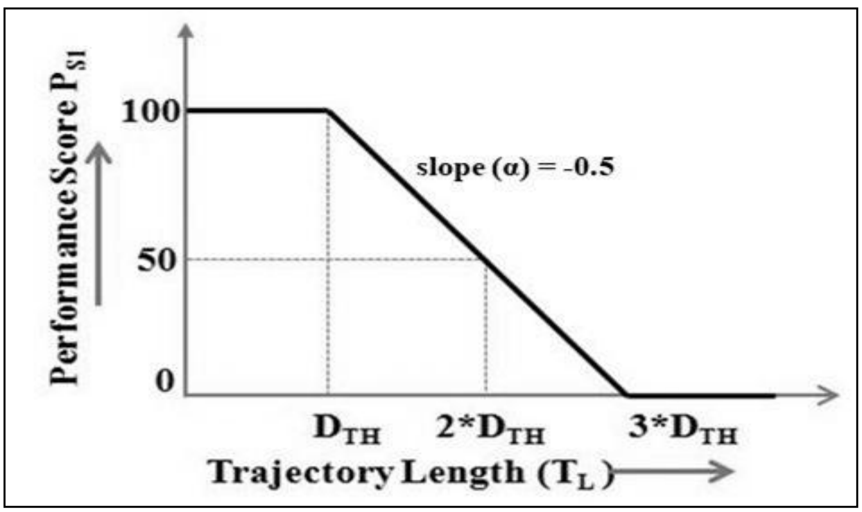


Figure 5.5. Evaluation of the participant's CoM trajectory for body sway while shifting weight (P_{S1})

function of D_{TH}) was chosen as an initial approximation. This can be changed based on the study design. The value of D_{TH} was decided based on a pilot study with healthy participants (n=7; Mean (SD) = 39 (16.45) years). For details on computation of P_{S1}, please see Section 4.2.4.1 of Chapter 4.

The second metric (P_{S2}) evaluated a participant's quality of weight-shifting in terms of deviation (D_A) of CoM trajectory upon reaching the *Target* position along the instructed direction (defined by $\theta_x=0^0$ for *East*; 45^0 for *North-East*; 90^0 for *North*; 135^0 for *North-West*; and 180^0 for *West*) with a tolerance range (θ_{RANGE}) of $\pm22.5^0$ around the instructed direction.

$$P_{S2} = 100 - \propto * \left(\frac{abs(\theta_x - D_A)}{\theta_{RANGE}}\right) * 100 \tag{5.7}$$

Here, $\alpha = ½$.

The third metric (P_{S3}) was considered to encourage stable weight-shifting by the participant.

$$P_{S3} = \begin{cases} 100 & ;\ if\ H_T \geq 1s \\ 0 & ;\ if H_T < 1s \end{cases} \tag{5.8}$$

The fourth metric (P_{S4}) was used to penalize the participant for heel lift during weight-shifting.

$$P_{S4} = \begin{cases} -100 & ;\ Ankle\ strategy\ 'NotFollowed' \\ 0 & ;\ Ankle\ strategy\ 'Followed' \end{cases} \tag{5.9}$$

Therefore, the weighted performance score (P_s^x) for each of the five directions (x = *North* / *East / West / North East / North West*) was calculated as:

$$P_s^x = 0.5P_{s1}^x + 0.25P_{s2}^x + 0.25P_{s3}^x + 0.2P_{s4}^x \tag{5.10}$$

As suggested by the therapist in my team, a higher weightage to P_{s1} (0.5) than P_{s2} (0.25) and P_{s3} (0.25) was assigned. In activities of daily living that require one to do weight-shifting, reduced postural sway (indicated by P_{s1}) is often considered more important than being right at the *Target* location (P_{S2}) and holding the shifted-weight posture (P_{S3}). This was realized in the current study through higher weightage being assigned to P_{s1}. Also, to encourage the participants to follow the Ankle strategy, a penalty factor (0.2) was added in case Ankle strategy was *Not Followed* while shifting weight. The penalty factor of 0.2 was chosen as an initial approximation and it can be changed in future based on the study design.

The final performance score (P_s) for each task was computed as the average of the performance scores for all the five directions by using Eq. (5.11).

$$P_s = \frac{1}{5}\sum_{x} P_s^x \tag{5.11}$$

The participant pool had both left and right hemiplegic participants. Therefore, it can be expected that their ability to shift weight in a specific direction may depend on the side of hemiplegia along with the amount of residual balance. Thus, I wanted to quantify participants' residual balance in terms of weight-shifting ability in each direction *(N, E, W, NE,* and *NW)* when they came in for the study. To achieve this, the Normalized Equivalent Performance (NEP) was computed for each participant for each of the five directions. The individual-specific NEP was decided based on the performance score in the *First* Attempt (that is, the first task in each of the two difficulty levels) of DL1 and DL2 for each participant by using Equation (5.12).

$$NEP_x = \frac{\left(\frac{1}{3} \times P_{f_{DL1_x}} + \frac{2}{3} \times P_{f_{DL2_x}}\right)}{100} \; ; x = N, E, W, NE, NW \tag{5.12}$$

Where, $P_{f_{DL1}}$ and $P_{f_{DL2}}$were participant's performance score for each of the five directions (*N, E, W, NE,* and *NW*) in the *First* Attempt of DL1 and *First* Attempt of DL2 tasks, respectively. The system used NEP score to quantify the participant's residual balance, since the tasks of DL1 and DL2 were of different difficulty levels. The DL2 tasks were more difficult than those in DL1 and required larger weight-shifting than that in DL1 tasks for completing the VR-based tasks. Therefore, the NEP was estimated as a weighted average of the scores in the two subtasks belonging to DL1 and DL2.

5.2.7 Task Switching Unit

In this study, one of the major contributions was to make the Virtual CoMBaT system adaptive to a participant's ability of weight-shifting and accordingly offer him / her tasks of varying difficulty levels. For this, a task switching unit was developed that offered various tasks to the participants while being adaptive to their individualized performance score in a task. The task switching unit considered two switching conditions (C1 and C2; Table 5.1). While a participant interacted with VR-based tasks, I was interested to know the trajectory of

Table 5.1. Task switching criteria of Virtual CoMBaT system

Condition	Description	Action
C1	$P_f\ (CT)_{i+1} < P_f\ (PT)_i$ and $P_f\ (CT)_i > 70\%$; where, i=1....n	Move to higher difficulty level (except for DL2)
C2	$P_f\ (CT)_{i+1} \geq P_f\ (PT)_i$ where, i=1....n	Remain in same difficulty level

improvement in performance score in tasks of each difficulty level. Therefore, when a participant was interacting with a task of DL1, I designed the Virtual CoMBaT system to continuously monitor whether (i) there was an improvement in his/her task performance in the current task compared to that in the previous task (that is, $P_f\ [CT] – P_f\ [PT]) > 0$; P_f = Percentage performance score, CT=current task and PT= previous task) (Table 5.1 and Fig 5.6) and (ii) his/her performance in any of the task trials of DL1 before the CT trial was 'Adequate' (Condition 'C1'). Unlike that in the previous study (Chapter 4), here, the tasks were not switched only on the basis of whether a participant scored 'Adequately' in a task belonging to a particular difficulty level. Instead, even if a participant scored 'Adequately', the system offered him/her the tasks of same difficulty level to see his maximum possible ability of weight-shifting in that difficulty level. For example, if a participant had scored 'Adequately' in one of the tasks belonging to DL1, then, the system did not switch the participant to a task of DL2 immediately. Instead, he/she was offered tasks in DL1 only to see whether his performance score improved further. There might be a case in which after performing 'Adequately' and gradually improving performance score in the tasks of DL1, there might be a dip in the performance score. One of the possibilities of such a drop can be due to loss of interest that might be because of repeated exposure to tasks of the same challenge level. In that case, the Virtual CoMBaT system offered

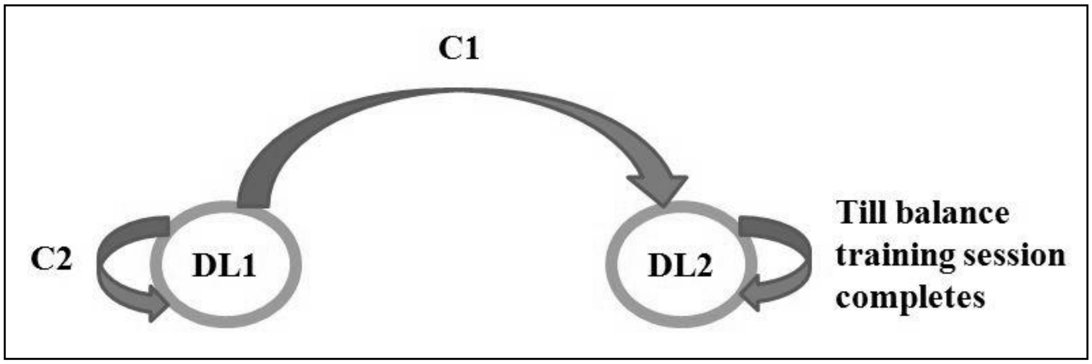


Figure 5.6. State machine representation for task switching in Virtual CoMBaT system

tasks of higher difficulty level (DL2) so as to regain the participant's interest in the balance training task. After being switched to DL2, the participant was expected to interact with the tasks of DL2 until the task completion time (20 minutes in this case) was over. In this study, the performance score ≥ 70% was chosen as 'Adequate' performance while <70% was considered as 'Inadequate'. The threshold value for 'Adequate' was chosen to be 70% since, literature indicates 70% as the average initial exercise performance for rehabilitation tasks [21], for outpatient clinics [22] and technology-assisted skill learning [23].

5.2.8 System Usability related Questionnaire

When interacting with a system or tool that helps users to achieve desired goals for which the system is designed, measuring user's experience on the usability of the system is critical [24], [25]. In the research presented in this chapter, I have designed a Usability study with the Virtual CoMBaT system. The idea was to understand the usability of the developed system using structured questionnaire, unlike the semi-structured feedback questionnaire that was used in the previous study (Chapter 4). Researchers have used different questionnaires such as System Usability Scale (SUS) [26], [27], VR usability (VRUSE) diagnostic tool [28], and others to evaluate the usability of various systems. These questionnaires have found wide usage in the evaluation of usability of security software [29], mobile phones [30], [31], Social Network sites [32], wiki sites [33], E-Yoga system [34], and others. These questionnaires though useful, have not been designed keeping perspectives of rehabilitation in mind. Specifically, in recent times, with rapid technological progress, researchers are exploring technology-assisted solutions to address issues of rehabilitation. For such technological solutions to succeed, proper usability measures need to be explored. Among the alternate technological solutions in rehabilitation, VR-based rehabilitation is an emerging field [35], [36], [37], [38]. Often, investigators claim that the VR-based systems can show promising results as far as rehabilitation is concerned [39], [40], [41] and can work as complementary tool in the hands of clinicians involved in rehabilitation [42]. Given the importance of checking the usability of such VR-based rehabilitation platforms before these can be deployed in real settings, Gil Gomej et al. [43] have designed User Suitability Evaluation Questionnaires (USEQ) based on a five-point Likert scale [44].

In order to understand the usability of the Virtual CoMBaT system, a set of questionnaires based on five point Likert scale [44] was used to get the participant's feedback post the study. Specifically, a set of five questions was framed in order to understand participants' views on their interaction with the Virtual CoMBaT system. Three out of the five questions were chosen (relevant for the current study) from the User Suitability Evaluation Questionnaires (USEQ) used by Gil-Gómez, José-Antonio, et al. [45]. The first question was "Did you face any difficulty in understanding the task?" (Q1). The idea behind asking this question was to understand whether the instructions provided by Virtual CoMBaT system during the VR-based task were clear. The second question was "Did you find the task interesting?" (Q2). This question was asked to know how the participants felt in interacting with Virtual CoMBaT system. Although the usability study was conducted for one day only, yet, this system was designed with an aim to develop a balance rehabilitation system in mind that might need its usage over extended period of time. Thus, I asked them the third question, namely, “Do you think that the repeated usage of this system would be beneficial to you?” (Q3). In addition to these three questions, I also wanted to understand whether the use of Virtual CoMBaT system was motivating to the participants. For this, I asked them two more questions, namely, “Will you agree to interact with the system again?” (Q4) and “Will you refer others to participate in the study?” (Q5).

5.3 Experiment and Methods

5.3.1 Participants

The study was carried out after informed consent at CMP college of nursing, Gandhinagar and Sadbhavna charitable trust clinic, Ahmedabad where post-stroke hemiplegic patients were

Table 5.2. Participants' metadata for usability study with Virtual CoMBaT system

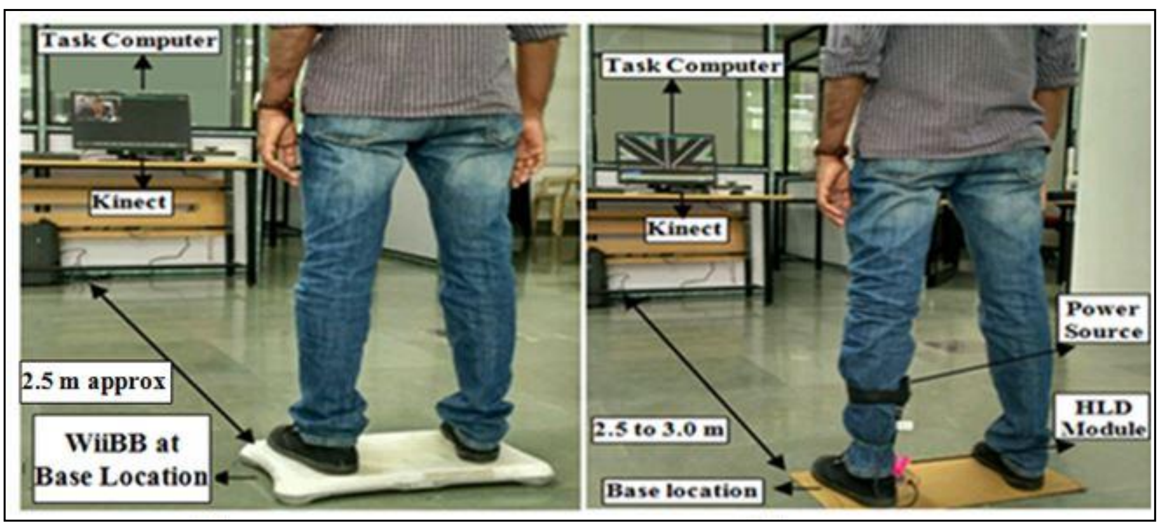
ID	Age/Sex	Affected Side	Post-stroke Period	BBS Score
S1	51/Male	Left	2 months	31
S2	47/Male	Right	1.5 years	21
S3	57/Male	Right	7 days	46
S4	70/Male	Left	3 years	31
S5	58/Male	Right	3 years	53
S6	36/Male	Left	1 years	46
S7	60/Male	Right	5 months	30
S8	56/Male	Left	1 month	36
S9	74/Male	Left	8 days	51
S10	52/Male	Right	2.5 years	23
S11	57/Male	Left	3 months	40
S12	45/Female	Left	8 months	25

BBS= Berg Balance Scale

undergoing treatment. The study followed institutional research ethics approved by ethics review committee of Indian Institute of Technology Gandhinagar. In the present usability study, twelve hemiplegic post-stroke survivors (S1-S12) (mean (SD)=55.25 years (10.34)) with varying post-stroke period and residual balance (based on the availability) participated. The BBS score of the participants varied between 21 to 53 indicating a wide spectrum of balance ability. Table 5.2 shows the participants' metadata. The participants did not have any prior exposure to computer-based tasks. The inclusion criteria for the current study were (i) ability to follow the instructions (ii) ability to stand and shift weight without orthopedic aids and (iii) should not have gone through any surgery in recent past that may interfere with their capability to do the weight-shifting tasks.

5.3.2 Experimental Setup

The experimental setup shown in Fig 5.7 consisted of a Kinect, a WiiBB, a HLD unit and Task Computer (PC). This study consisted of three stages, namely (i) SESC Identification (ii) Threshold Estimation (for individualized weight-shifting) and (iii) Task Execution stages. During the SESC Identification stage (as described in Section 5.2.1), the experimental setup consisted of a Kinect, a WiiBB and a PC. The WiiBB was kept on the ground at a distance of approximately 2.5 meter in front of the PC (Fig. 5.7(a)). The location of the WiiBB during SESC



(a) (b)

Figure 5.7. Experimental setup for (a) SESC Identification stage, (b) Threshold Estimation and Task Execution stages.

Identification stage was marked as Base Location for next stages. The participant was supposed to stand on the WiiBB placed at the Base Location. The Kinect sensor (for tracking the participant's movement) was connected to the PC kept on a table top and positioned near the PC with the Kinect camera facing towards the person standing on the top of the WiiBB.

In the Threshold Estimation stage, the WiiBB was removed and the participant was expected to stand on the ground at the Base Location with a HLD unit attached to the participant's shoe worn on the Affected leg (Fig. 5.7(b)). The experimental setup during the Task Execution stage was kept the same as it was for the Threshold estimation stage (Fig 5.7(b)).

5.3.3 Procedure

The study required a commitment of approximately forty minutes from each participant. Once the participant arrived in the experiment room, he/she was asked to sit on a chair and relax for 5 minutes. After that, a physiotherapist in the team assessed the participant’s residual balance using Berg Balance Scale Score [46] and also made sure that the inclusion criteria for the study (similar to that in the previous study) were satisfied. This took around 10 minutes for each participant. If the inclusion criteria were satisfied, then the experimenter explained him/her the

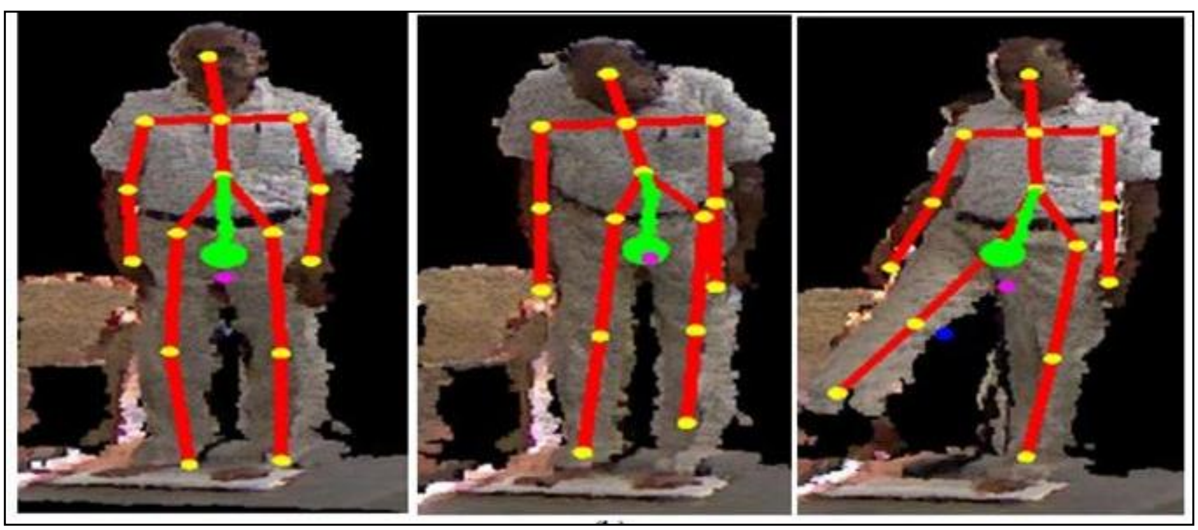


Figure 5.8. Typical SESC Identification stage; the participant was asked to hold a series of static postures while standing on top of a WiiBB.

experimental setup and demonstrated two VR-based tasks, one in each difficulty level. Then the experimenter ensured that the participant understood the task followed by signing of consent form. The participant was told that he/she can quit from the study at any time if he/she felt uncomfortable. Before starting the study, the experimenter asked the participant for his/her verbal consent.

Once the participant was ready, the experimenter started the experimental study that comprised of three stages (Section 5.3.2). First stage was SESC Identification stage in which, the participant was asked to stand upright on the top of WiiBB (Fig. 5.7(a)) kept at the Base Location and facing towards Kinect sensor placed at the bottom of the PC. Then the experimenter started the Kinect device and made sure that participant's skeleton was detected by the Kinect as well as WiiBB was providing CoP values of the participant. This was necessary for proper functioning of SESC algorithm. After that, the participant was asked to make few static postures, such as lean forward and lift one leg at a time away from the body (Fig. 5.8). These static postures were used to identify the SESC parameters in real-time. This stage was followed by the Threshold Estimation stage (Section 5.3.2). Before the start of the Threshold Estimation stage, the participant was asked to sit on a chair and the experimenter mounted the HLD unit (Section 5.2.5) on the shoe worn on the Affected leg of the participant (Fig. 5.4(b)). Also, the WiiBB was removed from the Base Location. This stage started by asking the participant to stand on the ground at the Base Location (Fig. 5.7 (b)) while facing the Kinect sensor. Once the Kinect sensor connected to the PC started tracking the participant's CoM position, the experimenter instructed the participant to shift his / her weight in the *N* (forward), *E* (right side) and *W* (left side) directions to his maximum possible capability while following Ankle strategy and not stepping out from the Base Location. This stage was used to determine direction-specific individualized threshold of weight-shifting (Section 5.2.3). Finally, the Task Execution stage was performed in which the participant was asked to stand on the ground at the Base Location and interact with the VR-based balance tasks by shifting his / her weight in the instructed direction. The VR-based balance tasks (Fig 5.2 and Fig. 5.3) required them to shift their weight in different directions so as to maneuver the VR_{Obj} in the VR environment from $Central_{Hold}$ to *Target* position (Section 5.2.2). After the completion of the Task Execution stage, the participants were asked to respond to a system usability questionnaire administered by the experimenter to get the participant's feedback on the usability of the Virtual CoMBaT system.

5.3.4 Statistical Analysis

During the Task Execution stage, the Virtual CoMBaT system computed the participant's performance score (P_S^x; Section 5.2.6) corresponding to each direction in the tasks of DL1 and DL2. I wanted to to understand whether interaction with the Virtual CoMBaT system which offered repeated exposure to tasks in each difficulty level, contributed to any statistically significant improvement in participant's performance score. Specifically, I wanted to know whether the improvement (if any) in the participants' performance score in their *First* Attempt task to that in the task (within each difficulty level) in which they performed best (*Best Case* henceforth) in each direction (such as. *N, W, E, NE,* and *NW*) was statistical. For this, the normality of the data was checked by performing Shapiro-Wilk test of normality [47] on the participants' performance score for each of the five directions of the tasks corresponding to their *First* Attempt and *Best Case* Attempt tasks of DL1 and DL2. For the sample size of 12 participants, a W value was calculated and found that the data was not normally distributed with significance level of p-value=0.05. Therefore, a non-parametric statistical hypothesis testing, namely, Wilcoxon signed rank test [48] was performed to know the significance of improved performance (if any) in the *Best Case* Attempt from the *First* Attempt for all the directions in DL1 and DL2. The Wilcoxon signed rank test was performed by keeping the performance score in the *First* Attempt and *Best Case* Attempt as between-subject factor and difficulty level (DL1 and DL2) as within-subject factor. Also, for paired difference between performance scores obtained in different directions (*N, W, E, NE,* and *NW*), a multiple comparison correction using Holm correction method [49] was performed. The test was performed with the significance level set at $p\text{-value} < 0.05$.

5.4 Results

In this study, I have designed a Virtual CoMBaT system and conducted a usability study to understand the users' view on the usage of Virtual CoMBaT system. In the section below, I present the findings on the participants' perspective on the usage of the system. Also, since, Virtual CoMBaT was designed with an aim to present a balance rehabilitation platform in the long run, I wanted to understand the potential of Virtual CoMBaT system to contribute to improving participants' performance in VR-based balance tasks even over a limited duration of exposure. Therefore, while participants interacted with the Virtual CoMBaT system, I monitored their ability to shift weight in different directions by following Ankle strategy. Also, to make sure that the Ankle strategy was followed, I have used an HLD unit. In the following sections, I will present the observation on the implication of the HLD-assisted Virtual CoMBaT system on the hemiplegic stroke participants' weight-shifting capability along with improved usage of Ankle strategy. In turn, I investigated whether the Virtual CoMBaT system augmented with a personalized CoM can (i) quantify participant's residual balance ability based on his/her ability to shift weight in different directions and (ii) improve participant's balance in terms of improved weight-shifting capability. Here, I present the result of the Usability study, reported in one of my published manuscript [50], in which 12 stroke participants volunteered.

5.4.1 Participants' Feedback on System Usability Questionnaires

At the end of the study, the participants were asked to respond to the system usability questionnaires based on a five point Likert scale [44]. The Table 5.3 shows the group average of the response to the system usability questionnaires. The responses indicate that they (except participant S6 and S10) did not face any difficulty in understanding the tasks and were in fact very interested in interacting with the system. Though, the participants were provided only one-day exposure to Virtual CoMBaT, yet, they could realize the potential benefits that such a system can bring to them as far as balance rehabilitation was concerned. Therefore, they expressed their

Table 5.3. Participants' feedback for Virtual CoMBaT system

Q. No	Suitability Evaluation Questions	Average Response Score
Q 1	Did you face any difficulty in understanding the tasks?	1
Q 2	Did you find the tasks interesting?	4
Q 3	Do you think you can benefit by using such a system?	4
Q 4	Do you want to play again with this system?	5
Q 5	Do you want to refer any of your acquaintance to the study?	5

Note: 1 = Strongly disagree, 2= Disagree, 3= Neutral, 4= Agree and 5= Strongly agree.

willingness to participate in the study again in future and also refer their known acquaintances to interact with Virtual CoMBaT. Thus, looking at all the feedback from the participants, I can infer that the Virtual CoMBaT system has the potential to be accepted by the target population.

5.4.2 Virtual CoMBaT System used for Quantification of Balance Ability based on Weight-shifting capability

In this study, one of the aims was to understand the potential of Virtual CoMBaT system to quantify an individual's direction-specific residual balance in terms of his / her ability to shift weight in different directions. For this, I analyzed the participant's Normalized Equivalent Performance (NEP) (section 5.2.6) while the participant interacted with the *First* Attempt tasks of DL1 and DL2. Fig. 5.9 shows each of the hemiplegic participant's *NEP* for all the five directions, their Berg Balance Scale (BBS) scores and Affected side. From Fig. 5.9, I can see that the participants showed lesser *NEP* while maneuvering the VR_{obj} in the direction (s) that required

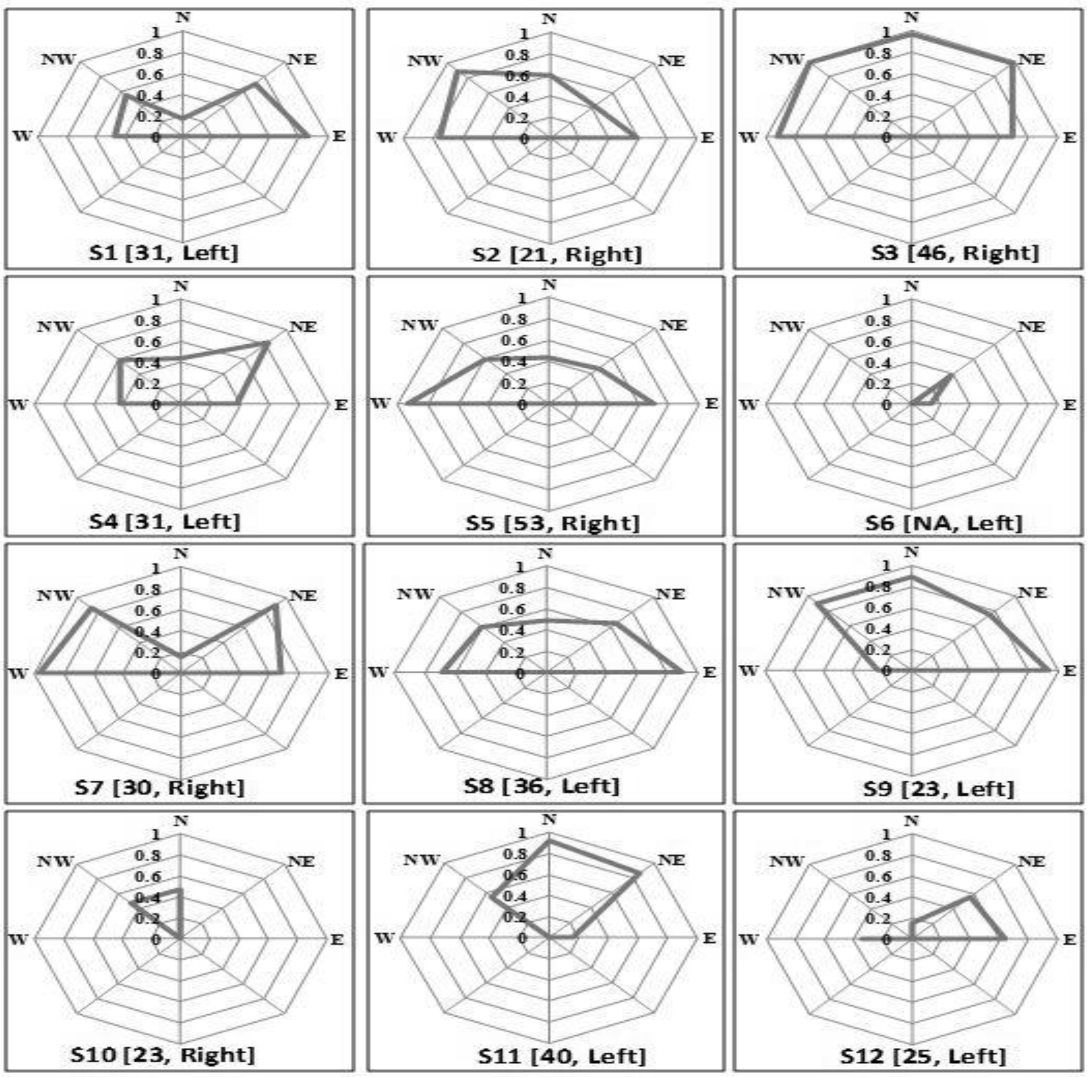


Figure 5.9. Individual Normalized Equivalent Performance score in all five directions.
Note: Notation written below each plot = Participant's ID [BBS score, Affected side]

more weight-shifting on their Affected leg. For example, participant S1 was left hemiplegic and therefore his weight-shifting capability on the left side was less compared to that in his right side. The left leg being comparitively weaker resulted in reducing the performance score in the VR-based tasks while shifting his weight to maneuver the VR_{obj} towards the *W, NW and N* directions. Specifically, directions *W and NW* required him to use his Affected leg to shift his body weight more towards his left side. Again, to maneuver the VR_{obj} in the *N* direction, one would ideally need equal contribution of both left and right legs to facilitate adequate weight-shifting. But, since S1 was left hemiplegic, his ability to shift weight was restricted towards the left side of his body that resulted in the VR_{obj} to be maneuvered more towards the *NE* direction rather than towards the *N* direction, causing him to score less along the *N* direction. The opposite was the case for the participant S2. He was right hemiplegic and thus as expected, his performance scores along the *E*, *NE* and *N* directions were lesser compared to that for the other directions due to restricted weight-shifting ability towards the right side of his body. However, there were some exceptions, such as participant S3 was right hemiplegic. Thus, he was expected to exhibit comparatively lesser performance score in the *E, NE* and *N* directions. But, participant S3 performed well in all the directions except for the *E* direction. This may be possibly due to S3 having most of his residual functional capabilities intact. His clinical report indicated that he suffered from a very mild stroke and his high BBS score suggested that he was not suffering from adverse balance disorders.

The weight-shifting profile shown in Fig. 5.9 can provide a quantified pictorial representation of a participant's direction-specific weight-shifting capability. This information can serve as a complementary information for a therapist to plan individualized rehabilitation exercise program. Though, I have tried to connect a participant's ability to shift weight in particular direction with the hemiplegic side, yet I do not want to generalize the findings due to lack of enough sample power.

Again, I was interested to see if there existed any correlation of the clinical measure of balance with task performance in the tasks offered by the Virtual CoMBaT system. As, in the current study, I had the participants' BBS data, I wanted to use this data as participants' quantitative clinical measure of balance score. However, BBS score is not just a measure of direction-specific weight shifting ability, rather it is a combination of 14 items including sitting to standing, standing unsupported, sitting unsupported and other tasks designed to measure complete balance of an individual in a clinical setting. Therefore, I decided to choose one item out of the 14 items (each scaled 1 to 4) of BBS score that can provide us direction-specific information. Specifically, I selected BBS task item number 8 that is 'Reaching Forward with Outstretched Arm while Standing' task of BBS (in which the participant needs to shift weight towards the front (*North*)). I tried to find out correlation of this quantitative clinical score with participants' performance score while maneuvering VR_{Obj} towards *North* direction in the *First* Attempt of DL1 task. During the computation of the correlation, only the BBS score for item 8 was considered for 8 (S1, S2, S4, S5, S7-S9 and S12) out of the 12 participants, since, only for these participants the BBS scores for each of the 14 items of BBS were available. For rest of the participants, only the total score (without breakup) was available. The correlation was found to be 0.84. From this I can infer that the correlation of the clinical measure of balance with participants' task performance in the tasks offered by the Virtual CoMBaT system was high.

5.4.3 Implication of Virtual CoMBaT System on participant's Performance

I designed the Virtual CoMBaT system with an ultimate long range aim that it can serve as a balance rehabilitation platform. Therefore, the second aim of this study was to understand the potential of Virtual CoMBaT system to improve a participant's weight-shifting capability in

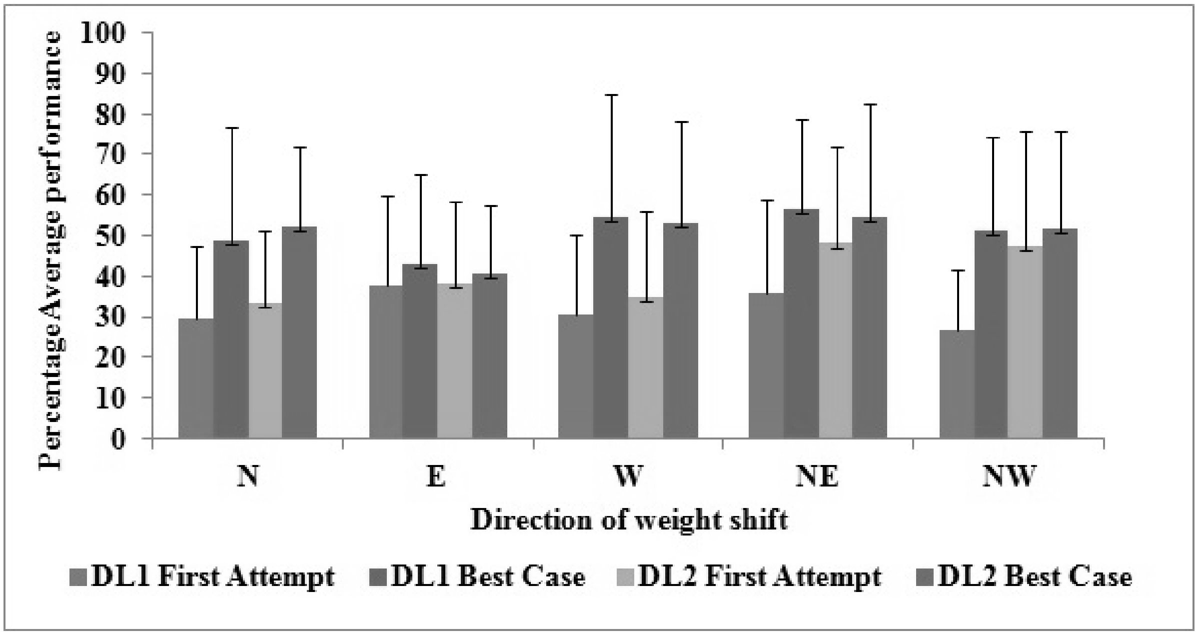


Figure 5.10. Group Average of % Performance score in *First* Attempt and *Best Case* Attempt in each direction
Note: DL1 = Difficulty level 1; DL2 = Difficulty level 2

terms of improved performance in the VR-based tasks. So, I evaluated participants' performance in the VR-based tasks to see whether there was any improvement in their performance score even during its usage for a limited duration that is for one session. Then only it is possible to judge its worthiness to be used over extended period for rehabilitation. For this, I computed the participants' performance scores in the *First* Attempt and the *Best Case* Attempt in tasks belonging to DL1 and DL2. Fig. 5.10 shows the comparison of the group average of the participants' percentage performance score in all the five directions for *First* Attempt and *Best Case*. I could observe improvement (Δ) in the group average (irrespective of the hemiplegic side) of %performance score from the *First* Attempt to *Best Case* Attempt (Δ=61.85% for DL1 and 24.1% for DL2) averaged over all the five directions. Also, a high variation in the group average % performance in each direction can be observed from Fig. 5.10. This can be attributed to the participants having widely varying balance capabilities as evident from the wide range of BBS scores. This might adversely affect the group statistics.

Again, for each participant, the *First* Attempt and *Best Case* Attempt were separated by intermediate trials and the number of intermediate trials varied across participants. From Table 5.4, it can be seen that all the participants (except S2, S6 and S10) took more number of trials to reach the *Best Case* Attempt in DL2 than that in DL1. This was expected, as DL2 was

Table 5.4. Number of trials needed to reach *Best Case* Attempt

Participants' ID	Number. of trials	
	DL1 (No.)	DL2 (No.)
S1	3	18
S2	9	2
S3	2	18
S4	5	7
S5	9	13
S6	4	NA
S7	2	22
S8	2	13
S9	1	17
S10	8	NA
S11	5	11
S12	3	5

Note: The number of trials are inclusive of *First* Attempt in both DL1 and DL2. 'NA' indicates that 'none'.

comparatively more difficult than DL1. The participant S2 played less number of trials in DL2 than DL1 to reach the *Best Case* Attempt. This could be because, S2 had least BBS score and therefore his balance was considerably impaired. Consequently, he spent most of the time from the 20 minute balance training duration in interacting with DL1 tasks before switching to DL2, unlike other participants. For the participants S6 and S10, I find that both of them spent entire 20 minutes of balance training session in interacting with DL1 tasks. This might be due to the fact that, both of them were having issues in understanding the tasks and the instructions given by the experimenter. As a result, they took longer to finish each task in the DL1, as reported by the experimenter. Also, I tried to examine the trend in participants' performance in the task trials that they played before reaching to the *Best Case* Attempt. I found that for most of the participants, there was an improving trend in the % performance score in the trials between their *First* Attempt to *Best Case* Attempt. I also observed that the group variability in the % performance score across the trials executed before reaching the *Best Case* Attempt was approximately 10% and 8% for tasks of DL1 and DL2 respectively.

In the current study, the Virtual CoMBaT offered tasks of higher difficulty level when the system detected that performance was ‘Adequate’ coupled with a decrease in participant’s performance score from that in a previous task (Table 5.1). Here, I did not consider any tolerance or margin as far as the decrement in performance score was concerned before switching to a task of the next higher difficulty level. However, the results indicate that for all the participants (except S3) this decrement in performance score before switching to the next higher difficulty level was ≥ 2%. The Table 5.4 indicates the number of trials (inclusive of the *First* Attempt) needed by a participant before reaching the *Best Case* Attempt. From Table 5.4, I can see that the participant S9 needed only 1 trial to reach the *Best Case* in DL1. Specifically, S9 interacted with 2 trials in DL1 that is, one trial for the *First* Attempt that happened to be the *Best Case* for him and one trial in which his performance score was less than that of *First* Attempt before he was switched to the DL2 tasks. The percentage change (%Δ) in performance score of S9 in the tasks of DL1 from the *Best Case* Attempt to that in the next trial before switching over to the DL2 was 2.92%. Please note that getting even a small improvement over one-day exposure can be considered as a step towards achieving improved performance. Again, in the case of participants S1, S7, S8 and S12, the number of trials needed to reach the *Best Case* (in DL1) was only 3, 2, 2, 3 trials respectively. However, the corresponding %Δ in the performance score from their *Best Case* Attempt to that in the next trial was approximately 23%, 11%, 2% and 3% for S1, S7, S8 and S12, respectively. For the participant S3, the %Δ in performance score from the *Best Case* Attempt to the next trial before switching to the DL2 was very less (approximately 0.2 %). However, the performance score of participant S3 was approximately 92% in all three trials (*First* Attempt, *Best Case* Attempt and the one before switching over to the DL2) and therefore the fourth task trial (that is first trial in DL2) helped to break the monotony. However, further modification in the Condition 1 (Table 5.1) of the Task Switching rationale is possible by choosing a specific value of %Δ in performance score (from the *Best Case* Attempt to that in the next trial before switching over to the DL2), say by x% change, with x = 2% (say, as a typical case) for switching a participant to tasks of higher difficulty level that is DL2.

Before performing the statistical test of the hypothesis, I performed Shapiro-Wilk test of normality to see if the distribution of the data on the participants' performance was normal. From the W statistics obtained from the Shapiro-Wilk test, I found that the average performance score (%) was not normally distributed for most of the directions, particularly for the performance score of the *First* Attempt for both the DL1 and DL2. Therefore, to assess whether the improvement in the participants' average performance score (%) from *First* Attempt to the *Best Case* Attempt was significant, I carried out a dependent sample Wilcoxon signed-rank test on the score in tasks belonging to DL1 and DL2. Also, multiple comparison correction using Holms method was applied on the p-values obtained from Wilcoxon signed-rank test. The statistical significance test was performed to assess the significance of the improvement in performance score for each of the five directions (*N, E, W, NE* and *NW*) separately. From the results, I observed a statistically significant improvement (p-value=0.024 for *N*, p-value=0.015 for *W*, p-value=0.016 for *NE* and p-value=0.042 for *NW*) in the % performance score in the *Best Case* Attempt from *First* Attempt for the different directions (except for direction *E*). For the *East* direction, the improvement in performance was not statistical. This may be because, in this study, 7 out of the 12 participants (Table 5.2) being left hemiplegic, performed better while shifting weight towards *East* direction since the beginning of the balance task than other directions irrespective of the task trial. In case of DL2 tasks, the improvement in the performance score from *First* Attempt to *Best Case* Attempt was not significant. A possible reason for this could be that the tasks of DL2 being more difficult, required more weight-shifting than that in the tasks of DL1 to bring in noticeable improvement in the performance score. However, a longitudinal study with more participants is required before generalizing such observations.

5.4.4 Implication of Virtual CoMBaT System on adherence to Ankle Strategy

In this study, I wanted to encourage the participants to follow Ankle strategy while interacting with the Virtual CoMBaT system. For this, the Heel Lift Detection (HLD) unit (Section 5.2.5) was used to monitor the participant's heel lift during Task Execution stage. While monitoring participant's heel lift, I recorded the total duration for which a participant lifted his/her heel (***Not Following*** Ankle strategy) in a VR-based task trial. Subsequently, for each task trial, I computed the total duration for which a participant lifted his/her heel in a task trial as a percentage of the total time taken to execute that task trial. The Fig. 5.11 shows the group average % heel lift time (out of the total time taken) for each of the *First* Attempt and *Best Case* Attempt for tasks belonging to both DL1 and DL2. The aim was to understand whether repeated exposure to the Virtual CoMBaT system augmented with HLD unit facilitated the participants to improve their weight-shifting capability while reducing the duration of heel lift (that is improved usage of Ankle strategy). From Fig. 5.11, it can be seen that there was a reduction in the group average percentage time (out of total duration of task trial) the participants had lifted their heels from their *First* Attempt to *Best Case* Attempt while performing weight-shifting tasks in each difficulty level. In case of DL1, the participants were frequently lifting their heels while performing *First* Attempt task. In contrast, while performing the *Best Case* Attempt in DL1,

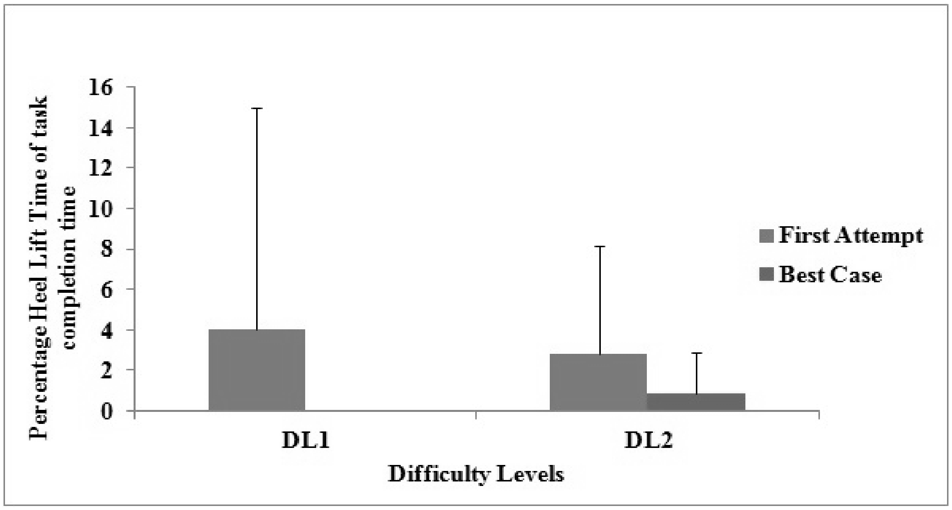


Figure 5.11. Group average of % heel lift duration

none of the participants lifted their heels from the surface of the ground. Also, for the more difficult tasks in DL2, a reduction in group average percentage heel lift in their *Best case* Attempt from the *First* Attempt can be seen. Specifically, in DL2 tasks, the amount of time of heel lift in the *Best Case* was 68.92% less than that for the *First* Attempt. These findings indicate that the Virtual CoMBaT system augmented with HLD unit helped the participants to improve their performance with improved adherence to Ankle strategy during weight-shifting.

5.5 Discussion and Limitation

The main contribution of the present work was the design of CoM-assisted VR-based balance training system augmented with Kinect to offer balance training exercises. This system was designed to be individualized and adaptive based on participant's performance capabilities. Additionally, to encourage the participant in using Ankle strategy during weight-shifting, an ultrasonic sensor based heel lift detection unit was used that can alert a participant with audio alarm if he/she did not follow the Ankle strategy. Finally, to test the Virtual CoMBaT system, a Usability study was carried out to understand the implications of such a system on the balance of stroke participants.

A Usability study with 12 stroke survivors was conducted. From participants' feedback to the system usability questionnaire, it can be inferred that the Virtual CoMBaT system has a potential to be accepted by the target population. Also, the results of the usability study indicate that the system has potential to contribute to improving stroke participants’ overall performance in tasks belonging to different difficulty levels. Though the study was carried out for a limited duration (one session), yet, there was statistical improvement in participants' performance score from *First* Attempt to the *Best Case* Attempt for the tasks of DL1. However, there was no statistically significant improvement in performance in tasks of DL2. Increased practice with such tasks might bring in statistical improvement in task performance even in DL2. Further, since the participants were hemiplegic, their direction-specific weight-shifting capability were restricted. The computation of Normalized Equivalent Performance enabled the system to quantify participants’ direction-specific weight-shifting capability.

Though the results of the Usability study are promising, yet, the study had some limitations. For example, the Task Switching Rationale (Table 5.1) did not have in-built tolerance while switching tasks making the task switching vulnerable to small fluctuation (that is, decrement) in % performance score. In future, I plan to fine tune the Task Switching Rationale by modifying the Condition 1 (Table 5.1) which might enhance the capability of the system as far as the balance training is concerned. Other limitations of the current study were small number (n=12) of participants having varied post-stroke periods, residual balance and different hemiplegic sides.

Also, in the present study, the participants were exposed to the Virtual CoMBaT system only for one session of balance training. Such a limited exposure may not be sufficient to judge the rehabilitation efficacy of the system. For this, one needs to carry out a longitudinal study, where a significant improvement in an individual's clinical measure of balance such as BBS score can be measured prior to and post the study. Also, in the current study, the participants had widely varying post-stroke periods as well as residual balance capability that might have affected the group average of the participants' performance scores. In future, I plan to carry out a more in-depth longitudinal study with larger patient population categorized based on residual balance capability, before such a balance training platform can be deployed in clinical settings. This would enable us to find out the rehabilitation efficacy of the system by carrying out in-depth statistical analysis on implication of Virtual CoMBaT system on the participants' balance ability. Another limitation of the system was the estimation of threshold values in the evaluation of participant's performance score in a task. Specifically, for the sake of simplicity and lack of available database on hemiplegic stroke patients, a specific threshold measure (such as D_{TH} in Eq. (5.4)) depending on pilot trials conducted with age-matched healthy participants was considered for the present study. This threshold value might differ in case of hemiplegic stroke participants and therefore can have implications on the performance score of the stroke group. However, this threshold value was chosen as an initial approximation. In future studies with Virtual CoMBaT, I plan to modify the threshold measure by using the database that I have obtained in the current study for post-stroke hemiplegic patients.

Given the potential of the Virtual CoMBaT system in contributing to improvement in stroke participants' weight-shifting ability, this system can be extended to be used as an alternate individualized rehabilitation platform at clinics and home-based settings. Though the results of the Usability study indicate the potential of the Virtual CoMBaT system to contribute to balance rehabilitation, yet this study was conducted in a controlled environment. Thus, questions still remain on the transferability of the weight-shifting skills learned from the controlled environment to real-life situations outside the simulated environment.

The current study presented the possible implications of Virtual CoMBaT system on an individual's direction-specific weight-shifting ability with improved usage of Ankle strategy. This system uses an individual's CoM measured using a WiiBB and Kinect to interact with VR-based weight-shifting tasks. This information on an individual's CoM was derived from the resultant effect of both the lower limbs while standing on the ground surface. However, I was not able to identify the contribution of each leg to the overall balance while maneuvering VR_{Obj} integrated with the overall CoM. While addressing the balance impairment in the hemiplegic post-stroke participants who often suffer from unequal weight distribution on both sides of the body, the information on the contribution of each leg towards overall balance in terms of weight-shifting ability is critical. This is because having information on the contribution of each leg to an individual's overall balance can allow us to condition the balance rehabilitation effort in such a way that the participants will be encouraged to increase their usage of the Affected leg. Thus, in the next chapter, I have tried to address this issue by using double WiiBB, one for each leg, while hemiplegic stroke participants performed VR-based balance tasks. Also, I will study the implication of variation in the weightage of the contribution of each leg towards the execution of the VR-based tasks.

Reference

[1] D. A. Winter, "Human balance and posture control during standing and walking," *Gait Posture*, vol. 3, no. 4, pp. 193–214, 1995.

[2] L. E. Cofré Lizama, M. Pijnappels, N. P. Reeves, S. M. Verschueren, and J. H. van Dieën, “Centre of pressure or centre of mass feedback in mediolateral balance assessment,” *J. Biomech.*, vol. 48, no. 3, pp. 539–543, 2015.

[3] G. I. Yatar and S. A. Yildirim, “Wii Fit balance training or progressive balance training in patients with chronic stroke: a randomised controlled trial,” *J. Phys. Ther. Sci.*, vol. 27, no. 4, pp. 1145–1151, 2015.

[4] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer, “Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: A pilot randomized clinical trial in patients with acquired brain injury,” *J. Neuroeng. Rehabil.*, vol. 8, no. 1, 2011.

[5] B. S. Rajaratnam et al., “Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?,” *Rehabil. Res. Pract.*, vol. 2013, pp. 1–6, 2013.

[6] K. H. Cho, K. J. Lee, and C. H. Song, “Virtual-Reality Balance Training with a Video-Game System Improves Dynamic Balance in Chronic Stroke Patients,” *Tohoku J. Exp. Med.*, vol. 228, no. 1, pp. 69–74, 2012.

[7] R. Barclay-Goddard, T. Stevenson, W. Poluha, M. E. K. Moffatt, and S. P. Taback, “Force platform feedback for standing balance training after stroke,” *Cochrane database Syst. Rev.*, no. 4, p. CD004129, 2004.

[8] D. G. Sayenko et al., “Positive effect of balance training with visual feedback on standing balance abilities in people with incomplete spinal cord injury,” *Spinal Cord*, vol. 48, no. 12, pp. 886–893, 2010.

[9] J. H. Pasma, D. Engelhart, A. C. Schouten, H. van der Kooij, A. B. Maier, and C. G. M. Meskers, “Impaired standing balance: The clinical need for closing the loop,”

Neuroscience, vol. 267. pp. 157–165, 2014.

[10] L. E. Cofré Lizama, M. Pijnappels, G. H. Faber, P. N. Reeves, S. M. Verschueren, and J. H. van Dieën, "Age effects on mediolateral balance control," *PLoS One*, vol. 9, no. 10, p. e110757, 2014.

[11] A. González, M. Hayashibe, V. Bonnet, and P. Fraisse, "Whole body center of mass estimation with portable sensors: Using the statically equivalent serial chain and a kinect," *Sensors (Switzerland)*, vol. 14, no. 9, pp. 16955–16971, 2014.

[12] M. Hayashibe, A. González, and P. Fraisse, "Personalized Modeling for Home-Based Postural Balance Rehabilitation," in *Human Modeling for Bio-Inspired Robotics: Mechanical Engineering in Assistive Technologies*, 2016, pp. 111–137.

[13] Kinect for Windows Sensor Components and Specifications [Available online: https://msdn.microsoft.com/en-us/library/jj131033.aspx]

[14] H. Y. Lee, Y. L. Kim, and S. M. Lee, "Effects of virtual reality-based training and task-oriented training on balance performance in stroke patients.," *J. Phys. Ther. Sci.*, vol. 27, no. 6, pp. 1883–8, 2015.

[15] Y. Lee, J. G. Her, Y. Choi, and H. Kim, "Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors' Names," *J. Phys. Ther. Sci.*, vol. 26, no. 2, pp. 179–182, 2014.

[16] E. Bernard and R. Boulic, "On the Computation and control of the mass center of articulated chains," INRIA, 1998.

[17] S. Cotton, A. Murray, and P. Fraisse, "Statically equivalent serial chains for modeling the center of mass of humanoid robots," in *2008 8th IEEE-RAS International Conference on Humanoid Robots, Humanoids 2008*, 2008, pp. 138–144.

[18] A. González, P. Fraisse, and M. Hayashibe, "Adaptive Interface for Personalized Center of Mass Self-Identification in Home Rehabilitation," *IEEE Sens. J.*, vol. 15, no. 5, pp. 2814–2823, 2015.

[19] Cotton et al., "Estimation of the centre of mass from motion capture and force plate recordings: A study on the elderly," *Appl. Bionics Biomech.*, vol. 8, no. 1, pp. 67–84, 2011.

[20] Ultrasonic Ranging Module HC - SR04 [Available online: https://www.electroschematics.com/wp-content/uploads/2013/07/HCSR04-datasheet-version-1.pdf]

[21] Metzger et al., "Assessment-driven selection and adaptation of exercise difficulty in robot-assisted therapy: A pilot study with a hand rehabilitation robot," *J. Neuroeng. Rehabil.*, vol. 11, no. 1, 2014.

[22] K. Jack, S. M. McLean, J. K. Moffett, and E. Gardiner, "Barriers to treatment adherence in physiotherapy outpatient clinics: A systematic review," *Manual Therapy*, vol. 15, no. 3. pp. 220–228, 2010.

[23] B. M. Young et al., "Changes in functional connectivity correlate with behavioral gains in stroke patients after therapy using a brain-computer interface device," *Front. Neuroeng.*, vol. 7, 2014.

[24] J. Nielsen, *Usability Engineering*, vol. 44, no. 3. 1993.

[25] N. BEVAN, "International standards for HCI and usability," *Int. J. Hum. Comput. Stud.*, vol. 55, no. 4, pp. 533–552, 2001.

[26] J. Brooke, "System Usability Scale (SUS): A Quick-and-Dirty Method of System Evaluation User Information.," in *Usability Evaluation In Industry*, vol. 189, no. 194,

1996, pp. 4–7.

[27] J. Brooke, “SUS : A Retrospective,” *J. Usability Stud.*, vol. 8, no. 2, pp. 29–40, 2013.

[28] R. S. Kalawsky, “VRUSE - A computerised diagnostic tool: For usability evaluation of virtual/synthetic environment systems,” *Appl. Ergon.*, vol. 30, no. 1, pp. 11–25, 1999.

[29] A. J. DeWitt and J. Kuljis, “Aligning usability and security: a usability study of Polaris,” *SOUPS ’06 Proc. Second Symp. Usable Priv. Secur.*, pp. 1–7, 2006.

[30] T. Jokela, J. Koivumaa, J. Pirkola, P. Salminen, and N. Kantola, “Methods for quantitative usability requirements: A case study on the development of the user interface of a mobile phone,” *Pers. Ubiquitous Comput.*, vol. 10, no. 6, pp. 345–355, 2006.

[31] J. Vermeulen et al., “User-centered development and testing of a monitoring system that provides feedback regarding physical functioning to elderly people,” *Patient Prefer. Adherence*, vol. 7, pp. 843–854, 2013.

[32] T. H. Tsai, F. G. Wu, and Y. H. Hung, “The Survey of Usability Evaluation in Social Network Sites’ Reply Mechanism.,” in *International Conference on Universal Access in Human-Computer Interaction*, 2013, pp. 718–725.

[33] D. Mathew, K. A. McKibbon, C. Lokker, and H. Colquhoun, “Engaging with a wiki related to knowledge translation: A survey of whatiskt wiki users,” *J. Med. Internet Res.*, vol. 16, no. 1, 2014.

[34] D. Fitzgerald, D. Kelly, T. Ward, C. Markham, and B. Caulfield, “Usability evaluation of e-motion: A virtual rehabilitation system designed to demonstrate, instruct and monitor a therapeutic exercise programme,” in *2008 Virtual Rehabilitation, IWVR*, 2008, pp. 144–149.

[35] M. K. Holden, “Virtual environments for motor rehabilitation: review.,” *Cyberpsychol.*

Behav., vol. 8, no. 3, pp. 187-211-219, 2005.

[36] H. Sveistrup, “Motor rehabilitation using virtual reality,” *Journal of NeuroEngineering and Rehabilitation*, vol. 1. 2004.

[37] B. Lange et al., “Designing informed game-based rehabilitation tasks leveraging advances in virtual reality,” *Disabil. Rehabil.*, vol. 34, no. 22, pp. 1863–1870, 2012.

[38] G. N. Lewis and J. A. Rosie, “Virtual reality games for movement rehabilitation in neurological conditions: How do I meet the needs and expectations of the users,” *Disabil. Rehabil.*, vol. 34, no. 22, pp. 1880–1886, 2012.

[39] D. Green and P. H. Wilson, “Use of virtual reality in rehabilitation of movement in children with hemiplegia - A multiple case study evaluation,” *Disabil. Rehabil.*, vol. 34, no. 7, pp. 593–604, 2012.

[40] M. Dara et al., “Effectiveness of conventional versus virtual reality based vestibular rehabilitation in the treatment of dizziness, gait and balance impairment in adults with unilateral peripheral vestibular loss: A randomised controlled trial,” *BMC Ear, Nose Throat Disord.*, vol. 12, no. 1, 2012.

[41] S. Albiol-Pérez, J. a Gil-Gomez, R. Llorens, M. Alcañiz, C. C. Font, and J.-A. Gil-Gómez, “The Role of Virtual Motor Rehabilitation: A Quantitative Analysis Between Acute and Chronic Patients With Acquired Brain Injury,” *Ieee J. Biomed. Heal. Informatics*, vol. 18, no. 1, pp. 391–398, 2014.

[42] D. K. A. Singh, N. A. Mohd Nordin, N. A. A. Aziz, B. K. Lim, and L. C. Soh, “Effects of substituting a portion of standard physiotherapy time with virtual reality games among community-dwelling stroke survivors,” *BMC Neurol.*, vol. 13, 2013.

[43] Gil-Gómez et al., “USEQ: A short questionnaire for satisfaction evaluation of virtual

rehabilitation systems," *Sensors (Switzerland)*, vol. 17, no. 7, 2017.

[44] R. Likert, "A technique for the measurement of attitudes," *Archives of Psychology*, vol. 22 140. p. 55, 1932.

[45] José-Antonio et al., "SEQ: Suitability Evaluation Questionnaire for Virtual Rehabilitation Systems. Application in a Virtual Rehabilitation System for Balance Rehabilitation.," in *Proceedings of the ICTs for improving Patients Rehabilitation Research Techniques*, 2013.

[46] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[47] S. S. Shapiro and M. B. Wilk, "An Analysis of Variance Test for Normality (Complete Samples)," *Biometrika*, vol. 52, no. 3/4, p. 591, 1965.

[48] J. D. Gibbons and S. Chakraborti, *Nonparametric Statistical Inference*, vol. 15, no. 2. 2003.

[49] M. Aickin and H. Gensler, "Adjusting for multiple testing when reporting research results: The Bonferroni vs Holm methods," *Am. J. Public Health*, vol. 86, no. 5, pp. 726–728, 1996.

[50] D. Kumar, A. Gonzalez, A. Das, A. Dutta, P. Fraisse, M. Hayashibe, and U. Lahiri, "Virtual Reality based Center of Mass Assisted Personalized Balance Training System," *Front. Bioeng. Biotechnol.*, vol. 5, p. 85, 2017.

CHAPTER 6

VR-BASED BALANCE TRAINING PLATFORM AUGMENTED WITH OPERANT CONDITIONING PARADIGM

6.1 Introduction

In the previous chapters, I have studied the implications of the VR-based balance training systems that use participant's Center of Pressure (CoP) (as measured by a Balance Board (Chapter 4)) and Center of Mass (CoM) (that used a Balance Board and Kinect sensor (Chapter 5)) on the balance of hemiplegic post-stroke participants. However, both the systems used the balance-related information derived from the cumulative effect of both the lower limbs while standing on the Balance Board and / or ground surface. None of these systems could identify the contribution of each leg to the overall balance while maneuvering VR-based objects integrated with the overall CoP or CoM. Literature indicates that hemiplegic post-stroke patients suffer from unequal weight distribution on both sides of the body during standing balance task [1], [2]. So, getting information on the contribution of each leg to an individual's overall balance is critical. Additionally, it would be interesting to understand the implication of varying weightage allocated to the CoP contribution of each leg of a hemiplegic post-stroke patient performing a balance task. In this chapter, I wanted to address two research questions, namely, (i) whether it is feasible to quantify the contribution of each leg during an individual's weight-shifting task? and (ii) what is the role of operant conditioning through variation in weight distribution among both of the lower limbs during a balance task? While addressing these research questions, I exposed the post-stroke participants to VR-based balance tasks.

Studies report that asymmetric body weight distribution is often modified by neurological disorders that cause hemiparesis [3] and has been associated with postural instability causing fall in people with balance disorders [4], [5], [6]. In such cases, one of the lower limbs is overloaded for postural adjustments [3], [7], [8]. Studies with hemiplegic participants showed that the paretic (Affected *henceforth*) lower limb carries lower percentage of the body weight compared to the comparatively healthier (Unaffected *henceforth*) side [9], [10] even for those who are ambulatory [11], [12]. Consequently, to maintain stability during walking, post-stroke patients often show increased postural sway [13], [14] that can be quantified in terms of excursions of the center of pressure (CoP) when moving their weight around the base of support, especially in the direction of the weaker leg [11], [12]. Again, during standing balance task, synchronization of CoP trajectories between both the Affected and Unaffected legs of post-stroke patients is often less than their healthy counterparts [15]. This implies that both the legs of hemiplegic post-stroke survivors are not effectively utilized to the same extent during weight-shifting task.

Thus, balance rehabilitation coupled with weight-shifting efforts have been directed towards regaining the symmetry in body weight distribution lost due to balance impairment with an aim to improve the weight bearing on the Affected leg. Various rehabilitation techniques based on targeted weight-shifting training while using force platform has been used to reduce the asymmetry in body weight distribution [16], [17], [18], [19]. For example, Tripoli et al. [20] used two force platforms and directed the hemiplegic participants (who stood with one feet on each platform) to shift weight while they stepped forward and backward with their Affected lower limb being taken forward and backward, respectively. The participants were asked to do this while the experimenters provided verbal commands and also in presence of computer-based visual and auditory feedback. Also, other researchers have used two force platforms coupled with computer-based visual feedback based on the weight borne by each leg while doing weight-shifting tasks [16], [17], [18], [19]. These studies, though beneficial, suffer from limitations such as lack of motivating practice environment, individualization, and other factors. Specifically, the repetitive practice with such systems in the absence of variations in task environment and challenge levels might turn out to be monotonous for the participants. At the same time, quantification of contribution of each leg to an individual's overall weight-shifting followed by individualization through variation in weight distribution between the two legs was not pursued in these settings. But, for achieving symmetric weight distribution on both the limbs as far as standing balance is concerned, there must be a mechanism to (i) challenge participants through allocating increased weightage to Affected leg while reducing the weightage to the Unaffected leg in a weight-shifting task and (ii) adapt the challenge level based on user's individualized weight-shifting ability.

Such a system can deliver rehabilitation while using constraint therapy. Literature review indicates rehabilitation approach namely, Constraint Induced Movement Therapy (CIMT) [21] being used for upper limb rehabilitation. This therapy involves constraining movements of the Unaffected limb, usually for 90% of waking hours, while encouraging intensive usage of the Affected limb. In this therapy, concentrated, repetitive training of the Affected limb is usually performed for six hours a day over two to three weeks period [22] . Again, researchers such as Edward et al. [21] used conditioning of the Affected upper limb of post-stroke hemiplegic patients while constraining the Unaffected side. The idea was to overcome learned nonuse of the Affected limb so that the patient starts to use the Affected limb for performing tasks. However, these studies also lay emphasis on the use of such constraints over long periods of time that might be limiting its applicability in clinical practice [23] along with infeasibility of such an approach for lower limb rehabilitation that can adversely affect an individual's mobility. According to Page et al. [23], CIMT is not widely viewed as a useful therapeutic intervention by clinicians and is often considered unfeasible due to patients' concerns about the intensive schedule of treatment. In addition, therapists are concerned about patients' compliance, safety issues and clinical resources. Few studies have reported adverse effects associated with CIMT, such as minor skin lesions and muscle soreness (stiffness and discomfort) in the Affected upper extremity [24], [25], [26]. Given these adverse effects, investigators have been exploring alternate rehabilitation approach such as operant conditioning [27]. Operant conditioning approach relies on observing and modifying explicit behaviors using the antecedents (the surrounding environmental factors leading to the behavior) and consequences (the outcome of the behavior in terms of effects on that environment) [28]. The principle of operant conditioning has been used by different researchers in the field of rehabilitation medicine [29], [30]. In the area of rehabilitation, this approach has also shown promise. Specifically, Thompson et. al. [31] have shown that operant conditioning of spinal reflex can improve locomotion after spinal cord injury that can have clinical applicability. Recently, Kazuhiro et al. [32] showed the feasibility of using an implicit guidance method for conditioning the weight-bearing ability towards the Affected side of hemiplegic post-stroke patients while using explicit cues such as tactile feedback to direct the weight-shifting towards the Affected side.

Given the limitations of the constraint therapy and the promise with clinical applicability of operant conditioning approach, I implemented the operant conditioning approach for lower limb standing balance task in the current research. In contrast to the currently existing approach that aims to condition an individual's standing balance capability by using explicit cues to prompt the user to direct the weight-shifting towards the Affected side [32], I used an implicit and subtle approach in my research. Specifically, I used Individualized Threshold and Weight Distribution Estimator (Please see Section 6.2.3 below) to condition the usage of the Affected limb in a weight-shifting balance task. Based on individualized balance capability, I varied the weight distribution between the Affected and Unaffected legs while offering tasks of varying challenges in a VR-based environment. The aim was to motivate the participants to increase the usage of their Affected leg during weight-shifting task without explicitly directing them to do so.

To achieve this, I have developed a VR-based balance training platform (V2BaT) interfaced with two Wii Balance Board (WiiBB) to implement an operant conditioning paradigm for balance rehabilitation. The VR-based task required a participant to maneuver a VR object (VR_{Obj}) in the VR environment by shifting his / her weight in a specific direction while standing on the Balance Boards. Keeping in mind that an individual's day-to-day activity often needs one to shift weight in different directions to perform reaching tasks, I chose anterior direction for the present study for performing static balance bipedal weight-shift exercise. The V2BaT system needed that a participant should effectively use both the legs to complete a task in which each participant was asked to stand on two WiiBBs (two legs on the balance board, one on each WiiBB). I used two WiiBBs, since I wanted to measure the contribution of each leg towards the overall weight-shifting capability of hemiplegic post-stroke participants. In this study, I (i) developed a VR-based balance training system augmented with operant conditioning paradigm (V2BaT system *henceforth*) and (ii) conducted a Usability study with the V2BaT system, so as to understand the implications of operant conditioning paradigm on the task performance of individuals having balance disorders. The V2BaT system also featured an automated Heel Lift Detection unit that detected an individual's incorrect posture (that is lifting of the heel from the WiiBB surface) during balance training while using Ankle strategy. The V2BaT system offered a variety of VR-based balance tasks of different challenge levels in a controlled and systematic manner to the participants.

The aims of my present study were two-fold, namely, to understand (i) whether it is feasible to quantify the contribution of each leg during an individual's weight-shifting task and (ii) the role of operant conditioning through variation in weight distribution among both of the lower limbs during a balance task.

6.2 System Design

The V2BaT system consisted of six units, namely, (a) VR-based Task (b) WiiBB-VR Handshake (c) Individualized Threshold and Weight Distribution Estimator (d) Heel Lift Detection (e) Performance Evaluation and (f) Task Switching units.

6.2.1 VR-based Task Unit

There is evidence in literature that playing games can have many positive behavioral and physiological effects leading to meaningful improvements in an individual's cognitive, motor and affective measures [33]. In this study, I designed various games for weight-shifting tasks. Though exercises set in conventional settings for post-stroke patients have shown promising outcomes, literature indicates that the VR-based rendering of exercise environment can be potent to yield functional outcomes that can sometimes surpass the contribution of conventional therapy for stroke rehabilitation [34]. One of the reasons behind the success of VR in stroke rehabilitation is the element of motivation [35] that it can bring in the participants when they interact with the realistic situations presented through imagery and sound. Literature indicates that motivation is an important factor in rehabilitation and is often linked with improved therapeutic outcomes [36]. Here, I have used VR to offer various weight-shifting tasks to post-stroke participants while allowing them to interact with game environments. Keeping in mind that an individual's day-to-day activity often needs one to shift weight in different directions to perform reaching tasks, the VR-based tasks were designed to leverage the participants' directional weight-shifting capability while trying to bring in balanced contribution from both the lower limbs during the tasks. The tasks required the participants to shift their weight in anterior direction while they were asked to perform static balance bipedal weight-shifting exercise. Specifically, the system needed participant's effective use of both the limbs to complete a task. Exercises that emphasize participant’s effective use of both the limbs is particularly critical for hemiplegic patients who often show asymmetric weight distribution on both legs, with the Unaffected leg bearing most of the body weight during shifting of weight [37].

In the present study, the participants interacted with VR-based tasks that were categorized as (i) Stage 1 and (ii) Stage 2 tasks. The Stage 1 was the pre-task calibration stage. In this, I computed the individualized threshold (Section 6.2.3 below) and initial weight distribution as far as the contribution of both the legs towards task completion in Stage 2 was concerned. The Stage 2 was the VR-based task execution stage. In this, the participants started with the initial weight distribution (as obtained from Stage 1) and slowly progressed to task trials with increasing challenge (Section 6.2.6 below). The tasks required the participants to interact with the VR-based environment for about 20 minutes. This duration is similar to that used in conventional settings where the physiotherapists often recommend about 20 minute exercises for lower limb [38] consisting of static and dynamic balance, passive and active range of motion, stretching, gait training, muscle strengthening and activities of daily living exercises. I used Vizard software toolkit (from Worldviz Llc.) to design realistic VR environments with variations so as to make the weight-shifting exercise interesting.

6.2.1.1 Design of VR-based Tasks for Stage 1

In the Pre-task calibration stage (Stage 1), I wanted to estimate the participant's residual ability to shift his weight in the anterior direction in terms of the amount of load that can be borne by the legs (left and right legs separately) during the weight-shifting task. The aim was to estimate (i) the individualized threshold for VR-based task and (ii) participant's initial weight distribution as far as both the legs were concerned. For this, I designed a VR-based task environment such as forest environment (Fig. 6.1) comprising of a pair of boots (VR_{Obj}) placed in the forest. The position of each boot $((VR_{Obj})_L$ and $(VR_{Obj})_R)$ corresponding to the CoP position

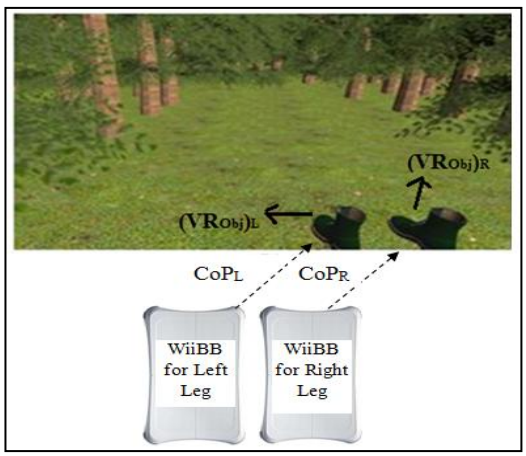


Figure 6.1. VR-based task for Stage 1 of V2BaT system

(CoP_L and CoP_R) measured by two WiiBBs placed under each of left (L) and right (R) legs was acquired by V2BaT in real-time. The participants were asked to shift weight in anterior direction to their maximum possible ability, without lifting heel.

6.2.1.2 Design of VR-based Tasks for Stage 2

For the VR-based task execution stage (Stage 2), I designed a database of 7 tasks. Fig. 6.2 shows some of the VR-based tasks designed using Google Sketch-up and Vizard softwares for Stage 2. The VR environments presented gaming tasks on land terrain (such as skating on road, playing on ground), water (such as swimming under water) and sky (such as flying helicopters) along with variations. The idea of offering variations within the tasks was to make the VR-based tasks interesting and not monotonous. The VR-based tasks required the participants to use both the legs to perform weight-shifting with their weighted CoP (discussed in Section 6.2.3) integrated to the VR_{Obj}. Examples of VR_{Obj} were avatars wearing skates on both legs for moving on a road, flying helicopters, skiing boards and others. Since the aim was to encourage the

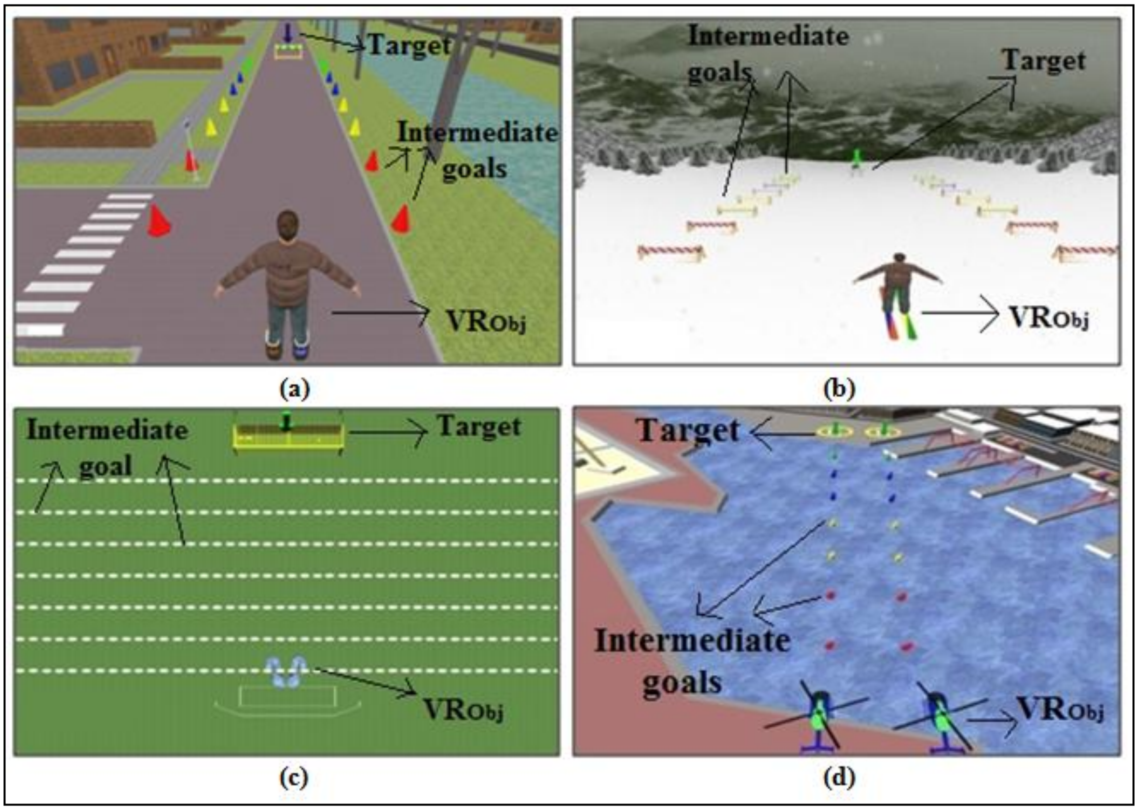


Figure 6.2. Various templates of VR-based balance rehabilitation tasks of V2BaT system

participants to use both of their legs as equally as possible (which is unusual for hemiplegic patients), the V2BaT featured VR_{Obj} presented in pairs (such as pair of helicopters, roller skates, skiing boards (one for each leg)) that might help the participants to visualize the usage of both their legs to complete a task. However, unlike Stage 1 (in which $(VR_{Obj})_L$ and $(VR_{Obj})_R$ were controlled by CoP due to each of the left and right legs separately), in Stage2, the VR_{Obj} (pair of objects, such as pair of helicopters, roller skates) was controlled by the resultant weighted contribution of the CoP due to each leg.

While designing the VR-based tasks, care was taken so that any action taken by the user should reflect meaningful change in the overall game environment, critical for a game to be meaningful [39]. First the participants were asked to stand straight on the WiiBBs as upright as possible without shifting weight ($Central_{Hold}$ position). Then for task execution, the participants were asked to shift their weight in forward (anterior) direction that in turn shifted their CoP due to each leg away from their $Central_{Hold}$ position, thereby causing the VR_{Obj} to move forward towards the Goal (*Target)* in the VR environment. Literature review shows that administering goal-directed movements are critical for stroke rehabilitation [40], [41], [42]. Thus, each VR-based task presented a *Target* and the participants were asked to take the VR_{Obj} from $Central_{Hold}$ position to *Target* by maneuvering the VR_{Obj} in the VR environment.

To bring in variety, the choice of *Target* differed for various VR environments. For example, for an Avatar skating on a road with a pair of skating shoes (VR_{Obj}), a milestone at the end of the road was used as the *Target* (Fig. 6.2 (a)). Again, for an Avatar skiing on the snowy track with a pair of skis (VR_{Obj}), a snowy mountain peak was chosen as the *Target* (Fig. 6.2(b)). Though the choice of *Target* differed based on the type of VR environment, in each case an arrow was shown as pointing towards the *Target* and it remained visible till the VR_{Obj} reached the *Target*. Apart from the *Target*, the VR-based task environment also presented relevant Intermediate Goals. For example, for an Avatar skating on a road, road dividers located at almost regular intervals were chosen as the Intermediate Goals (Fig 6.2(a)). Similarly, for the avatar skiing on the snowy track, multi-colored intermediate milestones were used as Intermediate Goals (Fig. 6.2(b)). These intermediate goals were kept so that the participants can gauge their improvement in performance that might serve as an incentive for the participant to keep playing even if he was not able to reach the *Target* in a single task trial.

Once a participant completed a task trial, the system provided feedback based on his performance. Care was taken to design the feedback so that the rehabilitation task can be meaningful [43] and engaging [44] to the participants. The feedback was both intrinsic and extrinsic in nature. Literature review indicates that intrinsic feedback can be mediated through vision (text, icons, scores), whereas, extrinsic feedback includes verbal encouragement [45], [46]. The intrinsic feedback was given using audio-visual feedback in which I showed the participants the image of single star (*) to five stars (*****) along with a coin-like sound with every star (*) based on their performance. Again, the extrinsic feedback was given to them using verbal encouragement as pre-recorded audio files that told either 'Well done, You are doing great' (in case they scored ‘Adequately’ (described in Section 6.2.5)) or 'Keep trying, you can do better' (in case they scored ‘Inadequately’).

6.2.2 Individualized Threshold and Weight Distribution Estimator Unit

There is evidence from literature that an optimal level of challenge in tasks is important as far as motor learning is concerned [47], [48]. In the study, the VR-based tasks were of varying difficulty based on the individualized weight distribution as far as both the legs of each participant were concerned. The stroke participants being hemiplegic with varying residual balance abilities (evident from their BBS scores (Table 6.1)), individualization in weight distribution was critical. Also, I wanted to make the game difficulty level (or challenge level) adaptive to individualized residual balance. Thus, for each participant, V2BaT started by offering task with comparatively lower challenge level that matched with the individual’s residual balance ability as indicated in other studies [33]. Subsequently, the V2BaT offered tasks with increasing challenge with an aim to improve the individual's balance skill in line with other studies [49]. In order to estimate participant's individualized residual balance ability when he/she came in to participate in the study, I designed Stage 1 task that assessed the participant's range of movement by asking him/her to shift weight in the anterior direction to his/her maximum ability (following Ankle strategy (Section 6.2.4)) while standing on the WiiBBs. This Stage 1 was used to decide the initial weight distribution as far as both the legs were concerned. In Fig. 6.1, the positions of left and right boots were controlled by the CoP due to participant's individual legs (measured by two WiiBBs, one each for each leg). The participant was asked to take part in three trials of Stage 1. Each trial of the task required the participant to move the boots ($(VR_{Obj})_L$ and $(VR_{Obj})_R$) as far as possible inside the forest by shifting weight. The range of movement of $(VR_{Obj})_L$ and $(VR_{Obj})_R$ was decided from the CoP displacement (ΔCoP) due to left (ΔCoP_L) and right (ΔCoP_R) boot from that in the Baseline condition (that is while standing upright without weight-shifting; $Central_{Hold}$). Then, the maximum of the three trials was computed to derive ΔCoP_{max_L} for left leg and ΔCoP_{max_R} for right leg. Using this data, I computed (i) individualized threshold (ΔCoP_{THRESH}) and (ii) initial weight distribution, namely, w_{L_ini} for left leg and w_{R_ini} for right leg.

The individualized threshold (ΔCoP_{THRESH}) was estimated by using the maximum (ΔCoP_{max}) of ΔCoP_{max_L} and ΔCoP_{max_R}. The idea behind extracting the maximum of the ΔCoP_{max_L} and ΔCoP_{max_R} was to find out the maximum contribution from the comparatively healthier leg of the hemiplegic participants in an individualized manner. Based on a preliminary pilot trial with 4 post-stroke hemiplegic participants (mean (SD) = 50.25 (7.84) years), I observed that mostly the participants tend to be conservative in shifting their weight at first, even after they are asked to maneuver the VR_{Obj} (such as boots as in Fig. 6.1) as far as possible in the anterior direction.

Again, for $\Delta CoP_{THRESH} = \Delta CoP_{max}$, most of the participants were able to achieve ~100% performance within fewer number of trials of the VR-based task execution stage (Stage 2). However, I wanted the VR-based tasks to be motivating with sufficient challenge while being adaptive to individualized weight-shifting capabilities. This is because, taking inputs from Flow Theory [50], a task should not be too easy to bring in a feeling of boredom in the participant. Again, the task should also not be too hard for the participant to be frustrated. Thus, I wanted to choose $\Delta CoP_{THRESH} > \Delta CoP_{max}$ by approximately 20% of ΔCoP_{max}. In this study, I chose,

$$\Delta CoP_{THRESH} = 1.2 * \Delta CoP_{max} \tag{6.1}$$

Please note that this factor of 1.2 was chosen as an initial approximation. One can easily modify this factor based on the study design.

As regards the estimation of initial weight distribution (w_{L_ini} and w_{R_ini} for left and right legs, respectively) was concerned, I used the values of ΔCoP_{max_L} and ΔCoP_{max_R} as follows:

$$w_{L_{ini}}(\%) = \frac{\Delta CoP_{\max\_L}}{\Delta CoP_{\max\_L} + \Delta CoP_{\max\_R}} \times 100 \;\; and \;\; w_{R_{ini}}(\%) = \frac{\Delta CoP_{\max\_R}}{\Delta CoP_{\max\_L} + \Delta CoP_{\max\_R}} \times 100 \tag{6.2}$$

6.2.3 WiiBB-VR Handshake Unit

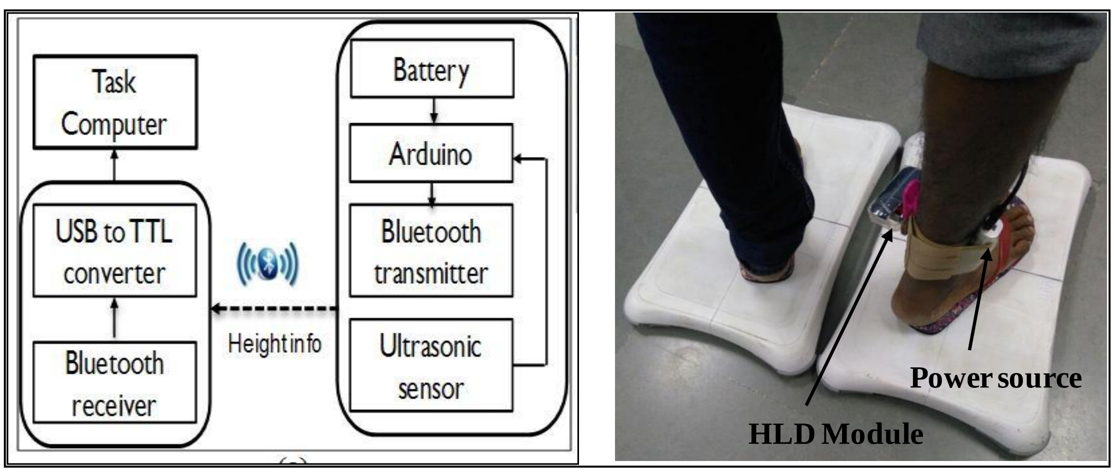
The VR-based tasks required the participant to maneuver virtual objects (VR_{Obj}) in the VR environment (shown on the Task Computer monitor) using their CoP while standing on the balance boards (WiiBBs). Since I wanted to know the contribution of each leg towards maneuvering of the VR_{Obj} in the VR environment, I used two WiiBBs, one for each leg. In the VR-based task excution stage (Stage 2), position of VR_{Obj} was controlled by the weighted sum of the CoPs obtained from two WiiBB (discussed in Section 6.2.1.2). Since the task was to shift weight in anterior direction, I used only y component of the CoP (that is, change in CoP along anterior direction) for navigating VR_{Obj} while I stored both the x and y coordinates of CoP in the backend for subsequent offline analysis. The raw CoP values acquired at 30 Hz were processed by a 5-point moving average filter. The position of VR_{Obj} was determined from the filtered CoP data by using equation (6.3).

$$[y]_{VR_{OBJ}} = w_L[y]_{CoP_L} + w_R[y]_{CoP_R} \tag{6.3}$$

Where, w_L and w_R are the weight factors for the left and right legs, respectively. $[y]_{CoP_L}$ and $[y]_{CoP_R}$ indicate the y coordinate of the CoP as measured by the two WiiBBs corresponding to the left and right legs, respectively.

6.2.4 Heel Lift Detection Unit

In the present study, I wanted to ensure that the participants followed Ankle strategy which is considered important during standing balance task [51]. In order to ensure that the Ankle strategy was followed, the participants were asked not to lift their heel from the ground while shifting their weight. Thus, to identify whether the Ankle strategy was 'Followed' or 'Not Followed' (Eq.



(a) (b)

Figure 6.3. (a) Block diagram and (b) Placement of HLD unit on the Affected leg of participant

(6.4)), I used a Heel Lift Detection (HLD) unit (Fig. 6.3) (similar to that in Chapter 5) that communicated the height of the heel above the base of support (WiiBB) wirelessly at a rate of 60 samples / sec to the VR-based system. In this study, the HLD unit was attached above the lateral malleolus of the Affected leg with the ultrasonic sensor facing downwards towards the surface of

WiiBB. If the Ankle strategy was *'Not Followed'*, then a penalty factor was added to the performance score (described below). Otherwise, no penalty factor was considered in the evaluation of the performance score.

6.2.5 Performance Score Evaluation Unit

Researchers have identified several CoP-based metrics, such as CoP displacement, Root Mean Square (RMS), and others to quantify an individual's postural movement [52], [53], [54]. While the participants performed VR-based tasks, the V2BaT system computed their performance scores using the information on the CoP displacement. Since execution of the task required one to shift weight in anterior direction only, I have used only *y*-coordinate (while masking the *x*-coordinate) of CoP (CoP_Y) for visual presentation of the trajectory of the VR_{Obj} from $Central_{Hold}$ position to *Target* position. Also, I computed the amount of the CoP displacement $(\Delta CoP)_Y$ as one of the performance measures (P_{s1}) (Eq. (6.6)).

$$P_{S1} = 100 - \left(\frac{T_L - T_D}{T_L}\right) \times 100 \tag{6.6}$$

Here, T_L is the length of straight line path between the $Central_{Hold}$ and *Target* position in the anterior direction; T_D is the amount (length) of participant's CoP displacement ($(\Delta CoP)_Y$) in the VR environment during a weight-shifting task.

Additionally, for rehabilitation related to weight-shifting to be effective, one needs to be (i) able to shift weight in the required direction and also (ii) adhering to Ankle strategy. To incorporate this factor that took the Ankle strategy into account, I quantified participants' ability to shift weight without lifting heel by using the second performance measure (P_{s2}). The P_{S2} (Eq. (6.7)) was used to penalize the participant for lifting heel during weight-shifting task. The penalty due to participant's lifting of the heel from the surface of WiiBB during weight-shifting task was designed to be proportional to the amount of time the participant lifted his heel (T_{Lift})

out of the total time a participant took to complete a task (T_{CT}) (similar to that in the previous study; please see Section 4.2.4.3 of Chapter 4).

$$P_{s2} = \frac{T_{Lift}}{T_{CT}} \times 100 \tag{6.7}$$

The final % performance score (P_s) for a task was calculated as

$$P_s = P_{s1} - P_{s2} \tag{6.8}$$

The V2BaT system was designed to be adaptive to the participant's performance score in a task. A participant's performance score was considered as either 'Adequate' or 'Inadequate' based on his/her percent performance score. For example, if a participant's score was ≥ 70%, then it was considered as 'Adequate', else 'Inadequate.' Please note that the threshold of 70% for the performance score was taken as an initial approximation since, literature indicates that a performance score of 70% can be considered as a satisfactory performance so far as the initial exercise performance for robot-assisted rehabilitation tasks [55], for outpatient clinics [56], technology-assisted skill learning [57], and others are concerned. This threshold can be easily adjusted based on the study requirement.

6.2.6 Task Switching Unit

I designed a Usability study with the V2BaT system. In this, the participants were exposed to Stages 1 and 2 of the tasks. In Stage 1, a participant's (i) individualized threshold (ΔCoP_{THRESH}) and (ii) initial weight distribution, namely, w_{L_ini} and w_{R_ini} for left and right legs, respectively (Section 6.2.2) were decided. The tasks in Stage 2 were of two types, namely (i) Catch Trial (*CT*) and (ii) Normal Trial (*NT*). Similar to that in other studies [58], the *CT* was one in which equal weight distribution was allocated to each of the Affected and Unaffected legs. Again, by *NT*, I refer to the trials in which the weight distribution allocated to each of the Affected and Unaffected legs were not equal. In the present study, I offered Normal Trials (*NTs*) of increasing challenge along with intermediate Catch Trials (*CTs*). The idea was (i) to help the participants learn to increasingly use their Affected leg during *NTs* along with (ii) facilitating the transfer of some of the residual effects from *NTs* to *CTs* (similar to that in other studies [58]).

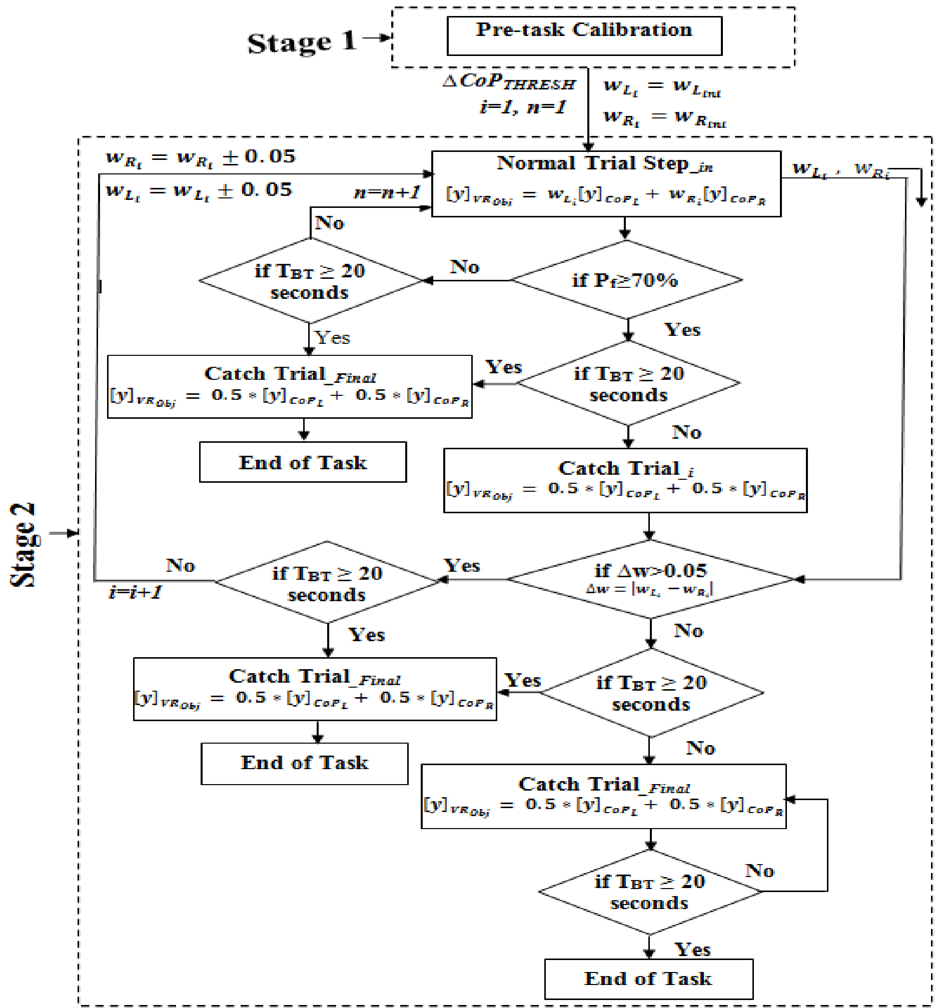


Figure 6.4. Flow diagram of task switching unit of V2BaT system

The overall aim was to understand the implication of operant conditioning on participant's ability to distribute weight as equally as possible on both sides of the body (measured by WiiBBs and quantified in terms of CoP displacement due to each of the left and right legs) during a weight-shifting task. To achieve this, I designed the VR-based tasks in which a VR_{Obj} can be maneuvered in the VR environment by a weighted contribution of the CoP displacement due to each leg (Section 6.2.3) by offering the tasks as *CT* or *NT*. In *CT* task, the weight factors w_L and w_R (Eq. (6.3)) corresponding to left and right legs, respectively were equal (that is, $w_L = w_R$). This implies that equal weightage was provided to the contribution of both the legs (Affected and Unaffected legs of post-stroke hemiplegic participants) in maneuvering the VR_{Obj}. In contrast, in the *NT* task, the weight factors w_L and w_R were manipulated keeping operant conditioning in mind. The values of w_L and w_R started with w_{L_ini} and w_{R_ini} (Section 6.2.2) and these were manipulated to different weights while offering *NT* tasks of varying task challenge levels. The V2BaT system offered *NT* tasks to expose the participants to an operant conditioning regime with *CT* tasks in between the *NT* tasks following a Task Switching Rationale.

In Stage 2, VR-based tasks were switched based on two conditions, namely, Condition_1 and Condition_2.

$$\left.\begin{array}{l}\text{Condition}_1\text{: \%Pf\_Score} \ \geq \ 70\% \text{ (‘Adequate’)} \\ \text{Condition}_2\text{: } \Delta \text{w } (= |\text{wL} - \text{wR}|) \ > \ 5\%\end{array}\right\} \tag{6.9}$$

The rationale for the task execution in Stage 2 is shown in Fig. 6.4. The Stage 2 started with NT_{in} tasks, with n =1, 2, …. with $w_L = w_{L1} = w_{L\_ini}$ and $w_R = w_{R1} = w_{R\_ini}$ (from Stage 1). Once, the Condition_1 was satisfied, the V2BaT used the Task Switching Rationale to offer a CT_i task (with $\Delta w = 0$ that is, $w_L = w_R$) with i = 1, 2, 3…Final task.

Before offering the CT_i task, if

Case1: the Condition_2 was also satisfied, then, the CT_i task for i = 1 was considered as First CT (CT_{First}) task else,

(ii) Case2: the CT_i task was considered as CT_{Final} task.

For example, if for a participant, $w_{L1} = w_{L\_ini} = 39\%$ and $w_{R1} = w_{R\_ini} = 61\%$, then $\Delta w > 5\%$ (Case1), the subsequent CT_i task that is CT_1 (with $w_{L1} = w_{R1}$) would be considered as CT_{First} task. On the other hand, if for a participant, $w_{L1} = w_{L\_ini} = 49\%$ and $w_{R1} = w_{R\_ini} = 51\%$, which shows that $\Delta w < 5\%$ (Case2), then the subsequent CT_n task (with $w_{L1} = w_{R1}$) would be considered as CT_{Final} task.

If Case1 was true, the V2BaT system offered NT_{2n} tasks, with $n = 1, 2$, and so on. Each of the NT_{2n} tasks, were designed to be more challenging than the NT_{1n} tasks in terms of altered weight distribution, such as $w_L = w_{L2} = w_{L1} \pm 0.05$ and $w_R = w_{R2} = w_{R1} \pm 0.05$. The increment / decrement (that is + 0.05 / - 0.05 in w_L and w_R) in the weightage was individualized. For example, if for the last task of NT_{1n}, $w_{L1} = 39\%$ and $w_{R1} = 61\%$, then, for the NT_{2n} tasks, $w_L = w_{L2} = 44\%$ and $w_R = w_{R2} = 56\%$ were chosen by the algorithm. The NT_{2n} tasks continued until the Condition$_1$ (Eq. 6.9) was satisfied. Thereafter, V2BaT system offered a CT_2 task (with $\Delta w = 0$). Subsequently, NT_{3n} tasks ($w_L = w_{L3} = 49\%$ and $w_R = w_{R3} = 51\%$) were offered. Once the Condition$_1$ was satisfied while interacting with NT_{2n} tasks, V2BaT offered CT_i tasks with $i = 3$. Again, if Case2 was true, then the V2BaT system offered CT_{Final} task.

For all the cases, the Stage 2 terminated with V2BaT offering a CT_i task with $i = Final$ if the total time of balance training (T_{BT}) exceeded 20 minutes. Thus, if a participant reached an NT step (NT_{in}) where both the Condition$_1$ and Condition$_2$ were satisfied before the end of the training session (that is $T_{BT} < 20$ minutes), then he/she was offered CT_{Final} tasks repetitively till $T_{BT} = 20$ minutes was over followed by a single trial of CT_{Final}. However, if a participant could reach to an NT_{in} where both Condition$_1$ and Condition$_2$ were not satisfied even to the end of $T_{BT} = 20$ minutes, then he / she was offered a single trial of CT_{Final} before exiting from the training session.

6.3 Experiment and System Design

6.3.1 Participants

The study was carried out after informed consent at an Institute of Neuroscience at Kolkata (West Bengal) and at a local civil hospital at Ahmedabad (Gujarat) where the stroke survivors were undergoing treatment. In the present study, based on the availability of the participants,

Table 6.1. Participants' metadata for usability study with V2BaT system

S. No.	Age (years)/Gender	Affected Side	Post-stroke Period (years)	BBS Score
S1	26/ Male	Right	1 years	48
S2	58/ Male	Right	3.5 years	46
S3	45/ Male	Left	1 years	53
S4	43/ Male	Right	0.3 years	52
S5	58/ Male	Right	1 years	46
S6	52/ Male	Left	5 years	48
S7	17/ Male	Right	1 years	54
S8	50/Female	Right	0.16 years	NA
S9	31/ Male	Right	2 years	53
S10	55/ Male	Right	1.5 year	53
S11	54/ Male	Right	0.08 year	43
S12	30/ Male	Right	3.5 years	49
S13	65/ Male	Left	0.08 years	51
S14	36/ Male	Right	1.5 years	45
S15	58/ Male	Right	0.16 years	50
S16	62/ Male	Right	0.33 years	55
S17	60/ Male	Right	1 year	51
S18	53/ Male	Left	0.58 years	54
S19	55/ Female	Right	4 years	46
S20	69/ Male	Left	2 years	45
S21	35/ Male	Right	0.58 years	55
S22	25/ Male	Right	0.25 years	53
S23	63/ Female	Right	2 years	41
S24	38/ Male	Right	2 years	55
S25	55/ Male	Right	0.16 years	52
S26	45/ Male	Right	7 years	54
S27	66/ Male	Right	0.16 years	50
S28	48/ Male	Right	0.16 years	52
S29	58/ Male	Left	0.02 years	53

Note: S= Stroke participant, BBS= Berg Balance Scale Score

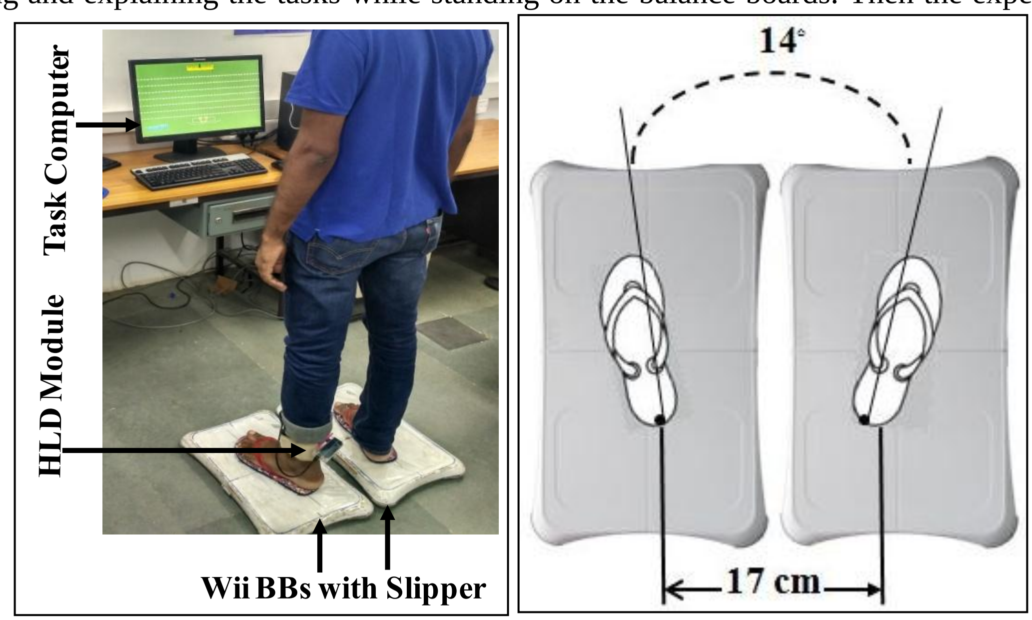
twenty-nine hemiplegic post-stroke survivors (S1-S29) (mean (SD) = 49.55years (13.89)) with varying residual balance and post-stroke periods participated. The Berg Balance Scale (BBS) scores ranged from 41 to 55 which show that the participants possessed a wide spectrum of balance disorder. Literature review indicates that individuals with BBS score < 45 are considered to have higher risk of fall and those with BBS score ≥ 45 have lesser risk of fall [59]. Table 6.1 shows the participants' metadata. As can be seen from this table, out of 29 participants, 2 participants (S11 and S23) had BBS score < 45. The participants did not have any prior exposure to computer-based task. The inclusion criteria were (1) ability to follow the instructions (2) ability to stand and shift weight without orthopedic aids and (3) should not have gone through any surgery in recent past that may interfere with their capability to do the weight-shifting tasks.

6.3.2 Experimental Setup

The experimental setup consisted of (i) two balance boards (WiiBBs), (ii) a pair of slippers (iii) an HLD unit and (iv) a Task Computer (PC) with a 2D computer monitor executing the VR-based balance training tasks (Fig.6.5 (a)). Each participant was asked to stand on the two WiiBBs, 1 mm apart (similar to that used in [60], [61]) and placed on the ground in front of the PC (Fig. 6.5 (b)). Each WiiBB was fitted with a slipper. The slippers were used to restrict the unwanted movement of participants' feet over the WiiBBs. This was necessary as the $Central_{Hold}$ position of the VR_{Obj} was calibrated to participants' initial position before the start of the tasks. Without the slippers, the stroke survivors may change their position in between the balance training sessions which can lead to unwanted fluctuation in the CoP values. The participant was expected to stand with each feet in each slipper attached to the surface of the WiiBB. The slippers were oriented by about 7° from the vertical direction (Fig. 6.5 (b)) and pasted on the surface of WiiBBs. The position of the slippers was maintained to aid the participants to stand with their feet oriented at 14° with respect to each other and heels about 17 cm apart, similar to the setup used by Mclloroy et al. [60] for their study with healthy individuals and Mansfield et al. [61] for their study with post-stroke individuals. An HLD unit (Section 6.2.4) was used to measure the height of the participant's heel above the surface of the WiiBB and thereby facilitate the participants in following the Ankle strategy. The Task Computer (PC) monitor was placed on a table of height 0.8 meter approximately, so that the participants do not face any difficulty in viewing the VR-based tasks presented on the monitor while standing on the balance boards.

6.3.3 Procedure

The study required a commitment of approximately 45 minutes from each participant. Once the participant arrived in the experiment room, he/she was asked to sit on a chair and relax for 5 minutes. Then, a physiotherapist in the team assessed the participant's residual balance using Berg Balance Scale [62] and also ensured that the inclusion criteria were satisfied. This took around 10 minutes for each participant. If the participants fulfilled the inclusion criteria, the experimenter explained the experimental setup and also demonstrated three VR-based tasks by executing and explaining the tasks while standing on the balance boards. Then the experimenter



(a) (b)

Figure 6.5. (a) Experimental Setup of V2BaT system, (b) Foot placement on double WiiBBs for V2BaT system

ensured that the participant understood the task followed by administration of signing of the consent form. The participant was told that he/she was free to quit the balance training session at any time if he/she felt uncomfortable. Before starting the study, the experimenter asked the participant for his/her verbal consent.

Once the participant was ready, the experimenter attached the HLD unit to the participant's Affected leg and asked the participant to stand upright with one leg on each of the slippers attached to the WiiBB kept in front of the PC (Fig.6.5 (a)). I chose only the Affected leg for attaching the HLD unit, since, I was interested to know whether the participants were following Ankle strategy (with emphasis on the Affected side) while shifting weight. Then the experimenter started the experimental study that consisted of two stages, namely, Stage 1 and Stage 2 (Section 6.2.1). In Stage 1 (Fig. 6.1; Pre-task Calibration Stage), the experimenter asked the participant to (i) Step1: stand upright as much as possible for approximately 10 seconds followed by (ii) Step2: shift weight as much as possible in the anterior direction while following Ankle strategy. In Step1, the V2BaT system recorded the baseline CoP (initial position of CoP) due to participant's left leg (CoP_{Base_L}) and right leg (CoP_{Base_R}). The V2BaT also recorded the baseline distance (d_{ini}) between US sensor of the HLD unit and the surface of WiiBB (Section 5.2.5 of Chapter 5). A 10 seconds window was chosen for baseline measurement. This duration was chosen as an initial approximation and this can be modified in future based on the study design. In Step2, the participant was asked to move the VR_{Obj} (the boots in Fig. 6.1) as far as possible from the $Central_{Hold}$ position in the VR environment (that is the forest) by shifting his/her weight in the anterior direction. While the participant interacted with the VR-based task in Step2 of Stage 1, the V2BaT system recorded the participant's CoP displacement (ΔCoP) separately for each leg. The participants were asked to repeat Step2 thrice and the maximum CoP displacement for each leg ($(\Delta CoP)_L$ and $(\Delta CoP)_R$) was used to estimate individualized threshold

(ΔCoP_{THRESH}) and initial weight distribution as far as both the legs were concerned (w_{L_ini} and w_{R_ini}) (Section 6.2.2).

The Stage 1 was followed by Stage 2. In this stage, the participants were exposed to VR-based tasks for 20 minutes. In this, the participants were offered VR-based tasks of different templates with an intent to keep them motivated and interested to perform the VR-based exercise tasks. Also, based on their performance score in the task trials, the challenge level of the tasks was modified. The VR-based tasks required the participants to maneuver the VR_{Obj} in the VR environment from the $Central_{Hold}$ position to *Target* position by shifting their weight in the anterior direction (Fig. 6.2). Once the participants completed the balance training session, the experimenter administered the System Usability Questionnaires (see Section 6.3.4 below) among the participants.

6.3.4 System Usability Questionnaires

To understand the usability of V2BaT system among stroke participants, I used a questionnaire based on a five-point Likert scale [63] to get the participants' feedback after they finished interacting with the V2BaT system. I framed five questions in order to understand the participants' views on their usage of the V2BaT system. For this, I took idea from the USEQ as proposed by Gil-Gómez, José-Antonio, et al. [64] and framed the questions (three questions based on USEQ and two additional questions to check the motivational component of the tasks) as were relevant for the study. Keeping in line with the previous study (Chapter 5), I asked similar Usability-related questions (Section 5.2.8 of Chapter 5) to the participants in connection with the V2BaT system. The first question was "Did you face any difficulty in understanding the task?" (Q1). The idea was to understand whether the information provided by V2BaT system during the task was clear. The second question was "Did you find the task interesting?" (Q2).

This was asked so as know how they felt while interacting with the system. Again, although the study needed participation from each individual for one day, yet, since this system was designed with rehabilitation in mind that might need its usage over extended period, I asked them the third question, namely, "Do you think that the usage of this system would be beneficial to you?" (Q3). In addition, I wanted to understand whether the use of the V2BaT system was motivating to the participants. For this, I asked them two more questions, namely, "Will you agree to interact with the system again?" (Q4) and "Will you refer others to participate in the study?" (Q5).

6.3.5 Statistical Analysis

While the participants interacted with the VR-based tasks during Stage 2, the V2BaT system computed the participant's performance (Section 6.2.5) and also recorded displacement in CoP (ΔCoP) due to each leg (left and right) during the tasks offered in various *NTs* and *CTs*. I was interested to understand whether the operant conditioning paradigm using V2BaT system contributed to any statistical improvement in participant's performance and displacement in CoP from their $CT_{Initial}$ trial to CT_{Final} trial. A Shapiro-Wilk test of normality was performed on the participants' data which suggested that the data was normally distributed. Thus, I performed student's t-test of statistical hypothesis testing [65] to determine whether the improvement (if any) in the participants' performance score and a corresponding improvement in CoP displacement was statistically significant. The test was performed with the significance level set at $p\text{-value} < 0.05$.

6.4 Results and Discussion

I conducted a Usability study with the V2BaT system with 29 hemiplegic participants. The aim was to understand whether the V2BaT system augmented with operant conditioning paradigm was capable of improving participants' performance by making them use both of their legs as much as possible to the same extent while maneuvering the virtual object (VR_{Obj}) by shifting weight. Additionally, I also wanted to understand whether the V2BaT system can be acceptable to the target population.

6.4.1 Participants' Feedback on System Usability Questionnaires

Table 6.2 shows the participants' responses to the System Usability Questionnaires asked to them after they finished interacting with the V2BaT system. From the participants' responses, I found that the participants did not face any difficulty in understanding the tasks and were in fact

Table 6.2. Participants' feedback for V2BaT system

Q. No.	User Suitability Evaluation Question	Average Response Score
1	Did you face any difficulty in understanding the tasks?	1
2	Did you find the tasks interesting?	5
3	Do you think you can benefit by using such a system?	5
4	Do you want to play again with this system?	5
5	Do you want to refer any of your acquaintance to our study?	5

Note: 1 = Strongly disagree, 2 = Disagree, 3 = Neutral, 4 = Agree and 5 = Strongly agree.

interested in interacting with the system. Also, even with limited duration of exposure to V2BaT, they were optimistic about potential benefits that the system can bring to them as far as balance rehabilitation was concerned. They expressed their willingness to interact with the system again in future and also refer their known acquaintances to use V2BaT system. Thus, from the participants' feedback, it can be inferred that the V2BaT system has potential to be accepted by the target population. Table 6.2 shows the average response scores of the participants.

6.4.2 Effect of V2BaT System on Participants' Performance Score

The Fig. 6.6 shows the group average of participants' percentage performance score ($\%P_{f\_Score}$) in their first catch trial (CT_{First}) and best of final catch trials (CT_{Final}). The CT_{First} was the catch trial immediately after the first step of the Normal Trial that is, NT_{1n}. The CT_{Final} was the last catch trial offered to the participant before exiting from the balance training session. From Fig. 6.6, it can be seen that the participants achieved marginally ‘Inadequate’ (69% approximately) performance score on an average in their initial catch trial. Though the mean $\%P_{f\_Score}$ during CT_{First} was found to be very near to the threshold for ‘Adequate’ performance score, yet a detailed look at the participants’ performance data indicates that about 50% of the

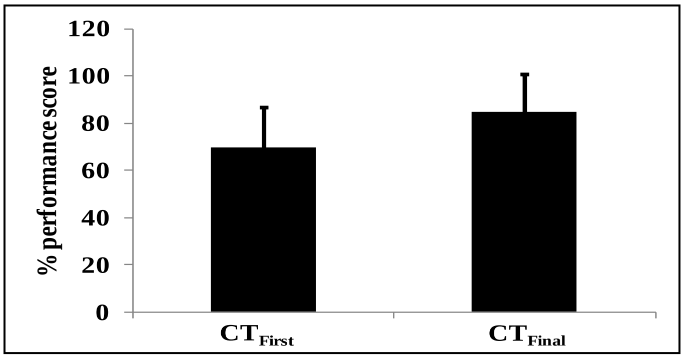


Figure 6.6: Group average of participants' performance score (%)

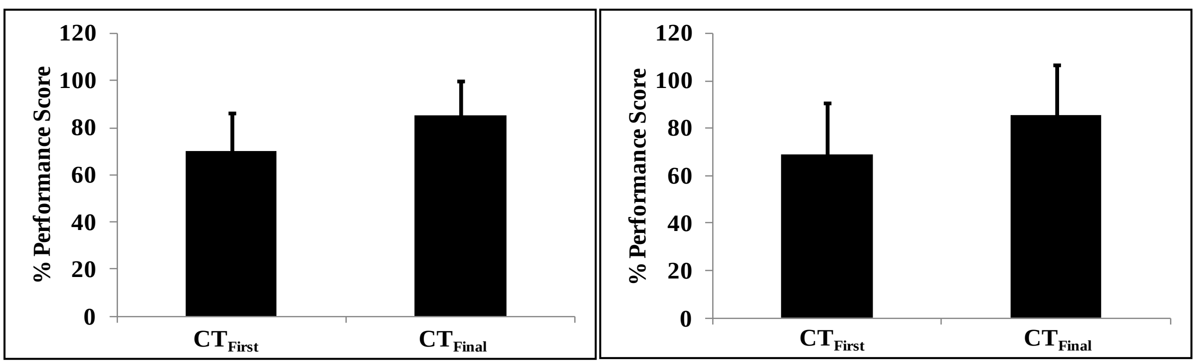


Figure 6.7: Average % performance score of (a) Left hemiplegic group, (b) Right hemiplegic group

participants had $\%P_{f\_Score}$ well below the threshold (70%) during CT_{First}. In contrast, the mean $\%P_{f\_Score}$ during CT_{Final} was well above the threshold with approximately 22.04 % of improvement in group average $\%P_{f\_Score}$ during CT_{Final} trial as compared to the CT_{First} trial. Almost 70% of the participant pool had a $\%P_{f\_Score}$ of > 80% during the CT_{Final} trial. Again, since the participant pool had a mix of left hemiplegic (n=7) as well as right hemiplegic (n=22) patients (enrolled based on the availability), I segregated the participants into two groups namely, left hemiplegic group (LH_{Group}) and right hemiplegic group (RH_{Group}). The Figs. 6.7 (a) and 6.7 (b) show a comparative estimate of the $\%P_{f\_Score}$ during CT_{First} and CT_{Final} trials for the LH_{Group} and RH_{Group}, respectively. In both of these figures, a similar improvement can be seen in the $\%P_{f\_Score}$ ($\Delta\%$=21.42% and 24.03% for the CT_{First} to CT_{Final} trials for LH_{Group} and RH_{Group}, respectively). For each participant, there was at least some improvement in the $\%P_{f\_Score}$ from CT_{First} to CT_{Final}.

A dependent sample t-test was carried out on the participants' $\%P_{f\_Score}$ and the result showed a significant improvement (p-value < 0.01) in performance score from the CT_{First} to the CT_{Final} trials. I also performed similar test for the LH_{Group} as well as the RH_{Group} and the result showed significant improvement in performance for both the groups (p-value < 0.01 for LH_{Group} as well as for the RH_{Group}).

To summarize, the V2BaT system with operant conditioning was able to motivate the post-stroke hemiplegic participants to increase the contribution of their Affected side during the weight-shifting task. This was evident from the overall statistically significant improvement in $\%P_{f\_Score}$ from the CT_{First} to CT_{Final} trials that used equal weight distribution as far as CoP trajectory due to individual leg towards maneuvering of the VR_{Obj} through weight-shifting was concerned.

6.4.3 Effect of V2BaT System on the displacement of Participants' Center of Pressure (CoP)

Having seen that there was an overall improvement in $\%P_{f\_Score}$ from the CT_{First} to CT_{Final} trials, I wanted to understand whether such an improvement was due to greater contribution of the Affected side alone at the expense of the reduced usage of the Unaffected side or majorly due to the Unaffected side alone. The aim of operant conditioning was to help the participant to improve the contribution of the Affected side for maneuvering the VR_{Obj} while enhancing the overall weight-shifting capability. The idea was to subtly encourage the participant to improve his/her ability on the Unaffected side as well. Thus, an in-depth analysis was carried out to find

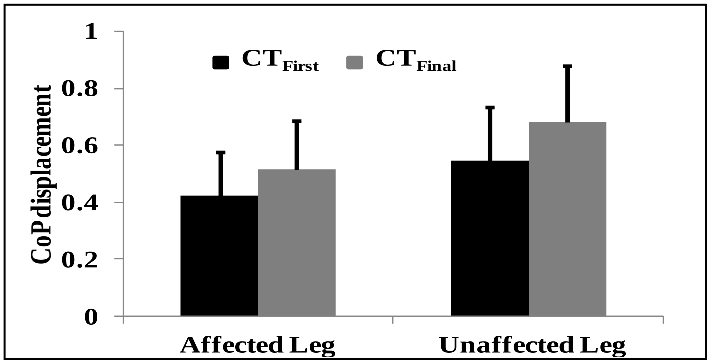


Figure 6.8: Group average of participants' CoP displacement (normalized)

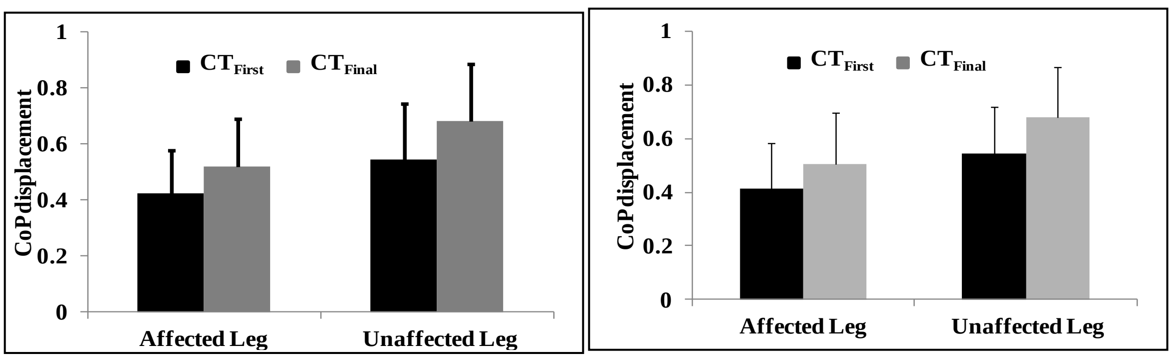


Figure 6.9: Average normalized CoP displacement of (a) left hemiplegic group, (b) right hemiplegic group

the contribution of each of the Affected and Unaffected legs of the hemiplegic participants to the overall improvement in their weight-shifting capability.

The Fig. 6.8 shows the group average of participants' normalized CoP displacement (ΔCoP) while interacting with CT_{First} and CT_{Final} trials. The ΔCoP was normalized with respect to the maximum ΔCoP showed by the participant pool. There was an improvement in the normalized ΔCoP from CT_{First} to CT_{Final} trials for both the Affected and the Unaffected sides (Δ%=21.29% and 25.25% for Affected and Unaffected legs, respectively). As far as the individual groups were concerned, there was a similar trend of improvement in CoP_{disp} with Δ% = 21.63% and 25.23%

for Affected and Unaffected legs, respectively for LH_{Group} and $\Delta\%=21.74\%$ and 25.31% for Affected and Unaffected legs, respectively for RH_{Group} (Figures. 6.9 (a) and (b)).

A dependent sample t-test was carried out on the CoP_{disp} contributed by the participants' Affected and Unaffected legs. The result showed a significant improvement (p-value < 0.01) in ΔCoP from the CT_{First} to the CT_{Final} trials for both the Affected and Unaffected legs. A similar test was performed for the LH_{Group} and the RH_{Group}. Results showed significant improvement in ΔCoP (p-value < 0.01) for both the Affected and Unaffected legs for the RH_{Group} as well as for the LH_{Group}.

To summarize, it can be seen that interaction with V2BaT system having operant conditioning can contribute to improvement in the ΔCoP for both the LH_{Group} and RH_{Group} as far as individual legs (Affected and Unaffected) were concerned. However, the amount of improvement in ΔCoP was nearly similar for both the legs. From this, it can be inferred that the operant conditioning have resulted in the improvement of the weight-shift capability on both the sides instead of improving the capability of one side at the expense of the other side.

6.4.4 Effect of V2BaT System on Participants' Performance during different Catch Trials and Normal Trial Steps

So far I have presented the contribution of each of the Affected and Unaffected legs towards overall improvement in $\%P_{f\_Score}$ during initial and final catch trials. I was also interested to see whether there was any consistency in the improvement in the $\%P_{f\_Score}$ across all the catch trials (inclusive of the intermediate catch trials). Added to studying the participants' performance at each *CT* trial, I was also interested to see how the participants responded to the operant conditioning when they were offered Normal Trials (*NTs*) with weight distribution (for both the

Affected and Unaffected legs) being modified from their comfortable setting (the weight distribution used during NT_{1n}).

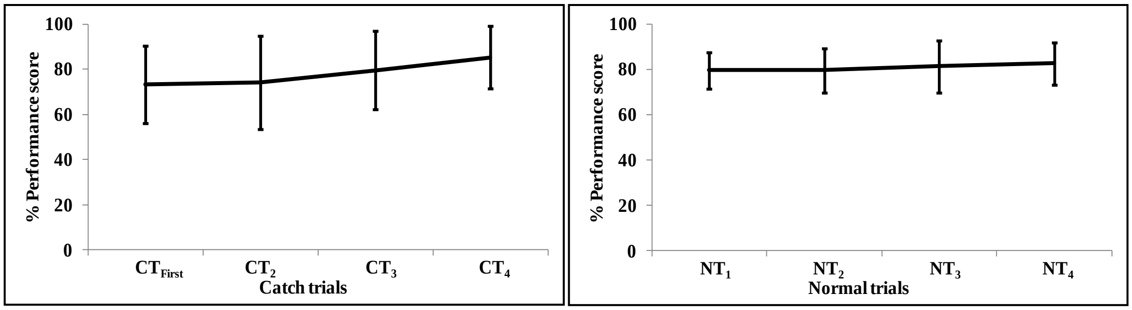


Figure 6.10. (a) Group average of performance score (%) at each CT, (b) Group average of performance score (%) at each NT ; Please Note: NT1 indicates NT Step 1, and others

The V2BaT system offered *NT* tasks to expose the participants to an operant conditioning regime with *CT* tasks in between the *NT* tasks. I offered *NTs* of increasing challenge along with intermediate *CTs*. The *CTs* were offered in between the *NTs* (without the participants' knowledge of whether the task was *NT* or *CT*). Please note that the number of intermediate *CTs* and number of *NT steps* (each stage composed of a number of *NTs* depending on how quickly a participant achieved 'Adequate' performance in tasks belonging to a particular difficulty level) offered to the participants was dependent on the individualized initial weight distribution (Section 6.2.2) as far as both the Affected and Unaffected legs of a participant were concerned.

Figs. 6.10 (a) and (b) show the group average of participants' $\%P_{f\_Score}$ at different *CTs* and trials of *NTs*, respectively. For each *NT*, the number of trials depended on the individualized performance. From Fig. 6.10 (a), an increasing trend in $\%P_{f\_Score}$ from CT_{First} to CT_i (where i = 1, 2, 3,...and so on) can be seen. Please note that based on the individualized performance capability, participants were offered different exit points. Specifically, before exiting from interaction with the V2BaT system, each participant was offered a CT_{Final} task trial. Based on the individualized weight distribution across both the legs, a task trial belonging to any one of the

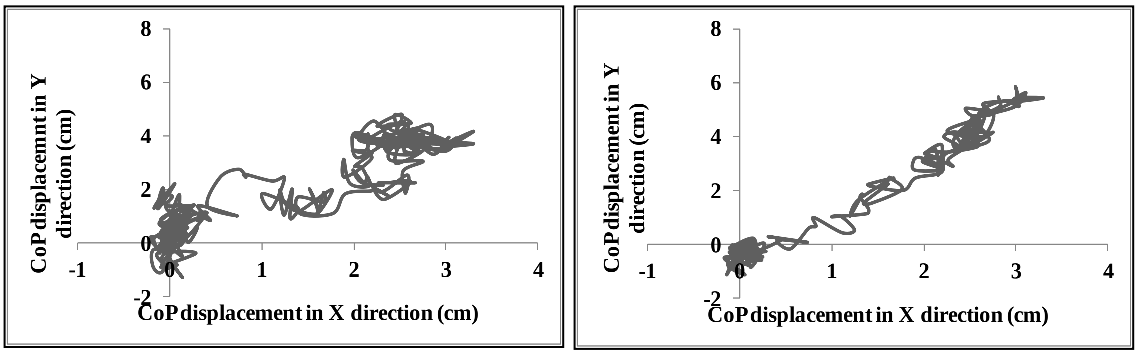
CT_i can be considered as the CT_{Final}. Again, from Fig. 6.10 (b), it can be seen that the $\%P_{f\_Score}$ was almost constant across NT_{1n} to NT_{4n} in spite of the fact that every *NT* was of increased challenge compared to the previous one as far as the weight distribution was concerned. Since the mean performance scores at different *NTs* were nearly similar (with small improvement = $\Delta\%$ = 4% approximately) across NT_{1n} to NT_{4n}, it can be said that each *NT* with inherent operant conditioning have helped the participants to actively compensate the difficulty introduced by the tasks of increased challenge offered by the V2BaT system at least partly through increased usage of the Affected leg.

In short, it can be said that for a limited exposure of one session, there was some improvement in terms of increased usage of Affected leg in the weight-shifting tasks of increasing challenge. This observation can be possibly attributed to the improved use of Affected leg by the hemiplegic participants as a result of the contribution of operant conditioning offered by V2BaT system during *NTs* through the use of modified weight distribution presented in a subtle manner. Also, the gradual improvement in the $\%P_{f\_Score}$ across the *CT* trials indicates the residual effect of operant conditioning provided by the *NTs*. I bilieve that increased exposure to such an environment over a prolonged duration (instead of a single session) might contribute to further improved $\%P_{f\_Score}$ across the balance training tasks.

6.4.5 Understanding the Implication of Operant Conditioning by a Case Study

Till now I have presented the group analysis of the participants' performance ($\%P_{f\_Score}$) along with the ΔCoP due to each leg (Affected and Unaffected) of individuals. The quality of weight-shifting in a balance task depends not only on the extent of COP_{disp} but also on the smoothness of the trajectory of the CoP (used to maneuver the VR_{Obj}) [66]. Thus, I wanted to make an in-depth comparative analysis into the trajectory of the CoP due to Affected leg during CT_{First} and CT_{Final} task trials, while a participant maneuvered the VR_{Obj} during a weight-shifting task.

I wanted to investigate the data particularly for participants for whom the BBS score was small enough to be in the high fall risk category. As a typical case, I chose participant S23 having the least BBS score < 45 (Table 6.1) among all the participants. I chose S23 since a BBS score of 41



(a) (b)

Figure 6.11. Affected leg's CoP trajectory for a typical case in (a) CT_{First} , (b) CT_{Final}
Note: Typical case is of participant S23

indicates high risk of fall [59]. As a manifestation of reduced balance capability, for S23, the weightage distribution of Affected:Unaffected legs was 0.39:0.61 for the NT_{1n} and 0.48:0.52 for NT_{4n} with CT ranging from CT_1 to CT_{Final} (CT_4 in this case).

Since each CT task used equal weight distribution as far as both the Affected and Unaffected legs were concerned, this was expected to pose greater challenge to the participants as compared to the NT tasks. Thus, I wanted to dig deeper into the CoP data for the Affected leg, particularly for the CT tasks. Figures 6.11 (a) and (b) show the CoP trajectory of the Affected leg for the participant S23 during CT_{First} and CT_{Final} task trials. It can be seen that S23 showed improvement (%Δ = 21%) in terms of greater ΔCoP (computed from the maximum distance between the start and end points in the anterior direction) in their CT_{Final} task trial compared to the CT_{First} task. Also, for S23, the spread of CoP trajectory (possibly due to sway along medio-lateral direction) was reduced by 5.86% from CT_{First} to CT_{Final} task trials.

To summarize, for S23, the change in the CoP trajectory from CT_{First} to CT_{Final} task trials might indicate that S23 has not only improved the amount of ΔCoP but also acquired improved control over his weight-shifting capability along the anterior direction. This is evident from the reduced sway of the CoP along the medio-lateral direction. Again, this improvement in the CoP trajectory did not come at the expense of performance. Specifically, there was an improvement in % performance score of approximately 16% of S23 from CT_{First} to CT_{Final} task trials. From this, it can be inferred that the system was potent to encourage the participant to use her Affected leg in tandem with the Unaffected leg, as much as possible thereby contributing to the overall improvement in the performance score in the VR-based balance tasks.

6.5 Conclusion

In this chapter, I have presented the V2BaT system with an aim to encourage the participants to increase the usage of their Affected leg during weight-shifting without explicitly directing them to do so and without constraining the abilities of the Unaffected leg. Results obtained through this study indicate that the contribution of participants' Affected leg increased almost in tandem with that of the Unaffected leg. The effect of this implicit conditioning could be seen as improved performance from CT_{First} to CT_{Final} trials for which the participants were required to use both of their legs as equally as possible while maneuvering the VR_{Obj}. This might infer that the operant conditioning can be applied to help the hemiplegic participants to move towards symmetrical weight distribution. The participants' feedback to the System Usability Questionnaires indicate that the V2BaT system was acceptable to the target population.

Though the preliminary results of V2BaT system are promising, yet, the study had some limitations such as patients with varied post-stroke period and different hemiplegic sides, and limited duration of exposure to the system. This study was used to administer exercises among participants only for one session of balance training. Such a limited exposure may not be sufficient to speak on the rehabilitation efficacy of the system. To see a significant improvement in an individual's clinical measure of balance, one needs to carry out a longitudinal study. Also, this must be associated with the clinical assessment of balance by measuring BBS score prior to and post the study. Also, the recruited participants had different residual balance (as evident from the BBS scores) based on the availability. This might have affected the group average performance scores. In future, I plan to carry out a more in-depth longitudinal study by enrolling a larger patient population, categorized based on residual balance before exposing them to the V2BaT. Also, questions remain on the transferability of the skills learnt from the VR-based controlled setting to real-life situations. However, this current study might serve as a stepping stone before moving to a full-fledged real-life task environment.

In the current study, I could understand the possible implications of interacting with VR-based setting augmented with operant conditioning on participant’s (i) usage of both the limbs and (ii) improvement in overall balance in a weight-shifting task. However, since oculomotor signature is a critical component related with neurological disorders such as stroke, there still remains an open question on the possible implications of operant conditioning on participant’s oculomotor signature while participating in a VR-based balance task. To search an answer to this question, I have designed a small extended study that I will present in the next chapter. In this I investigate the implications of conditioning paradigm on stroke participants' gaze behavior during a VR-based balance task.

Reference

[1] G. S. A. F. Verheyden et al., “Interventions for preventing falls in people after stroke.,” *Cochrane database Syst. Rev.*, no. 5, p. CD008728, 2013.

[2] S. F. Mackintosh, K. D. Hill, K. J. Dodd, P. A. Goldie, and E. G. Culham, "Balance Score and a History of Falls in Hospital Predict Recurrent Falls in the 6 Months Following Stroke Rehabilitation," *Arch. Phys. Med. Rehabil.*, vol. 87, no. 12, pp. 1583–1589, 2006.

[3] P. R. Rougier and N. Genthon, "Dynamical assessment of weight-bearing asymmetry during upright quiet stance in humans," *Gait Posture*, vol. 29, no. 3, pp. 437–443, 2009.

[4] P. T. Cheng, M. Y. Liaw, M. K. Wong, F. T. Tang, M. Y. Lee, and P. S. Lin, "The sit-to-stand movement in stroke patients and its correlation with falling," *Arch. Phys. Med. Rehabil.*, vol. 79, no. 9, pp. 1043–1046, 1998.

[5] L. C. Pereira, A. C. Botelho, and E. F. Martins, "Relationships between body symmetry during weight-bearing and functional reach among chronic hemiparetic patients.," *Rev. Bras. Fisioter.*, vol. 14, no. 3, pp. 229–266, 2010.

[6] J. F. Kamphuis, D. De Kam, A. C. H. Geurts, and V. Weerdesteyn, "Is weight-bearing asymmetry associated with postural instability after stroke? A systematic review," *Stroke Research and Treatment*. 2013.

[7] E. H. F. van Asseldonk et al., "Disentangling the contribution of the paretic and non-paretic ankle to balance control in stroke patients," *Exp. Neurol.*, vol. 201, no. 2, pp. 441–451, 2006.

[8] A. C. H. Geurts, M. De Haart, I. J. W. Van Nes, and J. Duysens, "A review of standing balance recovery from stroke," *Gait Posture*, vol. 22, no. 3, pp. 267–281, 2005.

[9] Y. Laufer, D. Sivan, R. Schwarzmann, and E. Sprecher, "Standing Balance and Functional Recovery of Patients with Right and Left Hemiparesis in the Early Stages of Rehabilitation," *Neurorehabil. Neural Repair*, vol. 17, no. 4, pp. 207–213, 2003.

[10] I. V. Bonan et al., "Reliance on Visual Information after Stroke. Part I: Balance on

Dynamic Posturography," *Arch. Phys. Med. Rehabil.*, vol. 85, no. 2, pp. 268–273, 2004.

[11] I.-C. Chen, P.-T. Cheng, C.-L. Chen, S.-C. Chen, C.-Y. Chung, and T.-H. Yeh, "Effects of balance training on hemiplegic stroke patients.," *Chang Gung Med. J.*, vol. 25, no. 9, pp. 583–590, 2002.

[12] C. B. de Oliveira et al., "Balance control in hemiparetic stroke patients: Main tools for evaluation," *J. Rehabil. Res. Dev.*, vol. 45, no. 8, pp. 1215–1226, 2008.

[13] G. Yavuzer, F. Eser, D. Karakus, B. Karaoglan, and H. J. Stam, "The effects of balance training on gait late after stroke: A randomized controlled trial," *Clin. Rehabil.*, vol. 20, no. 11, pp. 960–969, 2006.

[14] Z. Sawacha, E. Carraro, P. Contessa, A. Guiotto, S. Masiero, and C. Cobelli, "Relationship between clinical and instrumental balance assessments in chronic post-stroke hemiparesis subjects," *J. Neuroeng. Rehabil.*, vol. 10, no. 1, 2013.

[15] A. Mansfield, C. J. Danells, E. Inness, G. Mochizuki, and W. E. McIlroy, "Between-limb synchronization for control of standing balance in individuals with stroke," *Clin. Biomech.*, vol. 26, no. 3, pp. 312–317, 2011.

[16] K. H. Mauritz, "Gait training in hemiparetic stroke patients.," *Eura. Medicophys.*, vol. 40, no. 3, pp. 165–178, 2004.

[17] A. M. De Nunzio et al., "Biofeedback rehabilitation of posture and weightbearing distribution in stroke: A center of foot pressure analysis," *Funct. Neurol.*, vol. 29, no. 2, pp. 127–134, 2014.

[18] M. Sheikh, M. R. Azarpazhooh, and H. A. Hosseini, "Randomized comparison trial of gait training with and without compelled weight-shift therapy in individuals with chronic stroke," *Clin. Rehabil.*, vol. 30, no. 11, pp. 1088–1096, 2016.

[19] P. T. Cheng, S. H. Wu, M. Y. Liaw, A. M. K. Wong, and F. T. Tang, "Symmetrical body-weight distribution training in stroke patients and its effect on fall prevention," *Arch. Phys. Med. Rehabil.*, vol. 82, no. 12, pp. 1650–1654, 2001.

[20] F. Tripoli, S. Moreira, T. Oberg, and N. Lima, "Guided tasks and biofeedback: effects on weight transfer in hemiparetics," *Physiocratic Act*, vol. 15, no. 4, pp. 220–224, 2016.

[21] E. Taub et al., "An operant approach to rehabilitation medicine: overcoming learned nonuse by shaping.," *J. Exp. Anal. Behav.*, vol. 61, no. 2, pp. 281–93, 1994.

[22] A. P. Reiss, S. L. Wolf, E. A. Hammel, E. L. McLeod, and E. A. Williams, "Constraint-induced movement therapy (CIMT): current perspectives and future directions," *Stroke Res. Treat.*, 2012.

[23] S. J. Page et al., "Stroke patients??? and therapists??? opinions of constraint-induced movement therapy," *Clin. Rehabil.*, vol. 16, no. 1, pp. 55–60, 2002.

[24] E. Taub et al., "Technique to improve chronic motor deficit after stroke.," *Arch. Phys. Med. Rehabil.*, vol. 74, no. 4, pp. 347–354, 1993.

[25] J. H. van der Lee, R. C. Wagenaar, G. J. Lankhorst, T. W. Vogelaar, W. L. Devillé, and L. M. Bouter, "Forced use of the upper extremity in chronic stroke patients," *Stroke*. pp. 2369–2375, 1999.

[26] S. Hakkennes and J. L. Keating, "Constraint-induced movement therapy following stroke: A systematic review of randomised controlled trials," *Australian Journal of Physiotherapy*, vol. 51, no. 4. pp. 221–231, 2005.

[27] S. A. McLeod, "Skinner - Operant conditioning," *Simply Psychol.*, vol. 1, no. 1, p. 2, 2007.

[28] "Applied Behavior Analysis in Post-Stroke Therapy." [Online]. Available:

https://www.appliedbehavioranalysisedu.org/post-stroke/. [Accessed: 16-Jan-2018].

[29] R. Gatzounis, M. G. S. Schrooten, G. Crombez, and J. W. S. Vlaeyen, "Operant Learning Theory in Pain and Chronic Pain Rehabilitation," *Curr. Pain Headache Rep.*, vol. 16, no. 2, pp. 117–126, 2012.

[30] W. E. Fordyce, R. S. Fowler, J. F. Lehmann, B. J. Delabeur, P. L. Sand, and R. B. Triesch-mann, "Operant conditioning in the treatment of chronic pain," *Arch. Phys. Med. Rehab*, vol. 54, pp. 399–408, 1973.

[31] A. K. Thompson, F. R. Pomerantz, and J. R. Wolpaw, "Operant conditioning of a spinal reflex can improve locomotion after spinal cord injury in humans.," *J. Neurosci.*, vol. 33, no. 6, pp. 2365–75, 2013.

[32] K. Yasuda, K. Saichi, Y. Kitaji, H. Harashima, and H. Iwata, "Development of an implicit method for directing weight shifting to the affected side in patients with stroke: a proof of concept study," *ROBOMECH J.*, vol. 4, no. 1, p. 26, 2017.

[33] K. Lohse, N. Shirzad, A. Verster, N. Hodges, and H. F. M. Van der Loos, "Video games and rehabilitation: using design principles to enhance engagement in physical therapy.," *J. Neurol. Phys. Ther.*, vol. 37, no. 4, pp. 166–75, 2013.

[34] K. Laver, S. George, S. Thomas, J. E. Deutsch, and M. Crotty, "Virtual reality for stroke rehabilitation," *Stroke*, vol. 43, no. 2. 2012.

[35] Y. X. Hung, P. C. Huang, K. T. Chen, and W. C. Chu, "What do stroke patients look for in game-based rehabilitation: A survey study," *Med. (United States)*, vol. 95, no. 11, 2016.

[36] N. Maclean, "Qualitative analysis of stroke patients' motivation for rehabilitation," *BMJ*, vol. 321, no. 7268, pp. 1051–1054, 2000.

[37] H. J. Jeon, M. Y. Kim, J. U. Lee, and J. Kim, "Differences in leg length discrepancy and

weight distribution between the healthy and unhealthy sides of hemiplegic stroke patients," *Toxicol. Environ. Health Sci.*, vol. 5, no. 4, pp. 221–226, 2013.

[38] B. S. Rajaratnam et al., "Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?," *Rehabil. Res. Pract.*, vol. 2013, pp. 1–6, 2013.

[39] K. Salen and E. Zimmerman, "Rules of Play: Game Design Fundamentals," *Nihon Ronen Igakkai Zasshi.*, p. 672, 2004.

[40] M. Stokes, *Physical Management in Neurological Rehabilitation*. 2004.

[41] I. J. Hubbard, M. W. Parsons, C. Neilson, and L. M. Carey, "Task-specific training: Evidence for and translation to clinical practice," *Occupational Therapy International*, vol. 16, no. 3–4. pp. 175–189, 2009.

[42] J. M. Veerbeek et al., "What is the evidence for physical therapy poststroke? A systematic review and meta-analysis," *PLoS ONE*, vol. 9, no. 2. 2014.

[43] B. Bates, *Game Design*, vol. Masters. 2003.

[44] R. Koster, "Theory of Fun for Game Design," *A Theory Fun Game Des.*, p. 256, 2005.

[45] A. A. Timmermans, H. A. Seelen, R. D. Willmann, and H. Kingma, "Technology-assisted training of arm-hand skills in stroke: Concepts on reacquisition of motor control and therapist guidelines for rehabilitation technology design," *Journal of NeuroEngineering and Rehabilitation*, vol. 6, no. 1. 2009.

[46] P. van Vliet and G. Wulf, "Extrinsic feedback for motor learning after stroke: What is the evidence?," *Disability and Rehabilitation*, vol. 28, no. 13–14. pp. 831–840, 2006.

[47] M. A. Guadagnoll and T. D. Lee, "Challenge Point: A Framework for Conceptualizing the Effects of Various Practice Conditions in Motor Learning," *Journal of Motor Behavior*,

vol. 36, no. 2. pp. 212–224, 2004.

[48] E. L. Bjork and R. A. Bjork, “Making Things Hard on Yourself, But in a Good Way: Creating Desirable Difficulties to Enhance Learning,” *Psychol. Real World Essays Illus. Fundam. Contrib. to Soc.*, pp. 55–64, 2009.

[49] R. J. Pagulayan, K. Keeker, D. Wixon, R. L. Romero, and T. Fuller, “User-centered Design in Games,” in *The Human-computer Interaction Handbook*, 2003, pp. 883–906.

[50] C. Klimmt, C. Blake, D. Hefner, P. Vorderer, and C. Roth, “Player performance, satisfaction, and video game enjoyment,” in *Lecture Notes in Computer Science (including subseries Lecture Notes in Artificial Intelligence and Lecture Notes in Bioinformatics)*, 2009, vol. 5709 LNCS, pp. 1–12.

[51] Y. Lee, J. G. Her, Y. Choi, and H. Kim, “Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors’ Names,” *J. Phys. Ther. Sci.*, vol. 26, no. 2, pp. 179–182, 2014.

[52] A. Darekar, B. J. McFadyen, A. Lamontagne, and J. Fung, “Efficacy of virtual reality-based intervention on balance and mobility disorders post-stroke: A scoping review,” *J. Neuroeng. Rehabil.*, vol. 12, no. 1, 2015.

[53] H. Y. Lee, Y. L. Kim, and S. M. Lee, “Effects of virtual reality-based training and task-oriented training on balance performance in stroke patients.,” *J. Phys. Ther. Sci.*, vol. 27, no. 6, pp. 1883–8, 2015.

[54] D. Lin, H. Seol, M. A. Nussbaum, and M. L. Madigan, “Reliability of COP-based postural sway measures and age-related differences,” *Gait Posture*, vol. 28, no. 2, pp. 337–342, 2008.

[55] J. C. Metzger et al., “Assessment-driven selection and adaptation of exercise difficulty in

robot-assisted therapy: A pilot study with a hand rehabilitation robot," *J. Neuroeng. Rehabil.*, vol. 11, no. 1, 2014.

[56] K. Jack, S. M. McLean, J. K. Moffett, and E. Gardiner, "Barriers to treatment adherence in physiotherapy outpatient clinics: A systematic review," *Manual Therapy*, vol. 15, no. 3. pp. 220–228, 2010.

[57] B. M. Young et al., "Changes in functional connectivity correlate with behavioral gains in stroke patients after therapy using a brain-computer interface device," *Front. Neuroeng.*, vol. 7, 2014.

[58] J. Babič, E. Oztop, and M. Kawato, "Human motor adaptation in whole body motion," *Sci. Rep.*, vol. 6, no. 1, p. 32868, 2016.

[59] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[60] W. E. McIlroy and B. E. Maki, "Preferred placement of the feet during quiet stance: Development of a standardized foot placement for balance testing," *Clin. Biomech.*, vol. 12, no. 1, pp. 66–70, 1997.

[61] A. Mansfield, G. Mochizuki, E. L. Inness, and W. E. McIlroy, "Clinical correlates of between-limb synchronization of standing balance control and falls during inpatient stroke rehabilitation," *Neurorehabil. Neural Repair*, vol. 26, no. 6, pp. 627–635, 2012.

[62] K. O. Berg, B. E. Maki, J. I. Williams, P. J. Holliday, and S. L. Wood-Dauphinee, "Clinical and laboratory measures of postural balance in an elderly population.," *Arch. Phys. Med. Rehabil.*, vol. 73, no. 11, pp. 1073–1080, 1992.

[63] R. Likert, "A technique for the measurement of attitudes," *Archives of Psychology*, vol. 22

140. p. 55, 1932.

[64] J. A. Gil-Gómez, P. Manzano-Hernández, S. Albiol-Pérez, C. Aula-Valero, H. Gil-Gómez, and J. A. Lozano-Quilis, "USEQ: A short questionnaire for satisfaction evaluation of virtual rehabilitation systems," *Sensors (Switzerland)*, vol. 17, no. 7, 2017.

[65] T. W. Anderson, "An introduction to multivariate statistical analysis," *Wiley series in probability and statistics*, vol. 14, no. 9–12. p. xx, 721 , 2003.

[66] M. Roerdink, M. De Haart, A. Daffertshofer, S. F. Donker, A. C. H. Geurts, and P. J. Beek, "Dynamical structure of center-of-pressure trajectories in patients recovering from stroke," *Exp. Brain Res.*, vol. 174, no. 2, pp. 256–269, 2006.

CHAPTER 7

INVESTIGATORY STUDY INTO GAZE BEHAVIOR OF HEMIPLEGIC POST-STROKE PARTICIPANTS DURING VIRTUAL REALITY BASED BALANCE TRAINING AUGMENTED WITH OPERANT CONDITIONING PARADIGM

7.1 Introduction

In the previous chapter, I have studied the implications of the V2BaT system which used operant conditioning paradigm by varying the weightage allocated to the Center of Pressure (CoP) contribution of each leg of a hemiplegic post-stroke patient performing a balance task. The V2BaT system encouraged hemiplegic participants to use both of their legs as much as possible to the same extent in maneuvering the virtual object through weight-shifting while participating in the standing balance tasks. The idea was to encourage the hemiplegic participants to maintain balance while using both of their legs as equally as possible, critical for performing activities of daily living. There is evidence from literature that an individual employs sensory inputs such as from eyes to maintain balance [1], [2]. Also, I know from the previous study using SmartEye system (Chapter 3) that individuals with neurological disorder such as stroke often exhibit oculomotor abnormalities that is not present in their healthy counterparts. Such oculomotor abnormalities can affect the sensory input derived from the vision system necessary for maintaining balance. This is because, sensory inputs from vision system via different eye movements is an integral part of human balance system (along with inputs from vestibular and somatosensory systems) [3], [4]. Therefore, the gaze behavior of stroke patient may have implications on their balance. This motivated us to extend the previous study (Chapter 4) to understand (i) the gaze behavior of stroke survivors while they perform balance related tasks and

(ii) the implication of operant conditioning on gaze behavior of hemiplegic post-stroke participants performing balance tasks. Thus, this chapter addresses two research questions, namely, (i) what is the gaze behavior of hemiplegic stroke patients when they perform goal-directed VR-based weight-shifting tasks? and (ii) what are the implications of operant conditioning on the gaze behavior of post-stroke participants? While addressing these research questions, the post-stroke participants were exposed to VR-based balance tasks augmented with operant conditioning (V2BaT) (similar to that in the previous chapter (Chapter 6)) and recorded their gaze data in a time-synchronized manner.

There is a rich body of literature connecting an individual's vision with vestibular system. The roles of vision, vestibular system, and the somatosensory system are very important in maintaining balance. Among these systems, the vision provides important information on an individual's position in environment and postures [4]. Researchers have reported that we move our eyes in different ways followed by fixation and that an individual's gaze behavior can be understood in the context of a particular task [5], [6], [7]. While being stationed at a location, an individual fixate on a visual scene to retrieve information. However, the control and timing of eye movement and associated body movements vary widely with the task [8]. Researchers such as Land et al. [5] and Hayhoe et al. [6] have studied an individual's gaze fixation pattern during the performance of a well-learned task in a natural setting (such as making tea), and to classify the types of movement that the eyes perform. They have reported that during a given task, an individual's eyes usually fixate on the same object throughout the action although they often move on to the next object in the sequence before completion of the preceding action. Given a particular action of maneuvering an object from an initial position to a target position, the specific roles of individual fixations could be identified as locating the target position, planning a direction to the target prior to contact with the target, supervising the relative movements of other objects and checking whether some particular condition is met prior to the termination of the action [8]. For example, during visually-guided goal-directed manual tasks using upper limb, Sailer et al. [9] have showed that healthy individuals' gaze behaviour changed with the skill level while doing the task. Here, the authors have reported a task that was aimed to control a cursor by using a computer peripheral held in the hand so that the cursor can touch the target stimulus that appeared on the computer screen. During the initial phase of the task (exploratory stage), when a participant's performance was poor, the eye tracking data showed that the participant's gaze mostly followed the cursor. But, the gaze behavior was different during the later phase of the task, namely, the skill acquisition (that is when the participant's performance started to improve) and skill refinement (that is, when the performance score settled at a maximum value) stages. Specifically, during the later phase of the task, the participant's gaze tended to shift towards the target [9]. These research studies involved healthy participants and none from the pathologic group. Designing studies to understand the gaze behavior of pathologic group such as those with stroke is essential. This is because, individuals with neurological disorder such as stroke often demonstrate impairment in motor control as far as ocular and limb movement are concerned [10], [11]. Given these deficits, Rizzo et al. [12] have studied eye-hand coordination in patients with chronic cerebral injury. They have reported that post-stroke, the spatial and temporal relationships between the eye and hand are disrupted and these eye-limb coordination skills need to be specifically targeted during neurorehabilitation. Proper coordination between eyes and limbs is critical while performing activities such as climbing up the stairs. For example, when an individual performs visually-guided stepping while moving up the stairs, the individual consistently makes saccadic eye movements to the next target (the next upward step) prior to initiation of the swing phase towards that target [13], [14]. Inconsistencies in visually-guided stepping task have been reported for individuals with neurological disorders. For example, in a study of the eye movement and stepping characteristics of cerebellar patients during precision stepping, Crowdy et al. [15] demonstrated that delays in target fixation caused by oculomotor deficits were associated with delays in generating accurate steps. Furthermore, inaccurate eye movements were sometimes accompanied by missed steps [15]. These studies highlight the existence of interaction between the oculomotor and locomotor control systems during precision stepping and highlight how deficits in an individual's eye movement control can have a detrimental effect on an individual's locomotion.

Though numerous literatures can be found on individuals' gaze bahavior in relation to the activities requiring upper limbs and also locomotion tasks, yet, investigations on an individual's gaze behaviour during standing balance task are sparse. In one of the studies on eye movement and postural stabilization, Uchiyama et al. [16] revealed the role of fixation on postural stabilization while individulas were asked to maintain upright standing posture in a dark room. The authors reported that individuals who were given visual cue by showing a target position to fixate during the task, showed reduced body sway velocity compared to the individuals who were not provided with any visual cue. In a recent study, Dutta et al. [17] have conducted a visuomotor balance task with healthy individuals. The task was to maneuver a cursor on the computer screen from an initial central position to fixed Target positions presented on the screen. In this, the authors reported that the ratio of fixation duration towards a Target to that on the moving cursor increased with increase in performance score in the visuomotor balance task, inferring increased looking towards the target being associated with improved performance. These observations on an individual's gaze behavior during postural control in quiet standing task [16] and visuomotor balance task [17] were reported for healthy individuals. However, these reports suggest that the information on an individual's gaze fixation during standing balance task might play an important role in improving the existing balance rehabilitation techniques for individuals with balance disorders. Additionally, neurological disorder such as stroke can alter an individual's normative eye movement [18]. Thus, it is important to understand the gaze behavior of such participants during standing balance tasks. Motivated by this, in the current study, I wanted to investigate the gaze behavior of hemiplegic stroke patients while they performed balance tasks through weight-shifting. Also, I wanted to understand the implications of operant conditioning on their gaze behavior during goal-directed weight-shifting tasks.

To achieve this, I have used the previously developed V2BaT system (Chapter 6) to provide VR-based weight-shifting tasks coupled with operant conditioning to the participants. Additionally, the V2BaT system was augmented with a head-mounted gaze-sensitive assembly. This assembly features a head-mounted gaze-sensitive display unit to monitor participant's eye gaze while performing the balance tasks. The aim was to understand the gaze behavior of hemiplegic stroke participants while V2BaT system augmented with operant conditioning paradigm aimed to encourage them to use both of their legs as much as possible to the same extent in maneuvering the VR object through weight-shifting.

7.2 System Design

In this study, I wanted to understand a participant's eye gaze behavior while hemiplegic post-stroke participants performed weight-shifting in a Balance-related task. Also, I wanted to understand the implications of operant conditioning on participant's gaze behavior. For this, the V2BaT system (discussed in Chapter 6) was used to offer VR-based weight-shifting task augmented with operant conditioning and this was coupled with a head-mounted gaze-tracking assembly.

This system consisted of eight units, namely, (a) VR-based Task (b) Individualized Threshold and Weight Distribution Estimator (c) Wii Balance Boards-VR Handshake (d) Heel Lift Detection (e) Performance Evaluation (f) Task Switching (g) Head-mounted Gaze Sensitive Display (h) Gaze Data Analysis units.

7.2.1 VR-based Task Unit

For this unit, I used Vizard software toolkit (from Worldviz Llc.) to design realistic VR-based balance training tasks with variations to make the weight-shifting exercise interesting. The tasks required the participants to shift their weight in anterior direction while maneuvering a context-relevant VR object (VR_{Obj}) in the VR environment. The VR-based tasks were categorized into (i) Stage 1 and (ii) Stage 2 tasks. Stage 1 was the pre-task calibration stage. This Stage was used to compute the individualized threshold and initial weight distribution as far as the contribution of both legs was concerned (Please see Section 6.2.3 of Chapter 6). The Stage 2 was the VR-based task execution stage. In this, the participants started with the initial weight distribution (as obtained from Stage 1) and then moved on to interact with task trials of increasing challenge that depended on the weight distribution between the Center of Pressure (CoP) contribution of both the legs as recorded by the two balance boards (similar to that in Section 6.2.1.2 of Chapter 6). Please refer to Section 6.2.1 of Chapter 6 for more details on the VR-based task unit.

7.2.2 Individualized Threshold and Weight Distribution Estimator Unit

This unit was used in Stage 1 of the study. This unit was designed to estimate a participant's individualized residual balance before interacting with V2BaT system. Subsequently, this information was used to vary the weight distribution as far as the CoP contribution of each leg was concerned thereby varying the difficulty of VR-based tasks. In Stage 1, the participant's range of movement was assessed by asking him/her to shift weight in the anterior direction to his/her maximum ability (following Ankle strategy) while standing on the Wii Balance Boards (WiiBBs). This unit was same as that used in the previous study. For details, please see Section 6.2.3 of Chapter 6.

7.2.3 WiiBB-VR Handshake Unit

The VR-based tasks required the participant to maneuver virtual objects (VR_{Obj}) in the VR environment (shown on the stimulus display monitor (Described below in Section 7.2.7.2) using their CoP while standing on WiiBBs. In the VR-based task execution stage (Stage 2), the position of VR_{Obj} was controlled by the weighted sum of the displacement of the CoPs obtained from two WiiBBs due to participant's shifted weight while participating in the task. This unit was the same as that used in the previous study. For more details, please see Section 6.2.2, chapter 6.

7.2.4 Heel Lift Detection Unit

Similar to that in the previous studies reported in Chapters 5 and 6, here also a Heel Lift Detection Unit (HLD) unit was used to ensure that the participants followed Ankle strategy which is considered important during standing balance task [19]. To ensure that the Ankle strategy was followed, the participants were asked not to lift their heel from the surface of the WiiBB while shifting their weight. For details on this unit, please see Section 6.2.4 of Chapter 6.

7.2.5 Performance Evaluation Unit

While the participants were interacting with the VR-based weight-shifting tasks, the system computed their performance score based on the CoP displacement and also included penalty factor in case the Ankle strategy was '*Not Followed*'. The rationale for computing the performance score was similar to that as described in Section 6.2.5 of chapter 6.

7.2.6 Task Switching Unit

I designed a Usability study with the developed system with an aim to understand the gaze behavior of the hemiplegic post-stroke participants while they interacted with the weight-shifting task offered by the V2BaT system. The task switching criteria used in the study presented in this chapter was similar to that used in the previous study (For details, please see Section 6.2.6 of Chapter 6). However, the duration of tasks being presented to the participants was restricted to 10 mins instead of the 20 mins duration (Section 6.2.6 of Chapter 6). This was because, in the present study, the participants needed to (i) stand on the WiiBB and (ii) also use a Head-mounted assembly (Described in Section 7.2.7.3 below) while performing a weight-shifting task instead of only doing weight-shifting while standing on the WiiBB (as that in the previous study (Chapter 6)).

7.2.7 Head-mounted Gaze-Sensitive Display Unit

In the present work, I wanted to study the eye movements of post-stroke participants while they interacted with VR-based balance training tasks offered on a stimulus presentation screen. For recording the eye movements, a remote Eye Tracker from EyeTribe [20] (that was the same as that used in the study presented in Chapter 3) was used. The VR-based tasks were presented on a stimulus display monitor. Both the (i) Eye Tracker and (ii) stimulus display monitor were attached to an in-house fabricated (iii) helmet assembly. The entire unit will be referred to as **He**ad-**mo**unted **Ga**ze **S**ensitive **D**isplay *(Hemo-GaSiD)* unit. The sections below give a description of each of the component of the *Hemo-GaSiD* unit.

7.2.7.1 Eye Tracker Unit

In this study, a commercially available low-cost remote Eye Tracker from EyeTribe [20] was used. Technically, the EyeTribe tracker has been reported to be comparable with other state-of-the-art Eye Tracker such as EyeLink 1000 with regard to spatial precision and accuracy, thereby making it suitable for recording an individual's fixation, doing point-of-regard and pupilometry-related analyses [21]. The EyeTribe Eye Tracker is non-intrusive and tracks one's eye movement

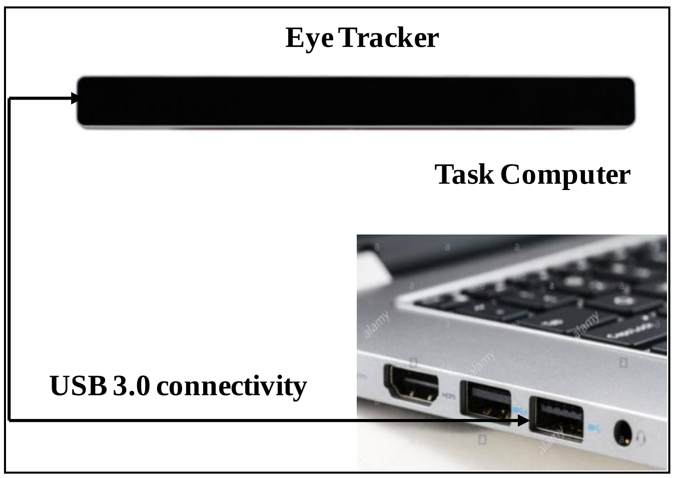


Figure 7.1. Eye Tracker system for *Hemo-GaSiD* module

by using a camera along with a high-resolution infrared source. This Eye Tracker has sampling rates of 30 Hz and 60 Hz with accuracy varying from 0.5^0 to 1^0 [20]. This Eye Tracker can be powered through USB 3.0 connectivity that interfaces it with computers and tablets. This is capable of recording gaze parameters, such as gaze coordinates, pupil size, and pupil center. This tracker can be used in presence of eye glasses or contact lens. It is portable and light-weight (70 grams approximately). In the present study, the Eye Tracker unit was attached to the *Hemo-GaSiD* unit (Fig. 7.4) and connected to the USB 3.0 of the Task Computer through a wired connection.

7.2.7.2 Stimulus Display Monitor

In this study, a small and light-weight (approximately, 150 grams) stimulus display monitor was used to present the VR-based tasks to the participants. This monitor comes with an LCD screen with display size of 7 inches (diagonal) and needs 5 V, 2 A power source. The monitor

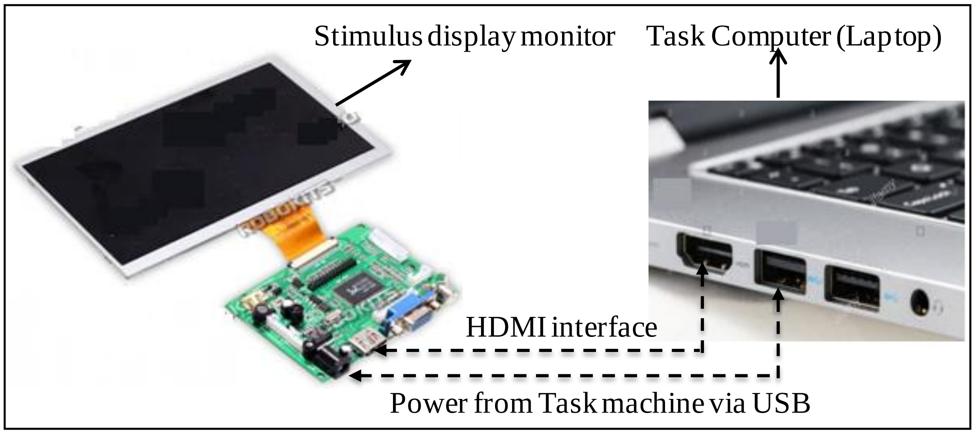


Figure 7.2. Connectivity of stimulus display monitor with task Computer

has a screen resolution was 800*480 pixels along with one number each of HDMI and VGA ports. The monitor was powered from the Task Computer using a USB 2.0 port. Also, the Stimulus Display Monitor was connected to the Task Computer machine using HDMI port (Fig. 7.2). The Stimulus Display Monitor was attached to the *Hemo-GaSiD* unit (Fig. 7.3 below).

7.2.7.3 Helmet Assembly

The helmet assembly, fabricated in-house, comprised of (a) a helmet, (b) a pair of telescopic channels, (c) Screen Support Structure, (d) a pair of J-shaped hooks, and (e) an adjustable Pelvic Support Harness. The Helmet used in this study was an open face motor sport type helmet weighing approximately 1 Kg. The pair of telescopic channels (of dimension 45 cm x 3 cm) with partial extension runner (adjustable length between 38 cm to 60 cm) was attached to the helmet at one end using rigid joints on both the sides (Fig 7.3). The other end of each telescopic channel was hinged to the Screen Support Structure by using Hinge Joint. The telescopic channels were meant to offer an adjustable distance of 32 cm to 54 cm between the participant's eyes and the

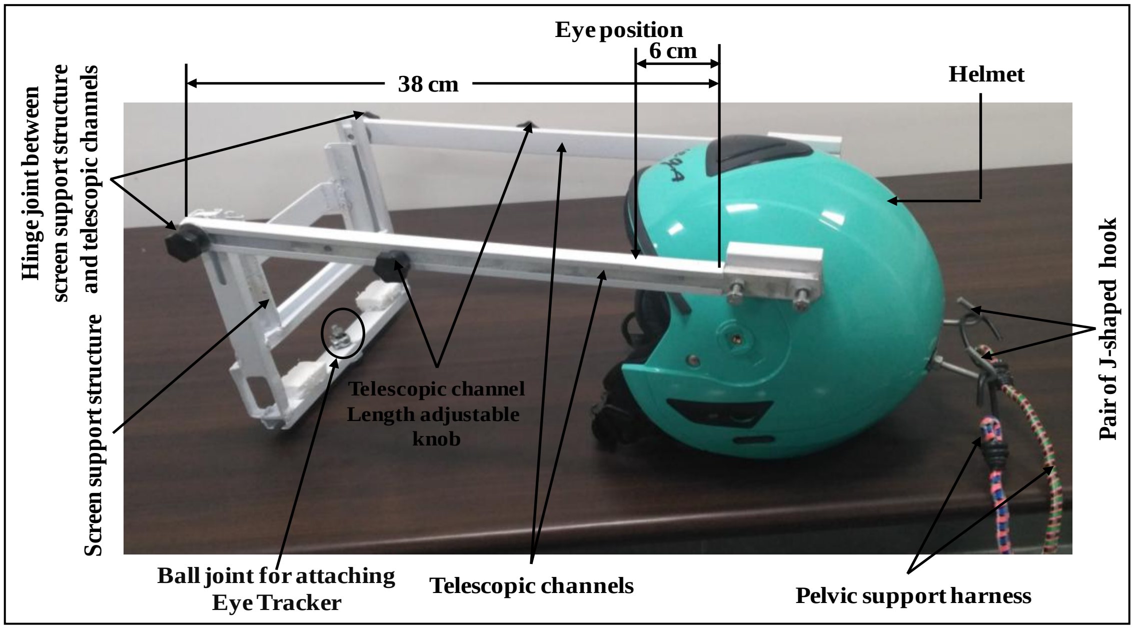


Figure 7.3. Helmet assembly

assembly of Stimulus Display Screen (Section 7.2.7.2) along with the Eye Tracker Unit (Section 7.2.7.1) housed in the Screen Support Structure. This variable distance of 32 to 54 cm was maintained following the technical specifications of the Eye Tracker [20]. The Hinge joints connecting the telescopic channels and the Screen Support Structure allowed us to adjust the angular position of the Support Structure. The Support Structure consisted of two sections, one meant for the Stimulus Screen Monitor (that was held in place with the help of an in-built channel) and the other meant for mounting the Eye Tracker Unit. The Eye Tracker Unit was mounted with the help of a ball joint that allowed angular adjustment of the Eye Tracker so that the participant's eyes can be properly detected by the Eye Tracker. The weight of the *Hemo-GaSiD* unit having the helmet along with the Screen Support Structure, Eye Tracker Unit and Stimulus Display Monitor was approximately 1.9 Kg. Using a light-weight plastic rod carrying a level meter (weighing approximately, 200 grams) and the method of balance [22], the center of gravity (CG) of the *Hemo-GaSiD* unit was obtained as indicated in Fig. 7.4. An adjustable Pelvic

Support Harness hooked with the helmet assembly was used to reduce the possible relative movement between participant's head and the helmet assembly. The Pelvic Support Harness comprised of a belt (to be worn on the pelvis) and pair of ordinary nylon elastic ropes with hooks connected at both ends. This harness was connected with the helmet assembly fitted with two J-shaped hooks screwed at the rear side of the body of the helmet. The entire assembly was set so that the center of the Stimulus Display Monitor was collinear with the participant's eyes.

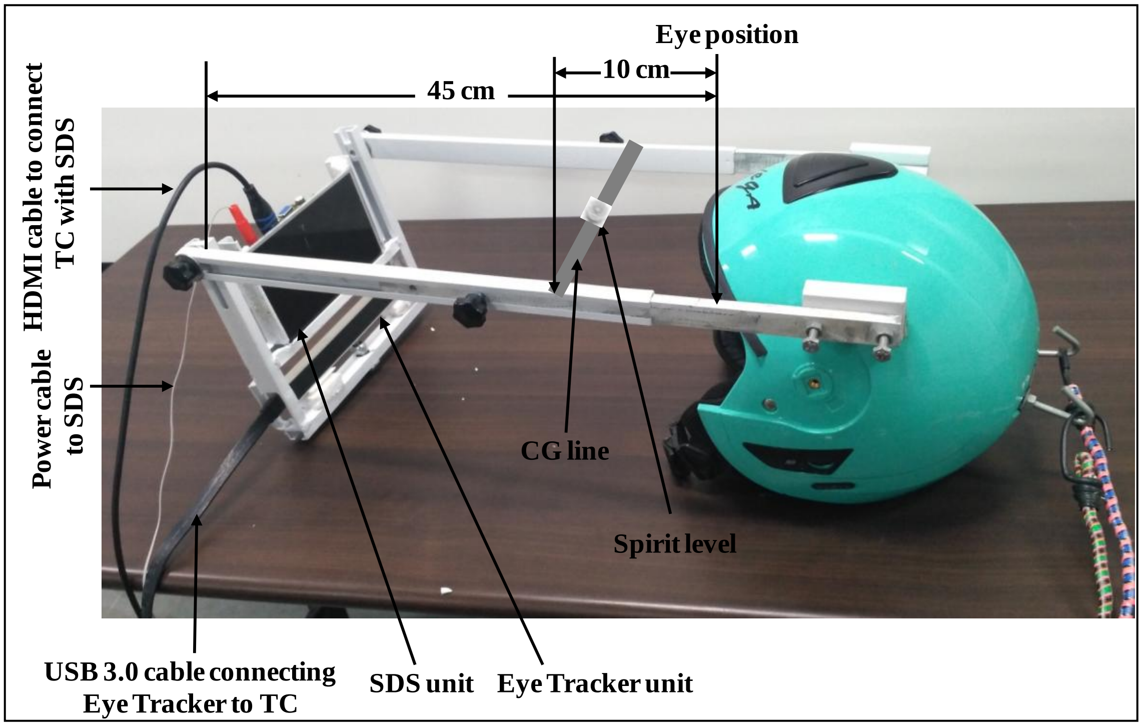


Figure 7.4. *Hemo-GaSiD* module

Note: TC= Task Computer, SDS= Stimulus Display Screen, CG= Center of Gravity

7.2.8 Gaze data Analysis Unit

In this study, I wanted to understand the gaze behavior of hemiplegic post-stroke participants while they interacted with V2BaT system augmented with operant conditioning paradigm. The V2BaT system encouraged them to use both of their legs as much as possible to the same extent in maneuvering the VR_{Obj} during weight-shifting. Prior work on gaze behavior during visually-guided manual task using upper limb [9] with healthy individuals suggests that, in goal-directed tasks, an individual's gaze behaviour changes with the skill level while doing the task. In this study by Sailer et al., the participants were asked to control a cursor on the screen while using a device (referred to as a rectangular tool). The participant was asked to grasp two cylindrical handles situated one at each end of the box. Subsequently the participant could move the cursor by applying bimanual force and torque about the long axis of the rectangular tool. The task was to move the cursor to touch a target stimulus presented on the computer screen. During the initial phase of the task (exploratory stage), when a participant's performance was poor, the eye tracking data showed that the participant's gaze mostly followed the cursor. In contrast, during the later phase of the task when the participant's performance started to increase (skill acquisition stage) and settled at maximum performance (skill refinement stage), the participant's gaze shifted towards the target [9]. Again, there is evidence of such variation in an individual's gaze pattern during a visuomotor balance task. Specifically, Dutta et. al. [17] have reported such

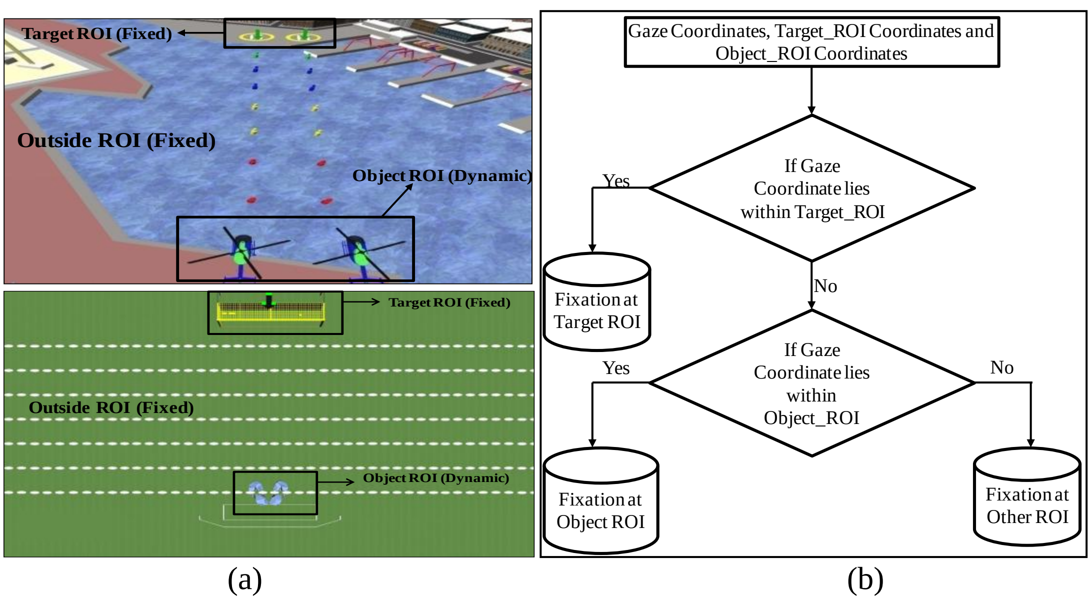


Figure 7.5. (a) Typical stimulus with different ROIs for eye tracking, (b) Algorithm to find the gaze fixation in different ROIs.

gaze behaviour while participating in a visuomotor balance task. The task was to maneuver a cursor on the computer screen from an initial central position to fixed target positions presented on the screen. In this, the authors reported that the ratio of fixation duration towards a target (that is the duration for which the participant looked towards the target) to that on the moving cursor increased with increase in performance score in the visuomotor balance task inferring increased looking towards the target. In the study, instead of the cursor, I had a VR object (VR_{Obj}) and the target was a position that was situated in the anterior direction to that of the central position ($Central_{Hold}$). The task was to maneuver the VR_{Obj} from the $\mathrm{Central}_{\mathrm{Hold}}$ position to the Target position through weight-shifting while using both the legs to the same extent as much as possible. In the present study, I was interested to investigate the gaze fixation pattern of post-stroke patients when being exposed to the visual stimulus presented to them in the form of VR-based balance training tasks. To achieve this, the stimulus (VR-based task presented on the stimulus display monitor) was divided into three regions of interest (ROIs). Specifically, these were *Target_ROI*, *Object_ROI* and *Outside_ROI* (Fig 7.5 (a)) that indicated the static *Target* region, the dynamic VR_{Obj} region in the VR-based task environment and all regions outside that covered by the *Object_ROI* and *Target_ROI*, respectively.

While the participant interacted with VR-based balance tasks presented on a stimulus display monitor, the Eye Tracker unit (mounted in the *Hemo-GaSiD* unit) connected to the Task Computer recorded the participant's time-synchronized gaze coordinates. The 2-D screen coordinates of the *Target_ROI* and *Object_ROI* along with the gaze coordinates were then processed offline to extract total fixation duration towards different ROIs. The flow diagram for calculating total fixation duration towards different ROIs is shown in Fig. 7.5 (b).

7.3 Experiment and Methods

7.3.1 Participants

The study was carried out after informed consent at an Institute of Neuroscience at Kolkata, West Bengal, where the stroke survivors were undergoing treatment. In the present study, five hemiplegic stroke survivors (S1-S5) (mean (SD) = 44.16 years (14.60)) with varying residual balance and post-stroke period volunteered. All the participants were right hemiplegic. The Berg Balance Scale (BBS) score > 45 for all the participants indicating that all of them were at a lower risk of falling [23]. Table 7.1 shows the participants' metadata. Since, (i) the participants were all

Table 7.1. Participants' metadata for investigatory study of gaze behavior during goal-directed balance task

S. No.	Age (years)/ Gender	Hemiplegic Side	Post-stroke Period	BBS Score	BBS_north Score
S1	31/ Male	Right	2 year	53	4
S2	66/ Male	Right	2 months	50	3
S3	55/ Male	Right	1.5 years	53	3
S4	30/ Male	Right	3.5 years	49	4
S5	48/ Male	Right	2 months	52	2

Note: S= Stroke participant, BBS= Berg Balance Scale; BBS_north= Reaching

hemiplegic, (ii) the task was to shift weight in the anterior direction and (iii) the BBS score is not a measure of an individual's direction-specific weight shifting ability, I also report one measure (last column in Table 7.1) out of the 14 measures (each scaled 1 to 4) of BBS score that can provide direction-specific (anterior direction) information (BBS_north). Specifically, I chose the score for 'Reaching Forward with Outstretched Arm while Standing' task of BBS when the participant needed to shift weight towards the front (that is in the Anterior direction). The inclusion criteria were (1) ability to follow the instructions (2) ability to stand and shift weight without orthopedic aids and (3) should be able to see stimulus screen at a distance of 50 cm (4) should not have gone through ocular surgery in recent past and (5) should not have gone through any surgery in recent past that may interfere with their capability to do the weight-shifting tasks. The study was carried out by following institutional research ethics.

7.3.2 Experimental Setup

The experimental setup (Fig 7.6 (a)) consisted of (i) two balance boards (WiiBBs), (ii) a pair of slippers (iii) an HLD unit (iv) a task computer and (v) head-mounted gaze sensitive display unit with an Eye Tracker (*Hemo-GasiD*; Section 7.2.7.3). Each participant was asked to wear the

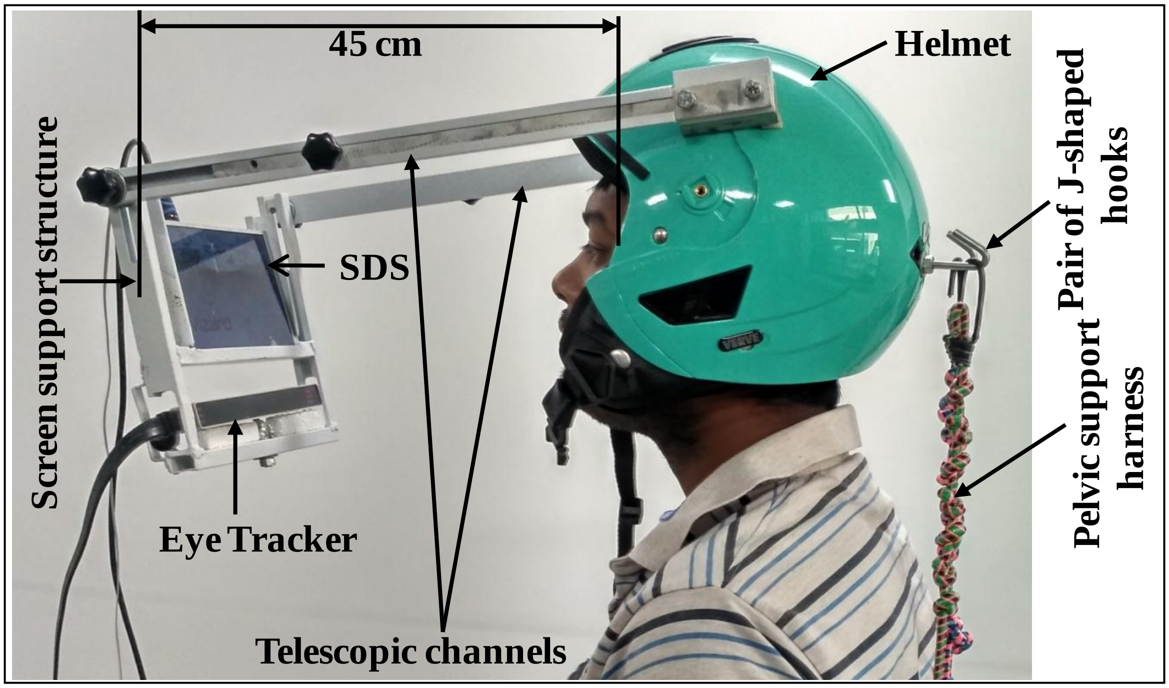


Figure 7.6. Participant wearing *Hemo-GaSiD* module
Note: SDS= Stimulus Display Screen, CG =center of gravity

Hemo-GasiD unit on his head and stand on the two WiiBBs placed on the ground 1 mm apart (similar to that used in [24], [25]). Each WiiBB was fitted with a slipper and the participant was expected to stand with each feet in each slipper attached to the surface of the WiiBB. The slippers were oriented by 7° from the vertical direction and pasted on the surface of WiiBBs (Fig. 7.6 (b), similar to the experimental setup described in Section 6.3.2 of Chapter 6). The helmet was connected with a Screen Support Structure that consisted of Stimulus Display Monitor and Eye Tracker unit (Fig. 7.3). The distance between the Screen Support Structure and the participant's eyes was adjustable and was kept at around 45 cm [20] for all the participants. The

VR-based tasks running on the Task Computer was projected on the Stimulus Display Monitor attached in the *Hemo-GasiD* unit using a 3 meter long HDMI cable. The time-synchronized gaze data measured by the Eye Tracker Unit was transmitted via a 2 meter long cable connecting the USB 3.0 ports of both the Eye Tracker and the Task Computer executing the VR-based balance tasks. To avoid the relative movement between participant's head and helmet, (i) we made sure that helmet should properly fit in the participant's head by using extra cushioning if needed and (ii) we used two elastic ropes (7.2.7.3) to connect the J-shaped hooks with the pelvic harness to balance the weight of the *Hemo-GaSiD* unit.

7.3.3 Procedure

The study required a commitment of approximately 25 minutes from each participant. Once the participant arrived in the experiment room, he/she was asked to sit on a chair and relax for 5 minutes. Then, a physiotherapist in the team assessed the participant's residual balance using BBS score [26] and also ensured that the inclusion criteria were satisfied. This step took around 10 minutes for each participant. If the participants fulfilled the inclusion criteria of the study, the experimenter explained the experimental setup and also demonstrated three VR-based tasks by executing and explaining the tasks while standing on the balance boards and wearing the *Hemo-GaSiD* unit on the head. Then the experimenter ensured that the participant understood the task followed by administration of signing of the consent form. The participant was told that he/she was free to quit from the study at anytime if he/she felt uncomfortable. Before starting the study, the experimenter asked the participant for his/her verbal consent.

Once the participant was ready, the experimenter helped him to wear the *Hemo-GaSiD* unit and also attached the HLD unit to the participant's Affected leg and asked him to stand upright with one leg on each of the slippers attached to the WiiBB placed on the ground. Subsequently, the experimenter started the study that consisted of two stages, namely, Stage 1 and Stage 2 (similar to that as mentioned in Section 6.2.1 of Chapter 6). In Stage 1 (Pre-task Calibration Stage), the experimenter asked the participant to (i) Step1: stand upright as much as possible for approximately 10 seconds followed by (ii) Step2: shift weight as much as possible in the anterior

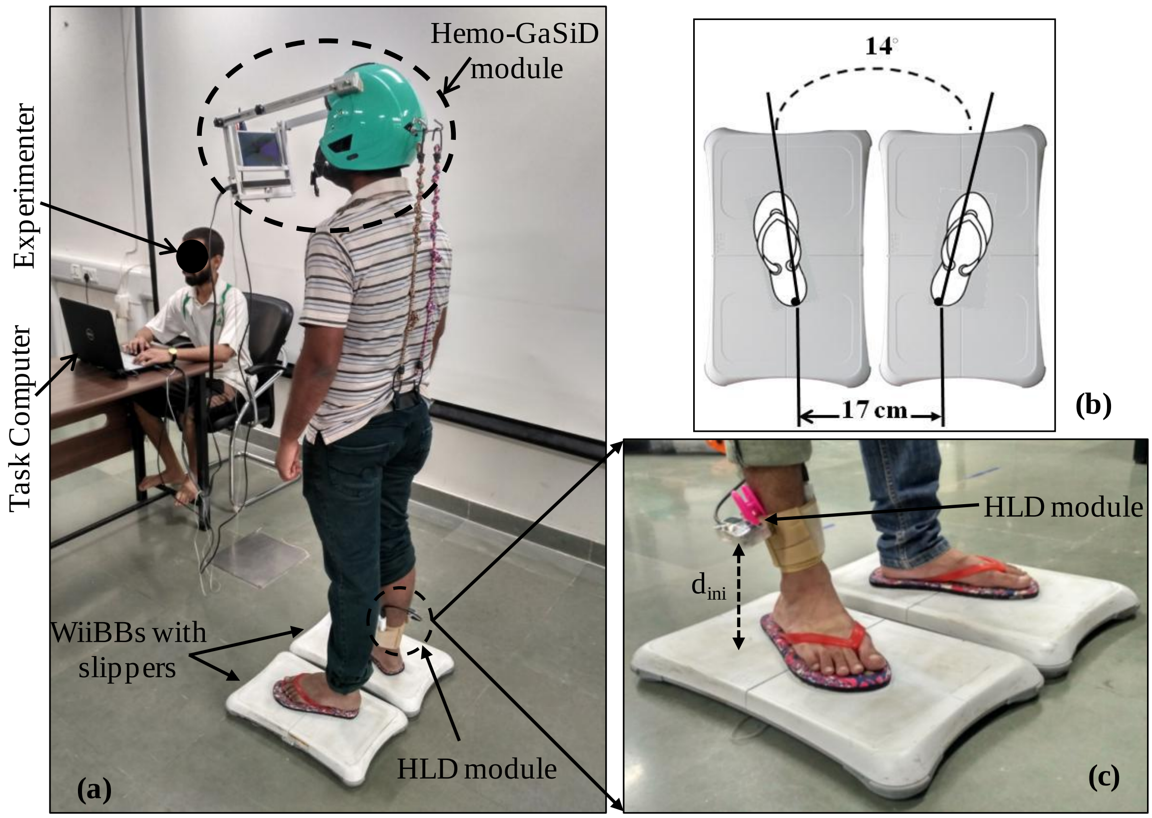


Figure 7.7. (a) Experimental setup, (b) placement of WiiBBs and (c) placement of HLD module for studying gaze behavior during goal-directed balance task

Note: d_{ini} =initial height between ultrasonic sensor of HLD module and surface of WiiBB

direction. In Step1, the V2BaT system recorded the baseline CoP position ($Central_{Hold}$: initial position of CoP) due to participant's left leg and right leg. The V2BaT also recorded the distance (d_{ini}; Section 6.2.4 of Chapter 6) between ultrasonic (*US*) sensor of the HLD unit and the surface of WiiBB which was used to detect any heel lift by the participant during Stage 2 (VR-based task execution stage). In Step2, the participant was asked to move the VR_{Obj} as far as possible from the initial position ($Central_{Hold}$) in the VR environment by shifting his/her weight in the anterior direction while following Ankle strategy. While the participant interacted with the VR-based task in Step2 of Stage 1, the V2BaT system recorded the participant's CoP displacement (ΔCoP) for each leg separately. Then, the maximum ΔCoP was computed to derive ΔCoP_{max_L} for left leg and ΔCoP_{max_R} for right leg. This data was used to compute (i) individualized threshold (ΔCoP_{THRESH}) and (ii) initial weight distribution, namely, w_{L_ini} for left leg and w_{R_ini} for right leg (same as that in previous chapter, Chapter 6 in Section 6.2.2).

The Stage 1 was followed by Stage 2. In this stage, the participants were exposed to VR-based tasks for 10 minutes. In this, the participants were offered VR-based tasks of different templates with an intent to keep them motivated and interested to perform the VR-based exercise tasks. Also, based on their performance score in the task trials, the challenge level of the tasks was modified. The VR-based tasks required the participants to maneuver the VR_{Obj} in the VR environment from the $Central_{Hold}$ position to *Target* position (Fig. 6.2 of Chapter 6) by shifting their weight in the anterior direction. While the participant performed the VR-based task, the Eye Tracker attached to *Hemo-GaSiD* unit recorded gaze data of the participant. Once the participant completed the study, the experimenter obtained feedback from the participant.

7.3.4 End-of-Study Feedback

After the participant finished interacting with the system, the experimenter asked two questions to understand the user's impression on the system. The participant's feedback was acquired to understand his/her view on the VR-based tasks and also on the experimental setup. Thus, the experimenter asked them (i) Did you face any difficulty in understanding the tasks? and (ii) Did you find the helmet assembly as inconvenient?

7.4 RESULT

The aim was to understand gaze behavior of hemiplegic stroke participants while V2BaT system augmented with operant conditioning paradigm encouraged them to use both of their legs to the same extent, as much as possible, to maneuver the VR_{Obj} through weight-shifting. The VR-based tasks offered by the system in Stage 2 were of two types, namely (i) Catch Trial (*CT*) and (ii) Normal Trial (*NT*). Similar to that used in previous chapter, the *CT* was one in which equal weight distribution was allocated to each of the Affected and Unaffected legs. Again, *NT* refer to the trials in which the weight distribution allocated to each of the Affected and Unaffected legs was not equal. While interacting with the first *CT* trial (CT_{First}) the participant was not exposed to any task with operant conditioning achieved through variation in weight distribution between the Affected and Unaffected legs as far as weight-shifting tasks were concerned. Again, in the course of interaction with the system, the participants were exposed to other *CT* trials that were offered intermediate between the *NT* trials (that came with operant conditioning). Among these *CT* trials, I considered the CT_{Final} trial that corresponded to the participant's best performance in terms of performance score to get an insight into the implication of transfer of some of the residual effects [27] from the completed *NTs* to the concerned CT_{Final}. Thus, I was interested to understand the variation in participant's performance and gaze pattern in CT_{Final} (having the maximum performance score) vis-à-vis the CT_{First} trial.

7.4.1 Acceptability of the System

Post the study, the experimenter administered an exit questionnaire (Section 7.3.4) in which the experimenter asked the participants few questions to get an idea on their impression about the system. In response to the question on the difficulty in understanding the VR-based tasks, all the participants expressed that they did not face any problem in understanding the tasks. In response to the second question regarding the experimental setup, all the participants (except S5) said that they did not face any inconvenience in interacting with the VR-based tasks while wearing the *Hemo-GaSiD* unit. The participant S5 reported that it seemed bit heavy.

7.4.2 Variation in Participants' Performance in VR-based Weight-shifting Tasks

There is evidence from literature that operant conditioning can improve participant's ability to perform a task [28] that can have clinical applicability. However, the role of operant conditioning to weight-shifting among post-stroke participants has not been adequately explored. In the

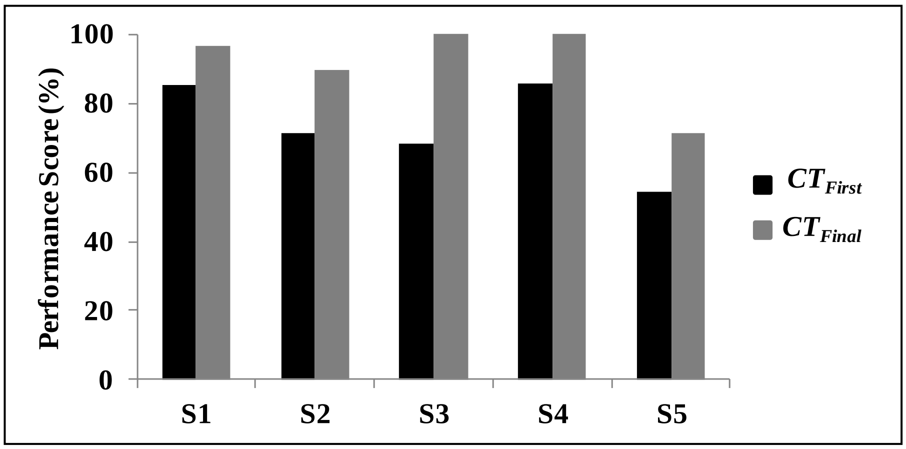


Figure 7.8. Participants' percentage performance score

present study, the hemiplegic post-stroke participants took part in VR-based weight-shifting tasks augmented with operant conditioning. The Fig. 7.8 shows the participants' percentage performance score *(*$\%P_{f\_Score}$*)* in their first catch trial (CT_{First}) and best of final catch trials (CT_{Final}). All the participants showed improvement in $\%P_{f\_Score}$ from CT_{First} to CT_{Final} task trial.

As far as the performance in CT_{First} task trial was concerned, three (S1, S2 and S4) of the five participants achieved 'Adequate' (>=70%) performance score in their CT_{First} task while others did not. The performance score (P_{f_Score}=69%) of S3 during CT_{First} was very near to the threshold performance for 'Adequate' score. However, that (P_{f_Score} =54.45%) of S5 during CT_{First} was well into the 'Inadequate' range. A possible reason for such a low performance score of S5 can be attributed to the lowest score for the 'Reaching Forward with Outstretched Arm while Standing' task of BBS (Table 7.1) indicating that S5 had issues with shifting weight in the anterior direction. Again, as far as the performance in CT_{Final} task trial was concerned, all the participants scored adequately.

There was an overall improvement (%Δ = 26.46%) in the group average performance ($\%P_{f\_Score}$) from CT_{First} to CT_{Final} trial to be close to that (%Δ = 24.05%) in the previous study (Chapter 6). One of the reasons behind greater improvement in $\%P_{f\_Score}$ from CT_{First} to CT_{Final} trials in the present study compared to that in the previous study might be that all of the participants (S1-S5) in the present study had improved residual balance (Group average BBS = 52) in comparison to that (Group average BBS = 49) in the previous study. From the overall improvement in the $\%P_{f\_Score}$ from CT_{First} to CT_{Final}, it can be inferred from both the studies that there is at least some potential of VR-based tasks augmented with operant conditioning to contribute to improved weight-shifting capability in post-stroke hemiplegic participants.

7.4.3 Variation in Gaze Behavior in VR-based Weight-shifting tasks

There is evidence from literature that an individual's gaze pattern varies with skill level acquired while doing a task. One of manifestations of improved skill acquisition can be performance score. Various investigators have reported this for tasks related to both upper and lower limbs. For example, Sailer et al. [9] studied the variation in gaze pattern in response to visually-guided manual tasks that required one to maneuver a cursor using upper limb to touch the target stimulus presented on the computer screen. The authors report that during the initial phase of the task, when an individual's performance was poor, the gaze pattern followed the object (cursor). However, a task of higher performance was characterized by increased looking towards the target. Also, Dutta et al. [29] have reported a visuomotor balance task in healthy participants where the task was to maneuver a cursor from a central location to pre-defined target positions by shifting weight using lower limbs. Here the authors report that improved task

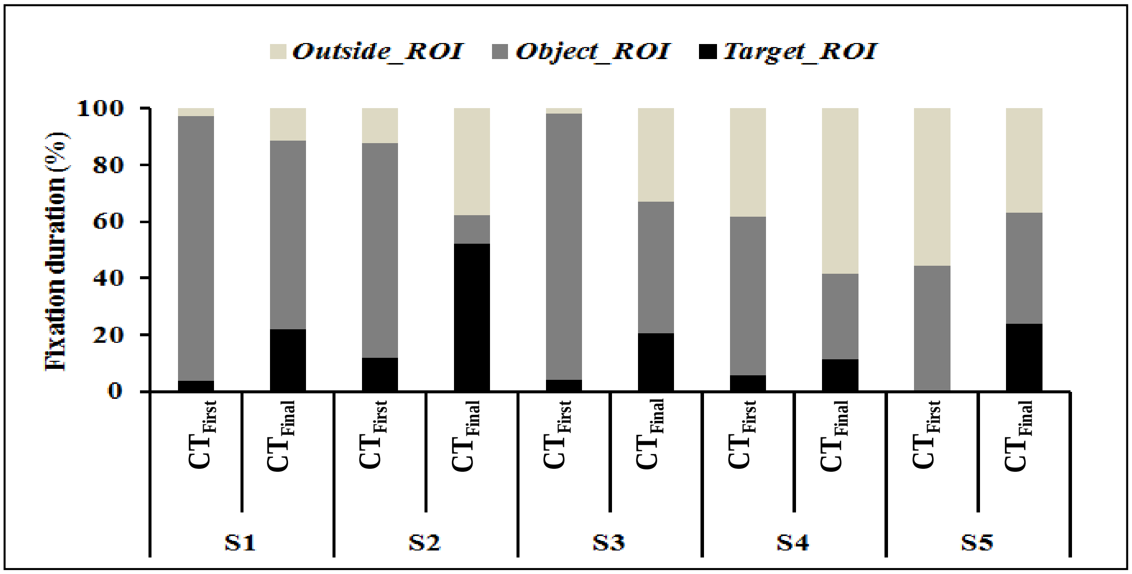


Figure 7.9. Participants' percentage fixation duration at different ROIs

performance was associated with increased viewing of the target quantified as increase in the ratio of fixation duration on the target to that on the cursor. In the present study, it can also be observed that there was an improvement in performance score ($\%P_{f\_Score}$) from CT_{First} to CT_{Final} (Section 7.4.2) that can be possibly due to interacting with the weight-shifting tasks equipped with operant conditioning. Thus, I wanted to relate the variation in the participants' gaze fixation pattern while they maneuvered a VR_{Obj} from $Central_{Hold}$ position to a fixed *Target* position. To achieve this, I calculated the total fixation duration towards *Target_ROI, Object_ROI and Other_ROI* (Section 7.2.8) for each trial of the VR-based weight-shifting tasks. Fig. 7.9 shows the participants' percentage fixation duration towards different ROIs in their CT_{First} and CT_{Final} task trials.

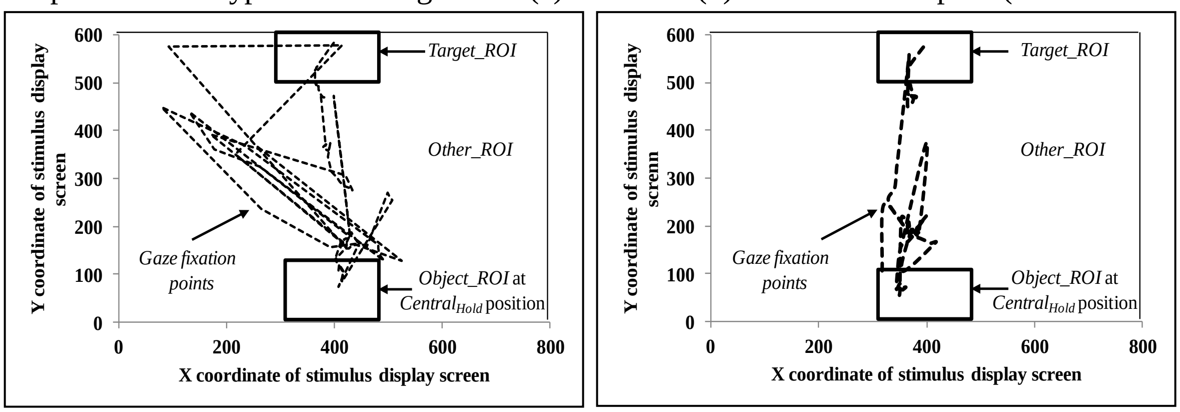
From Fig 7.9, it can be seen that during CT_{First} task trial (the initial phase of the VR-based task), all the participants fixated their gaze mostly towards *Object_ROI* and less often towards the *Target_ROI* (except S5 who did not look to the *Target_ROI*, but only looked towards

Object_ROI and *Other_ROI*), similar to that in literature. In contrast, in the CT_{Final} task trial, there was an overall (i) decrement in fixation duration towards the *Object_ROI* along with (ii) increment in the fixation duration towards the *Target_ROI*. From this it can be inferred that the increased viewing of the *Target_ROI* during CT_{Final} might be at least partially connected with the operant conditioning resulting in improved $\%P_{f\_Score}$. Overall, all the participants looked towards the three ROIs in varying proportions during CT_{First} and CT_{Final} tasks (except S5). Specifically, S5 did not fixate towards *Target_ROI* during CT_{First} while looking towards *Object_ROI and Other_ROI* in both the CT_{First} and CT_{Final} tasks. A possible explanation of such an atypical fixation towards the *Target* might be that S5 had the least residual balance in the anterior direction (least BBS_North score; Table 7.1). Consequently, he might be more concerned with maneuvering the VR_{Obj} while looking towards the *Object_ROI* and not towards the *Target_ROI*. This possibly resulted in S5 to score the least in the CT_{First} task among all the participants. Even his $\%P_{f\_Score}$ in the CT_{Final} task was marginally 'Adequate' (~71%) that was again the least among all the participants.

To summarize, preliminary findings on gaze behavior of hemiplegic participants suggested that during CT_{First} task (when the participants were not exposed to *NT* tasks with operant conditioning), they were more interested in controlling the VR_{Obj} with increased fixation towards the *Object_ROI*. However, once, the participants were exposed to *NTs* of different challenge levels augmented with operant conditioning, they were motivated to increase the usage of both of their legs (as evident from their improvement in performance score). Additionally, there was a variation in their looking pattern as regards the *Target_ROI* and *Object_ROI*. Specifically, in the CT_{Final} task, there was an increase in their fixation towards the *Target_ROI* coupled with reduction in that towards the *Object_ROI*.

7.4.4 An Analysis of the Scan Path: a Case Study

Till now I have presented the participants' fixation behavior towards different ROIs in terms of percentage fixation duration. The participants maneuvered the VR_{Obj} from the $Central_{Hold}$ to *Target* positions of the VR environment by shifting weight. I wanted to understand participant's (i) gaze switching pattern between different ROIs with the dynamic *Object_ROI* moving between the two static positions (such as the $Central_{Hold}$ and *Target* positions) and (ii) the variation in participant's scan path plot from CT_{First} to CT_{Final} task trials. Here, I present the scan path for participant S4 as a typical case. Figs. 7.10 (a) and 7.10 (b) show the scan path (while removing



(a) (b)

Figure 7.10. Scan path of a typical case in (a) CT_{First} task, (b) CT_{Final} task
Note: Typical case is for participant S4

the fixations outside the stimulus screen) of S4 during CT_{First} and CT_{Final} tasks. From Fig. 7.10 (a), it can be seen that during CT_{First} task, S4 fixated at many points lying intermediate between the $Central_{Hold}$ and *Target* positions possibly because, the trajectory of the VR_{Obj} was noisy due to reduced control of VR_{Obj} by S4. Also, he fixated on the *Target* position only a few number of times. In contrast, during CT_{Final} task, the trajectory of the VR_{Obj} as viewed by S4 is more prominent than that in the CT_{First} task. This might indicate that S4 had a better control on the movement of VR_{Obj} on the screen post operant conditioning. Also, during CT_{Final} task, S4 looked more towards the *Target* than that during CT_{First} task, possibly indicating the implication of operant conditioning.

In short, it can be said that the operant conditioning might have at least, to some extent, contributed to the variation in participant's task performance coupled with change in the scan path of S4 while interacting with the VR-based tasks.

7.5 Discussion and Conclusion

In this chapter, I investigated the gaze fixation pattern of post-stroke hemiplegic stroke participants while they interacted with VR-based weight-shifting tasks augmented with operant conditioning. I conducted experimental study with 5 hemiplegic participants and the preliminary results obtained from this study suggested that operant conditioning had implications on their gaze fixation pattern. Specifically, the change in gaze behavior can be seen as an increased fixation towards *Target* and decreased fixation toward VR_{Obj} in the CT_{Final} (post-operant conditioning) task compared to that in the CT_{First} (pre-operant conditioning) task. It was observed that while maneuvering VR_{Obj} from $Central_{Hold}$ to *Target* positions, their gaze scan paths became more controlled with increased viewing towards the *Target* position during CT_{Final} task compared to that in the CT_{First} task. These changes in gaze behavior in the hemiplegic stroke participants during CT_{Final} task were coupled with improved performance score from that in the CT_{First} task.

Though the preliminary results are promising, yet, the study had few limitations. The current study lacked sample power which restricted us to perform any statistical significance test and therefore it is difficult to generalize the findings. Also, in this study, I did not recruit any control group. Thus, care needs to be taken while drawing conclusions from the findings for the stroke population. Another limitation of the study was the weight of the in-house developed helmet assembly. None of the participants reported inconvenience in using the helmet assembly. One participant, S5 reported that he felt it bit heavy to use. To address these limitations, in future, I plan to extend this study to incorporate more participants with stroke and also age-matched healthy individuals. Also, I plan to reduce the weight of the helmet assembly by using strong but light-weight plastic materials for fabricating the telescopic channels and Display Screen Support Structure.

In the current study, I could understand the possible implications of operant conditioning based weight-shifting tasks on participant's gaze behavior. These findings need to be supported by extending this study with larger group of stroke and healthy participants. Also, the variations in participant's gaze behavior coupled with improved performance have been observed while the participants interacted with the VR world. Thus, questions still remain on the transferability of the oculomotor and balance skills learned from the simulated VR environment to real-life situations. However, this study can be a step towards improving our understanding into some of the subtle aspects associated with vision coupled with existing rehabilitation settings contributing to the improvement in balance for individuals with balance disorders.

Reference

[1] "The human balance system." [Online]. Available: http://vestibular.org/understanding-vestibular-disorder/human-balance-system. [Accessed: 13-Jan-2018].

[2] M. Shumway, Cook A Woollacott, *Motor Control: Theory and Practical Applications*. Philadelphia: Williams and Wilkins, 2001.

[3] R. J. Peterka and M. S. Benolken, "Role of somatosensory and vestibular cues in attenuating visually induced human postural sway," *Exp. Brain Res.*, vol. 105, no. 1, pp. 101–110, 1995.

[4] C. L. Horak, F. B., and Shupert, "Role of the vestibular system in postural control,"

Vestib. Rehabil., vol. 2, pp. 98–113, 1994.

[5] M. F. Land and M. Hayhoe, “In what ways do eye movements contribute to everyday activities?,” in *Vision Research*, 2001, vol. 41, no. 25–26, pp. 3559–3565.

[6] M. Land, N. Mennie, and J. Rusted, “The roles of vision and eye movements in the control of activities of daily living,” *Perception*, vol. 28, no. 11, pp. 1311–1328, 1999.

[7] M. M. Hayhoe, A. Shrivastava, R. Mruczek, and J. B. Pelz, “Visual memory and motor planning in a natural task,” *J. Vis.*, vol. 3, no. 1, p. 6, 2003.

[8] T. Foulsham, “Eye movements and their functions in everyday tasks,” *Eye (London, England)*, vol. 29, no. 2. pp. 196–199, 2015.

[9] U. Sailer, “Eye-Hand Coordination during Learning of a Novel Visuomotor Task,” *J. Neurosci.*, vol. 25, no. 39, pp. 8833–8842, 2005.

[10] K. Caeyenberghs et al., “Static and dynamic visuomotor task performance in children with acquired brain injury: Predictive control deficits under increased temporal pressure,” *J. Head Trauma Rehabil.*, vol. 24, no. 5, pp. 363–373, 2009.

[11] K. L. Gao, S. S. M. Ng, J. W. Y. Kwok, R. T. K. Chow, and W. W. N. Tsang, “Eye-hand coordination and its relationship with sensorimotor impairments in stroke survivors,” *J. Rehabil. Med.*, vol. 42, no. 4, pp. 368–373, 2010.

[12] J. R. Rizzo et al., “Eye control deficits coupled to hand control deficits: Eye-hand incoordination in chronic cerebral injury,” *Front. Neurol.*, vol. 8, no. JUL, 2017.

[13] M. a. Hollands, D. E. Marple-Horvat, S. Henkes, and a. K. Rowan, “Human Eye Movements During Visually Guided Stepping.,” *J. Mot. Behav.*, vol. 27, no. 2, pp. 155–163, 1995.

[14] M. A. Hollands and D. E. Marple-Horvat, “Coordination of Eye and Leg Movements

During Visually Guided Stepping," *J. Mot. Behav.*, vol. 33, no. 2, pp. 205–216, 2001.

[15] K. A. Crowdy, M. A. Hollands, I. T. Ferguson, and D. E. Marple-Horvat, "Evidence for interactive locomotor and oculomotor deficits cerebellar patients during visually guided stepping," *Exp. Brain Res.*, vol. 135, no. 4, pp. 437–454, 2000.

[16] M. Uchiyama and S. Demura, "Low visual acuity is associated with the decrease in postural sway.," *Tohoku J. Exp. Med.*, vol. 216, no. 3, pp. 277–85, 2008.

[17] A. Dutta, U. Lahiri, A. Das, M. A. Nitsche, and D. Guiraud, "Post-stroke balance rehabilitation under multi-level electrotherapy: A conceptual review," *Front. Neurosci.*, vol. 8, no. DEC, pp. 1–18, 2014.

[18] D. Kumar, A. Dutta, A. Das, and U. Lahiri, "SmartEye: Developing a Novel Eye Tracking System for Quantitative Assessment of Oculomotor Abnormalities," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 4320, no. c, pp. 1–1, 2016.

[19] Y. Lee, J. G. Her, Y. Choi, and H. Kim, "Effect of Ankle-foot Orthosis on Lower Limb Muscle Activities and Static Balance of Stroke Patients Authors' Names," *J. Phys. Ther. Sci.*, vol. 26, no. 2, pp. 179–182, 2014.

[20] "THEEYETRIBE." [Online]. Available: http://theeyetribe.com/theeyetribe.com/about/index.html. [Accessed: 01-May-2015].

[21] E. S. Dalmaijer, "Is the low-cost EyeTribe eye tracker any good for research?," *PeerJ Prepr.*, vol. 4, no. 606901, pp. 1–35, 2014.

[22] "Two Methods to Determine an Object's Center of Gravity." [Online]. Available: http://www.brighthubeducation.com/science-fair-projects/96255-center-of-gravity-experiment-balance-and-plumb-lines/#imgn_0. [Accessed: 16-Jan-2018].

[23] K. Berg, S. Wood-Dauphinee, J. I. Williams, and D. Gayton, "Measuring balance in the

elderly: Preliminary development of an instrument," *Physiother. Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[24] W. E. McIlroy and B. E. Maki, "Preferred placement of the feet during quiet stance: Development of a standardized foot placement for balance testing," *Clin. Biomech.*, vol. 12, no. 1, pp. 66–70, 1997.

[25] A. Mansfield, G. Mochizuki, E. L. Inness, and W. E. McIlroy, "Clinical correlates of between-limb synchronization of standing balance control and falls during inpatient stroke rehabilitation," *Neurorehabil. Neural Repair*, vol. 26, no. 6, pp. 627–635, 2012.

[26] K. O. Berg, B. E. Maki, J. I. Williams, P. J. Holliday, and S. L. Wood-Dauphinee, "Clinical and laboratory measures of postural balance in an elderly population.," *Arch. Phys. Med. Rehabil.*, vol. 73, no. 11, pp. 1073–1080, 1992.

[27] J. Babič, E. Oztop, and M. Kawato, "Human motor adaptation in whole body motion," *Sci. Rep.*, vol. 6, no. 1, p. 32868, 2016.

[28] K. Thompson, F. R. Pomerantz, and J. R. Wolpaw, "Operant conditioning of a spinal reflex can improve locomotion after spinal cord injury in humans.," *J. Neurosci.*, vol. 33, no. 6, pp. 2365–75, 2013.

[29] A. Dutta, D. Kumar, U. Lahiri, A. Das, and M. V Padma, "Post-stroke Engagement-sensitive Balance Rehabilitation Under An Adap- tive Multi-level Electrotherapy : Clinical Hypothesis and Computational Framework," pp. 68–80, 2014.

CHAPTER 8

CONTRIBUTIONS AND CONCLUSION

8.1 Contributions

Neurological disorder such as stroke is a medical emergency situation [1] that can be addressed at least to some extent if the treatment is given within a short time window post the onset of stroke symptoms [2]. Therefore, early screening of such cases is critical. The existing state-of-the-art techniques for stroke screening use high-end medical imaging devices such as Magnetic Resonance Imaging (MRI) [3], Computed Tomography (CT) [4] and others. Unfortunately, factors such as socio-economic status, limited availability of such medical imaging facilities in selected urban health centers and others are some of the major deterrents to restrict their accessibility in remote locations, particularly relevant to countries with developing economy such as India. As an alternative, conventional observation-based techniques for stroke screening are often followed. In the conventional techniques, the expert clinician looks out for the possible symptoms of stroke in an individual [5]. For example, expert clinicians often use Opto-kinetic drum [6] to observe the patient's eye movement during visual examination of an individual's oculomotor health. Clinicians also use a three-component bedside oculomotor examination, namely, Head Impulse test, Nystagmus and Test of Skew deviation (HINTS) that has been shown to diagnose stroke with more accuracy than diffusion-weighted magnetic MRI [7]. Such techniques though useful, yet, suffer from lack of adequately trained clinicians and subjectivity in deciphering the observed symptoms of stroke condition [8], [9]. To address these issues in stroke screening, in this research I have developed a screening device (SmartEye) that uses an individual's oculomotor signature as a potential biomarker of neurological disorder. While harnessing the rapid progress in computing technology and its widespread accessibility, the SmartEye system can offer a cost-effective, easy-to-use, easily-accessible and quantitative screening tool for probable neurological disorder such as stroke [10] Thus, this can bring in a paradigm shift in both the urban and rural healthcare.

Often, post-stroke vision problems accompany balance deficits. This is because, the sensory inputs on one's posture and the surrounding visual field, necessary to maintain balance [11], gets affected due to accompanying oculomotor deficits. The ensuing balance deficit often leads to falls in post-stroke patients. Literature indicates that about 73% of the stroke-survivors report falls in the first year of post-stroke condition [12]. The incidence of falls can have adverse effects on patients' mobility making them dependent on their caregivers. For addressing such balance-related issues, clinicians often deploy conventional balance rehabilitation techniques. Conventional balance rehabilitation involves repetitive movement exercises (devoid of variations) that often turn out to be monotonous for the patients. As a result, stroke patients often lose their interest to participate in balance rehabilitation exercises [13], [14]. Also, conventional techniques often require one-to-one services that face limitations such as scarcity of adequately trained physiotherapist [15], high cost of availing specialized healthcare settings [16] and others. To overcome these limitations, researchers have been exploring the use of technology-assisted balance rehabilitation such as robot-assisted, computer-based and Virtual Reality (VR) based systems [14], [17], [18]. Among the alternate available techniques, VR-based balance systems have been widely explored by researchers. This is because, VR offers the flexibility of design, controllability, and individualized approach to balance rehabilitation exercises [19]. Again VR offers a computer simulated environment that can be augmented with peripheral devices [20]. In recent years, researchers have used peripheral devices capable of measuring an individual's inertial characteristic of body segments related to balance (such as Center of Pressure (CoP) [21] and

Center of Mass (CoM) [22]) for designing balance rehabilitation systems. However, most of the existing VR-based balance rehabilitation systems have used off-the-shelf games for the balance rehabilitation tasks. These games are mostly designed with an entertainment perspective and have been used by young and healthy individuals. Also, the currently existing balance rehabilitation systems are not individualized. Specifically, these are not adaptive to one's performance quantifying one’s balance which is a critical requirement for rehabilitation to be effective. In my research, I have developed VR-based balance training systems augmented with Balance Board (WiiBB) and Kinect that can quantify an individual's residual balance and can accordingly plan out balance training tasks in an individualized manner.

Given the facts that patients with neurological disorder such as stroke exhibit deficits in oculomotor signatures [23] and also suffer from balance disorders [24], it is critical to understand their gaze behavior during balance training session. Thus, I augmented the VR-based balance training platfrom with an in-house built head-mounted gaze sensitive display unit to investigate the gaze behavior of post-stroke participants while they performed weight-shifting as required by the VR-based balance training tasks.

To summarize, the contribution of the dissertation is mainly categorized into (a) design, development and evaluation of a user-friendly, easily-accessible, and cost-effective gaze-based screening device (SmartEye) for probable neurological disorder (for addressing Objective 1 mentioned in Chapter 1) (b) develop and study the implication of intelligent adaptive VR-based balance training systems on a stroke patient’s balance (addressing Objective 2 mentioned in Chapter 1) and (c) investigate the gaze behavior of hemiplegic post-stroke participants during VR-based balance training (addressing Objective 3 mentioned in Chapter 1).

Design, development and evaluation of a user-friendly, easily-accessible, and cost-effective gaze-based screening device (SmartEye) for screening of probable neurological disorder

This involves developing a gaze-based screening device (SmartEye) with an aim to map the abnormality in gaze-related indices to one's probable neurological dysfunction. The SmartEye system (discussed in Chapter 3) consisted of a cost-effective Eye Tracker that monitored participant's gaze pattern while he/she followed the static and dynamic visual stimuli presented on a Task Computer screen. Subsequently, participant's gaze data corresponding to the visual task was analyzed to quantitatively measure participant’s eye fixation, smooth pursuit and blinking activities. A Usability study of SmartEye system with stroke survivors and age-matched healthy participants was carried out.

The result of this study indicates that the SmartEye system was able to identify some gaze-related indices, such as deviation in eye fixation, smooth pursuit length and blink per minute of the stroke survivors that were different from the normative pattern demonstrated by their age-matched healthy counterparts. Based on the participants’ feedback, I can infer that the SmartEye has a potential to serve as a user-friendly and easily-accessible oculomotor monitoring platform. Also, such a platform can serve as a complementary device in the hands of clinicians as far as screening of neurological disorders is concerned. In my present research, I have applied SmartEye for screening post-stroke cases. I believe that SmartEye can be used to assess oculomotor abnormalities in patients with various neurological disorders, of course with varying manifestations based on the type and intensity of neurological disorder.

Develop and study the Implication of Intelligent Adaptive VR-based Balance Training Systems on an individual's Balance

This involved developing VR-based balance training systems that offered a variety of VR-based balance tasks set in a variety of VR environments. These systems were augmented with peripheral devices such as force platform (Nintendo Wii Balance Board (WiiBB)) and motion capture device (Microsoft Kinect) to interact with the virtual world. The developed VR-based balance training systems were capable of understanding the patient's residual balance ability and accordingly offered the balance training tasks of varying challenge levels while adapting to the weight-shifting ability of the post-stroke patient in an individualized manner. Also, the developed systems had a mechanism to encourage the participants to follow Ankle strategy while carrying out weight-shifting task that is considered as an important postural strategy for standing balance. I have explored the applicability of different VR-based training systems through three studies.

In the first study, I have developed the VR-based balance training (VBaT) system coupled with a single WiiBB (discussed in Chapter 4). The VR-based tasks required one to shift his/her body weight in different directions while standing on the WiiBB and adhering to the Ankle strategy. The position of a virtual object in the VR environment was controlled with excursion of the Center of Pressure (CoP) measured by the WiiBB. The VBaT system offered tasks of different difficulty levels to the participants and the task switching was adaptive to participant's weight-shifting capability during the task execution. Result of this study showed that the VBaT system was able to improve the overall average task performance score over the course of the training in post-stroke hemiplegic participants. Also, the improvement in performance was coupled with increased speed of task execution.

In the second study, I developed a VR-based Center of Mass (CoM)-assisted Balance Training (Virtual CoMBaT) system (discussed in Chapter 5). The Virtual CoMBaT system was augmented with a Kinect device that was used to estimate a stroke participant's CoM position. Thus, in contrast to the VBaT system, instead of the CoP, I used an individual's CoM in Virtual CoMBaT system. This was because, one's balance can be quantified in terms of CoP as well as CoM-related measures [25]. Also, the CoM-based balance task can be more challenging than CoP-based task since, displacing of an individual's CoM can be more challenging than displacing CoP [26]. In this study, a participant was exposed to a VR-based task environment with tasks belonging to varying challenge levels. Also, the participant was required to shift his/her weight in different directions by standing on the ground while following the Ankle strategy. The position of a virtual object in the VR environment was controlled with dynamic CoM that was estimated using Kinect device. Results obtained from this study showed that the Virtual CoMBaT system (i) can quantify participant's direction-specific residual balance capability and (ii) was able to improve overall average task performance over the course of the training, thereby improving the weight-shifting capability of the participants.

In the next study, I designed a VR-based balance training system that can quantify the contribution of each leg of a participant towards his/her overall balance. Additionally, the system applied operant conditioning with an aim to encourage a hemiplegic stroke patient to use his/her Affected leg as equally as possible as that of the Unaffected leg. To achieve this, I have interfaced the VR-based Balance Training platform with two WiiBB (V2BaT) and augmented it with a paradigm that varied the distributed weight based on the principle of operant conditioning [27] (discussed in Chapter 6). In this study, a participant was required to shift his/her weight in the instructed direction while standing with one leg on each of the two WiiBB. During the weight-shifting task, the position of a virtual object in the VR environment was proportionally controlled with CoP excursion (measured by the two WiiBB, one for each leg). Results of a

Usability study conducted with V2BaT system by hemiplegic participants indicate that the V2BaT system was able to (i) quantify the contribution of each leg towards a participant's weight-shifting capability, (ii) improve a participant’s balance (weight-shifting ability) in the course of the training by increasing the usage of the Affected leg along with the Unaffected leg.

Investigate the Gaze Behavior of Hemiplegic Post-stroke participants during VR-based Balance Training augmented with Operant Conditioning

While addressing the third objective, I wanted to understand the gaze behavior of stroke participants when they performed goal-directed balance tasks. For this, I added a head-mounted gaze-sensitive stimulus display monitor (*Hemo-GaSiD*) unit to the V2BaT system augmented with a paradigm that varied the distributed weight (discussed in Chapter 7). Similar to the previous study (Chapter 7), here the participant was asked to stand on two WiiBB, with one leg on each WiiBB. Also, the participant was expected to maneuver a virtual object from a start location to a Target location through CoP excursion while using the operant conditioning. Unlike the previous study (Chapter 7), here the VR-based task was presented on the head-mounted stimulus display screen that was coupled with an Eye Tracker. While the participants performed the VR-based tasks, the Eye Tracker recorded their gaze data. The results of this study showed that after the participant was exposed to the VR-based balance task augmented with operant conditioning, the participant demonstrated an improvement in performance score coupled with change in gaze behavior. Specifically, the participant demonstrated increased fixation towards the static target location and decreased fixation towards the dynamic virtual object (maneuvered by the CoP excursion) in the VR-based balance task.

8.2 Conclusion

The research presented in this dissertation shows that the gaze-based oculomotor monitoring device (SmartEye) has a potential to serve as cost-effective, easy-to-use screening device for probable neurological disorder. The Usability study of SmartEye with post-stroke participants revealed that gaze-related indices can serve as potential biomarkers of neurological disorder, such as stroke. I believe that if such a screening device can be made available in the different healthcare centers, then it can serve as a potent complementary tool in the hands of clinicians to quantitatively estimate one’s probable neurological disorder through examination of oculomotor signatures.

Also, the intelligent adaptive VR-based balance rehabilitation systems discussed in this dissertation have potential to contribute to improvement in an individual's weight-shifting ability in an individualized manner. Though, the systems have not been developed as full-fledged rehabilitation systems, yet, the results indicate the potential of the systems to be able to offer rehabilitation platforms that can (i) understand an individual’s residual weight-shifting capability, (ii) offer variety of VR-based balance tasks of different challenge levels while being adaptive to the individual’s performance in a task, (iii) encourage the participants to follow Ankle strategy while performing the standing balance tasks so as to improve the quality of weight-shifting and (iv) provide quantitative estimates on an individual’s improvement in balance without the need for specialized knowledge for it to be operated, thereby enabling it to be used in home-based, community-based and hospital-based settings.

Additionally, the investigatory study on the gaze behavior of stroke participants during goal-directed balance tasks helped us to understand the possible implications of operant conditioning based weight-shifting tasks on stroke participant's gaze behavior. Specifically, I observed that post operant conditioning, the participant's fixation increased towards the target location and decreased towards the moving virtual object that was maneuvered by the dynamic CoP caused due to shifting weight. I believe that such information on a stroke participant's gaze behavior during a goal-directed balance task can lead to gaining deeper insights into the role of vision in improving the existing balance rehabilitation techniques for individuals with balance disorders. However, this study can be a step towards improving our understanding into some of the subtle aspects associated with vision coupled with existing rehabilitation settings contributing to the improvement in balance for individuals with balance disorders.

Though the technologies developed in my research have been tested through preliminary Usability studies, yet, there are bigger questions that need to be addressed. Specifically, though the results on the participants' improvement in the task performance in most of the Usability studies conducted by me were promising, yet these observations were mostly made within a predefined controlled setting with a limited participant pool. Also, in my research, the stroke participants were exposed to each of the different balance training systems (VBaT, Virtual CoMBaT and V2BaT) for one day. The outcome from the single session of balance training with stroke patients showed the potential of my systems to be used as an individualized balance rehabilitation platform with positive outcomes. To see a significant difference in a patient's balance capability, one might need a longer exposure (longitudinal study) to these systems. Even with prolonged exposure, concerns regarding the training time, pace and duration of exposure to the balance training that can be suitable for a stroke patient to gain permanent improvement in the functional outcome due to these rehabilitative intervention will still remain. This is particularly true in the context of formal therapy being often stopped when patients show no qualitative gains after a few weeks of treatment. Such a plateau in the recovery of an individual, however, does not necessarily imply the inability of the individual's capacity to have gains in functional improvement or in learning a new task. In case of balance rehabilitation, the opportunity to achieve maximal improvement can be assessed based on the quantitative measures of balance such as BBS score. Such a score can be measured pre and post the balance rehabilitation intervention. Also, functional neuroimaging may help to guide decisions about the type and duration of intervention by providing insight into the maximal cortical reorganization that can be achieved with a particular rehabilitation therapy over time. Additionally, the longevity of the balance-related skills learnt by an individual exposed to the balance rehabilitation tasks mapped to the maintenance of cortical reorganization needs to be addressed. However, exploring such options require further investigation. The ultimate aim will be to bring in functional improvement in patients outside the controlled settings to real-life situations, thereby helping these individuals to lead a quality life.

8.3 Future Work

In the research presented in this dissertation, the SmartEye system was shown as a gaze-sensitive platform that can be used to screen individuals with stroke based on oculomotor signatures quantified in terms of deviation in gaze fixation, smooth pursuit eye movement, and blinking. Also, I carried out a Usability study with SmartEye. However, due to limited sampling rate (60 Hz) offered by the Eye Tracker (from Eye Tribe) used in the SmartEye, I was not able to analyze the rapid eye movement such as saccadic activity. Picking up additional biomarkers such as saccades is critical in building up a robust screening platform. This is because, literature indicates that vast areas of brain network (that can be vulnerable to neurological disorder) is involved in controlling saccadic eye movement [28]. In future, I plan to augment the SmartEye system with an Eye Tracker having higher sampling rate (approximately 250 Hz and higher) to investigate the possibility of using saccadic eye movements as an additional potential biomarker for screening probable neurological disorder. Also, in future, I plan to implement the SmartEye system on other devices, such as Smartphone and tablets while using their inbuilt cameras as Eye Trackers and providing the visual task on the Smartphone/tablet screen. This will ensure increased accessibility of SmartEye system among users, even in remote locations.

Additionally, the research aimed to develop Center of Pressure (CoP) and Center of Mass (CoM)-assisted VR-based balance training tasks. While the participants interacted with the VR-based balance tasks, the systems quantified the residual balance capability of the participants and adaptively offered tasks of varying difficulty levels based on individualized task performance. Also, the systems offered them visual and auditory feedback. However, none of the systems offered an external stimulation such as Neuro-muscular Electrical Stimulation (NMES). This might be beneficial for the post-stroke patients since, such patients often suffer from muscle weakness and increased muscle spasticity that dampens the muscle activation necessary to perform a given task. Knutson et al. [29] have reported that electrical therapy can serve to help stroke patient to perform a task. Thus, in future, I plan to monitor the muscle activation by monitoring the electromyogram (EMG) signals from the ankle muscles such as from soleus, gastrocnemius medialis, tibialis anterior, peroneus longus along with the performance in a task, while one performs VR-based tasks. Subsequently, based on an individual's performance and muscle activation (measured from EMG signal), decisions on application of NMES to the participant's ankle muscle can be taken to facilitate weight-shifting.

Finally, in the present research, in each Usability study, the stroke participants were exposed to the VR-based balance training systems for only one session. Though the results of each of the studies are promising as far as the improvement in stroke participant's balance is concerned, yet, a limited exposure to the system may not be sufficient to speak on the rehabilitation efficacy of the system. Therefore, in future, I plan to augment the VR-based balance rehabilitation system with NMES therapy and conduct a longitudinal study with post-stroke participants. This will be coupled with clinical assessment of a patient's balance ability by using measures such as Berg Balance Scale (BBS) score prior to and post the study. This will help me get deeper insights into the clinical efficacy of the technology-assisted systems as far as improvement in an individual's balance is concerned.

Reference

[1] C. Lott, H. J. Hennes, and W. Dick, "Stroke--a medical emergency.," *J. Accid. Emerg. Med.*, vol. 16, no. 1, pp. 2–7, 1999.

[2] R. Advani, H. Naess, and M. W. Kurz, "The golden hour of acute ischemic stroke," *Scand. J. Trauma. Resusc. Emerg. Med.*, vol. 25, no. 1, 2017.

[3] J. a Chalela et al., "Magnetic resonance imaging and computed tomography in emergency assessment of patients with suspected acute stroke: a prospective comparison," *Lancet*, vol. 369, no. 9558, pp. 293–298, 2007.

[4] P. A. Barber, A. M. Demchuk, J. Zhang, and A. M. Buchan, "Validity and reliability of a quantitative computed tomography score in predicting outcome of hyperacute stroke before thrombolytic therapy. ASPECTS Study Group. Alberta Stroke Programme Early CT Score," *Lancet*, vol. 355, no. 9216, pp. 1670–1674, 2000.

[5] H. P. Adams et al., "Guidelines for the early management of adults with ischemic stroke," *Circulation*, vol. 115, no. 20, pp. e478-534, 2007.

[6] T. Brandt, M. Dieterich, and M. Strupp, *Vertigo and dizziness: Common complaints, second edition*. 2013.

[7] J. C. Kattah, A. V. Talkad, D. Z. Wang, Y. H. Hsieh, and D. E. Newman-Toker, "HINTS to diagnose stroke in the acute vestibular syndrome: Three-step bedside oculomotor examination more sensitive than early MRI diffusion-weighted imaging," *Stroke*, vol. 40, no. 11, pp. 3504–3510, 2009.

[8] J. Joubert et al., "Stroke in rural areas and small communities," *Stroke*, vol. 39, no. 6. pp. 1920–1928, 2008.

[9] J. D. Pandian et al., "Public awareness of warning symptoms, risk factors, and treatment of stroke in Northwest India," *Stroke*, vol. 36, no. 3, pp. 644–648, 2005.

[10] D. Kumar, A. Dutta, A. Das, and U. Lahiri, "SmartEye: Developing a Novel Eye Tracking System for Quantitative Assessment of Oculomotor Abnormalities," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 4320, no. c, pp. 1–1, 2016.

[11] "The human balance system." [Online]. Available: http://vestibular.org/understanding-vestibular-disorder/human-balance-system. [Accessed: 13-Jan-2018].

[12] G. S. A. F. Verheyden et al., "Interventions for preventing falls in people after stroke.," *Cochrane database Syst. Rev.*, no. 5, p. CD008728, 2013.

[13] B. S. Rajaratnam et al., "Does the Inclusion of Virtual Reality Games within Conventional Rehabilitation Enhance Balance Retraining after a Recent Episode of Stroke?," *Rehabil. Res. Pract.*, vol. 2013, pp. 1–6, 2013.

[14] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer, "Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: A pilot randomized clinical trial in patients with acquired brain injury," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, 2011.

[15] G. I. Yatar and S. A. Yildirim, "Wii Fit balance training or progressive balance training in patients with chronic stroke: a randomised controlled trial," *J. Phys. Ther. Sci.*, vol. 27,

no. 4, pp. 1145–1151, 2015.

[16] H. Bateni, “Changes in balance in older adults based on use of physical therapy vs the Wii Fit gaming system: A preliminary study,” *Physiother. (United Kingdom)*, vol. 98, no. 3, pp. 211–216, 2012.

[17] S. E. Fasoli, H. I. Krebs, and N. Hogan, “Robotic Technology and Stroke Rehabilitation: Translating Research into Practice,” *Top. Stroke Rehabil.*, vol. 11, no. 4, pp. 11–19, 2004.

[18] G. Saposnik and M. Levin, “Virtual reality in stroke rehabilitation: A meta-analysis and implications for clinicians,” *Stroke*, vol. 42, no. 5, pp. 1380–1386, 2011.

[19] A. Darekar, B. J. McFadyen, A. Lamontagne, and J. Fung, “Efficacy of virtual reality-based intervention on balance and mobility disorders post-stroke: A scoping review,” *J. Neuroeng. Rehabil.*, vol. 12, no. 1, 2015.

[20] R. V. Kenyon, J. Leigh, and E. A. Keshner, “Considerations for the future development of virtual technology as a rehabilitation tool,” *J. Neuroeng. Rehabil.*, vol. 1, 2004.

[21] B. Lange, S. Flynn, R. Proffitt, C.-Y. Chang, and A. “Skip” Rizzo, “Development of an Interactive Game-Based Rehabilitation Tool for Dynamic Balance Training,” *Top. Stroke Rehabil.*, vol. 17, no. 5, pp. 345–352, 2010.

[22] B. Lange, C. Y. Chang, E. Suma, B. Newman, A. S. Rizzo, and M. Bolas, “Development and evaluation of low cost game-based balance rehabilitation tool using the Microsoft Kinect sensor,” in *IEEE Engineering in Medicine and Biology Conference*, 2011, vol. 2011, pp. 1831–1834.

[23] J. Leigh and D. S. Zee, *The neurology of eye movements*, vol. 53, no. 9. 2013.

[24] K. Strong, C. Mathers, and R. Bonita, “Preventing stroke: saving lives around the world,” *Lancet Neurology*, vol. 6, no. 2. pp. 182–187, 2007.

[25] D. A. Winter, “Human balance and posture control during standing and walking,” *Gait Posture*, vol. 3, no. 4, pp. 193–214, 1995.

[26] L. E. Cofré Lizama, M. Pijnappels, N. P. Reeves, S. M. Verschueren, and J. H. van Dieën, “Centre of pressure or centre of mass feedback in mediolateral balance assessment,” *J. Biomech.*, vol. 48, no. 3, pp. 539–543, 2015.

[27] S. A. McLeod, “Skinner - Operant conditioning,” *Simply Psychol.*, vol. 1, no. 1, p. 2, 2007.

[28] B. Gaymard, C. J. Ploner, S. Rivaud, A. I. Vermersch, and C. Pierrot-Deseilligny, “Cortical control of saccades,” in *Experimental Brain Research*, 1998, vol. 123, no. 1–2, pp. 159–163.

[29] J. S. Knutson, M. J. Fu, L. R. Sheffler, and J. Chae, “Neuromuscular Electrical Stimulation for Motor Restoration in Hemiplegia,” *Physical Medicine and Rehabilitation Clinics of North America*, vol. 26, no. 4. pp. 729–745, 2015.