A USER-SPECIFIC APPROACH TO DEVELOP AN ADAPTIVE VR EXERGAME FOR INDIVIDUALS WITH SCI

by

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ABSTRACT

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Patients with Spinal Cord Injury (SCI) have limited time with supervised therapy in rehabilitation hospitals. This makes it imperative for them to continue regular therapy at home so they can maximize motor recovery especially for performing Activities of Daily Living (ADL). However, physical therapy can be tedious and frustrating leading to a lack of motivation. A novel upper extremity movement measurement tool was developed using a commercial VR system to rapidly and objectively measure an individual's range of motion, velocity of movement on an individual gesture basis, and frequency of movements in a three-dimensional space. Further, an exergame with varied and customizable gameplay parameters was developed. Through the analysis of participant interaction with the exergame, we identified gameplay parameters that can be adjusted to affect the player's perceived and physiological effort. We observed that VR has a significant motivational effect on range of motion of upper limbs in individuals with tetraplegia. The motion data and kernel density estimation is used to determine areas of comfort. Moreover, the system allowed calculation of joint torques through inverse kinematics and dynamics to serve as an analysis tool to gauge muscular effort. The system can provide an improved rehabilitation experience for persons with tetraplegia in home settings while allowing oversight by clinical therapists through analysis of mixed reality videos or it could be used as a supplement or alternative to conventional therapy.

1. INTRODUCTION

Research Problem

Each year there are 12,000 to 20,000 new cases of spinal cord injury (SCI) in the United States alone, and 238,000 to 332,000 individuals living with SCI [1]. The global incidence rate of SCI
1.1 ranged anywhere from 8.0 to 246.0 cases per million individuals per year. The prevalence ranged from 236.0 to 1298.0 per million individuals around the world. [2]–[4]

The 2014 SCI Model System reports a decrease in length of stays for rehabilitation over the last 40 years for patients with SCIs from a median stay of 98 days in 1973- 1979 to 36 days in 2010- 2014[5], a decrease of 63%. This implies that individuals with SCI spend lesser time under supervised therapy making it imperative to continue regular therapy when they return home [6].

Studies have shown that exercise regimens post SCI allow for improvements in functional capacity, bone density in upper limbs, endurance, muscle strength and psychological well-being[7]–[9]. Improving upper limb motor function is also crucial for improving functional independence after an SCI[10].

However, there exist several perceived barriers to individuals with SCI performing the prescribed therapy regularly[11]. These include physical barriers such as equipment, availability of resources as well as psychological or social barriers such as perceptions or attitudes towards disability, motivation and fear of injury. Moreover, it was identified that the desire to exercise did not necessarily match behavior and various studies have found that lack of motivation was a ubiquitous factor in reduced exercise[12], [13].

However, current forays into exergames using VR and other gaming platforms are designed for individuals who are able bodied and do not account for the limited manual dexterity of individuals with tetraplegia[14]. This lack of hand function of tetraplegics to hold and depress various buttons on a typical VR game controller restricts the type of exergames that can be played.

There is currently no accessible VR exergame that could be played by individuals with SCI that is engaging and doubles as an exercise that allows for therapy to be fun and thus reduce the barriers to using it regularly. We have developed a baseline tool and an exergame that uses a commercial off-the-shelf head mounted VR gaming system, the HTC Vive®. A method to calculate static and dynamic joint forces as a measure of muscular effort was also developed. Surveys were conducted at the end of the games to understand the perceived fatigue levels and feedback about gameplay mechanics. The baseline tool, coupled with kernel density estimation and static joint force calculation, was also used to identify ergonomic areas in the workspace of an individual with SCI.

Specific Aims

1.2 The accessible exergame was developed with the following specific aims in consideration:

- **Specific Aim 1:** Develop a VR-Based exergame that is both engaging and customizable
- **Specific Aim 2:** VR baseline tool and exergame can track and measure various physiological features and extract individualized parameters
- **Specific Aim 3:** User effort can be altered by changing gameplay parameters

1.3

Research Questions

- **Research Question 1:** Would a VR-based game be more engaging than performing the same gestures without VR?
- **Research Question 2:** Can an exergame track individual physiological movements that are clinically relevant?
- **Research Question 3:** How do changing gameplay parameters affect perceived effort (shoulder torque), range of motion, velocity of gestures and frequency of movements?
- **Research Question 4:** How can gameplay parameters be engineered to elicit a desired level of physical activity?

1.4 Background

1.4.1 Epidemiology of SCI

Each year there are 12,000 to 20,000 new cases of SCI in the United States alone, and 238,000 to 332,000 individuals living with SCI[1]. The global incidence rate of SCI ranged anywhere from 8 to 246 cases per million individuals per year. The prevalence ranged from 236 to 1298 per million individuals around the world. [2]–[4]

The economic consequences and impacts on the quality of life of living with a SCI are substantial. For an individual with tetraplegia, estimated yearly direct costs such as healthcare and living expenses, and indirect costs including loss of wages, fringe benefits and productivity are US$1.1 million and US$0.8 million per individual, respectively. SCIs have a considerable impact on the lives of the individuals who are injured as well as their families. Despite the low incidence of SCIs compared to health conditions such as heart disease and stroke, people with SCI are most likely to live with paralysis and other consequences of SCI for longer periods of time [15]. The most frequent age of injury is 19 and with a near normal life expectancy of individuals with chronic SCI due to advances in medicine in the past few decades [15].

1.4.2 Current methods and measures of rehabilitation

Inpatient rehabilitation involves physical and occupational therapists working with patients on an individual basis. The physical therapist first performs a functional evaluation to identify the motor functions which have been affected. They then set primary goals and determine exercises .and movements which are important to allow the patient to maximize recovery of motor function. The therapist periodically re-evaluates the patient to determine new directions, if needed. Primary goals set for patients include maximizing functional independence determined on a functional independence measure (FIM) scale.

Current clinical tests performed during rehabilitation to determine functional ability include the FIM, manual muscle testing (MMT), Range of Motion Scale (ROMS), and the Modified Ashworth Scale. All these tools require trained clinical therapists to obtain a good inter-rater reliability for these subjective tests [16].

1.4.2.1 Functional Independence Measure

The functional independence measure (FIM) is a scale that is used to record the severity of disability in individuals. There are eighteen FIM objects that are recorded during a measurement. These are divided into thirteen motor function and five cognitive function disability measures. FIM is usually performed during admission and discharge of a patient to assess the improvement in functional independence. [17]

1.4.2.2 MMT

MMT is used to evaluate muscles and their ability to generate forces. The traditional grading scale ranges from zero to five. A zero represents no contractions felt in the muscle and a five represents the individual's ability to hold the test position against strong pressure exerted by the therapist on different muscle groups. Scores below three are gravity eliminated, i.e. the individual does not need to fight gravity to make the movement and scores above three are against gravity[18].

1.4.2.3 ROMS

ROMS is used to evaluate the range of motion in degrees for each degree of freedom available at the joint under test. A goniometer is used to measure the maximum and minimum angles that an individual is able to achieve for a given joint[19]. While some studies validate the interrater reliability and validity of the ROMS, there some researchers who suggest the ROMS may not be the best tool to characterize joint function[17].

1.4.2.4 Ashworth Scale

The Ashworth scale is one of the most commonly used measurements of muscle spasticity and tone [20]. Clinicians are instructed to test various muscles. The muscle tone is assessed through feeling of the muscle and moving it through its range of motion and assessing muscle contraction. Therapists then rate it on a scale ranging from 0-4 where 0 representing no increase in muscle tone and 4 representing maximum muscle tone. There are some clinicians who strongly believe in the use of the Ashworth scale with some questioning the reliability and validity of the scale [21].

1.4.3 Limitations of current methods of rehabilitation

Currently there are no tools available which can objectively quantify clinical measures of rehabilitation. The measures are also limited by the raters' ability to accurately determine range of motion, muscle tone and various other performance metrics of a patient. Moreover, these methods are time consuming, labor and resource intensive and often require a lot of dependence on patient compliance [22], [23]. Lastly, these methods are often performed under the guidance of a clinician and can be repetitive and boring.

1.4.4 Virtual Reality

VR is a form of human-computer interfacing which allows users to interact within a multisensory simulated environment and receive "real-time" feedback on performance. It allows users to interact in a more natural manner relative to what is currently afforded by standard mouse and keyboard input devices [24]. Commercially available VR consoles have allowed the development of exergaming which has emerged as a new form of intervention for upper-limb rehabilitation. Exergames have increasingly been used to assist elderly individuals and persons with disabilities with regular exercise through the use of computer interfaced input devices such as Nintendo Wii™ (Nintendo, Kyoto, Japan) and Leap Motion (Leap Motion Inc, California) [25], [26]. Rehabilitation using VR often includes systems such as Computer Assisted Rehabilitation Environment (CAREN), the Interactive Rehabilitation Exercise System (IREX) Mieron, a commercial virtual reality neurotherapy system [27], and Virtual Test Track Environment (VIRTTEX). The CAREN system comprises a fully immersive VR environment with instrumented treadmills and multi-sensory real time feedback. The VIRTTEX is a driving simulator developed by Ford. All these systems are extremely expensive ($13,000- 1.5Million) [28].

It is important to note that the definition of VR is rather broadly defined in the literature. A majority of studies that mention a VR environment talk extensively about video capture VR or a flat screen VR [29]–[33]. Video capture VR systems take a real video of a user in a green screen environment and utilize chroma key compositing techniques to insert the user's video in a virtual environment. Chroma keying technique allows the use of a green screen to separate the background from the individual. The user then sees a video of themselves in a flat or a curved screen superimposed in a virtual environment (Figure 1). Various tracking techniques have been used by different systems to allow the user to interact with the presented virtual environment. These VR systems have a flat 2D screen. These are also extremely large setups that are generally not portable in nature. Their expensive nature also makes them impossible for use as part of a home-based rehabilitation system.

Figure 1: IREX Video Capture VR system.[34]

1.4.4.1 The HTC Vive®

The HTC Vive® uses a head mounted display to immerse the individual instead of a flat screen that has been used in previous systems such as the IREX. This head mounted display has two screens, one for each eye. The videos displayed in the two screens are offset by the inter-ocular distance to create a sense of depth in the virtual environment [35]. This allows for reaching tasks where distances in the virtual world correspond to the distances in the real world. Therefore the gestures and tasks performed in the virtual world are directly translatable to the virtual world. In a flat screen set up such as the IREX with an inherently 2D screen allows no depth perception. Completion of this visual feedback loop is vital for effective rehabilitation [36].

1.4.4.2 Virtual Reality in Rehabilitation

These VR-based exergames have emerged as a tool for rehabilitation for diseases such as stroke, traumatic brain injury, spinal cord injury, cerebral palsy, Parkinson's disease and other developmental issues. The use of VR in rehabilitation is attributed to some unique aspects of the technology such as the ability for an immersive experiential learning, and active learning in a motivating, challenging but safe environment [37], [38]. The VR system encourages the repetition of active movement, making it ideally suited as a tool for motor rehabilitation. VR is also emerging as a useful tool to facilitate rehabilitation with the potential to support home-based exercise programs[39]. The use of VR in rehabilitation allows the development of exercise games (exergames). Motivation through serious gameplay (exergaming) has been shown to raise patients' interest, improve their adherence in rehabilitation at home[40][29].

In addition to helping patients, VR's ability to automatically deliver stimulus at known timepoints allows clinicians and therapists to focus on the patients' performance and observe whether they are using effective strategies[41][42]. Clinicians can use VR to allow patients to achieve a variety of objectives through the varying of task complexity as well as type and amount of feedback[30].

However, one of the issues with the use of commercial affordable commercial exergaming consoles such as the Wii Fit and Microsoft Kinect™ is that these applications have not been designed as medical devices, with a primary focus as a rehabilitation tool or for use with individuals with limited motor function. This makes it difficult for them to be used as therapy tools with a high usability. These tools also cannot be easily modified to be appropriate for different levels of impairments[28], [41].

This section will discuss the role of VR in the rehabilitation of individuals with stroke, Parkinson's disease and traumatic brain injury presented in the literature. VR in the rehabilitation of persons with spinal cord injury, which is the focus of this research, has been investigated less.

1.4.4.2.1 Stroke

Rehabilitation is a crucial component of improving motor function in stroke survivors. Current paradigms for rehabilitation are often time intensive and difficult to implement and follow through for patients and clinicians. Stroke rehabilitation is rapidly evolving to incorporate novel techniques such as VR systems which have shown improvements in motor impairment, activities and social participation[22].

IREX VR and VMall virtual environments have been developed by researchers to determine the effect of VR on the cortical reorganization and locomotor recovery [43][44]. Researchers have also used the Wii Fit to develop games which could work best for the rehabilitation of motor functionality[45].

The VR training demonstrated neuroplasticity in the patients with stroke through fMRI studies. They also identified that there were significant improvements in motor functions in the VR exercise group compared to a control group which did not receive any treatment. VR systems have also shown an improvement in traditional clinical measures such as the Box and Block test and Functional Independence Measures (FIM) and strength[46] [47] in patients post-stroke.

The use of the VMall environment to allow post-stroke patients to engage in virtual shopping also showed improvement of upper extremity motor and functional ability[33]. Other studies show the improvement in reaching speed and reaching duration while performing a VR task which involved slotting envelopes in post-boxes [48]. There are also studies which allow a detailed documentation of the functional deficits post-stroke using VR gaming systems[49].

VR in stroke is a growing field wherein there is a need for larger scale studies with more randomized control trials. A review of 72 trials has shown that when VR was compared with the same level of conventional therapy, the results were not statistically significant for upper limb function. However, when combined and supplemented with usual care, there was a significant difference in the groups which received VR treatment[50]. Clinicians generally express positive experiences while using VR for treatment of people with stroke. However, the lack of time and knowledge to gain familiarity with the technology often poses a barrier in the wider acceptance of VR in the clinical setting[51].

1.4.4.2.2 Parkinson's disease

Parkinson's disease (PD) is a neurodegenerative disease which is best managed through a combination of medication and regular physiotherapy. Conventional physiotherapy in PD has been shown to have a positive impact on gait, endurance, balance and global motor function in individuals with PD[52], [53]. However, motor and non-motor symptom burden from rehabilitation affects the willingness of people with PD to participate and adhere to long-term exercise[54]. VR systems, both commercially available and customized tools, have been used as a rehabilitation tool with a potential added value over traditional physiotherapy approaches [55].

Studies involving VR applications in rehabilitation of individuals with PD involve commercially available technologies such as the Wii Fit [56], [57] or Motek. Several studies include a balance board , dancing movements or stepping in place as the actions being performed by the individuals[55]. These studies have shown that VR interventions in PD have led to a positive effect on gait, balance and cognitive function after training[56]–[59].

VR intervention studies for PD are still in their preliminary work stage with several studies not having randomized control trials. VR interventions may lead to greater improvements in step and stride length compared with traditional physiotherapy interventions. However, there was limited evidence that improvements in gait, balance, and quality of life were a result of primarily VR. At present, only a few studies have been conducted which makes it difficult to generalize the overall ability, but VR can definitely help improve rehabilitation in PD.

1.4.4.2.3 Traumatic Brain Injury

Traumatic brain injury (TBI) affects 1.7 million people in the United States alone. TBI leads to short- or long-term consequences which affect motor function including weakness in extremities, impaired coordination and balance. A large portion of the rehabilitation for individuals with TBI is done in a community setting with patients focusing on household independence[60]. Studies have indicated that intensive therapies lasting longer periods of time are more effective with training up to 20 hours/ week have the most positive outcome on motor recovery [28]. However, the shortage of resources for intensive recovery in TBI survivors prevents from a complete rehabilitation intervention from being completed. This has led to the adoption of VR for rehabilitation of individuals with TBI. VR therapies have been explored in the rehabilitation of motor functions such as balance, balance confidence, upper extremity function and arm-posture coordination[61]–[63].

Studies have used the Nintendo Wii Fit using a selection of games such as training balance, weight bearing, aerobics and yoga games[63]. Tasks ranged from using the VR environment to pour water from a cup [64] to using large arm movements to 'pop' a maximum number of balloons [32], [65].

The studies have shown that there is some impact of using VR training strategies in rehabilitation have led to an improvement in the some upper limb outcomes including increase in speed and some measurements of movement skill[65], [66]. Moreover, there are improvements in balance and upper extremity functions. While VR studies in rehabilitation of TBI have not been compared to a "gold standard", they have been adopted by several clinicians to provide precise performance measurements and exact replays of task performance[28]. VR therapy has some advantages including the ability to perform infinite repetitions of the same movement which can be made interesting through exergames. Studies have also shown that the use of VR therapy improves patient compliance and the patients' attitude towards VR therapy was quite positive[67].

1.4.4.2.4 Spinal cord injury

Studies have shown that exercise regimens post SCI allow for improvements in functional capacity, bone density in upper limbs, endurance, muscle strength and psychological well-being[7]–[9]. Improving upper limb motor function is also crucial for improving functional independence after an SCI[10].

However, there exist several perceived barriers to individuals with SCI performing the prescribed therapy regularly[11]. These include physical barriers such as equipment, availability of resources as well as psychological or social barriers such as perceptions or attitudes towards disability, motivation and fear of injury. Moreover, it was identified that the desire to exercise did not necessarily match behavior and various studies have found that lack of motivation was a ubiquitous factor in reduced exercise[12], [13].

However, current forays into exergames using VR and other gaming platforms are designed for individuals who are able bodied and do not account for the limited manual dexterity of individuals with tetraplegia[14]. This lack of hand function of tetraplegics to hold and depress various buttons on a typical VR game controller restricts the type of exergames that can be played. Design considerations of the type of exergames that are accessible to tetraplegics as well as engaging needs to be a significant consideration in the gameplay development. Most games available in the market do not work natively without the need for buttons. The controllers are generally designed to be grasped and individuals with limited hand function find it hard or impossible to use these without modifications.

When considering development of exergames, it is critical to track and quantify the progress of movements made by individuals performing at-home therapies. [19], [68]–[70]. There exists a need for objective testing based on these scores for at-home, which can quantify the progress made by individuals performing regular therapy.

Mieron is a commercially available VR exergame designed for individuals with SCI [27]. This system utilizes a portable VR headset to provide a virtual environment in conjunction with conventional means of rehabilitative therapy, including treadmill training and functional electrical stimulation cycling. This system is designed around changing environments using a VR headset to purely motivate individuals as there is no means to track the individual’s hand or other body parts in 3D space. However, such existence of a commercial system indicates that the motivational benefits of exercising while immersed in a VR system is significant enough to warrant a market presence.

1.5 Joint Forces as a measure of muscular effort

Joint reaction forces and torques have been studied with great interest in cyclists [71]. Joint reaction forces are calculated to understand the level of muscular exertion and even the technique used while cycling. This is valuable information for coaches and therapists in the field. In the study by Wangerin et al. 2007, joint forces at the knee and hip were calculated using an instrumented cycle's pedals with strain gauges to measure the pedal forces. Kinematic data was recorded using video cameras and retroreflective markers. Four important assumptions are put forward while performing inverse dynamics:

1. Anthropometric data
2. Link segment model of the human body
3. Kinematic data
4. External force measurement.

Anthropometric data is obtained from past studies that measure various parameters from cadavers. These parameters include an arm segment's center of mass, and weight of arm segment in proportion to total body weight. The link segment model of the body is a simplified representation of the complex joints that exist in the human body. The joints are modelled as simple revolute joints. The arm segments are modelled as masses and moment of inertias located at the center of mass of the segment [72].

In a VR setup, external forces are usually zero as the user is not interacting with any physical objects. Kinematic data will be obtained by using trackers that are part of VR systems. Complete kinematic data might not be available in classical setups. Commercial VR systems only track the hand. The position of the shoulder and the elbow are generally not tracked. Tracking the position of the shoulder and elbow might not be viable as VR trackers are not as light as passive retroreflective markers. It also increases the setup required as additional receivers need to be setup for each additional tracker used. Therefore, it is of interest to calculate joint forces based on only the end effector position. The HMD provides the position of the head which could be used to estimate the position of the shoulder. Calculating the position of the elbow from the position of the shoulder and the hand through inverse kinematics does not present a unique solution. Some of these solutions can be discarded due to the biomechanical constraints of the arm. The solution can be further optimized to obtain a natural "elbow down" orientation that is considered a comfortable pose for individuals [73].

It has been shown that shoulder torques are positively correlated with perceived muscular effort [74]. Calculation of shoulder torques could inform therapists of the perceived effort that the users likely felt during gameplay. Gameplay parameters could then be modified to possibly alter the level of perceived exertion and perhaps prevent frustration during gameplay.

Ergonomics and areas of comfort

Performing activities within the comfort areas of a person's workspace promotes
1.6 performance, efficiency and emotional wellbeing. Studies have been conducted to understand the comfort areas of able-bodied individuals using observations or software simulation [75]. However, there is a dearth of research investigating this aspect for individuals with upper extremity mobility impairments. The identification of comfort areas in 3D space for these individuals provides valuable insights for universal design and leads to a variety of applications, including workplace accommodations, customization of wheelchair controls, and accessible interface design for vehicles and other hardware systems [76].

Experimental studies have been conducted to quantify the level of comfort by mathematical functions of joint angles and applied forces [77], [78]. To determine the comfort areas within an individual's workspace, joint angles and applied forces need to be collected over the entire workspace. It is challenging and time-consuming to collect such data with human subjects, especially for people with upper extremity mobility impairments including those caused by SCI, stroke, multiple sclerosis, Parkinson's disease, or amyotrophic lateral sclerosis (ALS). Currently, comfort areas are determined through subjective scales with users performing iterative, trial-and-error experiments [75].

To tackle this challenge, VR techniques can be utilized that make the data collection process simpler and faster [79]. VR has been applied in the field of ergonomics to measure the reach envelope of individuals for workplace design [80]. Studies indicated that VR is efficient and cost-effective compared to prototyping with physical models [81]. VR simulation has been explored in ergonomics for able-bodied people to allow maximal comfort [40]. However, its potential has not been explored in setting up assistive technologies, which is typically performed by occupational therapists in a clinical rehabilitation setting [82].

2. METHODS

This section discusses the methods that were used to develop the virtual reality based rehabilitation system. This system comprises of the HTC Vive® (Taiwan), adaptations to the HTC Vive® participant controller known as a tracker, exergame development, mixed reality system, baseline physiology measurement tool, gesture extraction and segmentation, and joint force calculation and ROM calculation.

2.1 HTC Vive®

There were several virtual reality (VR) systems that were identified and explored to track an individual's movements, particularly the Oculus Rift VR system, Myo armband, Microsoft Kinect, Leap Motion, Nintendo Wii and the HTC Vive®. Ultimately the HTC Vive® was chosen as the preferred system as it combined an adaptable VR environment with a highly accurate sub-millimeter position tracking system [83].

The HTC Vive® comprises of two base stations, called lighthouses, which have spinning IR lasers that flash and sweep a beam of light alternatingly. The head mounted display (HMD) and trackers have a constellation of IR receivers that use flashes and beams of IR light to determine the position and orientation. The position data is calculated at a ~90Hz refresh rate [83].

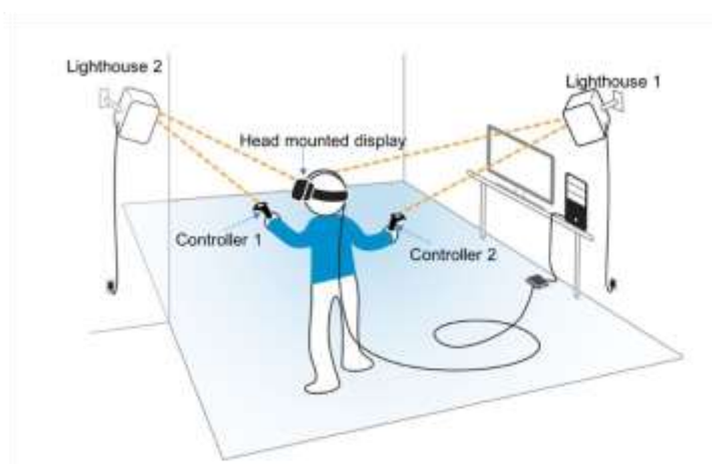


Figure 2: HTC Vive® Tracking System comprising the head mounted display, two controllers and two lighthouses.

2.1.1 Adaptations to the HTC Vive® and Game Development

None of the virtual reality systems explored accounted for the limited manual dexterity of individuals with tetraplegia [14]. The standard HTC Vive® controllers are inaccessible being large and unwieldy making it hard to mount to the wrist. They are also designed to require fine motor control to grip and use buttons for interaction (Figure 3A). Nearly all games commercially available for play assume using the standard HTC Vive® controllers and ability to interact with buttons.

Figure 3: A) Standard Inaccessible Vive Controller requiring finger dexterity to operate. B)Adapted Vive tracker worn on wrist of participant with no hand function. A Velcro strap is used to attach it to the palm of participants.

HTC® also sells Vive® trackers that were designed to be mounted to game objects to be tracked by the base stations. It is designed to be attached to racquets or to enable full body tracking by attaching it to a participant's shoes. A 3D printable shim was designed to allow the tracker to be secured firmly to a participant's end effector with a strip of Velcro (Figure 3B).

VR tools and games were developed using the Unity3D (Unity Technologies, San Francisco, California) game engine to work with the Vive trackers as the primary tracked device. The games involved designing 3D models to be rendered during gameplay. Each 3D model has a simple collider object associated with it that is used for the physics engine to detect collisions and perform the required calculations [84]. For instance, the balloon model has a detailed mesh of the balloon for rendering. However the collider associated for the balloon is a sphere that approximates the shape of the balloon (Figure 4). The physics engine performs a test for collision between objects each frame. Testing for collisions between several complex meshes is computationally very expensive, thus balloons are approximated with a sphere which allows for a smooth gameplay and any reduction in frame rate might cause nausea or motion sickness to the participant playing the game. Programs were written in C# to detect and handle collision events.

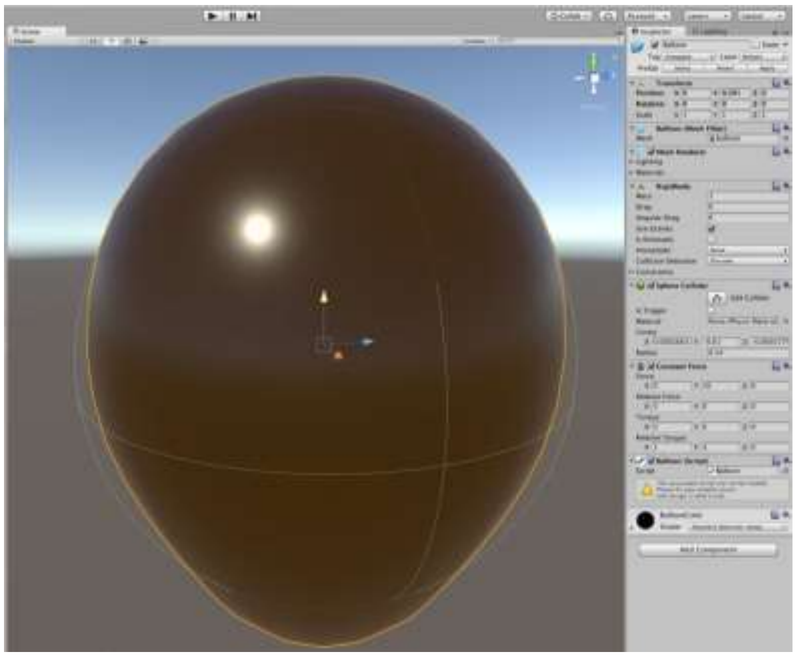


Figure 4: Balloon Model (outlined in orange) and a Sphere Collider (grey)

3.1.3 Mixed Reality System

A mixed reality system was developed to allow for a mixed view of the gameplay that incorporated both a real video with the virtual environment and virtual objects overlaid in the correct position and orientation. This allowed clinicians and researchers an insight into how the participant interacted with the game environment. This mixed reality system utilizes a position tracked camera using an HTC Vive® tracker, a green screen and the traditional HTC Vive® VR setup.

A virtual camera is setup in Unity3D in the exact same position the real camera is in the real world. A one-time calibration is required to determine the physical distance between the Vive tracker placed on the camera, and the camera itself (Figure 5) . The camera's field of view and orientation are also calibrated to match the representation in the virtual environment. Then the video feed from the real camera and the virtual camera are combined using the chroma key composting technique. This removes the background from the real video feed and separates the virtual objects into the foreground and background depending on their relative distances to the head mounted display [85]. The mixed reality video is generated through the Liv software (LIV Inc, San Francisco CA) and is generated in real time, which allows clinicians or researchers to move the physical camera and the virtual camera along with it during live gameplay to look at the participant and virtual environment from different angles. Video of the gameplay session can be saved for later assessment.

Figure 5: Mixed reality setup with HTC Vive® and Green Screen

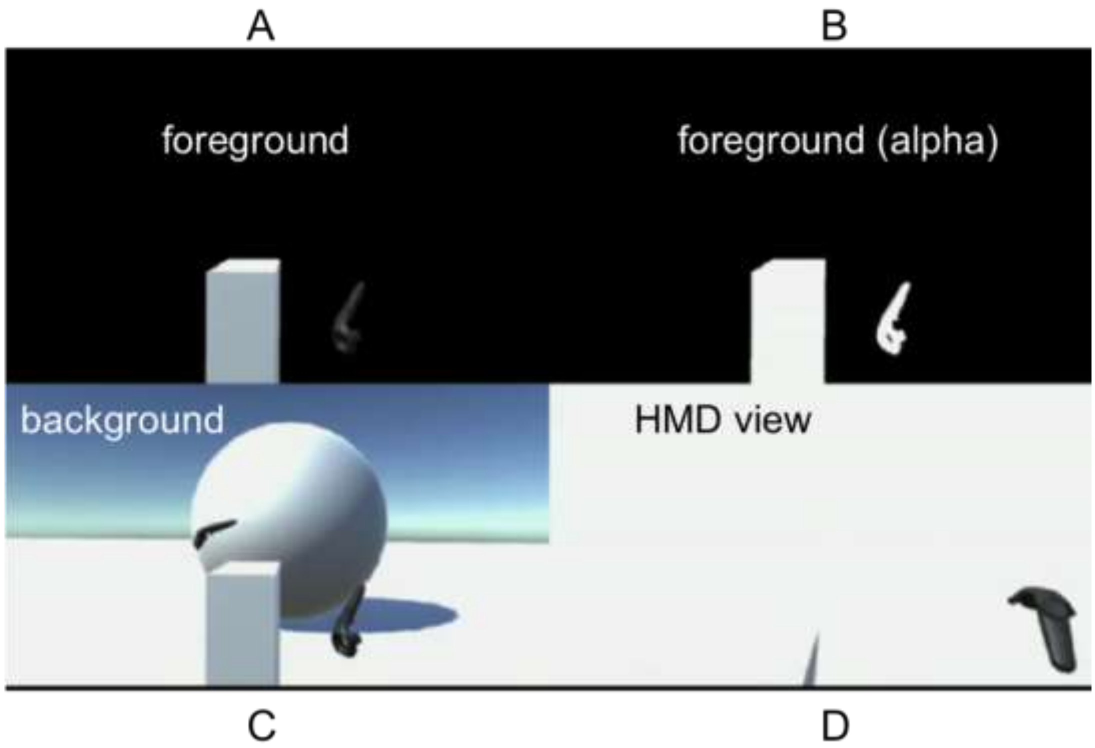


Figure 6: Mixed Reality Video Layers. A) Foreground objects in the virtual world B) Foreground(alpha) layer which serves as allow for combining with other layers C) Background layer is the first layer during composition of mixed reality D) First person view of the gameplay – showing the participant's interaction with the controller.
[image retrieved from https://medium.com/@dariony]

Figure 6 shows the various layers that are generated during gameplay in the Unity3D engine to enable mixed reality video. Figure 6A shows the foreground objects which are in front of the participant in the virtual world. The foreground objects are determined to be objects that are a given distance in front of the participant's HMD. This distance is programmable and can be varied based on the number of objects in the participants' virtual field of view ensuring an unobstructed view. This distance was not altered for the purposes of the games developed for this thesis as the number of game objects was not too high to cause serious occlusion.

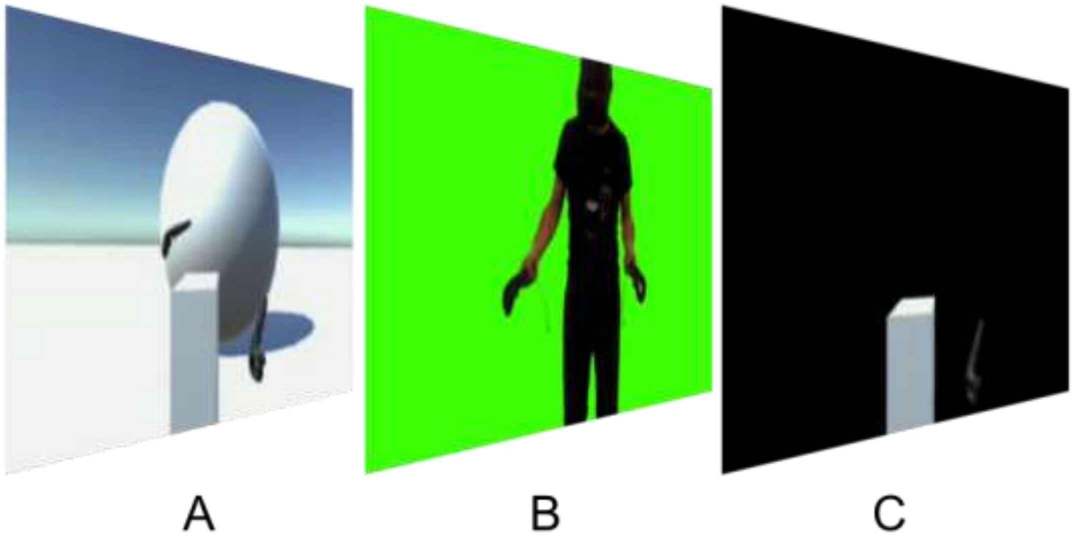


Figure 7: Order of compositing various layers for a mixed reality video.. A) Representation of background image B) Participant in front of a green screen C) Foreground layer.
[image retrieved from https://medium.com/@dariony]

Figure 7 shows the order of layers in a typical mixed reality set up. The background video (Figure 7 A) is the first layer to be added. The real video of the participant (Figure 7 B) is composited on top of the background video. The background video replaces the green screen to create a video of the participant in the virtual video of the background. Next the foreground video (Figure 7 C) is added on top using the foreground alpha layer (Figure 6 B) as a mask. The use of black coloring allows the underlying layer to show through and white is replaced with the foreground video. All these layers are combined to form a single stream of mixed reality video in real time as shown in Figure 8.

A major challenge in setting up mixed reality is calibrating the camera lens' optical parameters such as focus and field of view. Moreover, there is a translational and rotational offset between the camera and the tracker which must be accounted for. Several attempts were required to align the real and virtual videos and recalibration required large overhead times during participant studies. To prevent this, a tracker camera mount was 3D printed to keep the translational and rotational offsets fixed (Figure 9). This greatly reduced the time needed for re-calibration during each set up.

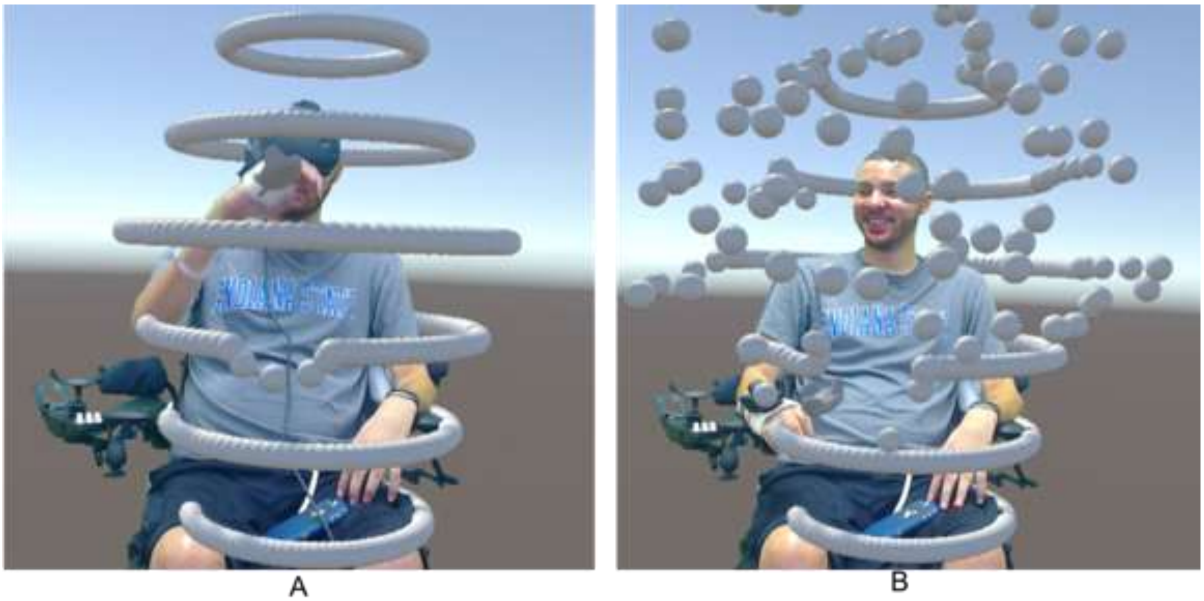


Figure 8. Screenshots of mixed reality video with A) Spawned Spheres at the start of a trial. B) Displaced spheres at the end of a trial.

Figure 9: 3D printed mount to fix camera and tracker geometry

There are certain drawbacks to this particular method of mixed reality composition. For instance the participant in the real video is treated as a 2D object. Complex 3D interactions might not be visible clearly and get compressed to 2 dimensions for instance, if the participant is reaching behind his body to interact with a virtual object it might not render correctly in the mixed video. However for the use case presented in this research, requiring a seated participant to interact with simple virtual objects, this method was sufficient to capture the nature of interaction.

Participant Recruitment

For this study, we recruited 6 participants from the Rehabilitation Hospital of Indiana and
2.2 Purdue University. The mean age of the participants was 37.5 ±9.9, with 1 female and 5 male participants. All participants had a cervical SCI ranging from C4- C7 level injuries. Prior to their participation in the study, the participants had been injured for 15 ±11.2 years. All study protocols were approved by the Institutional Review Board of Purdue University – protocol number 1705019528. Prior to the study, informed consent was obtained from all the participants.

Table 1: Participant Characteristics

Participant	Age	Gender	Weight Kg (lbs)	Height cm (ft, in)	Level of Injury	Complete/Incomplete	Years since injury
1	27	M	80.7 (178)	188 (6’ 2”)	C6/C7	Incomplete	2
2	27	M	81.7 (180)	180 (5’ 11”)	C5	Sensory Incomplete	2
3	35	M	79.4 (175)	188 (6' 2”)	C5	Sensory Incomplete	19
4	35	F	49.9 (110)	150 (4' 11”)	C5	Incomplete	28
5	47	M	81.7 (180)	188 (6’ 2”)	C4/C5	Complete	29
6	54	M	81.7 (180)	180 (5’ 11”)	C5	Complete	10

Baseline Physiology Measurement Tool

The baseline measurement tool is a VR game that was developed to function as a VR tool to map the participant's motion performance envelope and to determine movement clusters of comfort. The baseline game was developed in Unity3D. This game utilizes the modified trackers
2.3 to allow tracking of the participant's hand without use of any buttons or touchpads.

Upon launching the baseline tool, the program waits for the HMD and the tracker to be initialized and tracking. Once the HMD and trackers have been initialized and started tracking, 600 spheres are spawned around the HMD. They are arranged in 6 layers with 100 spheres each and spaced apart equally in each layer (Figure 8a). The spheres represent an isotropic stimulus that is encourages the participant to move in all directions. The spheres have zero resistance and are not affected by gravity. So they move when pushed by the participant but stop when the interaction ceases. This leaves the spheres at the farthest location the participant pushed them to during the gameplay. The participant is able to go back to spheres and try to push them a little further if they wish to.

The gameplay for the baseline measurement tool lasts approximately 2 minutes long. This duration was recommended by physical therapists at RHI as a typical length of time for an individual rehabilitation exercise for patients. The final coordinates of each sphere is logged when the game stops. The coordinates of the tracker strapped to the participant's hand is logged throughout the gameplay. The coordinates of the HMD is also logged throughout the game play.
2.3.1 The coordinates logged are all saved as comma separated values (CSV) files [79].

Experimental setup to measure engagement

The baseline tool was used in an experimental setup to measure engagement levels and motivational aspects of VR gameplay. The experiment involved the participant performing the baseline task three times. First with the HMD, second without the HMD, and third with the HMD again. The third trial with the HMD was to account for changes in performance due to fatigue.

Experimental Procedure

1. The participant first completed a short questionnaire to obtain their demographic data.

2. The participant was then requested to wear the HMD, hand mounted Vive tracker on the dominant hand and the Microsoft Band on the non-dominant hand.
3. The participant was then allowed two minutes to perform the baseline task. The baseline task begins with several spheres spawning around the participant (Figure 8). The participants' task was to push these spheres as far out as possible.
4. The HMD was removed and the participant was allowed to rest for five minutes.
5. The participant wore the HMD for roughly 10 seconds to familiarize themselves with the location of the spawned spheres, after which it was removed, and participants repeated the baseline task without the virtual spheres as motivational targets to hit.
6. The baseline task was repeated with the HMD.

2.4 Extracting Areas of Comfort

2.4.1 Kernel Density Estimation

3D heatmaps of clusters of movement were generated using the time series coordinates recorded during the VR exergames. Generating a heat map involved 'scoring' each point. Kernel density estimation (KDE) was used for this purpose [86]. KDE was selected over other clustering techniques that our used commonly for hotspot detection to better model the continuous nature of hand motion. Other techniques such as K-means clustering would model the hand's motion as discrete points therefore not chosen as a method to calculate a heatmap [86]. This approach using a KDE grants us with a smooth estimate of the probability density.

KDE attains this smooth estimate by placing a kernel or a hump with a known probability distribution function (PDF) at the center of each x_i. The average of all these kernels were averaged to find the estimated probability density function. Kernels can take many forms but they are generally unimodal and symmetrical in nature [87]. A gaussian kernel (eqn. 1) was used for the KDE implementation.

$$P(x) = \frac{1}{\sqrt{2\pi}} \exp^{-d^2/2} (1)$$

The KDE implementation was done using the `gaussian_kde` [88] function found in the `stats` module which is part of the `scipy` library in Python 3.6. This function automatically determines an appropriate bandwidth using the rule of thumb Scott's Rule [89].

$$d_i = \hat{f}_h(x_i, y_i, z_i) \quad (2)$$

The density scores d_i, where i ranges from 1 to the length of the time series data recorded from gameplay, are then used to color the 3D coordinates to plot a heat map of the gameplay.

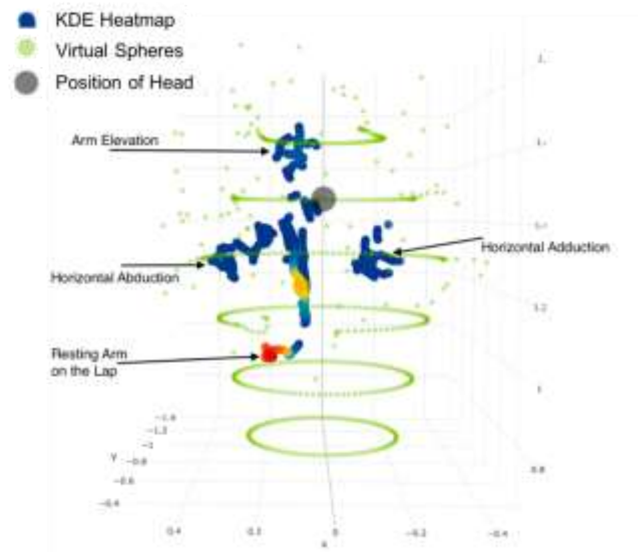


Figure 10: 3-D heatmap generated by kernel density estimation showing areas of high frequency movement of the arm. The green markers are the spheres from the baseline task. The position of the participant's head is indicated by the gray sphere.

In Figure 10 is a typical heat map of a participant generated by KDE. Many virtual spheres (shown in green) can be seen to have been displaced (out of plane) while some are untouched by the participant still lie in their rings. The area of rest has the largest density as generated by the KDE (Figure 10, shown in red).

2.4.2 Thresholding Density Values

The density value, d_i, associated with each 3D coordinate, (x_i, y_i, z_i), was used to cluster the data. Large density values were associated with areas that were visited most frequently by the participant. This was due to the hand spending a lot of time in this location, which meant there were several points logged leading to multiple kernels being averaged together in these areas.

The location associated with the highest density was always associated with an area of rest which was most often on the lap of the participant. This was verified through the use of the mixed reality video obtained during gameplay. Although this cluster had the highest density value, participants had the least motion within this cluster as they were usually simply resting their hand.

The 3D coordinates were sorted by their d_i score. A K-means algorithm was used to identify and separate various clusters with similar d_i scores but distinct 3D coordinates, implying that these clusters are equally frequented but are distinct areas in 3D space. These clusters could then be categorized as areas of comfort if the scores were high or areas of lesser comfort if the scores were on the lower end of the spectrum. This also allows us to rank various clusters by their order of comfort.

2.5 Gesture Extraction

Time series data of the hand was logged for approximately 2 minutes each time. With a ~90Hz sampling rate this generates 10,800 3D data points. Recognizing individual gestures and separating them has applications in rehabilitation [90].

The gesture extraction utilized *gesture spotting* techniques [91] to identify the start and end points of a deliberate motion from a continuous time series data. Due to segmentation ambiguity and spatio-temporal variability in the data, gesture extraction is often considered a difficult task. Moreover, there are supplementary movements which occur between two deliberate gestures making it difficult to distinguish them [90]. In this thesis, we aim to segregate a continuous stream of motion data into distinct and deliberate gestures.

There are several ways to separate gestures depending on the nature of data stream and type of gestures expected. As we understand the nature of interaction between the participants and the gameplay, which were mostly reaching and holding tasks. All participants had a rest position that could be extracted by the use of KDE.

A deliberate gesture for our scenario was defined as the hand starting from a position of *rest* where the velocity of the hand and the acceleration was zero. It could also start from a *turning* *point* which was part of the previous gesture where the direction was changing and therefore, the velocity of the hand was low. The end point for a gesture could be either a *rest* or a *turning point* as well. The common aspect of both these scenarios was that the velocity was very low or zero at the gesture extrema. This has been seen in literature on human arm trajectory formation describing velocity profiles of end effectors [92].

Separating the gestures or *gesture spotting* then could be achieved using the velocity as a metric to delimit the continuous data stream into gestures. This technique has been used to segment gestures in video games and for human computer interaction systems in literature [91], [93].

When performing gesture spotting of our data the rest position was first identified using KDE and K-means. This was verified using the mixed reality video recording of the gameplay. Time derivative of the 3D position data was performed to calculate the velocity. The local maxima in velocity within the rest position data was defined as the threshold velocity. The temporal data was then sliced up using the threshold velocity as the delimiter. This yielded individual gestures of variable length. All the extracted gestures satisfied the criteria that were established to start and end either at a *resting* position or a *turning point*.

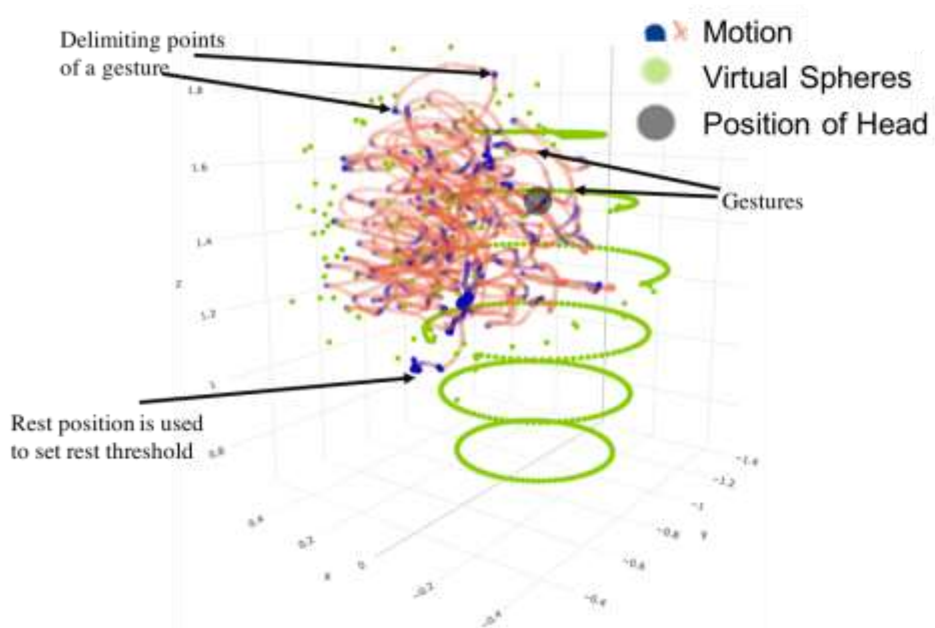


Figure 11: Separated gesture plot of a single gameplay by a tetraplegic participant using a gesture spotting approach. The grey sphere is the head and the green markers are the virtual spheres spawned during the baseline tool task.

In Figure 11 the time series data set was separated into gestures. The red lines are individual gestures and the blue parts are the areas where the velocity of motion is below the set threshold velocity.

Case Study

To validate the identified comfort areas, a case study was conducted with a user typing
2.6 using a two button method at each of the four areas of varying comfort levels. The time taken to complete the tasks and the typing accuracy was recorded for assessment.

Participants

2.6.1 One male participant with tetraplegia due to spinal cord injury at the C4/C5 level participated in this preliminary study.

2.6.2 Tasks

During each trial, the participant was asked to type a randomly chosen eight letter word, which were “Accuracy”, “Cellular”, “Document” and “Emerging”. The native switch access keyboard on an Apple iPad™ was utilized in this study, which is one of the standard methods of input for people with upper extremity mobility impairments. The participant performed sixteen trials in which the four words were randomly selected and typed at all four locations in random
2.6.3 order.

Experimental Setup

The experimental setup shown in Figure 6 included the Blue2™, an accessible two-button Bluetooth™ switch from Ablenet and an iPad for the typing task, a customizable rack to mount the switch, and the HTC® Vive Platform to correctly position the button. The two button method of typing uses one-button to scan groups of keys on the on- screen keyboard and the second button is used to select the key. The switch was placed at each of the four areas of highest and lowest levels of comfort. At the end of the experiment the participant was requested to complete a usability questionnaire to rate the level of comfort at each position.

A B

Figure 12: Experimental setup showing a comfort position (A) and a discomfort position (B).

2.7 Joint Force Calculation

In order to calculate joint forces for each gesture performed by participants during gameplay, the human arm was modelled kinematically as a serial-link robot following the Denavit-Hartemberg (D-H) notation [94], [95]. The MATLAB Robotic Toolbox [96] was used to implement the model. The toolbox contains methods to describe prismatic and revolute joints, their range of motion and the rotational and translational relationship from one link to another. This description takes the form of D-H parameters.

D-H parameters consist of five parameters that are used to describe each link to the previous link in the series [97]. These parameters are:

- d_i: link offset – distance from the origin of the previous frame to the x_i axis along the z_{i-1} axis.
- θ_i: joint angle – the angle between the x_{i-1} and x_i axes along the z_i axis.
- a_i: link length – the distance between the z_{i-1} and z_i axes along the x_i axis.
- α_i: link twist – the angle between the $z_{i-1}$and z_i axes along the x_i axis.
- σ_i: joint type – R for revolute and P for prismatic joint.

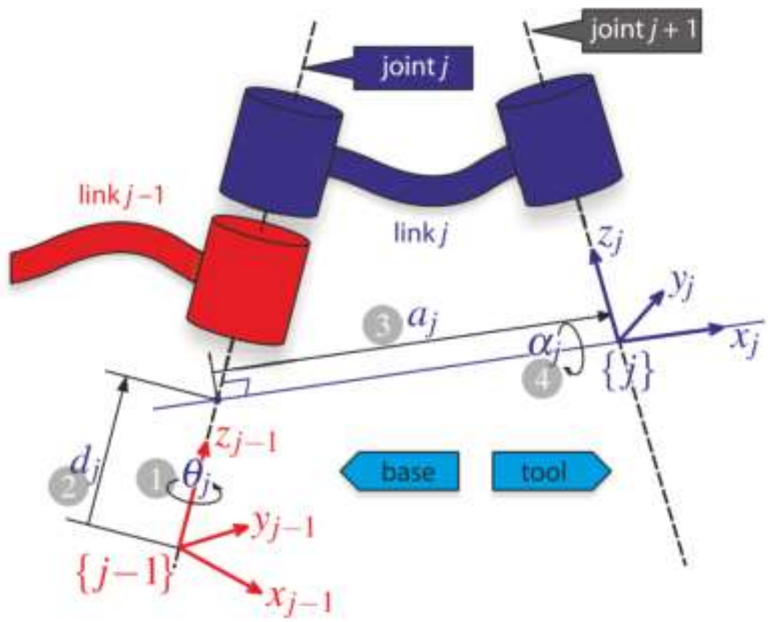


Figure 13: Geometry of D-H parameters. [98]

In addition to these default D-H parameters, absolute joint constraints (AJC_i) [99] were added to each joint to prevent orientations that are unachievable biomechanically. The D-H parameters used for the kinematic model are:

Table 2: D-H parameters used for the kinematic model of the arm

	$\theta\ (rad)$	d	a	α (rad)	AJC (deg)
1	q1	0	0	$-\pi/2$	-45 to 180
2	$q2 + \pi/2$	0	0	$-\pi/2$	-45 to 130
3	$q3 + \pi/2$	0	$L_{humerus}$	$\pi/2$	-60 to 180
4	q3	0	L_{ulna}	$-\pi/2$	0 to 150

Table 2 and Figure 14 illustrate that the first three joints are at the exact same point in 3D space but offset by 90º, these joints correspond to the degrees of freedom at the shoulder joint. The fourth joint is at the elbow offset from the first three joints by the length of the humerus or the upper arm. The fourth joint has a length of the forearm or the ulna. The wrist joint is not modelled, as the gameplay did not involve wrist motion.

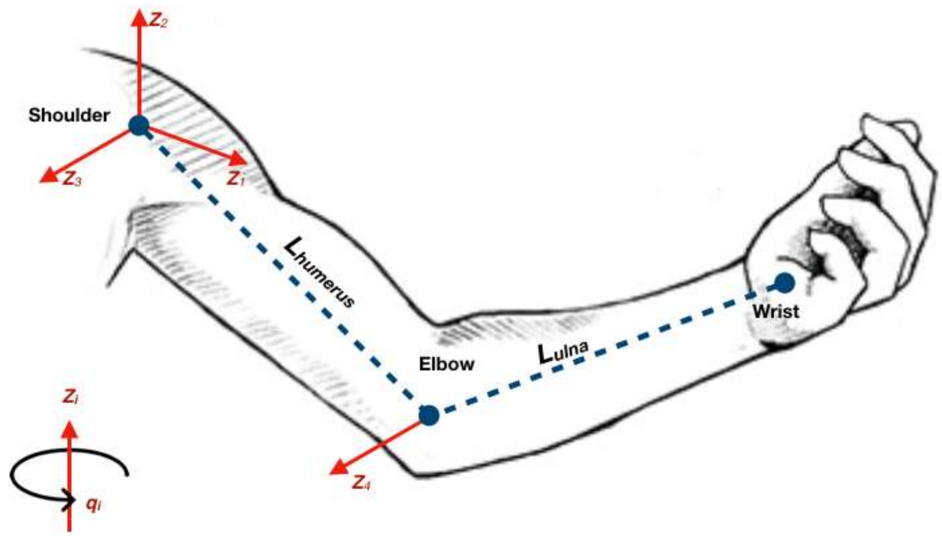


Figure 14: Kinematic model of the human arm with reference frames associated to the various degrees of freedom

The length of individual participant's upper and forearm were measured from participants' video recordings taken during the gameplay. Figure 15 shows how the anthropometric measurements were made using the open source physics video tracking tool Physlets Tracker [100] from the video recording of the participant during use of the baseline tool. The known value of the HMD's width was used to calibrate each frame of the recorded video, this can be seen as the blue arrow. This measurement was performed three times for each participant and the average length was used for data analysis.

The recorded coordinates of the tracker and the HMD are relative to a 'world space' that is determined by the position of the lighthouses that is used by the HTC Vive®. This means that the world space coordinates changed each time the entire virtual reality setup was torn down and moved. Thus, a new coordinate system was defined with the HMD coordinates as the origin. Therefore, the tracker coordinates and all the virtual objects were referenced to the HMD position.

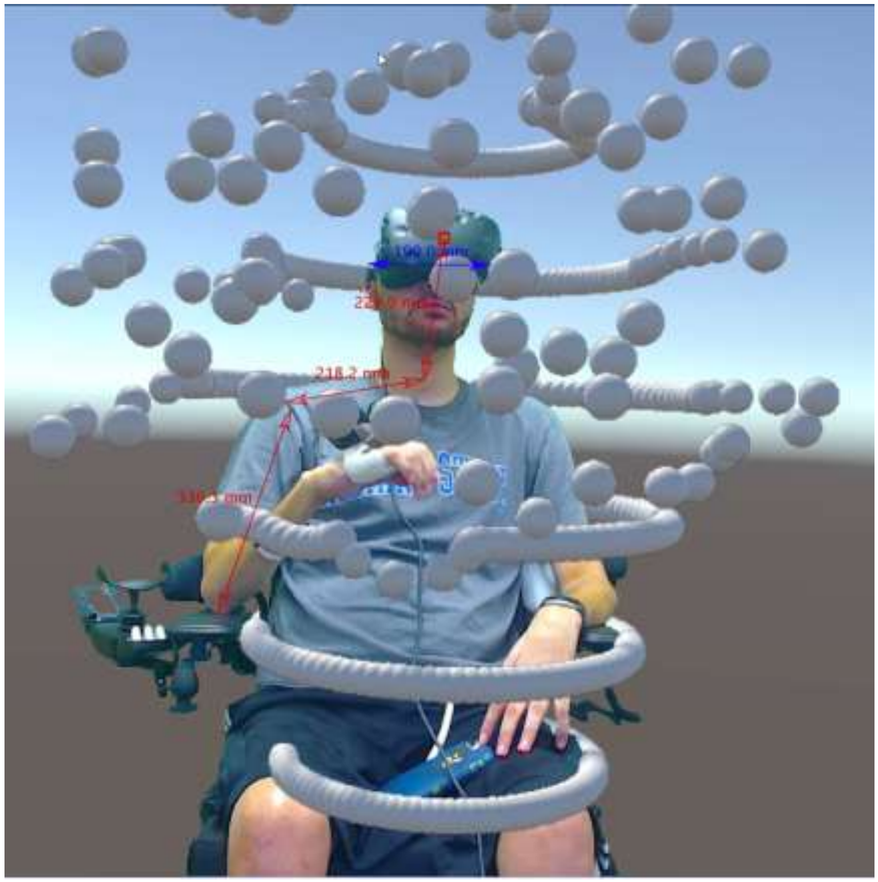
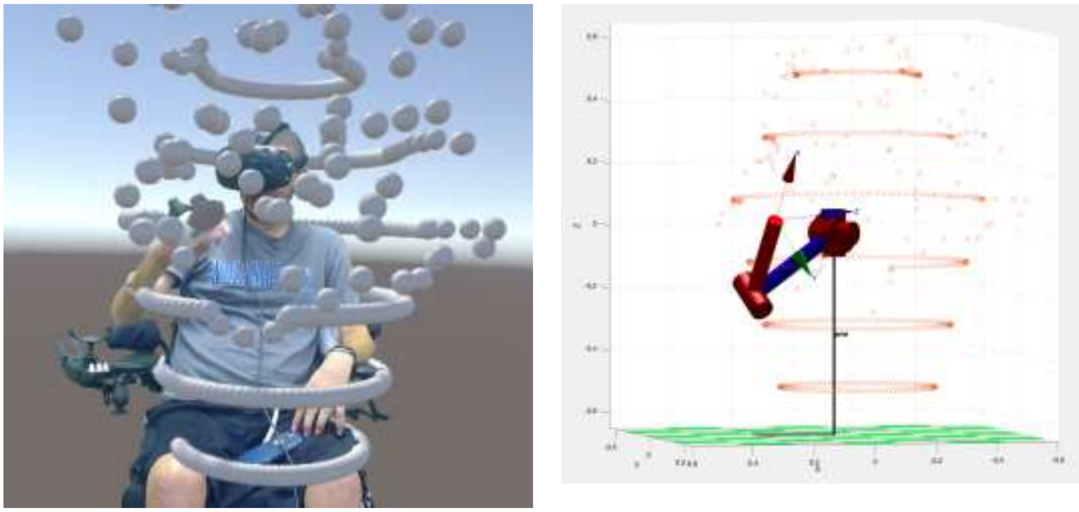


Figure 15: Anthropometric measurements of the participant from recorded mixed reality videos.

Inverse kinematics was performed using the kinematic arm model from the tracked and translated coordinates relative to the HMD coordinates of the end effector Figure 17. The inverse kinematics computation was performed in MATLAB using the robot toolbox. Inverse kinematics returned the joint angles necessary to achieve the specific pose. Orientation of the end effector was ignored as this was not a metric that was measured during game play. Inverse kinematics yields several possible solutions for the position of the elbow as there is no unique solution. Biomechanically not viable locations of the elbow are discarded based on joint constraints placed on the elbow. To obtain a conservative estimate based on comfort, an “elbow down” [96] start pose was determined empirically for each individual based on literature showing that this orientation is comfortable and natural [73]. The inverse kinematics tool accepts a start pose as an input argument to use as a starting point. This start pose based on the comfortable pose can be seen in Figure 16. This pose determined the final orientation of the calculated pose. Inverse kinematic algorithms have been developed in the past with maximizing human comfort by minimizing joint torques [101].



A B

Figure 16: Initial location of the elbow based on comfort A) mixed reality view with “elbow down” B) kinematic model of the arm with “elbow down” orientation.

An inertial model of the arm was also created by modelling the two arm segments as cylinders. The width of the arm segments was obtained using the same method as shown in Figure 15 using the physlets tracker. The mass and center of mass of the arm segments were calculated as a percentage of the body weight obtained from standardized anthropometric data. The upper arm and forearm mases were 2.66% and 1.82% of the entire body weight for men and 2.6% and 1.82% for women respectively. The distance of center of mass from the proximal joints were 48.5% and 44% respectively. [102], [103].

The inertial parameters for the principle axes for the arm segments modelled as solid cylinders were then calculated using the equations 3 and 4.44

$$I_{xx} = \frac{1}{2}mr^2 \ (3)$$

$$I_{yy} = I_{zz} = \frac{1}{2}m(3r^2 + \ h^2) \ (4)$$

Static torques were calculated by using the `SerialLink.gravload` method from the robot toolbox. This method calculates the joint gravity loading for a given model. The direction of gravitational acceleration and value can be set explicitly. This method returns the required joint torques for the modelled robot to attain a specific pose.

Dynamic torques were calculated by using the `SerialLink.rne` method from the robotic toolbox. This method calculates the joint torque required for the simulated arm to achieve the specified joint positions, velocities and accelerations for each joint possessed by the arm. The method utilizes a recursive Newton-Euler method. The number of iterations necessary to converge at a solution could be altered to allow for a convergence if the target pose was very far from the initial 'seed' pose.

The inverse dynamics solver was not a stable tool and often failed to converge. At times the inverse dynamics solver converges at extremely large meaningless values which had to be discarded. These large meaningless values were treated as outliers and removed from the data before analysis. A threshold of 55 Nm was picked to remove outliers, this threshold was picked as an average dynamic torque during external and internal rotations seen in able bodied subjects [104].

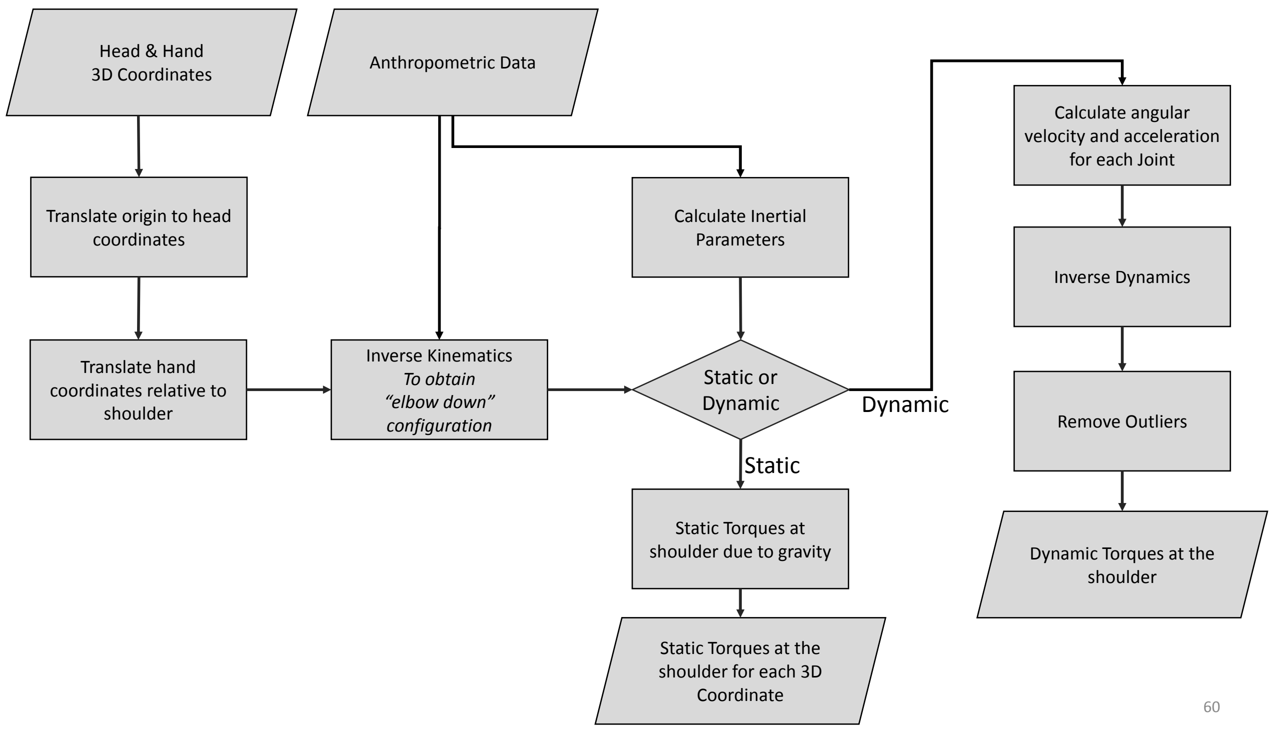


Figure 17: Flowchart showing all the steps in calculating static / dynamic torques at the shoulder

2.7.1 Sensitivity Analysis

Sensitivity analysis has been used in the past to understand how different methods of inverse dynamics are sensitive to various input parameters [105].

A sensitivity analysis of the inverse kinematics and joint torque calculation tool was performed by varying several input parameters such as arm segment length, segment radius, segment mass and segment center of mass (COM). A representative trajectory was picked from a subject to perform sensitivity analysis on. First static torques and dynamic torques were calculated for this trajectory to be used as the baseline value. Then static and dynamic torques were calculated by altering each aforementioned input parameters by ± 1% and ± 10%. The percentage difference in output due to the change in input was calculated. The results were then tabulated to understand which parameters the system was most sensitive to.

Balloon Exergame

An exergame was developed inhouse to test what gameplay parameters can be adjusted as a way to cause a participant's physiological measures to change over time. The purpose of increasing gameplay difficulty is to physically challenge the participant but not overly so as to discourage continued gameplay [106][107]. The exergame developed involved targeting virtual balloons that were spawned randomly around the participant. A virtual model of a light saber was attached to the participant's tracker that was used to target virtual balloons. These multicolored balloons were designed to pop when the lightsaber targeted them for a specific duration. The balloon's color would change to a fluorescent pink when it was successfully targeted, i.e. the light saber was inside the balloon's collider. However, to avoid inadvertent pops resulting from flailing motion or some other unplanned motion a small delay was added before a balloon would pop. The baseline delay was chosen to be 100ms. At the end of a successful pop, a popping animation and a loud realistic balloon pop sound was played as visual and auditory notifications to participants.

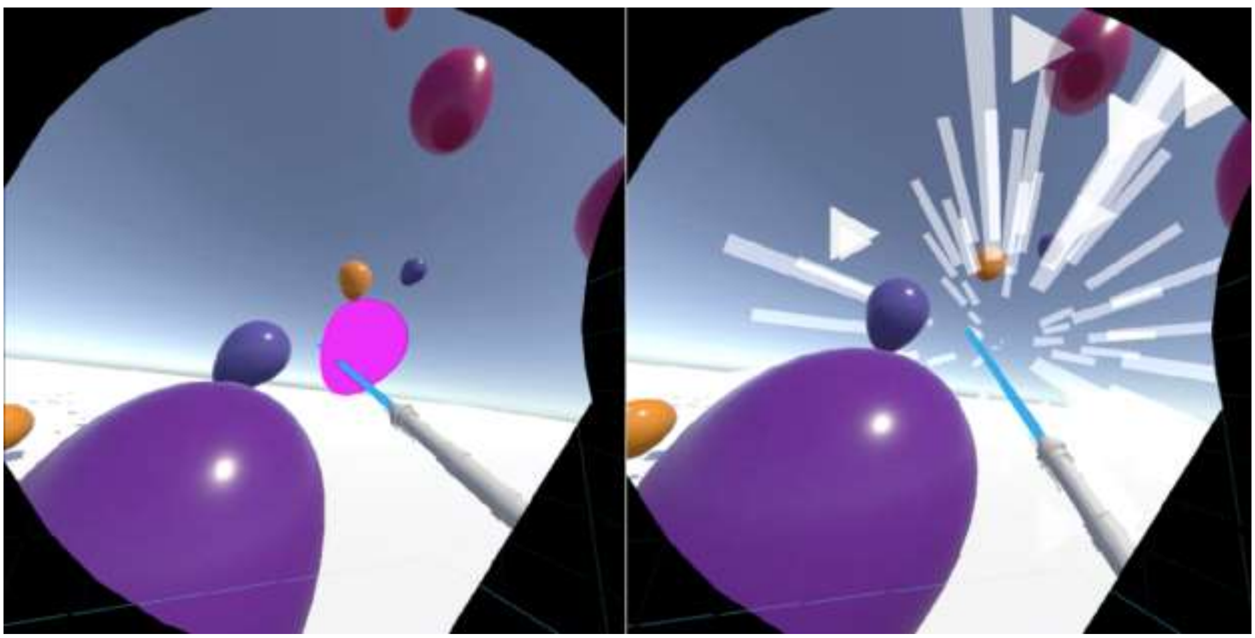


Figure 18: Left: balloon turning fluorescent pink indicating it was targeted. Right: Balloon popping animation indicates that the light saber targeted the balloon for a sufficient amount of time (>100 ms).

There were two gameplay parameters that were selected to be modified to measure the resultant change in participant biomechanical responses. The first gameplay parameter was changing the scale (size) of the balloons and the other was the delay required for the balloon to pop after targeting. Two different trials were designed to investigate how participants interacted with the game and how each of these parameters affected the participants' interaction/performance.

2.8.1 Scale Trial

A 2-minute-long gameplay with the balloon game was determined based on physical therapists' recommendations. The spawned balloons were either at full scale (1) or at half scale (0.5). The probability of a balloon spawning with either scale was 0.5. All the balloons had a pop delay threshold of 100ms. The participants were asked to play the game and try to pop as many balloons as they could during the duration of gameplay.

2.8.2 Delay Trial

A 2-minute-long gameplay with the balloon game where spawned balloons were all at full scale. However, half of the spawned balloons would pop at the default pop delay threshold of 100ms; whereas the other half of spawned balloons would only pop at the increased pop threshold of 300ms. The participants were asked to play the game and try to pop as many balloons as they could without knowing which balloons had the longer pop delay.

During both exergame trials, the coordinates of the participant's HMD, end effector tracker, number of popped balloons, and the location of the tip of the light saber were all logged throughout gameplay. In addition to these, the scale of the balloon and time taken to pop was also recorded. For the delay trial, failed attempts at popping the balloons were also recorded along with the time spent inside the balloon before the light saber left the balloon's collider.

2.9 Statistical Test

2.9.1 Permutation Test

A non-parametric permutation test was used to test for significance. The observed difference in means (T_{obs}) between the two test groups was first calculated. The two groups are then pooled together. This pooled data is shuffled and divided into two groups with the same length as the test groups. The difference in means is calculated again ($T_{permuted}$). This is repeated for the N = 10,000 times. Finally the number of times $T_{permuted} \geq T_{obs}$ is calculated and divided N to obtain the one sided P-value. This is described in the equation:

$$\mathrm{P-value} = \frac{T_{permuted} \geq T_{obs}}{N}$$

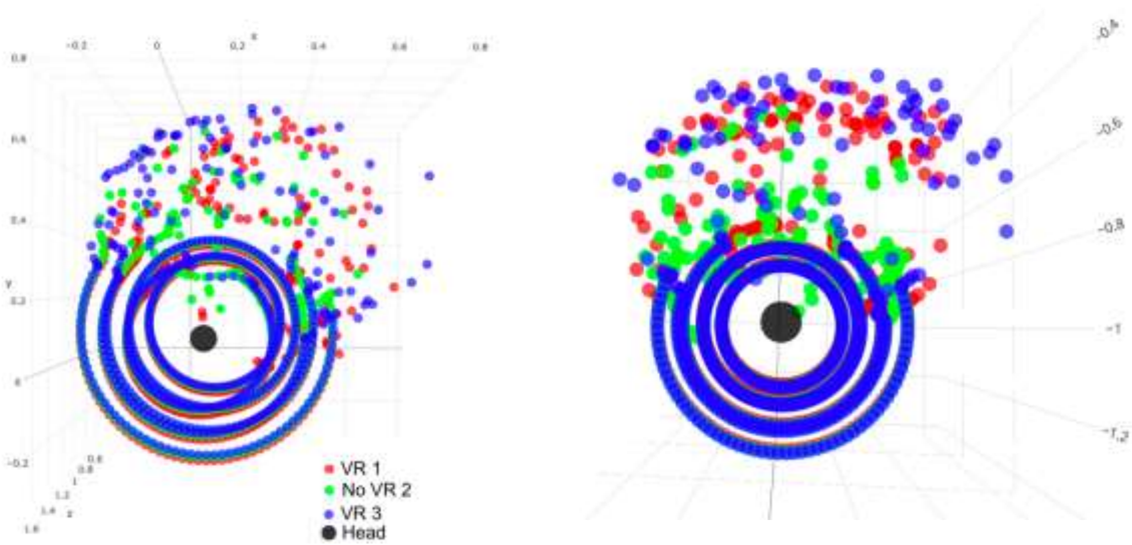
A significance level of 0.05 was chosen for this test [108]. The permutation test was used to test for significance in the average displacement of spheres in the baseline task, difference in velocity between balloons of different sizes, and balloons of different pop delays.

3. EXPERIMENTAL RESULTS

3.1 Baseline Physiology Measurement Tool

3.1.1 Range of Motion

When comparing range of motion through displacement of spheres between baseline tasks in VR and without VR, we observed a difference in performance of the participants. It was determined that with the VR headset, the participants were able to displace the spheres farther than in setups without VR (Figure 19). Displacement of spheres in trial 2 without the VR headset is represented by green spheres. Red and Blue spheres represent Trials 1 and 3 respectively, both performed with a VR headset. The participant's head is marked in grey. The green spheres can be seen to be closer to the head than the red or blue spheres.



A B

Figure 19: Top-down plot of virtual spheres displaced by two different participants (A and B) to compare performance across all three trials

The non-parametric permutation test showed that there was a significant difference ($p<0.05$) in the performance between trials with and without a VR headset. However, there was no significant difference between the trials with the VR headset (Trial 1 and Trial 3). It can be seen that on average the subject pushed the virtual spheres further during Trial 1 and Trial 3 than during Trial 2. This is despite the possible accumulation of fatigue in Trial 3.

Table 3: Permutation Test P-Values across the three trials. * Represents a significant Difference

	T1 (VR)	T2 (No VR)	T3 (VR)
T1 (VR)		p<0.05 *	p>0.05
T2 (No VR)	p<0.05 *		p<0.05 *
T3 (VR)	p>0.05	p<0.05 *	

Figure 20 shows the average displacement of virtual spheres during baseline task across all six subjects. The average displacement of virtual spheres is significantly larger in VR trials than in Non VR trials for all subjects. On average displacement of spheres with VR was 41.0% higher than without VR.

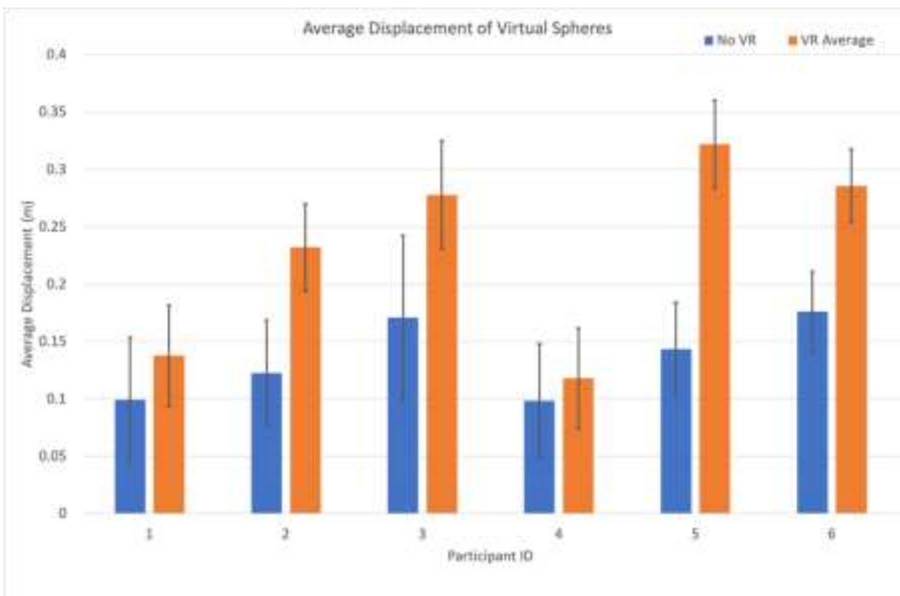


Figure 20: Average Displacement of Virtual Spheres across subjects. Results were significant for all subjects (p<0.05)

3.1.2 Velocity during Baseline Task

Figure 21 shows the combined average velocity of the hand for each trial for all participants. The average velocity was 13.9% and 14.2% higher in Trial 2 and Trial 3 respectively compared to Trial 1.

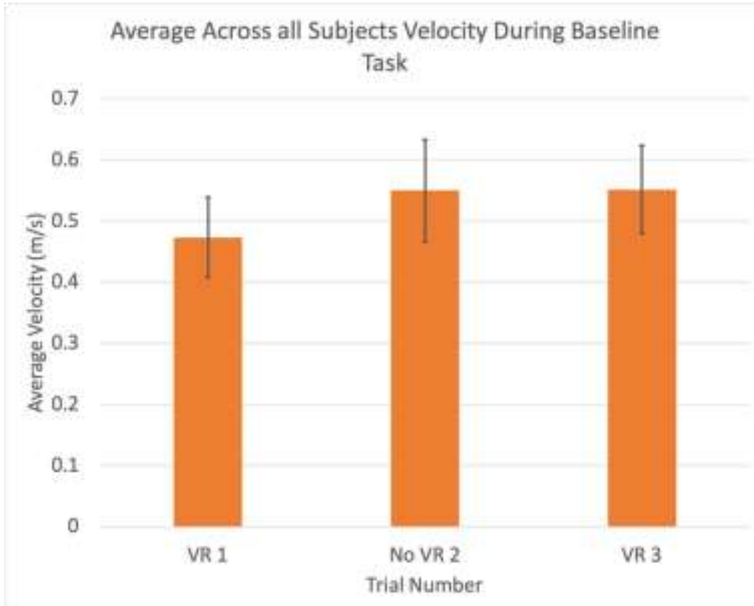


Figure 21: Combined average velocity during baseline task across all subjects.

3.1.3 Reported Fatigue Levels

On average participants reported the highest level of fatigue after the non-VR trial (3.42 ± 0.45). However, they also reported high fatigue after trial 3 which can be accounted for accumulation of fatigue across the three trials.

Table 4: Average fatigue reported by participants after each trial

	Average	SD
Trial 1 (VR)	2.67	0.75
Trial 2 (no VR)	3.42	0.45
Trial 3 (VR)	3.00	1.15

3.2 Extracting High Frequency Areas of Motion

Areas of comfort were extracted from tracked hand motion using the baseline tool during VR sessions. Areas of frequent motion were identified using kernel density estimation.

3.2.1 Kernel Density Heatmap

Figure 22 shows the heatmap generated using kernel density estimation. We can observe the various arm movements and the density of movement (color) associated with it. Density refers to how often the participant visited that 3D coordinate. The bright red area near the bottom of the figure has the greatest density and corresponds to a position of rest. This was also confirmed through the mixed reality video where we were able to observe the position of the subject's hand at various times and relative to the displacement of the virtual spheres in the baseline tool. Interactive web-based 3D graphs were generated to allow for easy sharing with clinicians.

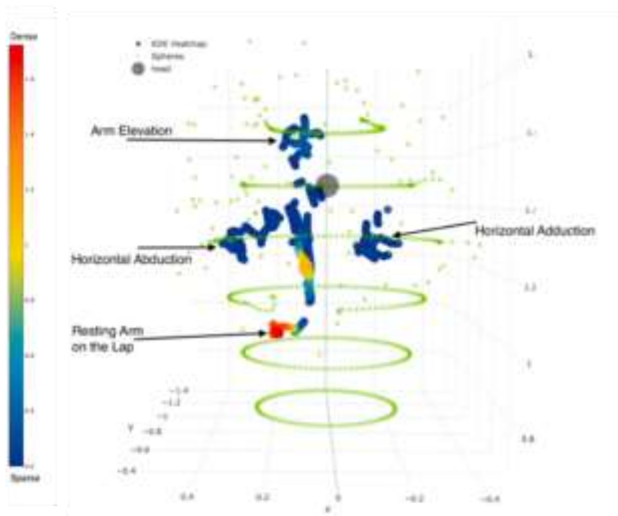


Figure 22: . Heatmap generated by kernel density estimation showing areas of high frequency movement of the arm

3.2.2 Extracting and Assessing Comfort Areas – A Case Study

K-means were applied to identify clusters within the kernel density heat map that represented areas of comfort areas and the areas of least comfort. Two clusters were observed within the high-density values of 30% were deemed areas of comfort. Areas of least comfort were identified as two clusters with the least 2% density values. The participant chosen for this study could reach all four areas, but found that movements in the lowest density areas to be more fatiguing and typically preferred not to make those movements. These areas henceforth are termed areas of discomfort. These four clusters are labelled in Figure 23.

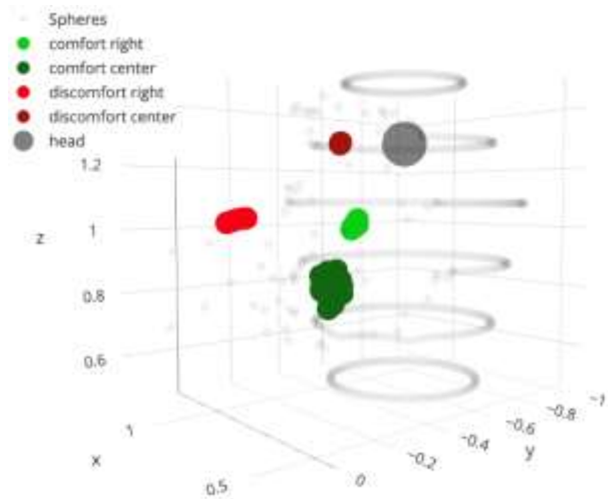


Figure 23: Clusters of comfort positions shown in shades of green and positions of least comfort in shades of red

The centroids of these clusters were calculated and Unity3D was used to render them as appropriately colored spheres, light green, dark green, light red and dark red (Figure 23). Using the mixed reality set up, rapid localization and verification of the target locations relative to the user could be performed (Figure 24).

Figure 24: A mixed reality photo showing a user and the identified areas of comfort and discomfort as virtual colored spheres

3.2.2.1 Torque Calculation for assessment of Comfort Areas

The total static torque at the shoulder was calculated for the four identified locations and presented in Figure 25. It was observed that the total torque calculated was higher in the areas of discomfort compared to the comfort areas. On average the calculated shoulder torque for the areas of discomfort was 47.5% higher than the comfort areas.

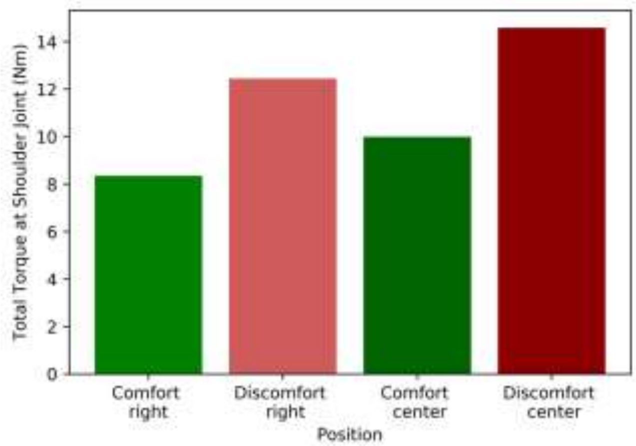


Figure 25: Total static torques at the shoulder for each of the four locations.

3.2.2.2 Performance at various areas of comfort

Figure 26a shows the average task completion time and Figure 26b shows the accuracy for each area. It took the participant approximately 40 seconds to type each word at the comfort areas, while the time taken doubled at the discomfort areas. The one-way ANOVA to compare the time taken to complete the task for comfort and discomfort groups showed a significant difference with a ($p=0.01$). There were zero mistakes made while typing at the comfort areas. Whereas half of the trials at the discomfort areas contained at least one typing mistake

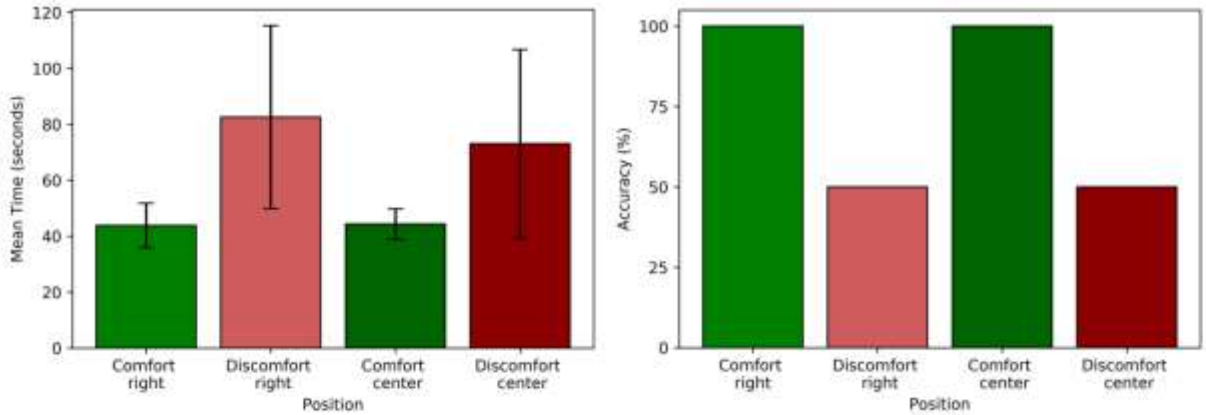


Figure 26: A) Average task completion time of four comfort areas. With standard deviation shown as error bars. B) Accuracy of performance at four comfort areas.

3.2.2.3 Participant Survey Results

Table 5 shows the survey results of the usability questionnaire where 1 is the lowest and level and 5 is the highest level. Based on these results, “comfort right” appears to be the area which is the most comfortable and least frustration.

Table 5: Survey Results

	Comfort	Ease of Reach	Frustration
Comfort Right	5	5	1
Discomfort Right	2	2	4
Comfort Center	4	4	2
Discomfort Center	3	3	3

3.3 Gesture Extraction Tool

Gesture extraction was done based on resting velocity. Figure 29 shows a 2D output of gesture separation over time. The red dashed line represents the resting threshold velocity for this participant. The intersection of the velocity profile with the red dashed lines, marked by grey lines were used as the delimiting points of gestures. These delimiters were then used to separate all the gestures for the participant’s gameplay.

The global gestures separated in a 3D plot is presented in Figure 30. We can see the resting position that was used as the reference point for resting velocity to calculate the threshold. In (Figure 27B) we can see the parts of the motion markers colored blue, indicating end points of a gesture occur closest to the displaced green spheres (indicated by the dashed black box in Figure 28B) , in the positive Y direction. The ends of gestures are clustered either close to the body (closest to the head) or at the bounds of motion away from the body (farthest from the head). These are likely the locations where the participant interacted with the virtual spheres.

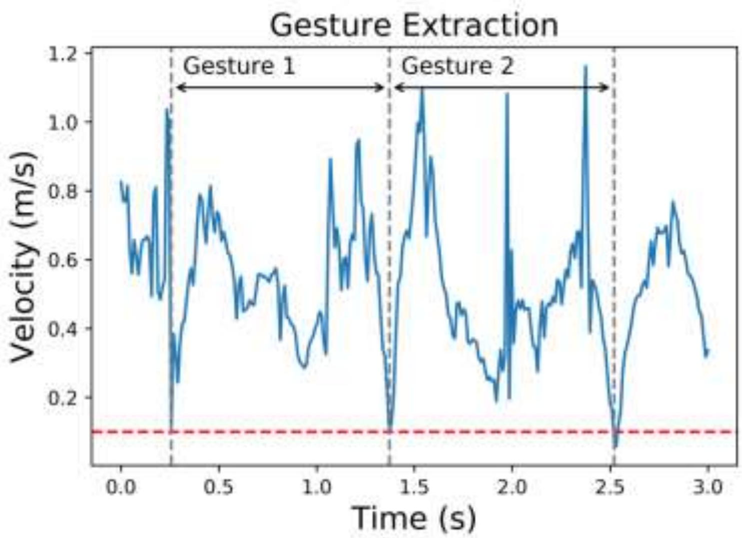


Figure 29: Gesture extraction based on resting threshold velocity.
Red dashed lines represent threshold velocity. Grey dashed lines represent end points of a gesture

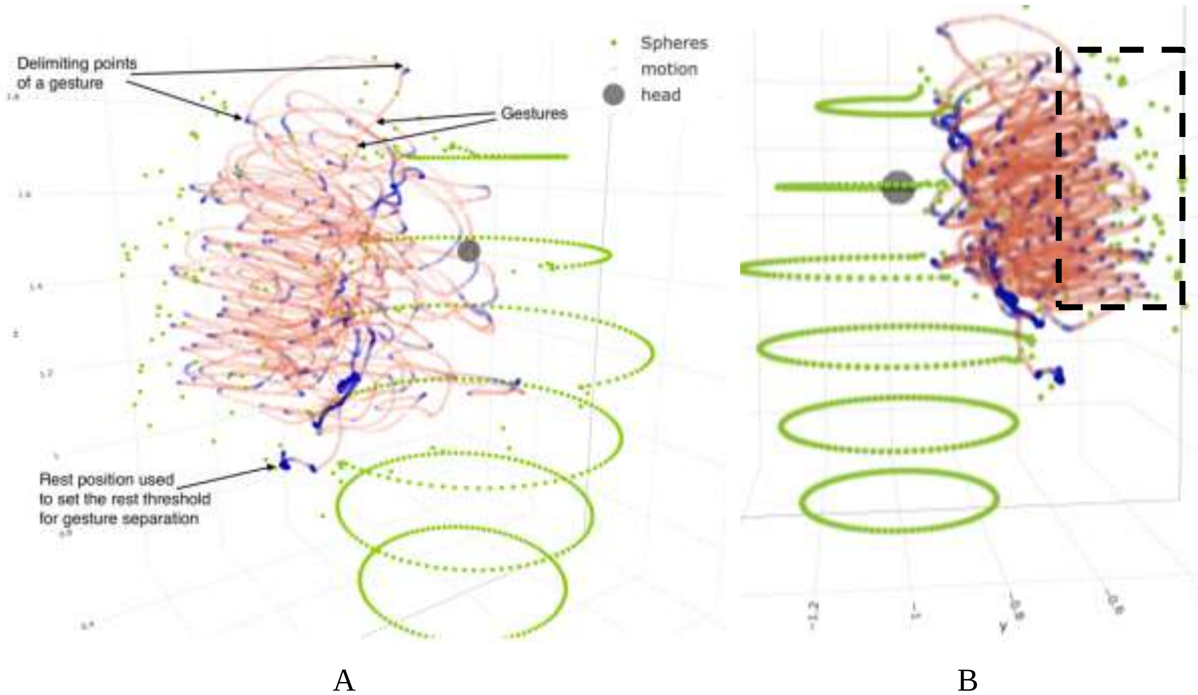


Figure 30: 3D Motion plot with separated gestures in red and delimiting points in blue.(A and B) show different angles of the same graph

3.4 Balloon Exergame

Two different gameplays were explored using the VR system with parameters that could be varied, such as the size of the balloons to be popped and duration or delay for the balloons to pop after being targeted.

3.4.1 Size of Balloon

3.4.1.1 Preference in size of balloons

On average 39.7% more large balloons were popped by the participants than the small balloons. This preference could be due to ease of targeting (Fig. 11). A single factor ANOVA was performed and the p value was found to be 0.07.

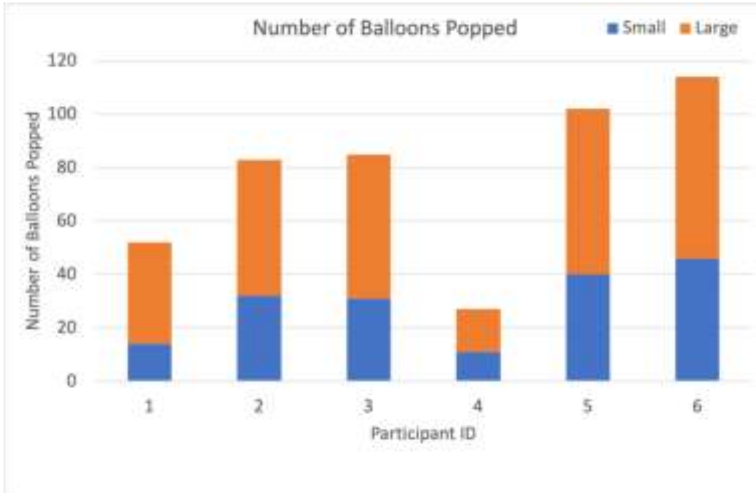


Figure 31: Preference of balloon based on number of balloons popped

3.4.1.2 Velocity changes due to size

Using the gesture extraction tool described in the previous section, we calculated the average velocities over a fixed time window of 800ms around a balloon pop event. The non-parametric permutation test showed a significant difference ($p<0.05$) in the velocities between popping small balloons compared to large balloons (Figure 3212). The average velocity was 25.4% higher for the larger balloons compared to the smaller ones. This suggests the need for more precise but slower movements when targeting small balloons.

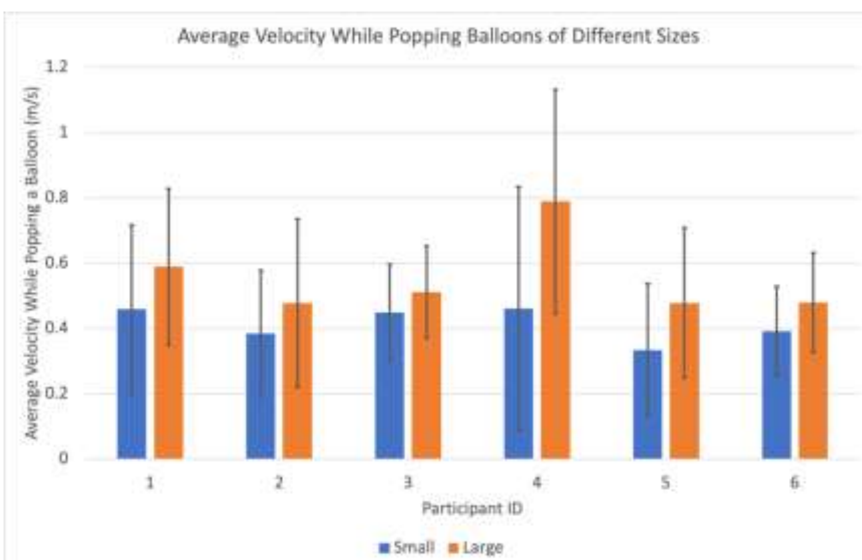


Figure 32: Average velocity while popping balloons of different sizes across subjects.

Figure 36 shows representative velocity profiles for popping events of small and large balloons. The profiles show a trend where the velocity change is sharper at or immediately prior to balloon popping event (indicated by red dashed lines) for larger balloons. In popping events for smaller balloons velocity changes happen over a longer period of time. This suggests a more deliberate approach to targeting of smaller balloons.

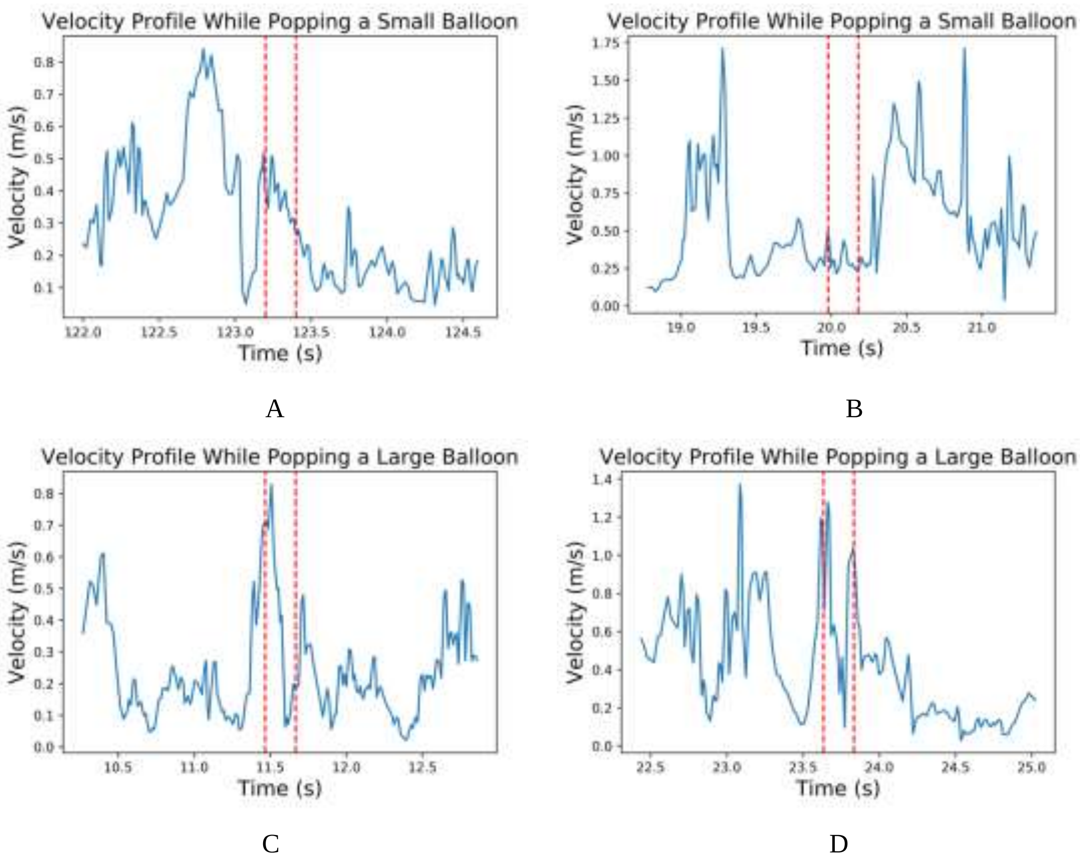


Figure 33: Representative velocity profiles while popping balloons of different sizes. Red dashed line indicates popping event. (A and B) Small Balloons for two different participants. (C and D) Large Balloons for two different participants.

3.4.1.3 Static Force Calculation

Static forces were calculated at the shoulder for the pose required for the hand to reach each balloon. We identified a positive correlation between the total torque at the shoulder and the distance of the balloon from the shoulder ($r^2 = 0.73$). The red markers and regression line represent large balloons whilst the blue markers represent small balloons.

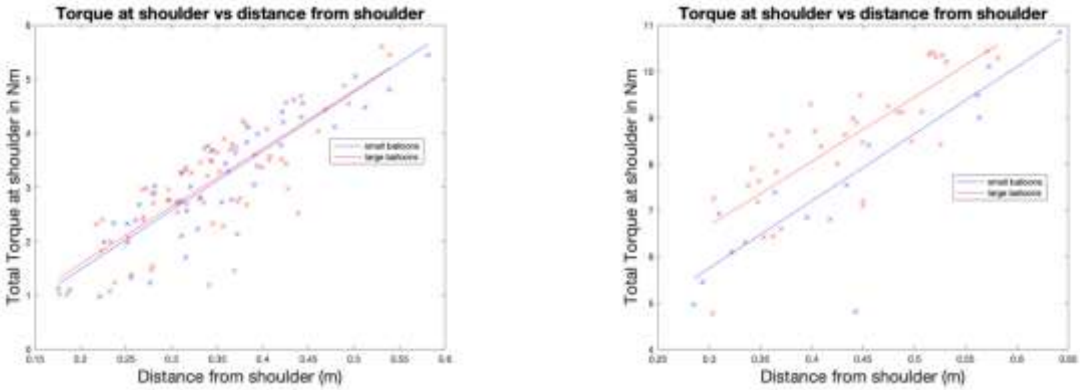
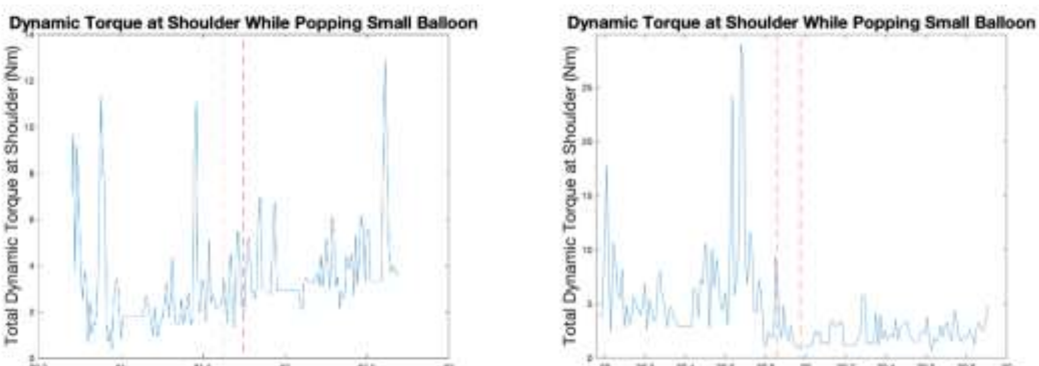


Figure 34: Static Torque calculations at the shoulder for different sized balloons versus distance from shoulder for two different subjects.

3.4.1.4 Dynamic Torque Calculations at the shoulder

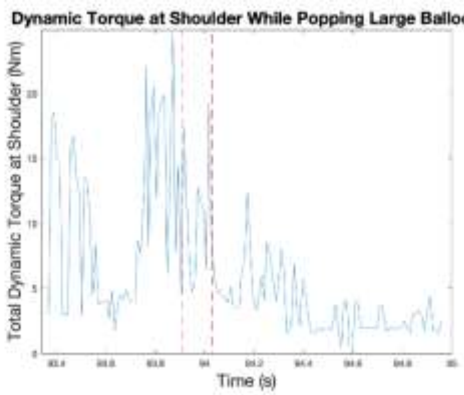
Total dynamic torques at the shoulder during a balloon pop event is presented in Figure 35. In the representative dynamic torque profiles associated with large balloons (Fig. 15 C & D) values were higher immediately prior to the red dashed lines indicating a balloon pop event. In popping events for small balloons dynamic torque changes happen over a slightly longer period of time. However, this was not consistent across all the participants' data.



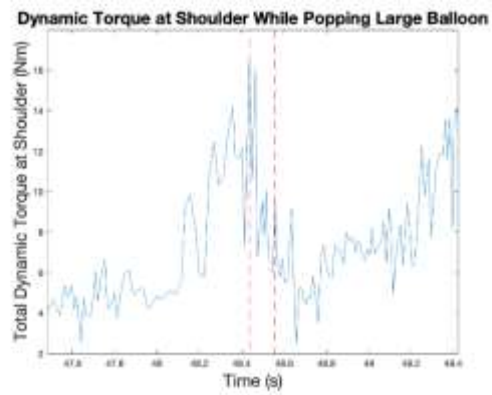
A B

Figure 35: Total dynamic torques at the shoulder while popping balloons of different sizes. Red dashed line indicates popping event. (A and B) Small balloons for two different participants. (C and D) Large balloons for two different participants.

Figure 35 continued



C



D

Reported Fatigue Levels

3.4.1.5 On average the participants reported a fatigue score of 3.0± 1.26 after playing the balloon popping game with different sized balloons. This fatigue score was similar to the fatigue experienced by the participants in the final trial of the baseline tool experiment.

3.4.2 Balloons with Different Pop Delays

3.4.2.1 Number of Balloons popped between different delays

From the data it was determined that on average participants successfully popped 10.0% more short delay (100 ms) balloons than long delay balloons. On average participants had a failure rate of 23.8% of all the long delay (300 ms) balloons attempted. The overall average duration a participant spent inside a long delay balloon before ultimately failing was 190.36ms. The average duration before failure for each participant is presented in Table 3.

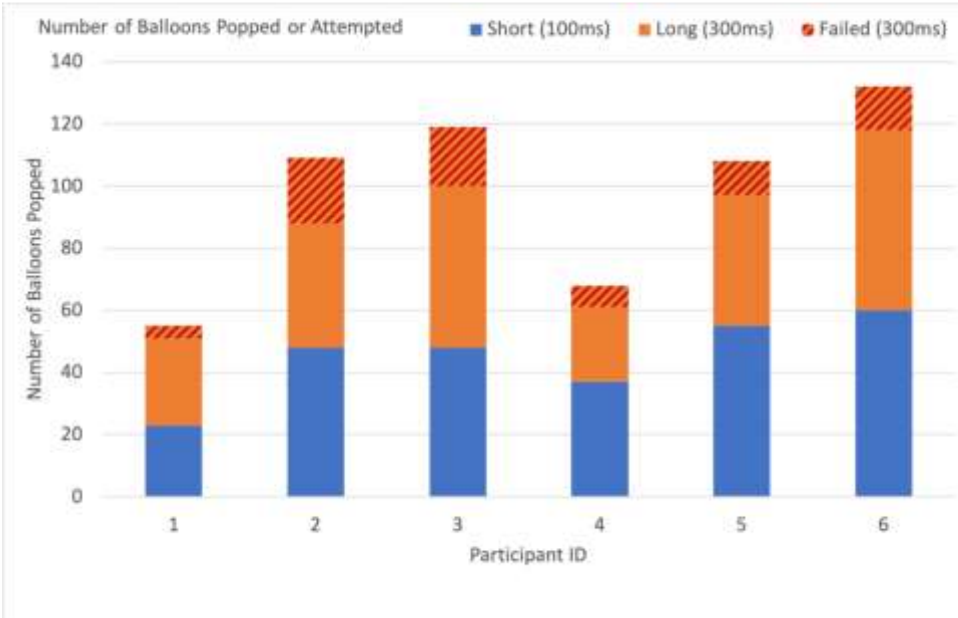


Figure 36: Number of balloons attempted/successfully popped with different pop delays

Table 6: Average duration spent inside a long-delay balloon before failure

	Average duration spent inside a long delay balloon before failure					
Participant ID	**1**	**2**	**3**	**4**	**5**	**6**
Duration (ms)	136.11	203.70	225.73	207.94	179.80	188.89

3.4.2.2

Differences in hand velocity while popping balloons with different pop delays

The non-parametric permutation test showed a significant difference in the velocities between popping balloons with short pop delays (100ms) and those with long pop delays (300ms) ($p < 0.05$) (Figure 37). The average velocity was 33.3% higher for the balloons with short pop delays.

The velocity profiles of the different pop delays (Figure 38) showed a trend wherein the velocity of the hand does not change significantly when popping the balloons with longer pop delays. This trend is due to a longer time spent inside a balloon waiting for the pop. However, in the case of balloons with shorter pop delays, the trend is consistent with previously observed

(Figure 33) velocity profiles with large balloons. In both these cases, the game parameters were identical.

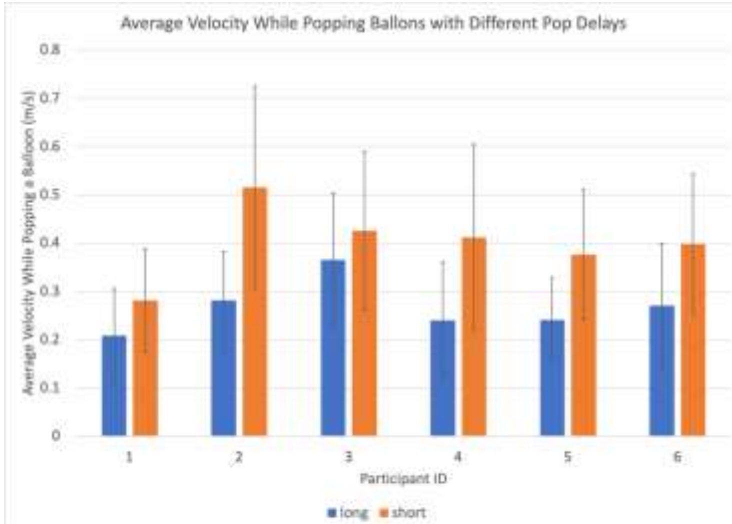
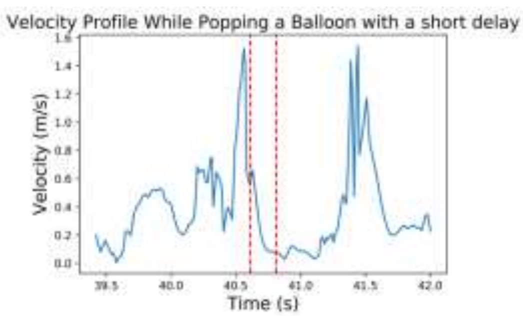
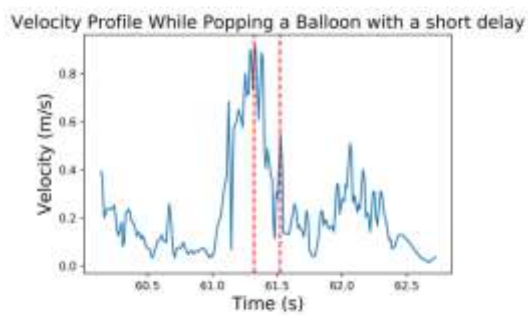


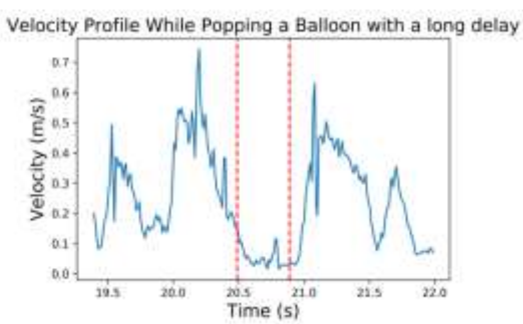
Figure 37: Average velocity while popping balloons with different pop delays



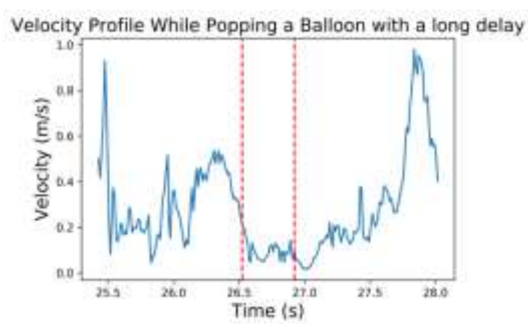
A



B



C



D

Figure 38: Representative velocity profiles while popping balloons of different pop delays. Red dashed line indicates popping event. (A and B) Profiles for short delay to pop. (C and D) Profiles for long delay to pop.

3.4.2.3 Static Force Calculation

We identified a positive correlation between the total torque at the shoulder and the distance of the balloon from the shoulder ($r^2 = 0.62$). The red markers and regression line represent balloons with a shorter pop delay whilst the blue markers represent longer pop delays.

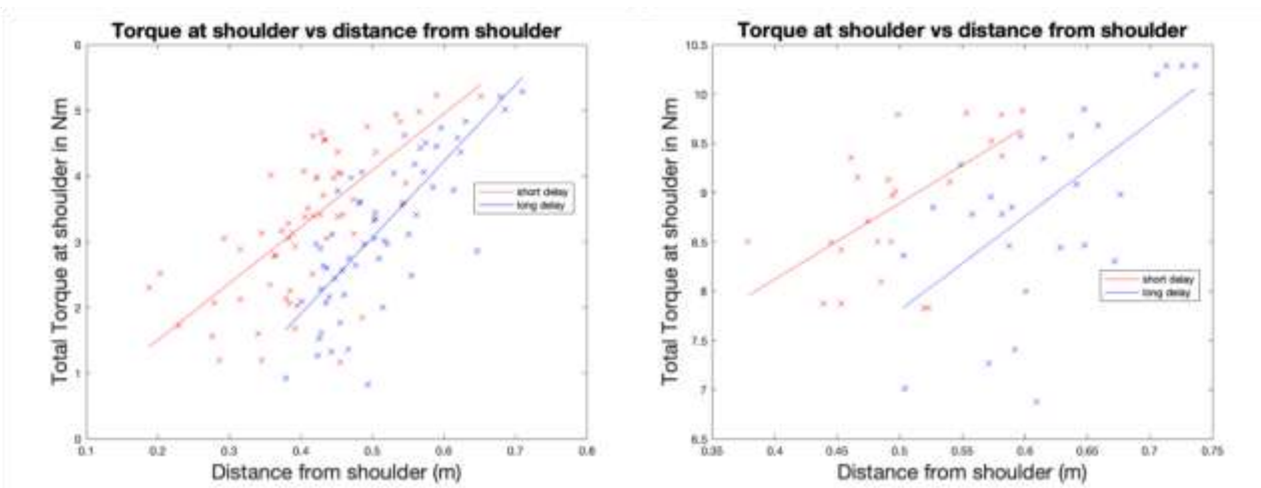


Figure 39: Static Torque calculations for balloons with different pop delays vs distance from shoulder for two different subjects.

3.4.2.4 Dynamic Torque Calculations at Shoulder

Total dynamic torque profiles at the shoulder for the short and long pop delays for different participants (Figure 40). Typically the total dynamic torque was lower during long balloon pop events. With shorter pop delays we saw a trend where the total dynamic torques at the shoulder is high close to the balloon pop event.

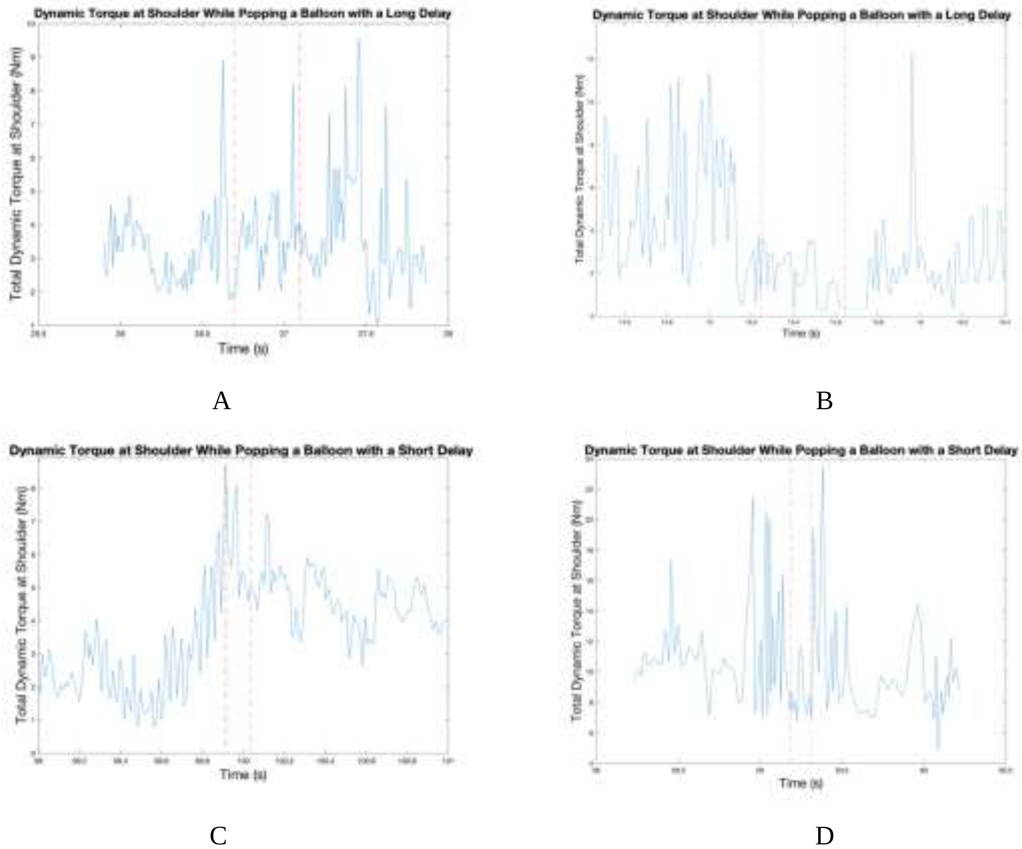


Figure 40: Representative profiles of the total dynamic torques at the shoulder while popping balloons of different pop delays. Red dashed lines indicate popping events. (A and B) Representative dynamic torques for long delay to pop. (C and D) Representative dynamic torques for short delays.

3.4.2.5

Reported Fatigue Levels

On average the participants reported a fatigue score of 3.83± 0.94 after playing the balloon popping with different pop delays. This is the highest fatigue experienced by all the subjects.

3.5

Sensitivity Analysis

A sensitivity analysis was performed on the static and dynamic torque calculation tool. The results are presented in the Table 7Table 8. We see that both the systems are very sensitive to changes in the lengths of both arm segments. The shoulder torques changed by (>30%) which is considerably larger than the corresponding input changes of (±10%).

Changes to the mass of the arm segments resulted in changes that were (>5%) but lower than the changes in input (±10%). Changes to the position of the radius of the arm segment had very little or no effect on both the systems. Changes to the center of mass (COM) of the upper arm caused a larger change than a change in the forearm COM.

Table 7: Sensitivity analysis of static torque calculations, all values percentage change from baseline sum of torques at the shoulder. Values are colored according to the magnitude of difference. Red indicating a large difference and green being small.

	Upper Arm				Foream			
Static	**+10%**	**+1%**	**-1%**	**-10%**	**+ 10%**	**+1%**	**-1%**	**-10%**
Mass	4.56	0.46	-0.46	-4.56	5.44	0.54	-0.54	-5.44
Length	17.84	1.36	-2.28	-35.09	-33.03	-2.23	1.37	15.05
Radius	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
COM (axially)	4.56	0.46	-0.46	-4.56	-0.78	-0.08	0.08	0.78

Table 8: Sensitivity analysis of dynamic torque calculations, all values percentage change from baseline sum of torques at the shoulder. Values are colored according to the magnitude of difference. Red indicating a large difference and green being small.

	Upper Arm				Foream			
Dynamic	**+10%**	**+1%**	**-1%**	**-10%**	**+ 10%**	**+1%**	**-1%**	**-10%**
Mass	3.44	0.34	-0.34	-3.44	6.56	0.66	-0.66	-6.56
Length	3.42	0.42	-7.32	-24.21	-24.91	-5.09	-1.51	8.08
Radius	0.15	0.01	-0.01	-0.14	0.10	0.01	-0.01	-0.09
COM (axially)	4.60	0.45	-0.44	-4.28	0.63	0.06	-0.06	-0.59

4. DISCUSSION

In this thesis, we developed a novel upper extremity movement measurement tool using a commercial VR system to rapidly and objectively measure an individual's range of motion, velocity of movement, and frequency of movements in a three-dimensional space. Further, we developed an exergame with varied and customizable gameplay parameters based on these measures for each individual. Through the exergame we aimed to understand the participants' interaction with the gameplay as well as its effects on their physiology (joint torques). The exergame also explored a number of gameplay parameters that could be adjusted to affect the player's perceived and physiological effort, which can used in the development of an adaptive VR exergame for exercise and rehabilitation.

4.1 Determining Range of Motion - Baseline Activity

The developed baseline movement measurement tool determines the physiological performance of the participants whether interacting in a VR and no-VR environment. In this section, we will discuss the impact of the baseline tool on the performance of participants, the ability of the tool to quantify movements, and potential clinical implications.

4.1.1 Impact of VR on Motivation

We showed that VR has a significant motivational effect on range of motion of upper limbs of individuals with tetraplegia. Participants were able to displace spheres 41% further with the VR headset than without the VR headset. This could be attributed to the visual feedback they received when using a VR headset. The visual feedback of a sphere in 3D space represents an achievable target allowing participants to perform repetitive, goal-oriented movements. This is in contrast to the haphazard motions performed by participants when they have no visual feedback through VR, observed in video footage of participants performing the task. Studies have shown that providing an achievable goal or a visual cue allows for an improved range of motion during upper limb rehabilitation [109][57].

In prior studies with individuals with stroke, researchers found that those who participated in VR enhanced rehabilitation showed greater improvement in upper limb rehabilitation outcomes including range of motion, dexterity and finger control than those who did not participate [110]. Through our baseline tool, we have validated this observation in individuals with SCI. We have also observed that despite repeated trials, the fatigue levels of the individuals during VR tasks was still lower than during the non-VR trial. The self-reported fatigue of the participants suggests that having feedback through VR might affect their perception of fatigue. Therefore, despite the possible accumulation of fatigue, the visual distraction provided by the VR system may have enabled participants to perceive lesser fatigue. This can make monotonous and repeated movements more tolerable and thereby less fatigue inducing [111].

The immersive nature of a VR environment coupled with dynamic virtual targets represented through the virtual spheres allow the individual to continually challenge themselves. The individual can observe the sphere which they pushed farthest and try to beat their own "record" [79]. This is similar to the approach of an arcade game wherein individuals will always try to best the highest score. By adding a personalized competitive element through exergaming adherence to routines such as rehabilitation regimes can be increased in individuals with SCI [55], [112].

4.1.2 Quantification of Movement

The 3D graphs generated from our baseline VR test allow us to visualize the movements of the arm that was tracked. We can observe arm elevation, shoulder abduction, and adduction. Through the observation of the movement, it is possible to conclude if the subject is able to perform motions against gravity which would be at least a 3 on the MMT scale. It is also possible to regularly quantify the changes in the degree of movement through at-home rehabilitation. In addition to this, we can also visualize and quantify the range of motion of individuals which would translate to data that could be comparable to data collected from ROMS.

Through the baseline tool we are also able to determine the velocity of gestures performed by the participants. Increased velocity and duration of gestures have been shown to be significantly correlated with Fugl-Meyer scores in stroke patients [36].

4.2 Comfort Area Detection

Our system allows for rapid identification of areas of comfort for individuals with upper extremity mobility impairments using kernel density estimation dispensing with the need to perform expensive and iterative prototyping. We have demonstrated that VR exergaming provides a quantitative estimation of areas of comfort without having to make inferences about locations of areas of comfort. The joint forces calculated at the areas of comfort were confirmed to be significantly lower than at areas of discomfort. Upon validation of the areas of comfort through joint force calculation this result was applied to a real-world application of scan-select typing, which is commonly performed by tetraplegics. We were able to rapidly position an input device at the various areas of comfort or discomfort to further validate during a case study of a tetraplegic user. This baseline tool would allow individuals with upper extremity mobility impairments to objectively determine the positioning of typical controllers, such as joysticks, accessible buttons, or touchpads, in the most comfortable areas.

4.3 Gesture Extraction Tool

We showed the ability to extract gestures based on the velocity of movement of the participant’s hand. Through gesture extraction we are able to visualize separated gestures in a 3D graph. This can be used by clinicians to better understand the type of motions that are performed by the individual. The tool would allow clinicians to correct the participant if there is an erroneous gesture or a compensatory gesture being performed. Compensatory gestures are often considered a ‘bad habit’ during rehabilitation as it does not allow patients to recover complete usability of the targeted limb/muscle [113].

Moreover, access to individual gestures would enable clinicians to track the progression of their patients’ motor outcomes when performing a known, specific gesture. Through this, the clinician can provide the required feedback to lead to more improved motor outcomes or prescribe further corrective measures.

The gestures associated with a balloon pop event also help in determining the torque at the shoulder. This allows for a more quick and computationally efficient calculation of the torque.

Most algorithms require extensive computational power and time to complete an inverse kinematics and dynamic torque calculation as these are iterative algorithms [98]. Through the gesture extraction method, we could speed up the process of analysis and possibly provide potential implementation of the calculations on a mobile device or portable computer.

4.4 Impact of gameplay parameters on the performance of users

We experimented with gameplay parameters, such as the size of balloons and the delay time in popping of the balloons, to determine the impact of changing these parameters on influencing the overall performance of the participants.

4.4.1 Size of Balloons

Fitts' law states that the time required to rapidly move to a target area is affected by the width of the target and the distance to the target [114]. Fitts' law has been widely used to describe reaching motions and has been applied to a variety of different upper-extremity exercises [114]. With this in mind, the size of the balloon in the exergame was manipulated to test participants to perform more difficult motor skills through the targeting of a smaller balloon. We observed that participants had a significant difference in preference for larger balloons over smaller ones. They targeted almost 40% more large balloons than small balloons. This could be accounted to the higher visibility of the larger balloons and thereby the ease of targeting by participants. Moreover, we observed changes in velocity wherein the participants' change in velocity was sharper at or immediately prior to balloon popping event for larger balloons. However, in popping events for smaller balloons velocity changes happened over a longer period of time. This suggests a more deliberate approach to targeting of smaller balloons. The participants also described targeting the smaller balloons as "requiring more finesse". Likewise, balloon sizes could be increased to improve the success rate of individuals with profound motor impairments to prevent frustration [107], [115].

4.4.2 Pop Delay of Balloons

The participants were presented with an equal number of balloons with short or long pop delays. From the data we observe that on average participants successfully popped 10% more short pop delay balloons than long delay balloons. On average participants had a failure rate of 23.8%

for all the long delay balloons attempted. The long and short delay balloons looked identical, thus balloon-popping differences could not be due to targeting preference. The lack of difference coupled with the failure rate suggests increased difficulty lie in holding the hand in position while tracking the balloon for 300ms to pop. This is in agreement with literature showing static holding task being harder than a dynamic task [116], [117]. It has been shown that dynamic tasks could be performed for a longer duration than static tasks even at higher relative intensities due to contraction induced ischemia leading to increased local lactate levels in shoulder muscles to cause greater fatigue [118].

The results showed that the overall average duration a participant spent inside a long delay balloon before ultimately failing was 190.36ms. This suggests that if the pop delay was lowered to this number, we would observe a greater success rate. We could alter the pop delay based on an individual’s failure rate to ensure a pre-determined success rate during gameplay. It has been shown that a difficult game can easily induce frustration [107], [115].

The results also showed a significant difference in the velocities between popping balloons with short pop delays and those with long pop delays The average velocity was 33.3% higher for the balloons with short pop delays. While popping a balloon with a long delay, the individual needs to slow down and use fine motor control to maintain the hand position inside the balloon. The velocity profile for balloons with short delay is consistent with previously observed velocity profiles as large balloons. In both these cases game parameters are identical.

Manipulating the aforementioned parameters could enable a more adaptive gameplay for individuals performing rehabilitation regimens. A user friendly and intuitive interface would also allow the clinicians to track the progress of their patients and ensure that they are improving their motor outcomes and are being sufficiently challenged. Customizable gameplay parameters can be included in the exergame to allow a personalized rehabilitation regime and future work could involve trials with variations in parameters.

Calculating Static Shoulder Torques

On average, a high positive correlation was determined between the total torque at the shoulder and the distance of the balloon from the shoulder. With increasing distance, the center of mass of the arm is farther away from the shoulder. The change in position of the center of mass 4.5 leads to a mechanical disadvantage due to the increased moment arm. Therefore, the torque required at the shoulder needs to be higher to support the arm at this extended pose.

Through the calculation of torque at the shoulder it is possible to estimate the level of exertion [74]. This would help clinicians develop a better understanding of how challenging a specific movement might be. Alternately, it would be possible to adapt the gameplay to either keep an individual at a constant level of exertion.

The data indicated that there was some difference between the torques at the shoulder for the different sized balloons, but this was not consistent among subjects. In participants where a difference was observed, it suggests that they had a bias to try to pop a certain size of balloon at certain locations or distances away. Therefore, these participants may have different levels of fine/gross motor control at different positions of their arm. In individuals without a difference between the sizes of the balloon it suggests that they did not have a bias towards small or large balloon at certain locations more than others, i.e. they were equally likely to pop large or small balloons at all locations. These individuals might have had better fine/gross motor skills across all locations.

Evaluating static torques for balloons with different delays, we saw a trend where balloons with long delays were generally popped at a longer distance from the individual. This could be due to balloons being popped in positions where skeletal loading could be high, thus reducing the torque on the shoulder. This could also be due to balloons slowly drifting upwards causing participants to track the balloons upwards and further away from the body.

4.6 Dynamic Force Calculations

Evaluating dynamic torques at the shoulder, we observed a similar trend as seen with velocity. The dynamic torques were higher immediately prior to a balloon pop event. In popping events for small balloons dynamic torque changes happened over a slightly longer period of time than for large balloons. For balloons with different pop delays, a trend where the total dynamic torque was lower during long balloon pop events was evident. This finding was not unexpected since while holding the position to pop the balloon, the torques experienced by the shoulder should be lower as there is little movement.

However trends seen in calculated dynamic torques at the shoulder was not consistent across all gestures, this could be due to the large differences in dynamic torques when the arm is moved with or against gravity. The iterative inverse dynamics solver for dynamic force calculations is a numerical tool that is not stable and often fails to converge. At times the inverse dynamics solver converges at extremely large meaningless values which had to be discarded.

Inverse dynamics allows calculation of shoulder torques. This can give clinicians an insight into the level of muscular exertion during gameplay. Through this exergames can be adapted to allow an increase in peak dynamic torque at the shoulder for individuals. This could offer less perceived pain for given intensities compared to tasks that require equivalent static loading of the arm [116][118].

4.7 Sensitivity Analysis of Torque Calculations

A sensitivity analysis of the torque calculation tool was performed by varying several input parameters including arm segment length, radius, mass and center of mass (COM). The input parameter that was shown to be most sensitive was arm segment lengths.

We observe that changes in the lengths of arm segments had the highest impact on the torque calculated at the shoulder. The reason for this change corresponding to change in arm segment lengths could be due to the inverse kinematics solver converging at a solution that is different from the initial pose geometry that was provided.

Two sources of errors in torque calculations that are not captured in the sensitivity analysis are tracked position of the hand and anthropometric data. Past studies have shown that tracking the position of the hand to obtain kinematic data has one of the largest effect on joint torques during a dynamic motion [119]. This could be due to the relative motion between the skin and the tracker [120]. This is unavoidable, it could be reduced my ensuring the hook and loop fastener is securely tightened. Masses and locations of center of masses were obtained from anthropometric data from literature. These are estimated values and could be a source of error.

4.8 Limitations of VR

Virtual reality is a beneficial intervention for various patient populations such as Parkinson’s disease, stroke, chemotherapy, etc. It has been used in various rehabilitation facilities and clinics and its impact has been explored in longitudinal studies. Moreover, assistance is required to wear the headset and supervision may be required in case of adverse events such as nausea due to motion sickness [42]. Despite these limitations, VR is becoming a popular tool for improvements in rehabilitation and developing regimens which complement traditional tools. Trackers used in VR are generally large and heavy. This might limit the duration of gameplay that can be performed by individuals. While VR provides an engaging environment for gameplay and interaction with virtual objects, these virtual objects provide no physical resistance. The user is not able to feel any force while interacting with them. This is unlike any real world interaction and thus performing a task in VR might be not completely transferrable to the real world.

5. CONCLUSIONS AND FUTURE WORK

The system developed in this thesis has multiple applications, including quantifications of range of motion, identification of areas of comfort and determining shoulder torques using gesture extraction.

The higher ROM shown by the participants through the VR system suggests that it might be possible to integrate VR into upper limb rehabilitation that is easily adapted for individuals to use in inpatient, outpatient and home-based care. It could be used as a supplement or alternative to conventional therapy. VR provides a method to encourage exercise and treatment compliance, provide safe and motivating therapy [29].

The system can also be used to determine areas of comfort. It can be used in numerous scenarios that would be difficult to physically replicate and train with, such as optimal placement of steering systems in a modified vehicle. Not only is it expensive to move a steering column or brake and accelerator in different locations, but dangerous to practice driving when experimenting with different placement areas. The system could also facilitate human-robot interaction for users with mobility impairments by providing a more comfortable overlap of human and robot workspaces. Wheelchair mounted robotic manipulators are programmed to manually defined positions to complete activities of daily living. Through use of this system these positions could be automatically generated to allow robotic arms to interact with their humans in a more comfortable fashion [6].

The system can provide an improved rehabilitation experience for persons with tetraplegia in home settings while allowing oversight by clinical therapists. The therapists can automatically receive patients' results and mixed reality videos for evaluation. This would allow patients to get regular feedback from their rehabilitation therapists on their upper limb movements more easily. Mixed reality videos would also allow clinicians to become aware of any compensatory movements, such as use of the shoulder to assist with arm raises, which could improve function in the short term but might be detrimental in the long term [121]. This can lead to telehealth applications wherein clinicians could observe their patients participate in rehabilitation through the VR exergame.

Future Work

In this thesis we present a baseline tool to capture an individual's range of motion rapidly. This data could be used in developing user-specific exergames that ensure virtual targets in the
5.1 game are within the individual's reaching ability. We also explore how gameplay parameters affect individuals' interaction with the exergame. The parameters explored, size of balloons and pop delay, could be altered to change the required exertion levels of players. Games are perceived to be boring if perceived to be too easy [122][106] and frustrating if it is found to be too difficult [107], [115]. Thus it is imperative to operate within their functional reach and within the peak static/dynamic torques to achieve maximal engagement and thus, the therapeutic potential of the exergame.

The baseline tool could be used to rapidly identify areas of comfort to allow gameplay to adapt to the individual by placing stimuli in areas of different levels of comfort to challenge the individual.

A more involved exergame should be developed in the future to allow for engagement over several weeks or months to study the rehabilitative outcomes from an adaptive exergame. An additional tracker at the elbow or a flex sensor to measure the angle at the elbow would allow for a more accurate elbow positioning. A VR system could be placed in a rehabilitation center with adaptive exergames for individuals to access and compete with each other to reap benefits of having a social gaming ecosystem.

APPENDIX A. PARTICIPANT QUESTIONNAIRE

Subject ID: Date:
……………………………………………………………………………………………………
…………………………………………………

Age: Gender:
Weight: Height:
Level of Injury?
Incomplete/Complete (ASIA scale)?
Time post-injury?
Dominant Hand?
After baseline (pushing balls)

1. Was there any fatigue? On a scale of 0-5

	Trial 1 (VR)	Trial 2 (no VR)	Trial 3 (VR)
Fatigue level			

<u>After game 1 (different sizes)</u>

2. Was there a difference between smaller and larger balloons? Eg. Difficulty, exhausting?

3. Do you feel any fatigue? On a scale of 0-5?

<u>After game 2 (different delays)</u>

4. Did you notice a delay in the balloons popping? If yes, would you like to be longer or shorter?

5. Do you feel more motivated to hit the balloons which were further away or did you just wait for them to float to you?

6. What did you think about playing the game? Was it tiring or was it fun/engaging?

7. What do you think of the duration of the session? Can you imagine playing this game regularly? Eg: 3 times a week for about 6 months?

8. What would be a good motivation for you to play a more involved game? Eg: High score, puzzle style?

9. Any other comments/suggestions?

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