Modern Clinical Tools for Interactive Rehabilitation Exercises, Skill Development and Treating Balance Problems

by

Aimee Leigh Betker

A Thesis submitted to the Faculty of Graduate Studies of

The University of Manitoba

in partial fulfilment of the requirements of the degree of

DOCTOR OF PHILOSOPHY

Department of Electrical and Computer Engineering

University of Manitoba

Winnipeg

THE UNIVERSITY OF MANITOBA

FACULTY OF GRADUATE STUDIES

*****

COPYRIGHT PERMISSION

Modern Clinical Tools for Interactive Rehabilitation Exercises, Skill Development, and Treating Balance Problems

BY

Aimee Leigh Betker

A Thesis/Practicum submitted to the Faculty of Graduate Studies of The University of Manitoba in partial fulfillment of the requirement of the degree

Of

Doctor or Philosophy

ABSTRACT

This research focuses on the development of objective clinical tools to treat and evaluate balance and mobility problems via easy to use, motivating, inexpensive, and portable equipment. First, it is known that intensive, task specific exercises promote better rehabilitation outcomes. To enhance the repetitive, task specific exercises, an interactive virtual environment-based tool was developed, which coupled the center of pressure (COP) signal to an on-screen computer cursor (sprite). Two case studies were performed to investigate if our developed tool would improve dynamic balance control during standing and short-sitting (upright torso, sitting on buttocks and/or thighs, and shank hanging over the sitting surface), and would be motivational and challenging. Post-exercise, participants exhibited a lower fall count, decreased COP excursion limits for some tasks, and increased attention span during training. This research contributes a tool that was an improvement to conventional exercise regimes, which allowed even the severely disabled to play, be competitive and improve dynamic balance control.

Second, we investigated the strategies healthy participants used to learn a modified sensory motor transformation in a virtual environment-based task. Specifically, a rotation of 60° was applied to the on-screen cursor movement. We show that two different strategies emerged for adapting to this shift in the spatial mapping, which controlled how the body movement trajectory (biofeedback signal) moved the computer screen sprite (perceived visually). This result is important as it shows that changing the learning strategy can result in better recalibration of internal to external spatial reference frames required to learn task specific visual based rotations viewed in a virtual environment.

Lastly, a model was developed to estimate the center of body (COM) during walking on firm and irregular surfaces. The model was derived using proprioceptive somatosensory information, represented by body segment accelerations, and an external spatial reference, the ground support, represented by the COP. The model is novel in that it does not require any calibration and provides a reasonably accurate estimation of the COM, which can be compared to the brain's balance performance. Hence, this model could be used instead of the cumbersome method of video motion analysis for COM calculation.

for my parents

"When I'm operating under restrictions, I definitely feel constrained by them. But without those restraints it doesn't seem like my actions are accomplishing anything. I guess I'm right back to where I started from."

Motoko Kusanagi

Acknowledgements

First, my gratitude goes to my advisors Dr. Moussavi and Dr. Szturm, for providing me with an enjoyable working environment and always encouraging my professional development. I would like to thank my examining committee for providing helpful feedback on my research. I would like to recognize the Manitoba Health Research Council (MHRC) and Natural Sciences and Engineering Research Council (NSERC) for providing financial assistance. Lastly, I am sincerely thankful to Michael Riediger, who has been the maple syrup in my tea.

Table of Contents

List of Tables

List of Figures

List of Abbreviations

1D	one-dimensional
2D	two-dimensional
3D	three-dimensional
AP	anterior-posterior
BMI	body mass index
C3	third congruent game of balloon burst prior to translation
CA1	first congruent game of balloon burst after translation
CA2	second congruent game of balloon burst after translation
CC	correlation coefficient
CCW	counterclockwise
CD	coefficient of determination
CNS	central nervous system
COM	center of body mass
COP	center of pressure
CS1	standing case 1
CS2	standing case 2
CS3	standing case 3
CSS1	short-sitting case 1
CSS2	short-sitting case 2
CSS3	short-sitting case 3
CW	clockwise
EMG	electromyography
err	percentage error
FSA	force sensing array
IFD	indentation force deflection
L1	spinous process of the first lumbar vertebra
LCD	liquid crystal display
LED	light emitting diode
ML	medial-lateral
NC	not completed
PD	preferred direction
PL	peroneus longus
RCVA	right cerebro-vascular accident
RMS	root mean square
SD	standard deviation
SE	standard error
T1	first translated session of balloon burst
T2	spinous process of the second thoracic vertebra
T23	the average of translated sessions two and three of balloon burst

T45	the average of translated sessions four and five of balloon burst
T10	spinous process of the tenth thoracic vertebra
T11	spinous process of the eleventh thoracic vertebra
TA	tibialis anterior
VT	vertical

1 INTRODUCTION

1.1 MOTIVATION

Balance of the human body requires timely (i.e., a timeline that prevents a fall) control of the position and motion of the centre of body mass relative to the base of support. Many essential sensory and motor processes are involved. Feed-forward predictive controls, which initiate preparatory postural adjustments, are required to maintain balance and to anticipate potential future disturbances. Sensory feedback processes are used in response to unexpected disturbances or when preparatory movements fail; experience plays a role in being able to better anticipate disturbances [Kuo, 1995]. The posterior parietal cortex of the brain interprets sensory information from visual, vestibular and somatosensory inputs, from which the central nervous system (CNS) can take preventative and corrective actions, produce an internal model of the required/desired movement, and mediate between conflicting sensory cues [Shepard and Telian, 1992; Peterka, 2002]. Each sensory input provides unique information regarding the alignment and relative motions of our body and our environment, thus, describing the interaction and transformation of both egocentric (self-to-object) and allocentric (object-to-object) representations on which our movements are dependent. In the absence of one these inputs, balance can still be maintained; however, the compensatory actions become larger.

Many uncontrollable factors can contribute to the degradation of our balance system, such as a decrease in processing efficient sensory information with age and disabling neurological and musculoskeletal conditions [Boult et al., 1994; Kriegsman et al., 1997; Moore et al., 1999; Gill et al., 2001; Harris et al., 2005]. Compounding these conditions are the self-induced functional limitations that are produced from declines in self efficacy. In addition, unconsciously employed compensatory strategies may prevent the individual from recognizing impairment in its early stages [Fried et al., 1991]. Hence, for many of these populations, even small disturbances result in a fall, increasing the likelihood of an injury and solidifying patient dependency in instrumental and basic activities of daily living. This in turn results in reduced physical activity levels and quality of living.

Due to the problems associated with reduced mobility, an effective rehabilitation and assessment regime is an essential part of any prevention and recovery program. There are many aspects to consider when developing these programs. In order for the tools to be available to a wide range of clinics (for example low and moderately funded clinics), the equipment should be easy to use and should not incur a large cost. Flexibility in use, pliability, and portability will permit the equipment to be used on different surfaces, which can emulate outdoor environments. In addition, it allows the equipment to be transported to the patient, rather than requiring them to come to the clinic or to a fixed location. Another important factor to consider is the patient's interest; the use of rewarding, fun, and motivational learning techniques has been shown to improve a patient's motivation to practice and complete the rehabilitation program [Cogan et al., 1977]. Finally, as relationships between the self and the environment are continually changing, it is important that the tool has the ability to challenge this relationship, causing the user to recalibrate internal to external spatial reference frames.

1.2 OBJECTIVES

The ability to develop objective clinical tools to rehabilitate and assess human balance and mobility via easy to use, motivating, inexpensive, and portable equipment will allow for the development of a system that is available to a large population. To this effect, there are three primary objectives of this research.

The first objective investigates if coupling the center of pressure (COP) to interactive virtual environment-based tasks will improve dynamic balance control in standing and short-sitting (torso supported by buttocks/thighs with shank hanging over the sitting surface). It was hypothesized that the motivational and challenging aspects of the virtual tasks would increase the participant's desire to perform their exercises and complete the rehabilitation process. The developed tool would thus facilitate the restoration of weight bearing and balance control for persons with neurological or musculoskeletal disorders. To this effect, four biofeedback-controlled interactive tools were developed, namely Tic-tac-toe, Memory Match, Under Pressure, and Balloon Burst. The feasibility of using the tool was validated via two case studies.

Secondly, using the virtual task Balloon Burst, we investigated the different strategies employed when participants were trained and evaluated under congruent visuo-spatial conditions and those which systematically change the sensory motor transformation. In other words, the internal model used to produce a movement given the received sensory information must change as the sensory information will be changed to have a new spatial representation. The transformed condition changed the manner in which the COP trajectory (biofeedback signal) moved the computer screen sprite (perceived visually). Specifically, the computer sprite's movement was rotated counterclockwise by 60°. The strategies were investigated via movement errors and the preferred direction (i.e., direction sum of the muscle's average amplitude activity) of the main leg muscles responsible for the movements: the tibialis anterior, soleus, and peroneus longus.

The final objective concerns balance assessment, specifically the three-dimensional (3D) center of body mass (COM) trajectory, which provides us with a measure of movement performance and level of stability while performing standing and walking tasks. As an alternative to directly calculating the COM from motion trajectories and anthropometric data, we developed a model to estimate the 3D COM components from accelerometer and COP data, while walking on firm and irregular surfaces. The model output was validated against the video-motion derived COM signal.

1.3 ORGANIZATION

This introductory chapter has described the motivation and objectives of this research. Chapter 2 provides an overview of current rehabilitation regimes and the developed interactive virtual environment-based tool, which maps the center of foot pressure to an on-screen computer cursor (sprite). The validation of the developed tool via two case studies is then discussed. Using a visual rotation applied to the interactive tool's computer sprite, Chapter 3 investigates the strategies used to adapt to the new transformation. Stability analysis through centre of mass estimation while walking on firm and irregular surfaces is described in Chapter 4. Finally, conclusions and future research are discussed in Chapter 5.

2 Interactive Virtual Environment-based Exercise Regime

In this research, the coupling of the COP to control virtual environment-based tasks for dynamic sitting and standing balance exercises is investigated. Evidence from human studies show that goal-oriented, task-specific training improves function and increased training times produces better outcomes. It is thus hypothesized that the motivational and challenging aspects of the interactive tasks will increase the subject's desire to perform their exercises and complete the rehabilitation process. This chapter details the virtual environment and the two case-studies carried out to validate the developed rehabilitation regime.

2.1 BACKGROUND

Conventional therapy programs generally include standing activities, steppers and over-ground and treadmill walking (with and without body weight support). These exercises are performed at a given rate for a specified duration, dependent on the type of disorder [Carr and Shepherd, 1998]. For example, Messier et al. [2000] reported that aerobic walking and strength training programs helped decrease the postural sway (movements made to correct postural imbalances) of elderly subjects with knee osteoarthritis. This type of exercise becomes important when we move from standing balance control

This material is reprinted from: 1. Archives of Physical Medicine and Rehabilitation, 87/8, Betker AL, Szturm T, Moussavi Z and Nett C, Video Game-Based Exercises for Balance Rehabilitation: Case Study, 1141-9, Copyright 2006, with permission from Elsevier. 2. Betker AL, Desai A, Nett C, Kapadia N and Szturm T, Game-Based Exercises for Dynamic Short-sitting Balance Rehabilitation of Subjects with Chronic Spinal Cord and Traumatic Brain Injuries, Physical Therapy, 2007, 87, with permission of the American Physical Therapy Association. This material is copyrighted, and any futher reproduction or distribution is prohibited.

(stationary base of support) to situations where the COM must remain within a moving base of support (e.g., walking).

Recovery of function, long-term maintenance and neuro-adaptation are strongly influenced by regularity and volume of training and physical activity, and goal-oriented, task specificity [Carr and Shepherd, 1998; Dean et al., 2000; Remple et al., 2001; Richards et al., 2004; Kwakkel, 2006]. This process is governed by neural reorganization, which is the foundation for learning and acquiring new skills [Carr et al., 2006]. Following these guidelines can result in functional improvements in the spinal cord, even in the case of a complete lesion [Rossignol, 2000; Johansen-Berg et al., 2002; Dietz and Harkema, 2004; Edgerton et al., 2004; Schaechter, 2004]. For example, Dean et al., [2000] studied the effects of using a circuit of task-oriented exercises, along with group races and relays. They found that the subjects exhibited improvement in walking speed, endurance and force production in the affected limb; these findings remained consistent when subjects were retested two months post-treatment.

One method of applying intensive, task-orientated training to the upper or lower limbs is constraint-induced movement therapy; for many central nervous system disorders, this method of therapy is essential [Taub et al., 1999; Marklund and Klassbo, 2006]. Constraint-induced movement therapy restricts use of the unaffected limb, forcing the participant to use the affected limb. For example, Marklund and Klassbo [2006] applied constraint induced movement therapy to chronic stroke patients, targeting the lower limb. The knee of the less-affected limb was restricted, requiring the tasks to be performed with the affected-limb. Results demonstrated that the subjects showed improved dynamic balance and motor ability.

The potential effectiveness of time-intensive therapeutic exercises can be impaired by lack of interest, motivation attention span, causing a treatment program to go uncompleted. This is particularly true when a large volume of practice is essential, as is the case in many central nervous system disorders. Conversely, the use of rewarding activities has been shown to improve a patient's motivation to practice [Cogan et al., 1997]. It has been shown that learning complex tasks such as Tai Chi and golf can improve joint proprioception [Tsang and Hui-Chan, 2004]. Tai Chi and golf have the added advantage that they are enjoyable and motivating. The joint proprioception acuity and standing stability bounds of experienced elderly Tai Chi practitioners and golfers were compared to healthy elderly and young adults with similar physical activity levels. They found that both Tai Chi and golf improved the dynamic standing balance control. However, not all patient populations would be able to complete these tasks; thus, another method of providing motivation must be found.

Biofeedback augmented training presents a biological signal to the subject in a simplified format and has long been used clinically to enhance movement, weight bearing status (i.e., distribution of body weight over the lower limbs), and balance awareness [De Weerdt and Harrison, 1985; Glanz et al., 1997]. While the subject is performing a task, a biological signal is recorded and presented to them (in real-time). In most cases, the feedback of the input signal intensity is represented in a visual or auditory format. A commonly incorporated biofeedback signal and outcome measure is the center of pressure (COP) [Shumway-Cook et al., 1988; Lee et al., 1996; Cheng et al., 2001; Geiger et al., 2001; Bourbonnais et al., 2002]. Instabilities will be manifested in the resulting COP signal as a reflection of the global effects of a balance disturbance and reaction.

Thus, the amount of COP displacement and its trajectory are important, providing a measure of stability which can be used human postural control system evaluation. For example, in Zambarbieri et al., [1998], pressure insoles were used to determine the COP trajectory. A tactile stimulation device was then used to inform the subject when their COP fell outside an accepted normal range, in order to indicate proper posture. The amount of load for each leg measured via a force plate was displayed on a light emitting diode (LED) in [Lee et al., 1996; Cheng et al., 2001], with a line indicating if the weight was equally balanced or towards which leg the balance was skewed. The device served to promote symmetrical weight bearing between the two lower limbs. In Femery et al., [2001], vertical forces were measured below the foot using 8 force sensing resistors. A visual display gave a footprint of the sensors, with a sound being played if the amount of pressure exceeded a predefined threshold. The goal of this device was to alert patients with peripheral neuropathy of excessive pressures, in order to prevent ulceration. Similarly, in Urhan and Dincer, [2001] force transducers woven into a fabric heel insert were used to measure the force at the heels. The measured force was displayed on a liquid crystal display (LCD), with an audio tone indicating if the correct force was being applied. The goal of this device was to prevent children with cerebral palsy from solely walking on their toes. In Rougier, [2004], anterior-posterior (AP) and medial-lateral (ML) displacements of the COP were measured via a force platform during standing and their trajectories were displayed on a computer screen. Subjects were asked to keep the COP movements to a minimum. Different delays were also added to the display to determine their effects on balance. This association between the pressure signal and the visual/auditory feedback strengthens/creates awareness of a given activity or performance level; in most cases, to regain or learn an activity that would assist with daily life activities.

The biofeedback should be applied to functionally relevant (i.e., indicative of overall balance) dynamic movements, in order to be effective [Huang et al., 2006]. Novel and promising methods of applying biofeedback to rehabilitation are virtual reality (VR) and video games [Bach-y-Rita et al., 2002; Sveistrup, 2004]. In Webster et al., [2001], a virtual environment was created to help people with the control and mobility of their wheelchair, where the participants had to navigate through a virtual obstacle course. Navigation was first done using a hand-held 4 button controller, one button per movement direction; next, two foot pedals and a wheel were used. Post-treatment, participants exhibited a decrease in wheelchair accidents and falls; participants also had superior performance on an actual obstacle course, compared with subjects who did not have training with the virtual course. Using video games on a regular personal computer has the benefit of eliminating the need for elaborate and expensive VR display systems. The link to computer games engages the patient in the process of practice, in order to make it attractive and prevent boredom; i.e., the idea of *motivated use* to increase volume of practice and retain the patient's attention. In turn, this can increase practice time, volume and therefore, recovery [Malone, 1982; Cunningham and Krishack, 1999; Bach-y-Rita et al., 2002; Lazzaro and Keeker, 2004]. O'Connor et al. [2000] attempted to increase manual wheelchair participants' physiologic response and examine their effect on the participant's motivation to perform their exercises. The GAME$^{\text{Wheels}}$ system interfaced custom dynamometer rollers and optical sensors, which allowed stationary propulsion of the participant's manual wheelchair, to commercial video games [O'Connor et al., 2000]. The system setup allowed as input to the games the propulsion velocity and direction. Results of their study showed that 87% of the subjects found that the games motivated them to perform their exercises. NeuroCom's Balance Master (NeuroCom International, Inc. Clackamas, OR, USA) provides three center of gravity controlled games that are used to enhance practice and motivation: a puzzle, NeuroPong, and Solitaire. The center of gravity, representing the vertical projection of the COM, is measured via a custom strain-gauge force plate, where single-segment inverted pendulum dynamics are considered. NeuroCom suggests that these motivational rehabilitation exercises increase recovery. However, a drawback to these games is the need for additional control via a computer mouse, a device which provides additional tactile (touch) information and may not be usable by all patients.

Thus, the main goal of this study was the development of an interactive tool consisting of COP-controlled virtual environment-based tasks. The environment should be interactive, be cognitively engaging, involve motor learning tasks, be applicable to a wide arrange of dynamic situations (i.e., sitting, standing and walking tasks). Based on these concepts, four interactive virtual environment-based tasks were created, which are controlled via COP signal biofeedback. We hypothesized that the inclusion of motivational and functional tasks to rehabilitation and sport training should increase the participant's desire to perform their exercises; therefore, the participants should exhibit improved dynamic balance control post-exercise. Lastly, we employed equipment that was easy to use and portable, while minimizing the cost, such that the tool would be available to a larger range of clinicians in their daily practice.

2.2 INTERACTIVE COP-CONTROLLED VIRTUAL ENVIRONMENT-BASED EXERCISE REGIME DEVELOPMENT

The interactive tool consists of the developed virtual environment-based software (i.e. the tasks Tic-Tac-Toe, Memory Match, Under Pressure, and Balloon Burst), which is integrated with an off-the-shelf pressure mapping system. The following subsections describe the interactive tool, including details of the virtual environment-based exercise software's integration with the pressure mat and the development of the stand-alone software applications Tic-Tac-Toe, Memory Match, Under Pressure, and Balloon Burst.

2.2.1 SYSTEM INTEGRATION IN THIS STUDY

The developed interactive tasks are controlled via the participant's COP trajectory, which is acquired from an off-the-shelf thin, flexible pressure mat (Vista Medical Inc., Winnipeg, MB, Canada) (Fig. 2.1). Unlike force plates, the pressure mat can be made to any size (including insoles) and may be used on many different surfaces, which can challenge dynamic balance. The pressure mat selected for use in this study was of dimension 53 cm x 53 cm x 0.036 cm and contained a 16 x 16 grid of piezo resistive sensors (total 256 sensors) spaced 2.86 cm apart. The pressure mat uses a calibration file to maintain sensor calibration to vertical pressures of 300 mmHg. The pressure mat unit is portable and easy to use, making it an ideal system for use in routine clinical applications. Each task was developed as a stand alone application, which used a dynamic linked library (dll) file provided by the pressure mat manufacturer to acquire the COP signal from the data acquisition box. The position of the COP is calculated from the vertical loads distributed on the pressure mat. This physical COP position signal is then mapped as input to each of the tasks, in order to control the on-screen cursor (sprite).

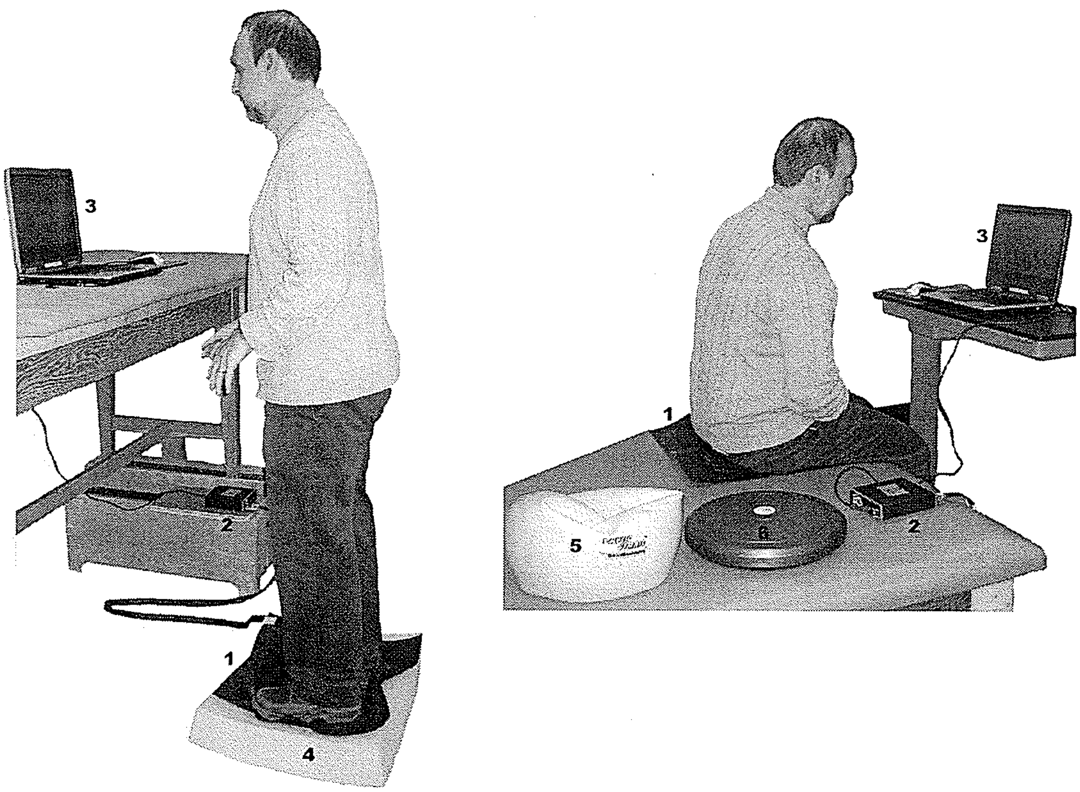

Fig. 2.1. System integration shown for standing and sitting exercises: (**1**) subject stands/sits on the pressure mat, which is connected via (**2**) the data acquisition box to (**3**) the computer. The computer stores and runs the developed interactive tool's software tasks. Example surfaces the pressure mat may be placed on top of are (**4**) a foam pad, (**5**) a PhysioGymnic ball, or (**6**) a SwisDisk.

2.2.2 SOFTWARE DESIGN

Four different interactive and dynamic tasks were created and programmed as stand alone applications: Tic-Tac-Toe, Memory Match, Under Pressure, and Balloon Burst. The programming of the tasks was done using the Visual Studio C++ compiler version 7.1 (Microsoft Corporation, Redmond, WA, USA). The software tasks acquire the COP signal via a *dll* provided by the mat manufacturer (Vista Medical Inc., Winnipeg, MB, Canada). Note that as the *dll* file was created specifically for this research, the subsequent code developed in this research is proprietary. The goals of each task are completed by controlling the on-screen position of the participant's COP trajectory, which is acquired from an off-the shelf pressure mat via a *dll* file. These applications were designed to fully exercise the participant's current range of movement, with the goal of increasing their movement range and speed. To this effect, each task has parameters that can be configured to appropriately map the participant's movement range to the on-screen coordinates, introduce different levels of difficulty, and produce cognitive distractions. Furthermore, additional difficulty levels were achieved through the placement of the flexible pressure mat on irregular and compliant surfaces, which increased the task's balance requirements. The following subsections describe the bound initialization routine (common to all tasks) and each of the four tasks.

2.2.2.1 Movement Range Mapping

To maximize the on-screen range of the participant's movement, the range of the participant's physical COP is mapped to the on-screen pixel range. The steps for mapping the participant's movement range are as follows, and are indicated to the participant via an on-screen instruction display. First, the center point coordinate of the participant's stance is determined. The participant remains still on the pressure mat in the position which they will complete the task for a duration of 5 seconds. Their COP coordinate is averaged over the 5 second time frame and recorded as the center point coordinate. Next, the peak amplitudes of self-induced oscillatory movement about the center point coordinate are detected, in the anterior-posterior (front to back) and medial-lateral (side to side) planes. The peaks are recorded as the average of five points on either side of the center coordinate in each plane. Finally, the bounds are displayed on the screen.

As the participant increases his/her movement capabilities, his/her movement ranges should increase to optimal levels. As the determined ranges are average values, with the speed of the oscillations playing a role, the ranges can be scaled by a percentage value. In cases where the participant's weight distribution between his/her legs is asymmetric, the COP will be biased towards the weight-bearing leg. Similarly, this can be true with improper balance between toes and heels. Thus, another feature offered is the ability to offset the center value. Lastly, the COP movements can be smoothed over a given number of samples to reduce jerkiness in the computer sprite movement.

2.2.2.2 Tic-Tac-Toe

The goal of Tic-Tac-Toe (Fig. 2.2a) is to create a line, with the computer serving as the opponent. The participant selects a square in which to place the computer sprite by shifting his/her weight to move the on-screen COP indicator to the desired square. The participant's movement range and speed are exercised in both the AP and ML directions concurrently. A display in the upper left hand corner of the screen contains light indicators, displaying what stage the game is at:

Red Signifies that it is the computer's turn.

Yellow Signifies that the participant may move the computer sprite to the square he/she would like to mark. The number of seconds he/she has left to move the sprite is also indicated. A sound is played for each elapsed second while the participant is selecting a square to mark.

Green If it is the participant's turn, the square that the computer sprite is on is selected and an X is placed.

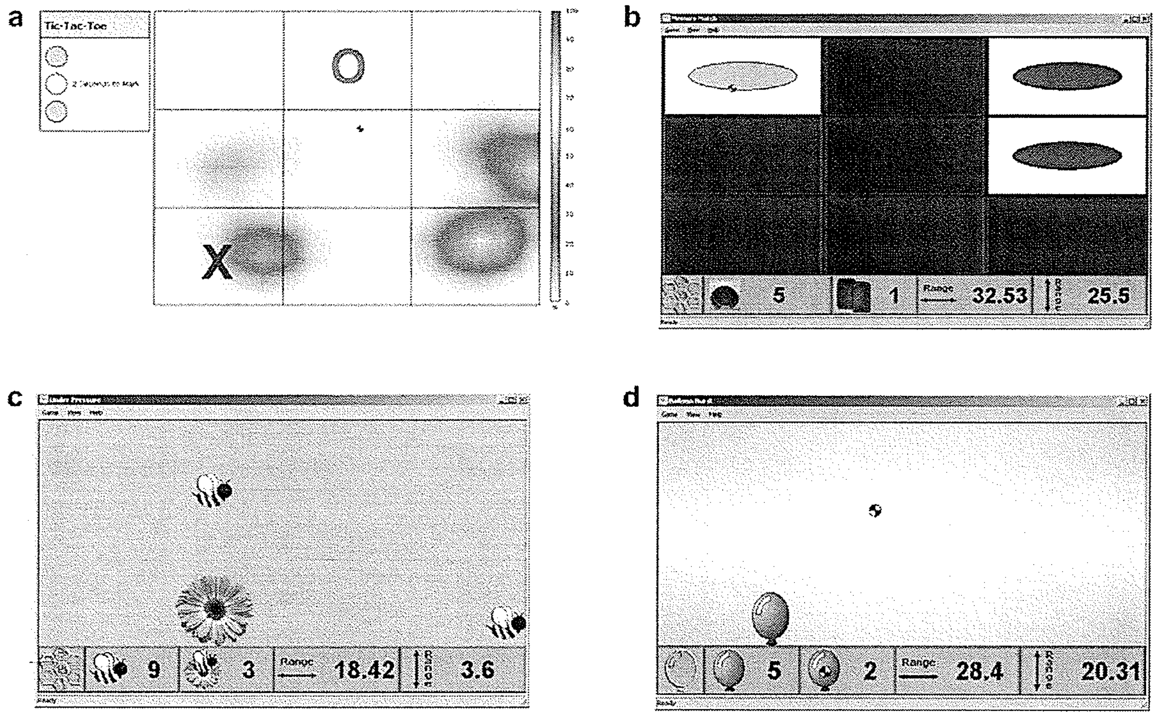

Fig. 2.2. Screenshots: (**a**) Tic-Tac-Toe: the participant has 2 more seconds to select their square, as indicated by the yellow light; (**b**) Memory Match: the number of pairs found and the ranges are displayed; (**c**)Under Pressure: the total number of bees, number of bees caught, and ranges are displayed; and (**d**) Balloon Burst: the total number of balloons, number of balloons popped, and ranges are displayed.

If the participant selects a square that has previously been marked, the game returns to the Yellow state. Positive reinforcement is provided to the participant via:

1. A sound played for each elapsed second while the participant is selecting a square.
2. If the participant wins, the statement "You Win!" appearing on the screen, with a red line drawn through the winning squares.

The number of seconds the participant has to select their card is configurable, in order to allow the participant/physiotherapist to select the appropriate exercise protocol. Performance is measured by completion of the game.

2.2.2.3 Memory Match

In Memory Match (Fig. 2.2b), the goal is to select two matching cards from a 3-by-3 or 4-by-4 array of squares. The participant selects a card (square) by shifting his/her weight to move the computer sprite to one of the nine or sixteen possible cards (squares). Once the COP is held still in a square, for a user selected duration, the card is revealed. The second card is then selected in a similar manner; if the cards match, they remain face up. This process is repeated until all of the card pairs are selected. Difficulty levels are configurable through:

1. The number of seconds the participant has to select his/her card.
2. The number of cards displayed: 9 or 16. A cognitive difficulty is added when 9 squares is selected, as 1 card will be without a match. Then, when the number of cards is increased to 16, the area the COP must be in to select the card is smaller and thus the COP movement must be more precise.

This game attempts to provide an increased range and speed of movement in the AP and ML directions concurrently.

The Memory Match display is similar to that for Tic-Tac-Toe, showing the number of pairs found and the light indicators for what stage the game is at:

Red Signifies that the game is checking if a pair has been found. When a pair is found, a sound plays and the score is updated.

Yellow Signifies that the participant may move the computer sprite to the square he/she would like to mark. The number of seconds the participant has left to move the sprite is also indicated. A sound is played for each elapsed second while the participant is selecting a square to mark.

Green The card that the computer sprite is on is selected and the object on the card is displayed.

If the participant selects a card that has previously been selected, the game returns to the Yellow state. The game is over when all the pairs have been found. Performance is then measured by successful completion of the game. The ML and AP movement ranges are also displayed and logged in a report at the end of the game.

2.2.2.4 Under Pressure

In Under Pressure (Fig. 2.2c), the participant shifts his/her weight to move a receptacle in order to catch an object. The game is comprised of three modes:

1. Horizontal: the participant must shift his/her weight side to side.
2. Vertical: the participant must shift his/her weight back and forth.
3. Both: the participant must shift his/her weight in all directions.

Thus, movement range and speed in all or targeted directions are exercised. Difficulty levels are configurable through:

1. The receptacle size.
2. The object speed.
3. The number of objects.
4. The option of multiple objects appearing at a specified interval.

A display shows the participant how many objects he/she has caught, the total number of objects that have appeared, the ML movement range, and the AP movement range. The game is over when the number of objects reaches a pre-specified maximum number.

Performance is then measured by how many objects are caught; ideally the participant wants to catch all of the objects. The ML and AP movement ranges are also displayed and logged in a report at the end of the game.

2.2.2.5 Balloon Burst

In Balloon Burst (Fig. 2.2d), the goal is to *pop* balloons. The stationary balloons appear at either random or pre-determined locations on the screen. The participant must shift his/her weight in all directions, in order to move the computer sprite over the balloon to pop it. Thus, movement range and speed in all directions are exercised. Difficulty levels are configurable through:

1. The size of the balloon.
2. The duration for which the balloon appears.

Performance is measured by how many balloons are popped; ideally the participant wants to pop all the balloons. A display shows the participant how many objects he/she has caught, the total number of objects that have appeared, the ML movement range, and the AP movement range. The game is over when the number of objects reaches a pre-specified maximum number.

2.2.2.6 Comparison of Developed Virtual Tasks

In the tasks Tic-Tac-Toe and Memory Match, participants move their body to produce quick, short movements, followed by a sustained hold in the position of the computer sprite for a user defined number of seconds. Memory Match has an additional cognitive aspect, in that the participant must remember the location of each card in order to match the cards. In order to select a square (Tic-Tac-Toe) or card (Memory Match), the participant must maintain the position of the computer sprite within the virtual bounds of the square/card. Thus, the accuracy of the movements has a low precision. Increasing the precision demands, while still requiring the participant to sustain his/her movements at a particular virtual location, is the task Under Pressure. In Under Pressure, the participant must move the receptacle in line to catch a moving object. Moving to a more dynamic task, the participant may play Balloon Burst. Here, quick and precise movements are required to move the computer sprite to intersect and pop the stationary balloon. Balloon Burst includes an additional mode to promote the learning of a new coordinate system and logs the balloon and on-screen COP movements during the task so the learning may be quantified. This additional mode will be discussed in Chapter 3.

2.3 TOOL VALIDATION: CASE STUDIES

The feasibility of using the developed interactive virtual environment-based exercise tool as a viable exercise regime was validated through two case studies:

1. A three person short-sitting case study [Betker et al., 2007].
2. A three person standing case study [Betker et al., 2006a].

These case studies are described in the following sub-sections.

2.3.1 CASE DESCRIPTIONS

2.3.1.1 Short-sitting Case Study (CSS)

Short-sitting balance was investigated, where participants maintained an upright position of the torso, while sitting-on the buttocks and/or thighs, with the shank hanging over the sitting surface. Three people consented to participate in this study and provide the following information:

CSS1 Participant CSS1 was a 26 year old male with a spina bifida (myelomenigocele) extending from the spinous process of the tenth thoracic vertebra (T10) to spinous process of the first lumbar vertebra (L1) resulting in complete paraplegia and poorly developed lower extremities. At the time of initial assessment, he demonstrated good static and dynamic short-sitting balance and was independent with all transfers, activities of daily living, and work. He is a paralympian who actively races for Team Canada and was actively training to improve dynamic balance control: an important requirement for high-speed wheelchair racing.

CSS2 Participant CSS2 was a 52 year old male with complete paraplegia (at the level of the spinous process of the eleventh thoracic vertebra (T11) to L1) and a transfemoral amputation; these injuries resulted from a motor vehicle accident 10 months prior to recruitment in this case study. Following the accident he received in-patient rehabilitation for 6 months. At the time of initial assessment prior to the present treatment program, he demonstrated: complete motor and sensory loss below T11 level; dependent short-sitting balance: he sat with a kyphotic (a convex spine curvature) posture with bilateral upper extremity support and was unable to perform any functional activity using upper extremities in unsupported short-sitting; required a moderate assist of one for transfers. His primary treatment goal was to regain independent short-sitting balance for return to office work.

CSS3 Participant CSS3 was a 41 year old male, who had a severe traumatic brain injury over five years ago. He received physical therapy several times in the past 5 years for trunk and lower extremity motor control and balance re-education. His upper limb function was good bilaterally, but he had poor trunk and lower limb motor control and high muscle tone which fluctuated from extensor to flexor tone, depending on his positioning. He had a progressive plantarflexion (toes flexed down) contracture of the right ankle secondary to spasticity. He was unable to maintain short-sitting balance independently without use of his hands for support, due to impaired balance and trunk control. As a result (and due to his size), he was transferred with a Hoyer lift. He used a powered wheelchair for mobility indoors and outdoors. He had no sensory loss and his intellectual and memory functions were also intact. However he was easily distracted during most activities and therapy, requiring constant cuing and verbal commands to stay focused on the task at hand.

2.3.1.2 Standing Case Study (CS)

Three people consented to participate in this study and provide the following information:

CS1 Participant CS1 was a 20 year old, who had a Cerebellar Tumor Excision, resulting in severe ataxia (muscles uncoordinated).

CS2 Participant CS2 was a 58 year old, who suffered a single right cerebro-vascular accident (RCVA). The cause was diagnosed as an infarction by a neurologist after clinical examination and computed tomography.

CS3 Participant CS3 was a 41 year old, who suffered a closed Traumatic Brain Injury. Prior to training with our COP video game-based system, CS3 would only stand and attend to balance exercises for 20-30 seconds, with the training sessions typically lasting for only 10-15 minutes. Due to mobility and balance limitations, the treating physiotherapist provided support while CS3 stood up. Once standing, CS3 held onto a solid and firm table in order to maintain upright stance and complete the tasks. During the tasks, the physiotherapist only intervened if a loss of balance occurred or if CS3 needed to sit down. Due to CS3's reliance on support during standing, the assessments were not performed (Section 2.3.3.2). Hence, a more qualitative assessment is provided for CS3.

2.3.2 TREATMENT REGIME

A single-participant design was used, and pre- and post-exercise changes in outcome measures were determined. Participants attended an out-patient Physical Therapy clinic for balance and gait re-education. The clinic was operated by the Department of Physical Therapy, University of Manitoba, partially for clinical training of undergraduate Physical Therapy students under the supervision of a Physiotherapist. Each participant partook in eight 45 minute exercise sessions over a three week period, with 2-3 sessions per week. The exercise regime consisted solely of our COP-controlled virtual environment-based exercises; the participants did not receive any other balance training exercises during the treatment period. For the standing participants, the tasks were performed while standing on a firm surface and on a dense foam pad. The foam pad was 50.8 x 61 x 10.16 cm in dimension, with density 22.66 kg/m^3 and a 25% indentation force deflection (IFD) of 13.64 kg, i.e. a weight of 13.64 kg will compress the foam by 2.54 cm (25% of the height). A 1.91 cm thick wooden board of dimension 40.64 cm x 25.4 cm placed on top to evenly distribute the participant's mass. For the short-sitting participants, the tasks were performed while sitting on their regular wheelchair seat cushion and progressed (as appropriate) to sitting on a deflated Physio Gymnic ball (Ledraplastic spa., Osoppo, UD, Italy) or SwisDisk (PI Professional Therapy Products Inc., Athens, TN, USA). The additional surfaces added uncertainty into the system, as they would randomly modify the surface reaction forces; for people with sensation, it would distort/delay the pressure information from ground-to-foot or seat-to-surface contact, respectively. A summary of each participant and his respective treatment regime is given in Table 2.1.

2.3.3 EVALUATIONS AND OUTCOME MEASURES

2.3.3.1 Questionnaire

Post-exercise, a questionnaire was administered to each of the participant, which included the following questions:

1. Were the tasks and exercises fun to play?
2. Did the tasks increase your motivation to perform your exercises?
3. Were the task exercises challenging?
4. Did the difficulty level enhance the exercises?

Table 2.1. Case and treatment summary.

Subject	Case		Treatment				Surfaces
	Age	Injury	Tic-Tac-Toe	Memory Match	Under Pressure	Balloon Burst	
CSS1	26	Spina Bifida, lesion T10- L1/L2	-	-	100%	-	Firm, Swiss Ball
CSS2	52	Complete paraplegic, lesion T11-L1, above knee amputee	-	19%	80%	1%	Firm, Swiss Disk
CSS3	41	Severe traumatic brain injury	-	30%	70%	-	Firm, Swiss Disk
CS1	20	Severe ataxia	15%	15%	70%	-	Firm, Foam Pad
CS2	58	Single RCVA	15%	15%	70%	-	Firm, Foam Pad
CS3	41	Closed traumatic brain injury	15%	15%	70%	-	Firm, Foam Pad

5. Did you prefer the virtual environment-based exercises to traditional balance exercises?

The response options were:

1. Strongly Disagree.
2. Disagree.
3. Agree.
4. Strongly Agree.

The participants were also encouraged to provide personal comments regarding each question.

2.3.3.2 Stability Measurements

Based on the Sensory Organization Test concept, Shumway-Cook and Horak [1986] devised a clinical tool to test the sensory component of balance: the Clinical Test of Sensory Interaction and Balance. The Clinical Test of Sensory Interaction and Balance used a compliant foam pad as an unstable support base to emulate the Sensory Organization Test in terms of somatosensory distortion, with an added advantage that it was not limited to the pitch plane; the disturbance could be multi-directional [Shumway-Cook and Horak, 1986; Allum et al., 2002]. In this study, a Physio Gymnic ball or SwisDisk was used for the short-sitting participants and a dense foam pad (previously described in Section 2.3.2) was used for the standing participants. In addition to the Clinical Test of Sensory Interaction and Balance, four functionally relevant movement tasks were performed in order to challenge the different components of balance, which required both feedforward and feedback control, covering a relatively wide dynamic scope.

For the short-sitting participants, all tasks were performed on:

1. The regular seat cushion.
2. A deflated Swiss ball (CSS1) or Swiss disk (CSS2 and CSS3).

Similarly, for the standing participants, all tasks were performed on:

1. A fixed, firm floor surface.
2. A dense foam pad.

In the first four tasks, each participant was instructed to maintain erect sitting/standing balance for 20 seconds, with eyes open and then eyes closed, without hand support, on each surface. Note that participant CS3 could not stand without support; therefore, participant CS3 could not perform the standing tests. The second set of tasks included the following four movements of increasing difficulty:

1. Left and right head horizontal rotation, to visual targets placed 120 degrees apart.
2. Arm lift while holding onto a 50 cm light-weight wooden pole, 1.91 cm in diameter, with both hands kept shoulder width apart. The pole was raised up to eye level and then back down to the legs, keeping elbows extended.
3. Left and right horizontal trunk rotations, to approximately 60 degrees in each direction.
4. Forward trunk bending to approximately 60 degrees, then returning to the upright (erect) sitting/standing position.

These movements were performed for 20 seconds, in a rhythmical oscillating pattern at a frequency of 0.4 Hz (paced by a metronome). For all participants, a fall was recorded for each task if the participant could not maintain independent balance for 20 seconds or if he could not perform the movements without holding on with his hands. A physiotherapist was positioned directly behind the participant to provide assistance if required.

For the standing participants, the COP trajectory was obtained for each task using the force sensing array (FSA) pressure mapping system. The COP position signal has long been used as an indicator of balance performance [Szturm and Fallang, 1998]. The following two parameters were calculated to index balance performance for each task:

1. The range of the COP excursion, i.e., the peak-to-peak magnitude of the COP position [Whipple et al., 1999].
2. The COP sway path: a linear parameter that quantifies the total amount of resultant COP (COP_R) movement over the 20 second time period, calculated as $\left(\sum \Delta COP_R\right)/t_T$, where t_T is the duration of the movement [Barratto et al., 2002].

The COP sway path has been demonstrated to be one of the most valuable clinical parameters for standing balance control analysis [Barratto et al., 2002]. All data analysis was performed in Matlab (The MathWorks, Natick, Mass., USA).

2.3.4 RESULTS

2.3.4.1 Questionnaire

The results of the questionnaires were very positive, with all participants answering "Strongly Agree" to all five questions. All participants reported that they had fun practicing with the interactive virtual environment-based tasks and often lost track of time. Participants stated that they greatly preferred the training sessions with the virtual tasks over the exercise programs they have performed in the past and that they would like to continue the treatment. The tool's configurable parameters and different modes offered sufficient difficulty levels; even CSS1 who is a paralympian found the games challenging.

2.3.4.2 Stability Measurements

The following paragraphs summarize the stability measurement results for each participant. For the standing participants, Tables 2.2 and 2.3 present the results for the COP excursion range and COP sway path, respectively, during each of the 12 tasks pre- and post-exercise.

Pre-exercise, participant CSS1 maintained independent short-sitting balance for the full 20 seconds, for all 6 tasks where he sat on his regular wheelchair cushion. In addition, short-sitting balance was maintained for the eyes open, head rotations and arm lift tasks, while sitting on the deflated Physio Gymnic ball. CSS1 clearly lost his balance and therapist intervention was required to prevent a fall for three conditions short-sitting on the deflated Physio Gymnic ball. Post-exercise CSS1 maintained independent short-sitting balance for the full 20 seconds, during all 6 tasks on both short-sitting surfaces.

For participant CSS2, nine falls were recorded pre-exercise. CSS2 was only able to maintain independent short-sitting balance (without use of his hands for support) while sitting on the wheelchair cushion for the eyes open, head rotation and arm lift tasks. Post-exercise, CSS2 was able to maintain independent short-sitting balance for the full 20 seconds, for all tasks on both short-sitting surfaces.

Prior to training with our interactive virtual environment-based system, participant CSS3 would typically only attend to balance exercises for 20-30 seconds at a time, with the training sessions typically lasting for only 10-15 minutes. After practice with the COP-controlled video game-based system, CSS3 was able to maintain concentration during the virtual tasks (balance exercises) up to 2-3 minutes at a time and would repeat this for 10-15 times. The duration of the exercises increased from short interval training (approximately 20 seconds for 10-15 minutes) to two-minute interval training for 20-30 minutes. Twelve falls were recorded pre-exercise for CSS3. In addition, hand support was required for all 6 tasks during both surface conditions. Post-exercise, CSS3 was able to maintain independent short-sitting balance for the 20 seconds for all tasks on both

Table 2.2. COP excursion range [cm]; values for medio-lateral and anterior-posterior directions are presented separated by a comma (NC: not complete).

Task	CS1		CS2	
	Pre- Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Floor, eyes open	1.3, 2.6	0.4, 0.7	0.4, 0.4	0.4, 0.4
Floor, eyes closed	2.7, 2.9	2.2,2.1	NC	NC
Floor, head rotation	2.3, 2.6	1.3, 1.5	1.0, 1.2	1.1, 1.6
Floor, arm lift	3.6, 2.7	1.2, 1.1	0.3, 0.5	0.7, 0.8
Floor, trunk rotation	5.2, 3.4	2.5, 2.8	2.6, 1.0	0.8, 1.3
Floor, trunk bending	NC	NC	0.8, 0.9	1.1, 1.0
Foam, eyes open	2.7, 3.1	0.7, 1.2	1.0, 1.5	1.3, 1.4
Foam, eyes closed	NC	NC	NC	NC
Foam, head rotation	1.1,1.8	1.0,1.5	3.2, 3.1	2.0, 2.0
Foam, arm lift	3.3, 3.1	1.3, 1.7	1.0, 1.0	1.1, 1.9
Foam, trunk rotation	NC	NC	2.2, 1.4	1.6, 1.5
Foam, trunk bending	NC	NC	1.5,1.8	1.1,1.5

Table 2.3. COP sway path per unit time (NC: not complete).

Task	CS1		CS2	
	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Floor, eyes open	1.4	0.5	0.5	0.4
Floor, eyes closed	1.1	1.4	NC	NC
Floor, head rotation	1.8	1.0	1.4	1.5
Floor, arm lift	2.1	0.8	0.5	0.7
Floor, trunk rotation	3.4	1.4	1.8	1.5
Floor, trunk bending	NC	NC	1.1	1.3
Foam, eyes open	2.7	0.8	1.4	1.5
Foam, eyes closed	NC	NC	NC	NC
Foam, head rotation	1.2	0.9	3.4	2.7
Foam, arm lift	2.5	0.7	1.0	1.2
Foam, trunk rotation	NC	NC	2.0	2.1
Foam, trunk bending	NC	NC	1.5	1.9

surfaces.

For participant CS1, 10 falls were recorded pre-exercise. CS1 only maintained independent standing balance during the eyes open and head rotation tasks, performed on the normal floor surface. During all other tasks, CS1 clearly lost balance and therapist intervention was required to prevent a fall. Post-exercise, CS1 was able to carry out all tasks for the total 20 second duration, except the foam surface, eyes closed standing task. In this case, vision was eliminated and ground reaction force information was distorted and delayed by the compliant foam pad. Note that for some tasks where CS1 fell, the fall occurred at the end of the trial and therefore the full 20s of data was recorded. Tasks where the full data was not recorded pre- or post-exercise are indicated by not completed (NC) in Tables 2.2 and 2.3 for both the pre- and post-exercise trials, as data was not analyzed post-exercise if incomplete data was present pre-exercise. When CS1 did fall, there were larger COP peak-to-peak amplitude and sway path values pre-exercise in all fall cases except the floor surface eyes closed task. CS1 maintained balance both pre- and post-exercise during the floor surface eyes open and head rotation tasks. For these two tasks, the post-exercise values for the COP excursion range and sway path were approximately half the value pre-exercise.

For participant CS2, five falls were recorded pre-exercise. Post-exercise, only two falls were recorded. CS2 could not independently maintain standing balance with eyes closed on either surface. With the exception of the eyes closed tasks, CS2 successfully performed the remaining five tasks on the floor surface. CS2 was successfully able to perform the oscillating head rotations, trunk bending and trunk rotation tasks on the foam pad; these were tasks CS2 was unable to perform pre-exercise. CS2 maintained balance both pre- and post-exercise during all tasks except the two eyes closed conditions and the head rotations, trunk rotations, and trunk bending tasks on the foam pad. Similar to CS1, note that for some tasks where CS2 fell, the fall occurred at the end of the trial and therefore the full 20s of data was recorded. Tasks where the full data was not recorded pre- or post-exercise are indicated by not completed (NC) in Tables 2.2 and 2.3 for both the pre- and post-exercise trials, as data was not analyzed post-exercise if incomplete data was present pre-exercise.

For the cases where balance was maintained, the COP excursion range values were similar for pre- and post-exercise; for the fixed, eyes open task they remained the same; for the fixed surface head rotations, arm lift and trunk bending tasks and the foam surface arm lift task there was an increase post-exercise; for the fixed, trunk rotation task there was a decrease in the ML direction and an increase in the AP direction; for the foam surface eyes open condition there was an increase in the ML direction and a decrease in the AP direction. In the cases where participant CS2 fell pre-exercise and maintained balance post-exercise, the results were lower post-exercise except for a slight increase in the AP direction for the foam surface trunk rotation.

We found similar results for the COP excursion range values. For the cases where balance was maintained both pre- and post-exercise, the values for the fixed surface eyes open and trunk rotation tasks were lower post-exercise; for the remaining tasks there was a slight increase post-exercise. In the cases where participant CS2 fell pre-exercise and maintained balance post-exercise, the results were lower post-exercise for the foam surface head rotations task and slightly higher for the foam surface trunk rotations and trunk bending tasks. Although there was less consistency in the COP excursion range and sway path values for CS2 when compared to CS1, recall that CS2 was able to independently maintain balance for five additional tasks post-exercise, which is a very important finding.

Participant CS3 was able to maintain concentration on standing balance exercises for up to 10 minutes at a time pre-exercise. The time was limited by foot pain rather than concentration. CS3 showed a marked improvement at maintaining an erect standing position while playing the COP-controlled virtual tasks. Post-exercise, CS3 was able to independently stand on firm ground without hand support for up to 20 seconds. The duration of training sessions increased three-fold from 10-15 minutes to 40 minutes.

2.4 DISCUSSION

The goal of this part of the study was to investigate the feasibility of using COP-controlled interactive virtual environment-based tasks for training of short-sitting and standing balance and whether they would result in improved dynamic balance. The main finding was that post-exercise, all participants exhibited a decrease in their fall rate. This finding is consistent with the observation that intensive, functionally relevant practice of a motor task can result in significant improvements. The COP excursion and path length values for participant CS1, for all tasks where standing balance was maintained pre- and post-exercise, decreased post-exercise. The values for participant CS2 were less consistent. For tasks where CS2 maintained balance pre- and post-exercise, the COP excursion and path length values were similar; in some cases, however, the post-exercise values were higher. For the tasks where CS1 and CS2 fell pre-exercise and maintained standing balance post-exercise, the post-exercise values were generally lower.

When the participant did fall, large COP excursions occurred. This was reflected in the generally larger COP peak-to-peak amplitude and sway path values pre-exercise. In some cases, however, the post-exercise values were higher. In these cases, large COP excursions were also produced, reflecting the compensatory strategies employed to maintain standing balance. In other words, although balance was maintained post-exercise, the large COP peak-to-peak amplitude and sway path values indicated that the participants still had difficulty with the tasks. All participants expressed that they enjoyed the virtual environment-based tool, preferring it to normal treatment regimes, and that they would like to continue the treatment. Thus, the results showed that our COP-controlled virtual tasks provided a motivational and challenging environment, which improved the participants' dynamic balance post-exercise.

Conventional exercises are often repetitive, causing patients to lose interest and not complete the rehabilitation process. Hence, studies are being done which include more motivating tasks. In [Tsang and Hui-Chan, 2004], the effects of Tai Chi and golf, complex and enjoyable tasks, were evaluated. In both of these tasks, the participant has an increased awareness of the motor control required to successfully perform the movements. This is what we were trying to achieve through the use of COP biofeedback coupled to dynamic exercise tasks. In addition, we showed that our therapy approach could be applied to people with:

- CNS lesions, hemiplegia, and bilateral deficits
- Severe balance and mobility limitations, including long standing, secondary musculo-skeletal problems.

When compared to biofeedback training applied in the same context, our interactive virtual environment-based tasks offer numerous advantages. Of primary importance was the fact that the developed system could be enhanced to meet the needs and performance levels of each participant [Roig et al., 2004]. This was possible through configurable difficulty levels, stability parameters, and the ability to place the pliable pressure mat on different surfaces. This allowed all participants, including CSS1 who was a paralympian, to find the games challenging. In Sihvonen et al., [2004], the COP trajectory was measured via a fixed, force platform and was displayed to a group of frail elderly women in order to motivate them. Training with the device was done over a four week period. The task set included standing balance movements where the participants had to stand as still as possible and dynamic balance (including stepping) movements, where they had to move the COP cursor to follow a displayed pattern. Foam support surfaces of different thicknesses were also placed on the force platform; however, due to the platform's inflexibility and fixed nature, their system is still limited in the amount of surfaces that may be used.

Our system uses a flexible pressure mat, which allows training to be conducted on compliant or uneven surfaces; i.e., the mat may be placed on top of a surface, rather than the surface being placed on the force platform. This can better prepare subjects to deal with more dynamic environmental conditions. Flexible pressure mats permit accurate recording of the COP, while eliminating the nonlinear distortions and damping effects incurred in the COP trajectory by the different foam materials [Betker et al., 2005].

Secondly, the movements in the virtual environment-based tasks were random and varied in direction, amplitude, and precision. In order to meet the goals of the game, the participants had to make timely, goal-directed shifts in the COP trajectory; this required active ML and AP weight shifts. Lastly, our virtual environment-based system provided a motivating and interactive environment for the purposeful and challenging dynamic balance exercises.

Biofeedback systems that do not rely on a force platform can also be performed on different surfaces. An encoded audio biofeedback signal representation of torso acceleration was used in Dozza et al., [2005] to inform patients with bilateral vestibular loss of their body movements. The sound was intended to represent information normally provided by the otoliths. In this case, the patients stood on a compliant foam pad with their eyes closed, thus adding the need for timely feedback adjustments. They found that the audio signal helped to reduce the overall torso sway area, torso acceleration and increase stability. Similar to current biofeedback systems, our video games provided the subject and therapist with instantaneous feedback about performance and goal attainment. The subjects were able to measure their successful progression to more complex tasks and support surfaces in real time.

The games developed in this research offer several advantages over the NeruoCom NeuroGames. Firstly, in Tic-Tac-Toe and Memory match, item selection is done independently of aid through use of the 'number of seconds to mark' parameter. Conversely, NeruoGames requires the mouse button to be clicked when the center of gravity marker is on top of the desired item. The mouse button must then be held down while the item is dragged to the desired location. Thus, in order for the participant to independently play the game, they must use the mouse. The use of a mouse in therapeutic games is not desired as:

1. In cases where the participant must hold on to an object to maintain stance, they would not be able to simultaneously operate the mouse; noting that eventually the participant should learn to bear the weight entirely in his/her legs, as support removes some of the balance task dynamics [Cogan et al., 1977].
2. In cases where the participant is able to maintain stance on their own, the mouse would provide an unwanted additional tactile input, which has been shown to reduce sway variance [Pai and Patton, 1997]. In addition, the particpant could offset some of his/her weight through the use of his/her hands [Cogan et al., 1977].
3. Some participants' hands and/or arms are paretic and they would therefore not be able to operate the mouse.

Secondly, the NeuroCom system has an increased cost compared to the pressure mat. Therefore, the pressure mat will be available to a wider range of clinics and will thus benefit a larger population. Lastly, the pressure mat is easily portable and can therefore be taken to where the subjects are, rather than requiring them to come to the clinic; this also makes our system available to a larger population. In addition, the Neurocom system assumes that the center of gravity and COP are equivalent for single-segment inverted pendulum dynamics; however, this is not always true as demonstrated in Barbier et al., [2003].

2.5 SUMMARY

An interactive virtual environment-based tool was developed, which coupled the COP movement trajectory to an on-screen cursor (computer sprite). Four different tasks were developed, which sought to improve dynamic balance control in both standing and short-sitting. The tasks Tic-Tac-Toe and Memory Match were designed to elicit quick, short movements followed by a sustained hold in the computer sprite position, in order to select a square/card. In Under Pressure, the participant had to shift their body weight to move a receptacle in line to catch a moving object. In Balloon Burst, quick and precise movements were required to move the computer sprite to intersect and pop the stationary balloon.

The benefits of the developed interactive virtual environment-based tool were evaluated via standing and short-sitting case studies, where the amount of recovery was quantified pre-exercise versus post-exercise. Six different tasks were performed on two different surfaces (Section 2.3.3.2). For the short-sitting case study, the number of falls was compared pre- and post-exercise. Post-exercise, all subjects exhibited a lower fall count and in fact, were able to complete all 6 balance tasks on both surfaces. For the standing case study, the number of falls, the range of COP excursion, and the COP path length outcomes were compared pre- and post-exercise. Post-exercise, the subjects exhibited a lower fall count and a decreased COP excursion limit for some tasks. In both the short-sitting and standing case studies, participants exhibited an increased desire to practice and an increased attention span during training. These findings demonstrated that graded, dynamic balance exercises on different surfaces can be effectively coupled to virtual environment-based tasks. In addition to the training program being enjoyable, all subjects improved their fall rate performance after using the virtual environment-based therapy. Hence, this approach was effectively applied to:

- People who had severe balance and mobility limitations, including long standing, secondary musculo-skeletal problems.
- People with CNS lesions, hemiplegia, and bilateral deficits.
- People who were actively participating in sports.

3 Strategies for Learning New Motor Tasks

In this research, the position of a visible computer sprite is controlled through the movement of the COP, which changes when a person produces body movements. The visually guided movements of the COP require timely (i.e., prior to falling) sensory-motor mappings of visually derived spatial information, which is acquired in a two-dimensional (2D) virtual environment, i.e. the computer sprite location relative to a target position. Recent investigations [Shadmehr and Mussa-Ivaldi, 1994; Krakauer et al., 1999; Jones et al., 2001; Patton and Mussa-Ivaldi, 2004; Viau et al., 2004] have examined sensory-motor learning with various visual-spatial tasks viewed inside a virtual environment. Movement errors and adaptations have been investigated during online visually guided control of arm reaching movements, which were perturbed by sudden forces. The perturbations form new relationships between the motor output and limb trajectories, in a manner that is beneficial for sensory-motor learning. Another way to systematically introduce random errors and graded difficulty levels in a visual-spatial task for specific training purposes is through transformations in the trajectories of the computer sprite motion in route to a target. In this research, we introduced a rotation in the mapping of the computer sprite trajectory relative to the normal COP movement trajectory [Jones et al., 2001]. This chapter details the virtual environment and the resulting outcome strategies adopted by participants to learn the visual-spatial task.

3.1 BACKGROUND

When performing daily life activities, appropriate sensory-motor transformations are required to successfully map the changing relationships between one's self, the environment, and objects moving in the environment. Our daily actions required to perform basic and instrumental activities involve varying combinations of head-eye (gaze), arm reaching, and whole body (stepping and walking) movements. These movements depend on the interaction and transformation of both egocentric (self-to-object) and allocentric (object-to-object) representations of the environment [Burgess 2006; Mou et al., 2006]. In order to successfully map these representations, appropriate sensory-motor transformations are required [Mussa-Ivaldi 1999; Singh and Scott, 2003; Mou et al., 2006]. For visually guided movements, the primary motor cortex and its interactions with the visual cortex, dominantly through the visual pathway called the dorsal stream [Ellerman et al., 1998; Fogassi and Luppino, 2005], are largely responsible for mapping the sensory-motor actions [Leon-Sarmiento et al., 2005; Culham et al., 2006]. The visually acquired movement amplitude and direction information is received by the parietal cortex, from which motor plans for voluntary motor actions are partially determined. These motor actions cause muscle contractions that move the body segment masses, velocities, and accelerations in a manner that will change the support forces of the feet. In turn, the COP and consequently the computer sprite move to the desired location in order to intersect the target object within the allotted time [Shadmehr and Mussa-Ivaldi, 1994].

It is important to train and evaluate people under congruent visual-spatial mapping conditions (i.e., physical and perceived actions move in sync with the normal, known pattern) and those which systematically change the sensory motor transformation. A number of recent investigations have examined visual-spatial learning through the use of prism glasses, which shift the visual world horizontally relative to the physical target location [Rossetti et al., 1993; Fernandez-Ruiz et al., 2003]. At the start of the reaching tasks, there were large errors in the participants' movement trajectories. As the participants adapted to wearing the prism glasses, the errors were reduced. After removal of the glasses, aftereffects were demonstrated, in which the participants still planned their movements as if they were wearing the prism glasses. As the effects of the prism glasses are limited in function, additional methods of disturbing the visual-spatial mapping have been explored.

Task specific and goal oriented rehabilitation programs are used to train people to produce appropriate multi-sequence motor responses under different environmental and cognitive conditions [Broeren et al., 2004; Perez et al., 2004]. Of particular interest are rehabilitative regimes incorporating biofeedback in an interactive virtual environment (which emulate egocentric and allocentric relationships), as they are motivational and provide the ability to perform task analysis through the tracking of biofeedback movements relative to intended goals/targets [Betker et al., 2006a]. Commonly used biological signals include electromyography (EMG), segment motion, and the center of foot pressure. Through muscle contractions (EMG), limb movement, or body movements (COP), the on-screen cursor (computer sprite) can interact with the on-screen virtual task objects. Thus, the biofeedback signal allows for natural and intuitive visually guided movements of varying accuracy and precision to be used in the virtual environment-based tasks.

The combined use of physical movements and virtual psychomotor tasks, having specific goals and graded spatiotemporal accuracy, represents functional training with a wide scope of sensori-motor and cognitive experiences, which can drive neuroadaptation sufficient for producing behavioral recovery. In the virtual environment, recorded biofeedback signals are transformed and slaved to an on-screen computer sprite pixel coordinate system; this allows body movements to be visually-guided through the virtual space. This can include predictable or random event sequences, which require movements of varying amplitude, speed, and precision. In the context of the virtual task, producing and controlling the motion and location of the computer sprite is a mixture of both allocentric and egocentric motion, i.e., object-to-object spatial relationships (computer sprite to target) and physical body to sprite movement spatial relationships, respectively [Burgess, 2006].

A method to systematically introduce random errors and graded difficulty levels in a visual-spatial task for specific training purposes is through transformations in the trajectories of the computer sprite motion in route to a target. In this research, we introduced a rotation in the mapping of the computer sprite trajectory relative to the normal COP movement trajectory [Jones et al., 2001]. The sensori-motor system will rely on visual-based feedback controls to adapt to the visual transformation. When first learning the task, the visual information is incongruent to the previously known mapping with the proprioceptive and tactile information received by the feet [Jones et al., 2001]. Learning to perform the new sensory-motor task in the allotted time will require adaptation to a mixture of egocentric and allocentric spatial relationships before producing the required body movements.

Previous studies have determined the effects of a transformed physical-to-virtual spatial relationship on perception-action through the quantification of errors in movement planning with respect to an ideal movement [Abeele and Bock, 2001; Henry et al., 2001; Frassinetti et al., 2002; Vasvada et al., 2002; Marotta et al., 2005]. Another method of investigating learning is through the quantification of key muscle activity during the selected task. For example, a two-dimensional robotic manipulandum has been used in an arm reaching task, under normal and modified force conditions [Shadmehr and Moussavi, 2000]. The modified condition consisted of a constant force applied to the manipulandum during the reaching task. The force produced was proportional to the speed of the hand and perpendicular to the position (direction) of the hand [Shadmehr and Mussa-Ivaldi, 1994; Shadmehr and Moussavi, 2000]. EMG signals were recorded from the biceps-triceps and anterior-posterior deltoid muscles during the task. Results showed that relative to the direction observed when no field was applied, the preferred direction of the muscle was rotated after training with the force applied to the manipulandum.

In this research, we explored the strategies used in the visual-based cognitive re-mapping that takes place during the transformed visual-spatial task. Recent developments strongly suggest that the combined use of physical movements and virtual psychomotor tasks represents functional training with a wide scope of sensory motor and cognitive experiences, which can drive neuroadaptation sufficient for producing behavioral recovery. We predicted that subjects would produce an optimal strategy that would emerge through a decrease in movement errors and specific, consistent changes in the preferred directions of three leg muscles. We determined the different strategies participants used to adapt to the task and which strategy resulted in optimal recalibration.

3.2 METHODOLOGY

3.2.1 PARTICIPANTS

Twelve healthy participants (5 males) aged 26 ± 2.98, of height 1.71 ± 0.11 m, weighing 63.66 ± 13.90 kg (± standard deviation), were recruited for this study among students at the University of Manitoba. Participants were screened by oral questionnaire to rule out any history of postural, neurological, musculo-skeletal, visual, or vestibular problems. Participants voluntarily gave informed consent. Ethics approval was granted prior to recruiting participants by The University of Manitoba, Faculty of Medicine, Ethics Research Board. The participants were not aware of the rotation applied to the on-screen sprite trajectory, nor had they previously played Balloon Burst with the rotation applied.

3.2.2 VIRTUAL ENVIRONMENT

A virtual environment-based task was produced, *Balloon Burst version 2*, the goal of which is to pop balloons which appear at random or predetermined locations on the video monitor (Fig. 3.1a,b) [Betker et al., 2006a]. The custom Balloon Burst software we developed interfaces with a flexible pressure mat (Verg Inc., Winnipeg, MB, Canada), from which it acquires the COP signal at a rate of 35.5 Hz, i.e. the rate equivalent to the instrumentation dependent high speed mode. The pressure mat used in this study was of dimension 53 cm x 53 cm x 0.036 cm and contained a 16 x 16 grid of piezo resistive sensors (total 256 sensors) spaced 2.86 cm apart. The pressure mat uses a calibration file to maintain sensor calibration to vertical pressures of 300 mmHg. This recorded instantaneous physical COP position is mapped to the game sprite, in a manner identical to the use of a computer mouse. The physical COP movement range of the participant is scaled to cover the onscreen movement range. To play Balloon Burst, the user first

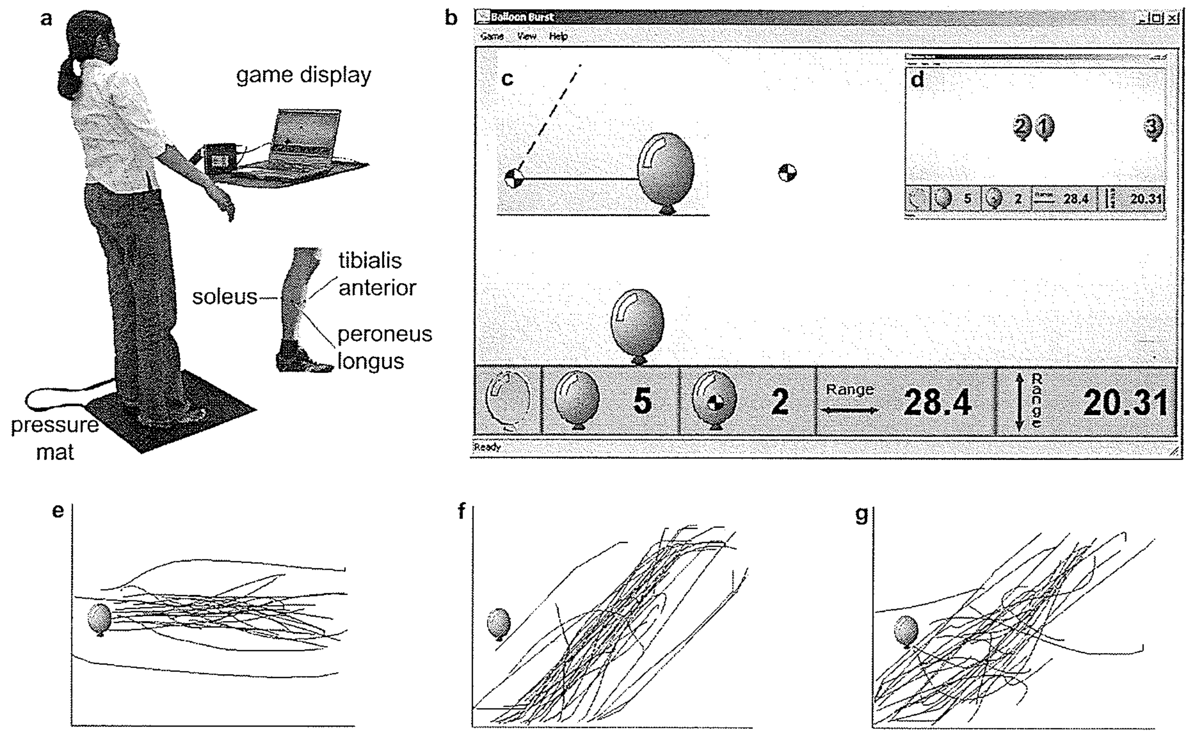

Fig. 3.1. Experimental setup. (**a**) the subject stands on the pressure mat, which is connected to the laptop via the interface box. The laptop currently displays the game Balloon Burst. Three surface electrodes were used to capture the muscle activity of the peroneus longus, soleus, and tibialis anterior. (**b**) a screenshot of Balloon Burst displaying the total number of balloons, the number of balloons popped, and the movement ranges (in cm). The player shifts their weight to control the on-screen COP marker (black and white circle). The balloon is popped when the COP marker area overlaps the balloon area. When the rotation is applied, the participant's on-screen trajectory is rotated 60° counterclockwise (**c**). Triads, the sequence of three balloons, appear to the right (**d**). Insets (**e**)-(**g**) show the COP trajectories for all participants, as they shift their weight to intersect the balloon with the on-screen COP marker, for (**e**) the congruent game, (**f**) the first translated game, and (**g**) the final translated game.

defines the number of balloons that will appear on the screen (one at a time) and the number of seconds for which each balloon will appear. Once the game begins, the participant must shift his/her weight in all directions, in order to move the onscreen COP sprite to intersect and burst the balloon. Difficulty levels are provided through a configurable balloon size.

The locations of some balloon appearances were preset in order to have a consistent starting position for all participants and between trials. A series of three fixed balloon locations, termed a triad event, were introduced at random times during the course of the session. Each balloon in the triad appears one at a time. The triad has the positions of the first two balloon locations fixed. The first balloon in the triad is positioned near the center of the screen; the second balloon in the triad is located close beside the first. Thus, participants will only have to make a small movement correction between the first and second balloon. The third balloon appears along the boundary of the screen and can be configured to appear in the directions: up, down, left, right, or one of the four corners. The number of triads per session is configurable.

In addition to having a congruent linear mapping between the motion of the COP position and the on-screen COP cursor (slaved to COP signal), a geometric transformation can be applied. A rotational transformation was added to induce a visual-spatial (cognitive) re-calibration during the task, which used body movements (COP) to move a cursor to intersect the randomly presented target balloons. The visual translation consists of a rotation of the on-screen COP marker. Specifically, the on-screen COP marker can be rotated counter-clockwise by $\theta \in \{30°, 45°, 60°\}$, using the rotation matrix $\mathbf{R}_\theta$

$$\mathbf{R}_\theta = \begin{bmatrix} \cos\theta & -\sin\theta \\ \sin\theta & \cos\theta \end{bmatrix}. \tag{4.1}$$

The rotation matrix is used to transform the current on-screen COP marker coordinate (COP_x, COP_y) to the coordinate $(COP_x^{'}, COP_y^{'})$ via

$$\begin{bmatrix} COP_x' \\ COP_y' \end{bmatrix} = \mathbf{R}_\theta \begin{bmatrix} COP_x \\ COP_y \end{bmatrix}. \tag{4.2}$$

The rotation occurs about the user's on-screen center point coordinate, which is set to the value (0,0). As there are multiple balloon locations in different directions that are rotated, the subjects should be better able to adapt to the new rotated reference frame [Krakauer et al., 2000].

3.2.3 EXPERIMENTS

Participants used the motion of their COP to change the position of a computer sprite relative to a target object in a virtual environment-based task, *Balloon Burst version 2*. The goal of Balloon Burst was to pop balloons appearing in random or predetermined locations on the video monitor by intersecting the balloon with the computer sprite (Fig. 3.1a,b). For all experiments, each session of Balloon Burst had a total of 30 balloon appearances and 3 triad sets (10% of all balloon appearances). The triads appeared to the right (Fig. 3.1d) and a there was a three second time period between balloon appearances. Participants stood on the pressure mat with their feet apart at their preferred normal position; this position was kept constant during all sessions. Participants were instructed to keep their feet in contact with the support surface at all times. The range of the on-screen mapping corresponded to a bounded area of the pressure mat, focused around the user's stationary COP point.

First, three sessions were played where a normal, preferred congruent mapping between the motion of the COP position and the on-screen COP cursor existed. These sessions were used to familiarize the participants with the game-based task and served as a baseline for movement error comparison. Next, a geometric transformation of the previously learnt computer sprite trajectory was introduced to produce a performance error, which would require a visual-spatial re-calibration in order to successfully pop the balloons during the task; participants were not told of the change. Five sessions were played where the computer sprite movement was rotated by 60° counterclockwise. As the directional mapping of the COP movement to computer sprite movement was altered, participants needed to alter how they moved the COP to successfully intersect the balloons. Specifically, with the congruent mapping, a shift of the participant's weight to the left resulted in the computer sprite moving left (Fig 3.1c, solid line). With the rotation applied, the computer sprite moved in a direction approximately north-north-east (Fig 1c, dashed line). Finally, an additional three games were played with the original congruent mapping, i.e., without the rotation applied, to determine if any aftereffects occurred. In other words, the trials investigated if the participant was still compensating for the rotation after it was unknowingly removed.

In summary, three experiments were performed:

1. Three sessions with a congruent mapping between the physical and onscreen movements.
2. Five sessions with a rotation of 60° counterclockwise applied to the onscreen sprite movements.
3. Three sessions with the congruent mapping applied in experiment 1.

3.2.4 DATA RECORDING AND ANALYSIS

The COP trajectory and balloon locations were logged to a file, at a rate of 35.5 Hz (i.e., the sampling rate of the pressure mat). Raw EMG recordings, using surface EMG probes model SX 230 (NexGen Ergonomics Inc., Pointe Clare, QC, Canada), were recorded at a sampling rate of 1 kHz from the tibialis anterior, peroneus longus, and soleus using the Datalink DLK 900 EMG amplifier and data logger. The SX230 EMG probes have a built-in amplifier with gain set to 1000, highpass filter (18 dB/octave) to remove the DC offset from the signals and a lowpass filter with a cutoff frequency of 450 Hz. For each balloon, the start of the movement was determined at the time of balloon appearance plus an offset of approximately 400 ms, which accounted for the delay in visual processing of the balloon appearance [Rossetti 1998].

Movement errors were calculated during the initial attempt to move the on-screen cursor to the third balloon of each triad. The displacement angle ϕ was calculated between the trajectory of the initial COP movement and the direct line path between the starting COP position and the balloon position (Fig. 3.2). The maximum perpendicular displacement was calculated between the trajectory of the initial COP movement and the direct line path (Fig. 3.2). Note that the COP trajectory often curved towards the balloon and hence, the maximum perpendicular distance does not necessarily occur at the end point of the COP trajectory.

The muscle tuning function [Shadmehr and Moussavi, 2000] was investigated during movement initiation, i.e. the first 200 ms of the movement [Szturm and Fallang, 1998], for all balloon appearances. The muscle tuning function is a measure of how the muscle activity amplitude and direction changes with learning/time. For that purpose, the superposition of the muscle activity (root mean square (RMS) value within a particular time frame of the movement) vectors in all directions is calculated, and called the preferred direction (PD). Changes in the PD vector can reveal learning and adaptation [Shadmehr & Moussavi, 2000]. In this study, the RMS of the muscle's EMG was

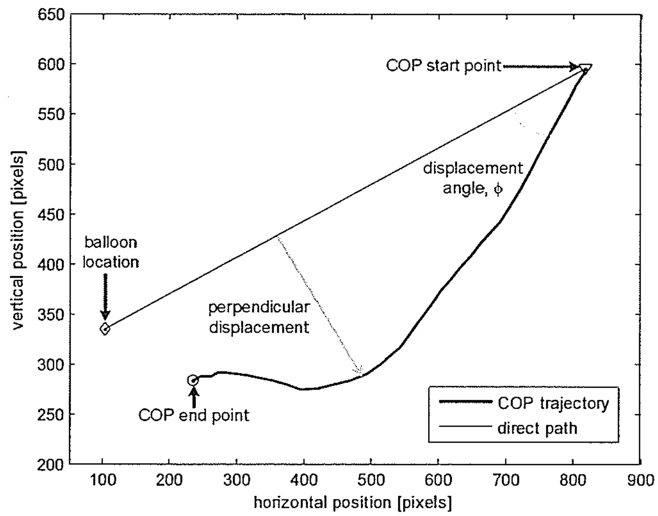

Fig. 3.2. Movement errors calculated between the true COP trajectory (solid line) and the ideal direct path to the balloon (dashed line). The displacement angle is indicated by ϕ and the arrow indicates the maximum perpendicular displacement.

calculated for each balloon in a given session, for the first 200 ms of each movement, and averaged. The movement space was sequestered into four quadrants, centered around $\Phi = \{0°, 90°, 180°, 270°\}$.

The EMG activity was recorded for the peroneus longus, tibialis anterior, and the soleus. We are interested in the peroneus longus and tibialis anterior muscle activity in the quadrants $\theta = \{0°, 90°\}$, i.e. movements to the front and right. In these directions, both of the muscles should be active and play a large role in creating the movements: the tibialis anterior will decrease the angle between the foot and lower limb and the peroneus longus will attempt to turn the lateral edge of the foot outward (eversion). Similarly, we are interested in the peroneus longus and soleus muscle activity in the quadrants $\theta = \{0°, 270°\}$, i.e. movements to the back and right. In these directions, both of the muscles should be active and play a large role in creating the movements: the soleus and peroneus longus will increase the angle between the foot and lower limb. Note that in these directions, the gastrocnemius performs a similar function to the soleus. However, we selected the soleus as it spans a single joint (the ankle), whereas the gastrocnemius spans two joints (ankle and knee). For each muscle pair, the EMG recording was normalized by the amplitude of the sum of the two muscle signals. This normalization procedure was selected to normalize the EMG amplitudes relative to each other and to represent the changing dynamics of the Balloon Burst task. The direction of the movement was calculated as the angle required to intersect the target balloon at the beginning of movement initiation. The preferred direction, PD, of the muscle was then calculated as the directional sum of the RMS values in the desired quadrants θ

$$PD = \sum_{\theta} RMS . \tag{4.3}$$

For the five transformed sessions, the first session was analyzed to investigate the subjects' initial reaction to the rotation. The results from transformed sessions 2 and 3 were averaged (T23) to see how they react as they familiarize themselves with the rotation, and the results from sessions 4 and 5 were averaged (T45) to obtain their final reactions. The outcome measures were averaged over all subjects, on a per session basis, and the mean and standard error (SE) were calculated.

3.3 RESULTS

To identify the strategies used to adapt to the transformation, we quantified the changes that took place during and after the spatial transformation: movement errors were calculated with respect to an ideal trajectory (Fig. 3.2) and muscle activity was investigated for the tibialis anterior, soleus, and peroneus longus. Movement errors give an overall task outcome perspective on the learning of the rotation, whereas the muscle activity serves to provide insight on the internal remappings.

The maximum perpendicular displacement and displacement angles, along with the standard error, were calculated between the trajectory of the initial COP movement and the direct line path to the balloon (Fig. 3.2). Prior to applying the rotation, the movement errors for the congruent sessions were low (Fig. 3.3a,d). When the rotation was first applied, there were large movement errors. As the participants adapted to the rotation, we saw a decrease in the movement errors. Specifically, across all subjects, a decrease of $21.43 \pm 9.96\,\%$ (±SE) was observed in displacement angle and a decrease of $15.01 \pm 5.14\,\%$ (±SE) was observed in the perpendicular displacement (Table 3.1). Next, the initial congruent mapping was again applied. The perpendicular displacements quickly became the same as before the rotation was applied. For the displacement angle, decreased errors were observed compared with the rotated sessions. However increased errors were seen for the first congruent session following exposure to the rotated spatial transformation when compared to the original congruent session (Fig. 3.3a,d); in other words, aftereffects were observed for the displacement angle. This suggests that it was easier for participants to correct the computer sprite trajectory during the movement, as opposed to predicting the amount of rotation required prior to the movement.

The muscle tuning (i.e. the muscle's preferred direction, measuring how muscle activity amplitude and direction changes) was investigated during movement initiation (moving computer sprite to burst balloon) for the tibialis anterior and peroneus longus muscles during movements to the front and right and for the soleus and peroneus longus muscles during movements to the back and right. For movements to the front and right,

Table 3.1. Absolute and percentage change in outcome parameter between the first translated session (T1) and the final translated sessions (T45), across all participants and per group.

Parameter	Change in parameter between T1 and T45					
	All subjects		Group CCW		Group CW	
	(°)	(%)	(°)	(%)	(°)	(%)
PL Tuning	4.52 ± 1.03	9.32 ± 2.11	2.87 ± 1.97	10.33 ± 4.11	4.23 ± 1.12	8.48 ± 2.18
TA Tuning	-	-	-0.69 ± 0.23	-1.33 ± 0.53	2.89 ± 1.21	6.17 ± 2.50
Angle	9.36 ± 3.82	21.43 ± 9.96	17.88 ± 5.32	44.06 ± 11.87	0.18 ± 2.79	1.87 ± 11.16
	(pixels)	(%)	(pixels)	(%)	(pixels)	(%)
Displacement	66.66 ± 23.11	15.01 ± 5.14	123.64 ± 39.43	27.50 ± 8.45	31.71 ± 18.43	7.31 ± 4.67

Parameter values along with the standard error mean are given
PL – peroneus longus, TA – tibialis anterior, CCW – counterclockwise, CW - clockwise

all subjects exhibited the same strategy, a clockwise shift in the preferred direction of the peroneus longus from the initial exposure to the rotation (Fig. 3.4a). The average total shift, along with the SE, for the peroneus longus tuning function was 4.52 ± 1.03 ° or 9.32 ± 2.11% (Table 3.1). For movement to the back and right, no consistent pattern in the peroneus longus tuning was observed. Similarly, for the soleus, no consistent tuning pattern was observed.

For the tibialis anterior, we saw two distinct strategies emerge (Fig. 3.4b,c). Half of the participants exhibited a clockwise shift in the preferred direction with successive exposure to the task, similar to the shift for the peroneus longus (Fig. 3.4a). This consistent shift in the tuning angle is what we hypothesized would occur. However, not all participants exhibited this strategy. The remaining participants had a counterclockwise shift in the preferred direction for the average of the final two sessions, when compared to the first session. For five of these participants, a clockwise shift in preferred direction was observed for the average of sessions 2 and 3, followed by a counterclockwise shift in

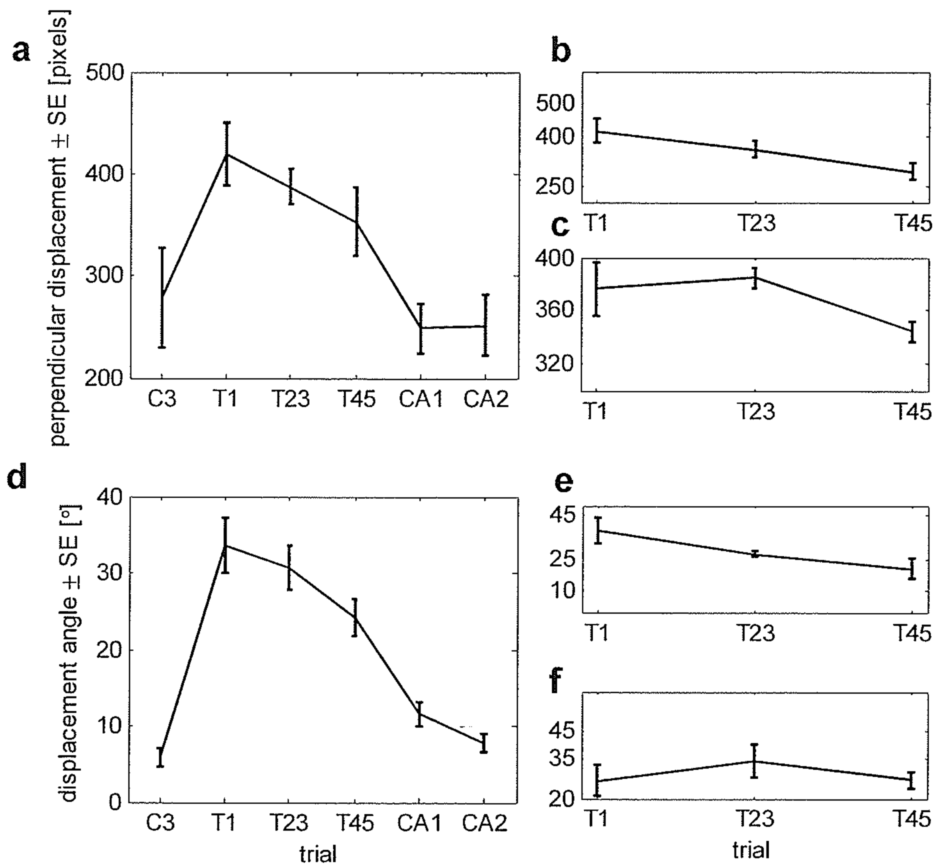

Fig. 3.3. Movement error outcome measures. The mean values across all participants are displayed for the (**a**) displacement angle and (**d**) maximum perpendicular displacement, for the third (final) congruent game (C3), the games with the translation applied (T1, T23, and T45), and the two congruent games played after the translated tasks (CA1 and CA2); error bars indicate the standard error mean. The perpendicular displacement for group CCW for the translated games is depicted in (**b**) and for group CW in (**c**). The displacement angles for group CCW for the games with the translation applied are depicted in (**e**) and for group CW in (**f**).

the average of the final trials (Fig. 3.4b). The final subject had a consistent counterclockwise shift in the preferred direction.

As two distinct strategies appeared for the tibialis anterior muscle tuning, it was interesting to investigate which one was the more optimal strategy. We term the group which exhibited a counterclockwise shift in the preferred direction group CCW and the group which exhibited a clockwise shift in the preferred direction group CW. For group CCW, we observed an average decrease in the displacement angle of 44.06±11.87% (±

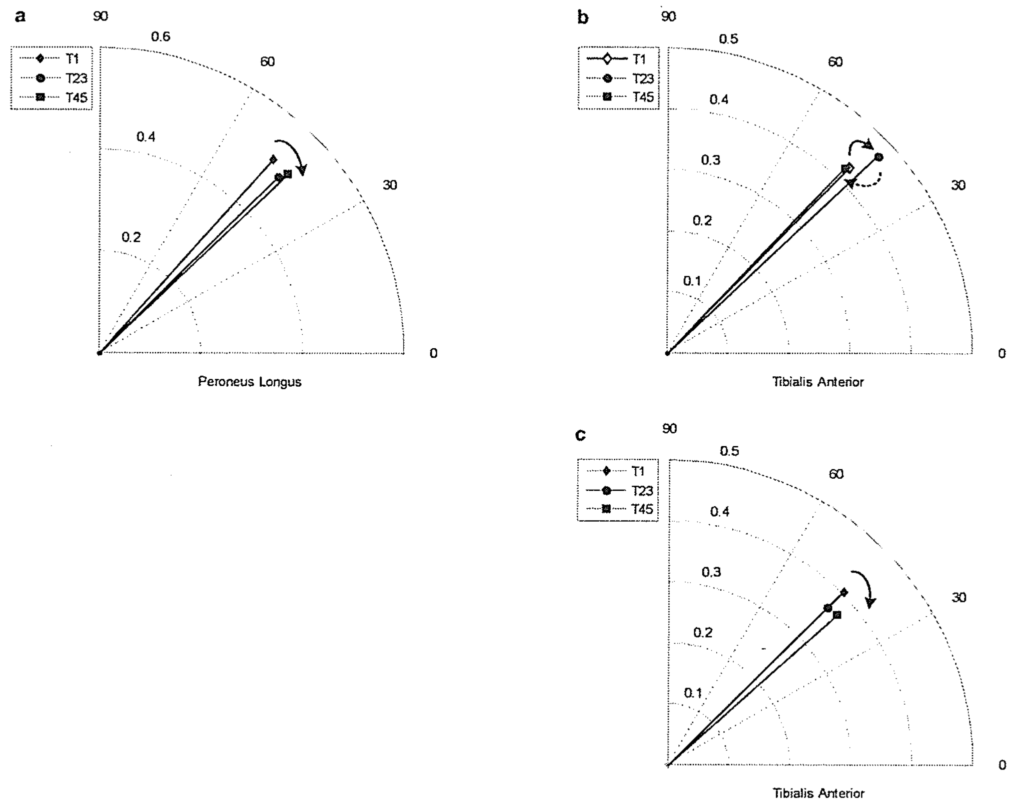

Fig. 3.4. Muscle tuning outcome measures. The average tuning curves, showing the preferred direction of the muscle tuning (Equation 4.3) across the game trials are depicted in (**a**)-(**c**). (**a**) the tuning of the peroneus longus muscle is shown for trial 1 (♦), the average of trials 2 and 3 (•), and the average of trials 4 and 5 (■). As indicated (arrow), there is a clockwise shift in the preferred direction of the muscle. (**b**) the average tuning curve for the tibialis anterior muscle of participants in group A for trial 1 (◊), the average of trials 2 and 3 (•), and the average of trials 4 and 5 (■). As indicated (solid arrow), there is a clockwise shift in the preferred direction of the muscle between T1 and T23. However, in T45, the preferred direction shifts in the counter-clockwise direction (dashed arrow). (**c**) the average tuning curve for the tibialis anterior muscle of participants in group B for trial 1 (♦), the average of trials 2 and 3 (•), and the average of trials 4 and 5 (■). As indicated (arrow), there is a clockwise shift in the preferred direction of the muscle.

SE) (Fig. 3.3e) and in the perpendicular displacement of 27.50 ± 8.45 % ($\pm$ SE) (Fig. 3.3b and Table 3.1). For group CW, the decreases were much less than those of the CCW group, with an average decrease in the displacement angle of 1.87 ± 11.16 % ($\pm$ SE) (Fig. 3.3f) and in the perpendicular displacement of 7.31 ± 4.67 % ($\pm$ SE) (Fig. 3.3c and Table 3.1).

3.4 DISCUSSION

Here we show that participants adopted one of two distinct strategies in order to recalibrate their spatial reference coordinates between the physical COP position and the computer sprite. Participants played a total of five sessions with the rotation applied to the mapping of COP motion to the computer sprite motion. Prior to the rotational transformation, the participants quickly learnt how to shift their body position (COP) to move the cursor and successfully intersect the balloons. When the rotation was applied to the mapping of the COP to computer sprite position, the participants were required to reformulate the spatial relationship between physical space and virtual pixel coordinates of two objects.

When first learning the task, there was no evidence of preplanning for the rotation, which was most likely compensated for by a second, planned movement that occurred after seeing that the computer sprite movement was offset from the target balloon. As participants became familiar with the direction and extent of the errors (seeing the sprite miss the balloon), they began to preplan their movements with the proper offset angle (Fig. 3.3a). This resulted in a consistent clockwise shift in the preferred direction for the peroneus longus for movements to the forward and right for all subjects (Fig. 3.4a). This was the expected result, which would compensate for the counterclockwise rotation applied to the virtual COP movement trajectory. For example, if the subject must hit a balloon target at 90°, they would have to move at an angle of $90^\circ - 60^\circ = 30^\circ$. If they did not, the computer sprite would move at an angle of $90^\circ + 60^\circ = 150^\circ$. This suggests that task accuracy within the time constraints (i.e. bursting of balloon) in the forward and right directions for the peroneus longus muscle was reestablished by a preplanned cognitive re-mapping, expressed by a clockwise tuning.

In terms of the muscle tibialis anterior, two distinct tuning strategies were observed for adapting to the on-screen rotation. Compared to a clockwise shift, a counterclockwise shift in the tuning function between the final and first sessions was consistent with a significantly larger decrease in movement errors. This can be interpreted as the participant first attempting to offset the rotation by tuning the tibialis anterior in the direction opposite to the rotation. Although movement errors were reduced, subsequent reversal of the tuning to a larger value than the original angle resulted in an even larger decrease (Fig. 3.3a,d). This suggests that the counterclockwise shift in the tibialis anterior muscle tuning was a more effective movement strategy, involving a preplanned cognitive re-mapping between the physical body motion and virtual environment. The change in the tuning direction most likely coincided with the ability to preplan the movement trajectory. This result is supported by the fact that the CCW group had larger decreases in their displacement angle than the CW group.

The participants in the CW group were likely unaware of the transformation mechanism, but were prepared to do quick corrections after the movement began. Conversely, the CCW group probably recognized that the movement field was rotated and used this information to preplan their movements. As the CCW group adapted to the task, the degree to which they could preplan (offset) the rotation increased. This is consistent with [Abeele and Bock, 2001], which investigated virtual object tracking using a joystick under congruent conditions and the condition where the virtual trajectory was rotated. Their results found that the adaptation to the rotation was gradual with repeated exposure to the task. These results study suggest that learning a visual transformation is similar for both single segment and full body tasks. However, due to the increased degrees of freedom and dynamic mechanical demands of full body movements, it may take more exposure to fully adapt to the transformation. A further study could be done where the number of trials required to fully adapt to the exposure would be determined.

No consistent pattern was observed for the tuning of the soleus muscle. This could partially be due to the increased task difficulty when moving backwards and to the rear diagonals while the feet are planted. In these directions, our limits of stability are much less, as the horizontal distance from the ankle joint center to the posterior extent of the foot (heel) is less than the distance from the joint center to the anterior extent of the foot (toes) [Liston and Brouwer, 1996]. In addition, the perceived threat of falling backwards is greater than falling forwards, as we cannot see where or what we will fall on. Thus, when required to quickly move in a backwards direction, in addition to focusing on the balloon location and movement direction, a greater awareness is required to deal with increased task demands (i.e., reduced limits of stability and greater consequences of losing balance). This could cause the participants to switch from a single-link inverted pendulum (ankle movement strategy) to a multi-segment movement strategy, using active knee flexion or trunk extension as a counter in order to decelerate the backward movements. Multi-segment movements would cause a different control problem for ankle muscles, which would influence the tuning of the soleus. Finally, note that the soleus along with the gastrocnemius comprise the triceps surae, a multi-joint muscle configuration. Hence, the tuning may be different for this mechanical arrangement.

3.5 SUMMARY

This study showed that two different strategies emerged for adapting to a counterclockwise visual rotation, applied to virtual movement trajectories within a virtual environment based task. The two strategies consisted of a clockwise tuning of the peroneus longus muscle along with either a clockwise or counterclockwise shift of the tibialis anterior muscle. This adaptation occurred after gaining knowledge of the computer sprite's position relative to the target under the new spatial relationship. The amount of learning was quantified by the perpendicular displacement and the displacement angle between the actual and ideal sprite movement trajectories. While a decrease in perpendicular displacement was exhibited, the ability to preplan corrections (i.e. decrease displacement angle) to account for the rotational offset resulted in better learning. This is supported by the fact that no aftereffects were observed for the perpendicular displacement. The decrease in the displacement angle was optimally achieved through a clockwise tuning of the peroneus longus along with a counterclockwise tuning of the tibialis anterior muscles. The counterclockwise tuning of the tibialis anterior was observed after the participants originally tuned the muscle in a clockwise manner. This result is important as it shows that changing the learning strategy can result in better recalibration of internal to external spatial reference frames required to learn task specific visual based rotations viewed in a virtual environment.

4 Stability Assessment and Center of Mass Estimation During Walking

The COM, COP, and body segment acceleration signals are commonly used to indicate movement performance and stability during standing activities and walking. For balance maintenance and restoration, the human brain is capable of estimating and predicting the COM even in the absence of visual or vestibular information. Thus, we hypothesized that the COM may be acquired through the processing of proprioceptive somatosensory information, represented by body segment accelerations, and an external spatial reference, the ground support, represented by the COP. To investigate this hypothesis, we modeled the relationships that exist between the COP and accelerometer data with the 3D COM trajectory, during walking on firm and irregular surfaces. This chapter describes the model developed to estimate the 3D COM trajectory and its validation.

4.1 BACKGROUND

Poor balance and consequently mobility restrictions are limiting factors in a person's health, confidence, ability to perform activities of daily living, and overall quality of life. These factors are serious problems that many older adults and people with neurological and musculo-skeletal disorders face in their day to day lives [Moore et al., 1999]. Thus, the ability to maintain balance becomes an important factor in gait analysis. Common movement signals used to indicate movement performance and balance are the COM, COP, and body segment acceleration signals [Patla, 2003].

During walking, leg swing and foot placement contribute significantly to balance. Adjustments to the COM motion can be made as necessary during the stance phase by intrinsic foot muscles, ankle muscles, and trunk or arm counter-movements [Jian et al., 1993]. Balance control has been investigated during large body disturbances which require a step (change in base of support) to restore stability [Maki and McIlroy, 1999; Tripp et al., 2004]. In [Maki and McIlroy, 1999], compensatory stepping was investigated in the presence of a large perturbation and the displacement and velocity stability margins were determined. The results of [Maki and McIlroy, 1999] indicated that a tradeoff existed between step speed and stability, with stability taking priority for single step recoveries. In [Tripp et al., 2004], it is noted that if the perturbation occurs prior to toe-off of the swing leg, the direction of the compensatory step can be altered in the initial stages of stepping.

When walking, the step is generally broken down into three stages [Henriksen et al., 2004]:

1. *The initial double support phase;* this is the time prior to taking the step, where both feet are planted and adjustments are being made for preparation of swing limb unloading. During this time, the COP moves back towards the swing leg and then forward to the stance leg. This in turn causes the COM to accelerate towards the stance leg [Jian et al., 1993].
2. *The single support phase;* this is the time from toe-off of the swing leg until the heel strike of the swing leg. During this stage, the COP is completely under the stance foot. The COM is moving forward along with

the step, while shifting laterally towards the stance leg [Jian et al., 1993].

3. *The final double support phase*, where both feet are again planted after having taken a step.

During walking, while maintaining a constant forward COM progression, the ML motion must be restrained within the mobile single support base. The mobile base of support changes in size; it is smaller during the single support phase and larger in the double support phase [Patla, 2003]. The ML motion and stability can become increasingly larger and more difficult to control when irregular, unpredictable and unstable surfaces are encountered; this is due to the smaller base of support width [Marigold and Patla, 2005]. Walking outdoors on varied terrains is a high risk task for many older adults and people affected with balance impairments, and is often avoided.

The COM is generally calculated using a video-based motion analysis system, via motion trajectories of markers placed on each body segment and anthropometric models. The resultant three-dimensional (3D) COM trajectory provides us with a picture of the overall movement and the coupling that exists between the movements in different planes. During standing activities and walking, important balance parameters for the COM trajectory are:

- *the amplitude mean* [Marigold and Patla, 2005],
- *the peak-peak amplitude* [Marigold and Patla, 2005],
- *the phase lag*, relative to step initiation, representing the time to peak amplitude across each step [MacLellan and Patla, 2006],
- *the signal variability*, due to differences in biomechanical configurations and balance strategies/gait patterns, from step to step and across subjects

[Moe-Nilssen and Helbostad, 2006].

These parameters are generally calculated for the resultant 3D COM as well as the medial-lateral (ML) and vertical (VT) COM trajectories [Chou et al., 2003; Patla, 2003]. Another parameter that is also calculated is the horizontal distance between the COM and COP. However, as the horizontal projection of the COM can be accurately obtained through appropriate filtering of the COP [Lafond et al., 2004]; hence, this parameter was not investigated in this study.

Another widely used method of evaluating balance is the analysis of body segment accelerations [Smidt et al., 1971; Menz et al., 2003; Henriksen et al., 2004]. However, a study [Hahn and Chou, 2003] demonstrated that the COM was more accurate than acceleration data in distinguishing between dynamic instabilities. As the cost of the video motion analysis system and setting up the markers for the video analysis can be cumbersome and time consuming, we sought an alternative method to predict/estimate the COM.

The human brain controls and maintains the body's balance through the processing of sensory information received by the visual, vestibular, and somatosensory systems, each having a specific reference frame. By analyzing the sensory information, it is believed that the brain estimates the COM in relation to the body's base of support. The corrective actions required to maintain control of balance during planned movements and to restore balance during stumbles are then determined. Given that a healthy individual can maintain his/her balance during standing or walking even in the absence of visual or vestibular information, high weight must be placed on the internal reference frame provided by the proprioceptive somatosensory information and an external reference frame perceived from the ground support of the COP [Peterka, 2002]. Therefore, we hypothesized that the COM may be acquired through the processing of proprioceptive somatosensory information, represented by body segment accelerations, and an external spatial reference, the ground support, represented by the COP.

In our initial investigation, we fit the COM to acceleration data during standing and a mathematical model was presented [Betker et al., 2006b]. However, the coefficients of that model were found through a calibration process, which relied on experimental data. Thus, that model could not replace COM calculation derived from video-based motion systems, as motion data were required during the calibration stage. In this study, we have developed a novel model for COM estimation/prediction using body segment accelerations and the COP position signals. The method does not require any calibration stage (Section 4.3.1). The following sections elaborate on the developed model and its use in stability analysis.

4.2 METHODOLOGY

4.2.1 SUBJECTS

Fifteen healthy subjects (8 females) aged 28.9 ± 4.5, of height 170.1 ± 11.6 cm, weighing 67.3 ± 16.7 kg, with no history of postural problems, volunteered to participate in this study and gave informed consent. Ethics approval was granted prior to recruiting subjects by The University of Manitoba, Faculty of Medicine, Research Ethics Board. All subjects gave their informed consent and were briefed about the tasks and instrumentations before the experiments.

4.2.2 EXPERIMENTAL SETUP

A diagram of the 7 m long walkway, consisting of a fixed floor surface, the test surface, and a firm surface, is given in Fig. 4.1. The length of the fixed floor surface was selected such that two steps could be taken prior to a single right foot step on the test surface. Two different test surfaces were used:

1. A firm surface, which was the same as the remainder of the walkway.
2. An irregular doweling surface. The doweling surface measured 1.2 m x 0.8 m (width x length) and had pieces of doweling 2.4 cm in diameter of length 2.5 cm spaced in an grid 5.6 cm apart.

The irregular surface was used to emulate environmental uncertainty. Foot contact with the irregular surface produces an unpredictable reaction force acting on the body, causing a disturbance to the planned segmental trajectories; this sudden balance disturbance requires a feedback compensatory reaction. The walkway was covered in an outdoor carpet to keep the test surface unknown from the participant. This setup was selected as knowledge of and experience with a surface enables one to anticipate the surface's destabilizing effects; if the surface type can be predicted, one will be more cautious and may reduce walking speed and exhibit anticipatory strategies [MacLellan and Patla, 2006].

Two tri-axial accelerometers (model S2-10G-MF, Biometrics Ltd., Cwmfelinfach, Gwent, UK) were affixed to the subjects using double-sided tape:

1. Representing the trunk segment's movement placed on the spinous process of the second thoracic vertebra (i.e., vertebra T2).
2. Representing the swing leg's movement, placed on the lateral malleolus (ankle).

The 3D sway was recorded by each accelerometer using the analog channels of the Vicon system, at 1080 Hz. Kinematic data was obtained using the VICON 460 video motion analysis system, with six digital video cameras. The data was sampled at 120 Hz. The data rate of 120 Hz is sufficient to capture the walking movements [Winter, 1982] and is the lowest available data rate with the Vicon system (6 cameras). The reflective markers were placed on the end points of each segment and the coordinates were captured via VICON's Plug-in Gait software, according to the Helen Hayes Model [Kabada et al., 1990; Gutierrez-Farewik et al. 2006]. From the coordinate data, the 3D COM position was computed and used for validating the model's COM estimation. COP data was collected via an FSA pressure-sensing insole mat (Vista Medical Ltd., Winnipeg, MB, Canada), which is placed into the shoe of the participant. The insole mat consisted of 128 sensors, arranged in an 8 x 16 grid. Pressure data was captured at a rate of 75 Hz for the combined sensors. The pressure mat used a calibration file to maintain sensor calibration to vertical pressures of 1550 mmHg. In order to synchronize the data collection, a custom trigger was developed, which used the trigger output from the FSA interface box to activate the VICON system.

4.2.3 PROTOCOL

The subjects stood on the fixed floor surface at the beginning of the walkway, with their elbows flexed 90 degrees at their sides to ensure VICON marker visibility (Fig. 4.1). Subjects wore their own comfortable running shoes. The subjects began walking at a comfortable pace beginning with the right foot, with the second right footfall landing on the test surface (firm or irregular). For each subject, ten trials were performed on the firm

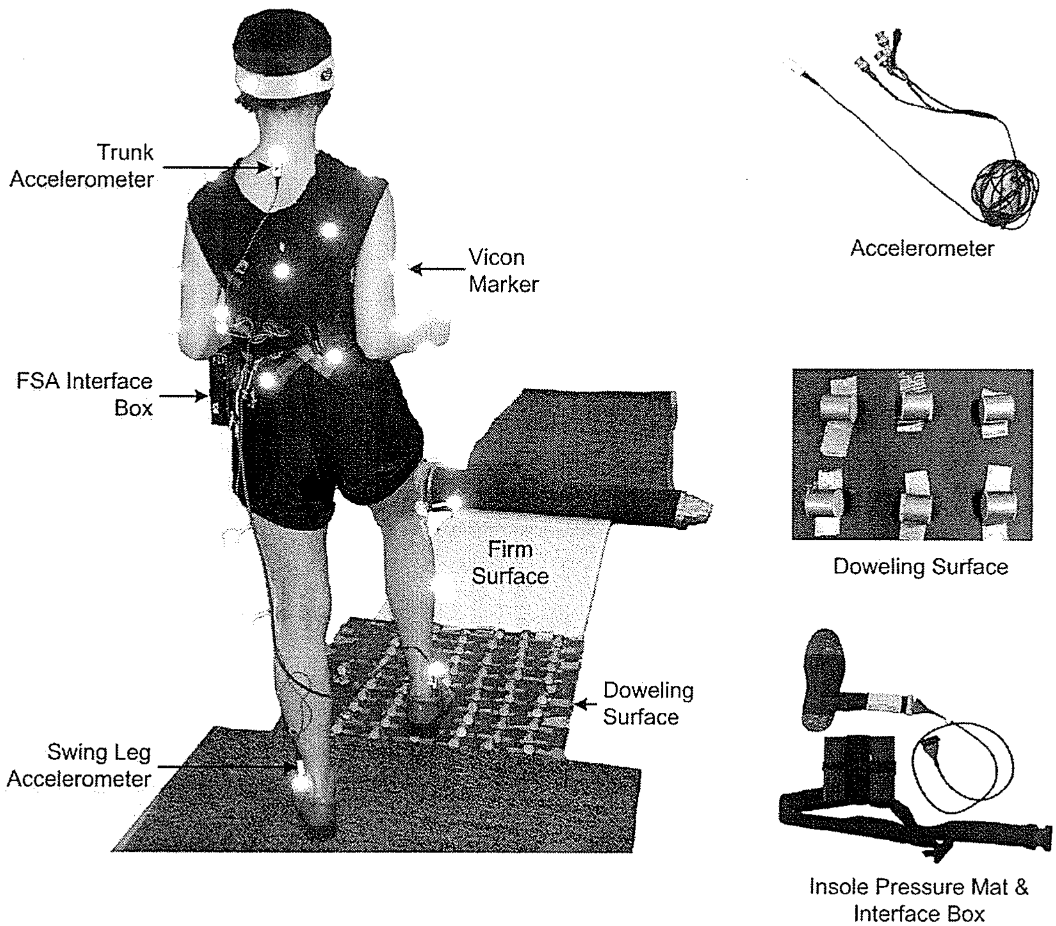

Fig. 4.1. Experimental setup and walkway. The subject first takes a right and left step on the floor. Next, a right step is taken on the test surface, which is either the firm or doweling surface. In this figure, the doweling surface is shown as the test surface. The remaining steps are taken on the firm surface. The walkway is covered in outdoor carpeting, which has been partially removed for illustrative purposes. The placement of the trunk accelerometer at T2 and the placement of the swing leg accelerometer above the lateral malleolus are indicated.

surface and four trials were performed on the irregular surface, with other compliant surface trials conducted in between. The surfaces were randomly presented to the subjects in order to ensure the subjects did not anticipate the disturbance caused by the irregular surface. Each walk trial was separated by a one minute rest period.

4.2.4 DATA CONDITIONING

The total number of data sets collected was 150 for the firm surface, i.e., 10 per subject, and was 20 for the doweling surface, i.e., 4 per subject. The accelerometers' data were downsampled to a rate of 120 Hz, i.e., to that of the kinematic data. Similarly, the COP data was interpolated from 75 Hz to 120 Hz. Next, the right step on the test surface was extracted from the data by applying a threshold to the pressure information of the stance foot, i.e. there will be no pressure when the foot is in swing.

The COM is dependent on the endpoint coordinates of each segment (i.e. segment lengths and orientations in space) and the mass center of each segment. These segment lengths and masses are unique for each person and are incorporated into the COM computation. Hence, the data was then divided by the subject's body mass index (BMI), $BMI = weight/height^2$ [kg/m^2], in order to normalize the data across subjects. This was done via a custom written script in Matlab (The MathWorks, Natick, Mass., USA). Thus, the data for the duration of the single-stance phase for the right step on the test surface was obtained.

4.3 COM APPROXIMATION

The following subsections describe the relationships between the COP, segment accelerations, and COM signals during walking on the two test surfaces, followed by the best fit model between the resultant 3D COM trajectory and the COP and acceleration signals.

4.3.1 MOTION SIGNAL RELATIONSHIPS

Scatter plots describing the relationships between the COP and acceleration signals and the COM components were created to gain insight for model development. A scatter plot displays corresponding data from two signals, where the first signal is plotted along the horizontal axis and the second signal is plotted along the vertical axis. The COM relationships with COP and acceleration data in anterior-posterior (AP) and VT directions were found to have a polynomial relationship (approximately one-dimensional (1D)) as expected [Jian et al., 1993]. However, the relationships in the ML direction was clearly made of a 2 segment piecewise polynomial relationship; that is, the relationship is composed of two segments, where each segment could be described by a polynomial. The following paragraphs describe the found relationships for the ML, AP, and VT COM components, respectively.

COM_{ML} Trajectory. As mentioned in the Introduction, a step can be broken down into three stages. Taking advantage of how the COP and COM move during these stages, the ML component of the COM trajectory (COM_{ML}) can be segmented into two parts. Specifically, a movement valley occurs in the COM_{ML} trajectory; this valley coincides with a valley in the ML ankle acceleration of the swing leg (Fig. 4.2). Hence, the ankle acceleration of the swing leg is used to divide the COM_{ML} trajectory into segments A and B: segment A is from the beginning of the step until the time at which the movement valley occurs and segment B is from the time at which the movement valley occurs until the end of the step (indicated by ◊ in Fig. 4.2). For the segment A COM_{ML} trajectory, no consistent relationship was found between the trunk or swing leg accelerations or the COP. However, we recorded position data for 30 motion analysis segment markers. This

position data was double differentiated to see whether the given segment's acceleration would be a more appropriate location for future accelerometer placement. The ML acceleration of the right hip marker was found to have the most consistent relationship with the COM_{ML} trajectory in segment A. Therefore, the hip marker acceleration was used as an input to the model. In segment B, the COP_{ML} trajectory was found to have the most consistent linear relationship with the COM_{ML} trajectory.

COM_{AP} Trajectory. The AP component of the COM trajectory (COM_{AP}) reflects the forward progression of the movement and accounts for the largest COM component

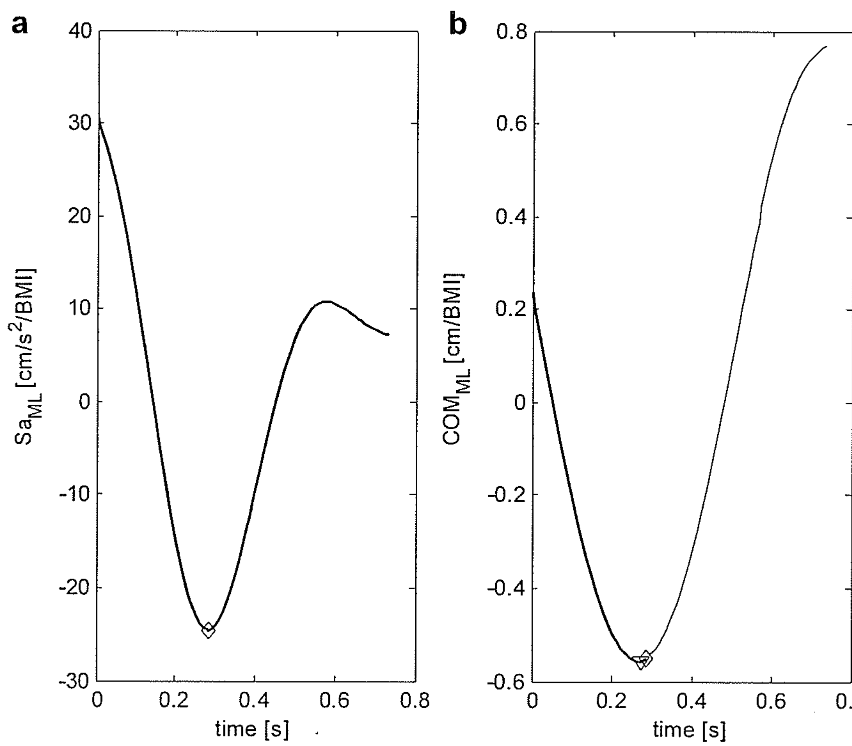

Fig. 4.2. Segmentation plots for a typical subject's step on the firm surface: (**a**) medial-lateral (ML) swing leg acceleration (Sa_{ML}), with the minimum amplitude indicated (◊); and (**b**) the ML center of mass (COM_{ML}), with the minimum amplitude of both the Sa_{ML} (◊) and COM_{ML} (∇) indicated. The time at which the minimum amplitude of the Sa_{ML} occurs is used to segment the COM_{ML} into parts A (-) and B (-.-).

movement. A consistent linear relationship was found between the AP component of the COP trajectory (COP_{AP}) and COM_{AP}.

COM_{VT} Trajectory. For the VT component of the COM trajectory, COM_{VT}, a consistent linear relationship was found with the vertical trunk acceleration.

4.3.2 3D RESULTANT COM (COM_R) MODEL

The resultant 3D COM trajectory, $\mathbf{COM}_R$, accounts for the overall COM motion and the coupling that occurs between each movement direction. The $\mathbf{COM}_R$ is calculated according to

$$\mathbf{COM}_R = \sqrt{\mathbf{COM}_{AP}{}^2 + \mathbf{COM}_{ML}{}^2 + \mathbf{COM}_{VT}{}^2}\,. \tag{4.1}$$

The model requires inputs from a pressure mat (to obtain the COP) and from three accelerometers (to get segment accelerations for the swing leg, hip, and trunk). Note that in this study only two accelerometers were used to obtain data; however, during the analysis it was determined that a third accelerometer would be required. Hence, we used the double differentiated hip marker trajectory (from the video motion system) as the hip acceleration signal. In future studies, an accelerometer should also be placed on the hip to provide this signal. The relationships determined in Section 4.3.1 provided the basic model for each of the COM components, from which the overall resultant COM could be estimated (Fig. 4.3). Statistical properties of the signals, such as the standard deviation, were used to appropriately scale the inputs to the model, i.e., to scale the input amplitudes to the output amplitudes. Thus, no calibration is required for the model. The equations developed for the COM components are described in the following paragraphs and in Fig. 4.3.

The estimated $\hat{\mathbf{COM}}_{ML}$, was calculated according to

$$\begin{aligned}
\hat{\mathbf{COM}}^{A}_{ML} &= \frac{1}{2\sigma_{\mathbf{Ha}^{A}_{ML}}}\mathbf{Ha}^{A}_{ML} \\
\hat{\mathbf{COM}}^{B}_{ML} &= \frac{1}{2\sigma_{\mathbf{COP}^{B}_{ML}}}\mathbf{COP}^{B}_{ML} \\
\hat{\mathbf{COM}}_{ML} &= \left[\hat{\mathbf{COM}}^{A}_{ML} \quad \hat{\mathbf{COM}}^{B}_{ML}\right]
\end{aligned} , \tag{4.2}$$

where the A and B superscripts indicate segments A and B, respectively, as segmented by the swing leg acceleration in the ML direction, $\mathbf{Sa}_{ML}$; $\mathbf{Ha}^{A}_{ML}$ is the hip acceleration in ML direction, $\mathbf{COP}^{B}_{ML}$ is the COP trajectory in the ML direction, and σ is the standard deviation of a given signal. The estimated $\hat{\mathbf{COM}}_{AP}$ component was calculated according to

$$\begin{aligned}
y &= a\mathbf{COP}_{AP} + b \\
y^{N} &= \frac{y-\mu_y}{\sigma_y} \\
\hat{\mathbf{COM}}_{AP} &= y^{N} \cdot \left[\frac{1}{2}\frac{\mathrm{pp}(\mathbf{Ha}_{ML}) + \mathrm{pp}(\mathbf{Ta}_{AP})}{\mathrm{pp}(y^{N})}\right]
\end{aligned} , \tag{4.3}$$

where y is a line fit to the $\mathrm{COP_{AP}}$ component via coefficients a and b, μ is the mean of a given signal, σ is the standard deviation of a given signal, y^N indicates the normalized fit y, $\mathbf{Ta}^{A}_{AP}$ is the trunk acceleration in the AP direction, and $\mathrm{pp}(\cdot)$ indicates the peak-peak amplitude of a given signal. The estimated $\mathrm{COM_{VT}}$ component was calculated according to

$$\hat{\mathbf{COM}}_{VT} = \frac{1}{1.4\sigma_{\mathbf{Ta}_{VT}}}\mathbf{Ta}_{VT}, \tag{4.4}$$

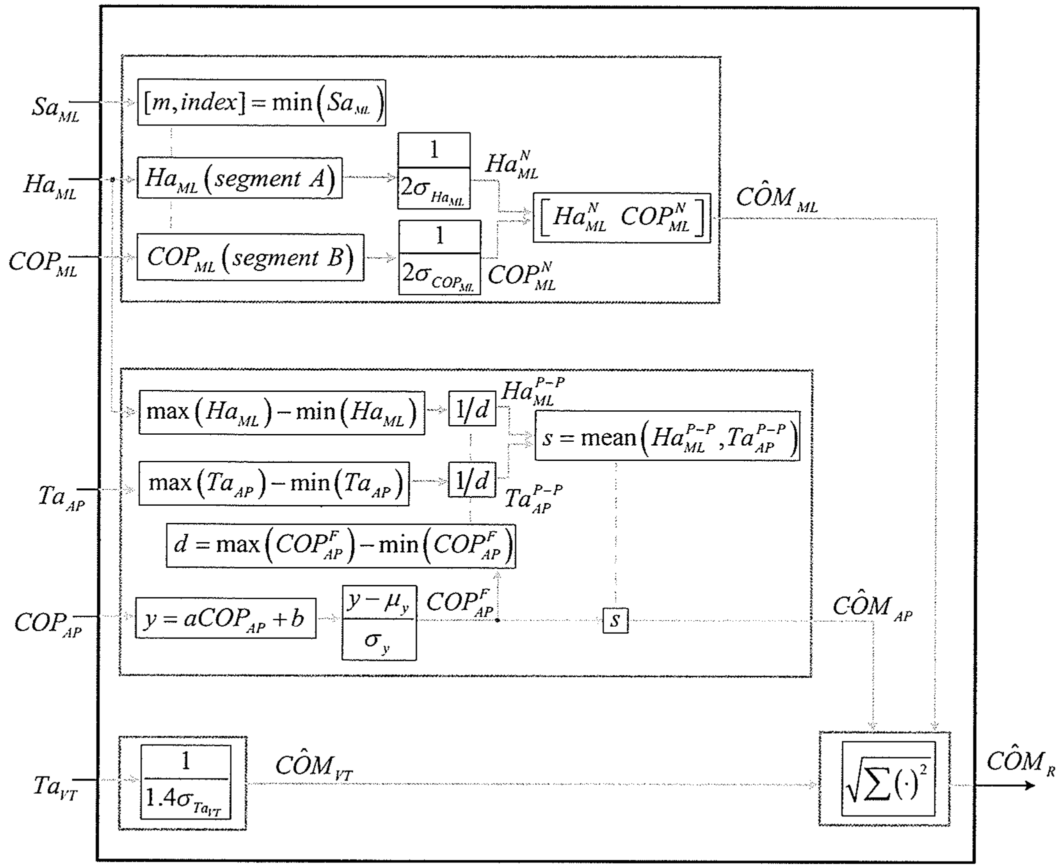

Fig. 4.3. Block diagram of resultant COM trajectory model. Inputs to the block estimating the medial-lateral (ML) center of mass (COM) trajectory ($\hat{COM}_{ML}$) are: the swing leg ML acceleration (Sa_{ML}), the hip ML acceleration (Ha_{ML}), and the center of foot pressure (COP) ML trajectory (COP_{ML}). Inputs to the block estimating the anterior-posterior (AP) COM trajectory ($\hat{COM}_{AP}$) are: Ha_{ML}, the AP COP trajectory (COP_{AP}), and the AP trunk acceleration (Ta_{AP}). The input to the block estimating the vertical (VT) COM trajectory ($\hat{COM}_{VT}$) is the VT trunk acceleration (Ta_{VT}). The 3D estimated resultant COM ($\hat{COM}_{R}$) is then determined using $\hat{COM}_{ML}$, $\hat{COM}_{AP}$, and $\hat{COM}_{VT}$.

where $\mathbf{Ta}_{VT}$ is the trunk acceleration in VT direction. The estimated resultant COM trajectory, $\hat{\mathbf{COM}}_R$, was calculated from the estimated components according to (1), for each subject and for each of the two surfaces.

4.3.3 OUTCOME MEASURES

Model performance. For each test surface, the following outcome measures were calculated:

1. The correlation coefficient (CC) R between the actual ($\mathbf{COM}_R$) and estimated ($\hat{\mathbf{COM}}_R$) resultant COM trajectories.
2. The coefficient of determination (CD) between the $\mathbf{COM}_R$ and $\hat{\mathbf{COM}}_R$, calculated as R^2, representing the amount of variability the model accounts for.
3. The percentage error (err) between $\mathbf{COM}_R$ and $\hat{\mathbf{COM}}_R$, according to

$$err = \left[\left(\hat{\mathbf{COM}} - \mathbf{COM} \right) \big/ \mathbf{COM} \right] \cdot 100\%. \tag{4.5}$$

4. The absolute error between $\mathbf{COM}_R$ and $\hat{\mathbf{COM}}_R$ in centimeters.

Stability outcomes. The following trajectories were calculated for the actual COM (derived by the motion analysis system) and the estimated COM in the ML, AP, and VT directions and for the resultant: $\mathbf{COM}_{ML}$, $\hat{\mathbf{COM}}_{ML}$, $\mathbf{COM}_{VT}$, $\hat{\mathbf{COM}}_{VT}$, $\mathbf{COM}_R$, and $\hat{\mathbf{COM}}_R$. Commonly used measurements to quantify the subject's walking performance are:

1. *Signal amplitude mean,* calculated across all subjects [Marigold and Patla, 2005; Segers et al., 2007].

2. *Signal peak-to-peak amplitudes,* calculated across all subjects [Marigold and Patla, 2005].
3. *Signal phase lag,* calculated across all subjects using the procedure described in [MacLellan and Patla, 2006]. The phase lag equivalent to step periodicity was calculated from the unbiased autocorrelation sequence, as the time index representing the first dominant peak [Moe-Nilssen and Helbostad, 2004].
4. *Signal variability:* a signal representing the mean step trajectory for each signal was calculated across all subjects. The correlation coefficient was then calculated between each step and the mean trajectory.

These stability outcomes were calculated for each of the aforementioned trajectories, in order to determine whether or not the estimated signals would provide the same stability information as the actual signals. Note that these outcomes were only used from a model validation perspective and a participant stability assessment was not performed.

4.3.4 VALIDATION

The proposed model was validated from two aspects:

1. The percentage and actual errors between the actual and estimated resultant COM were compared.
2. The outcome measures derived from the actual and estimated resultant, ML, and VT COM trajectories were examined using a one-way repeated measure analysis of variance (ANOVA). The independent variables were the trajectory type: actual or estimated. The dependent variables were each

of the stability outcome measures. The analysis was performed in Matlab (The MathWorks, Natick, Mass., USA), with significance level $p = 0.05$.

4.4 RESULTS

The system inputs and the actual resultant COM trajectory (system output) of a typical subject for the firm and doweling surfaces are shown in Fig. 4.4. Note that for the model input signals (Fig. 4.4a-f), the initial 200-300 ms of the signals are quite different between the firm and doweling surfaces. The initial difference in these trajectories reflects the mechanical balance perturbation produced by unexpectedly stepping on the irregular doweling surface. In general, the recovery from the surface perturbation did

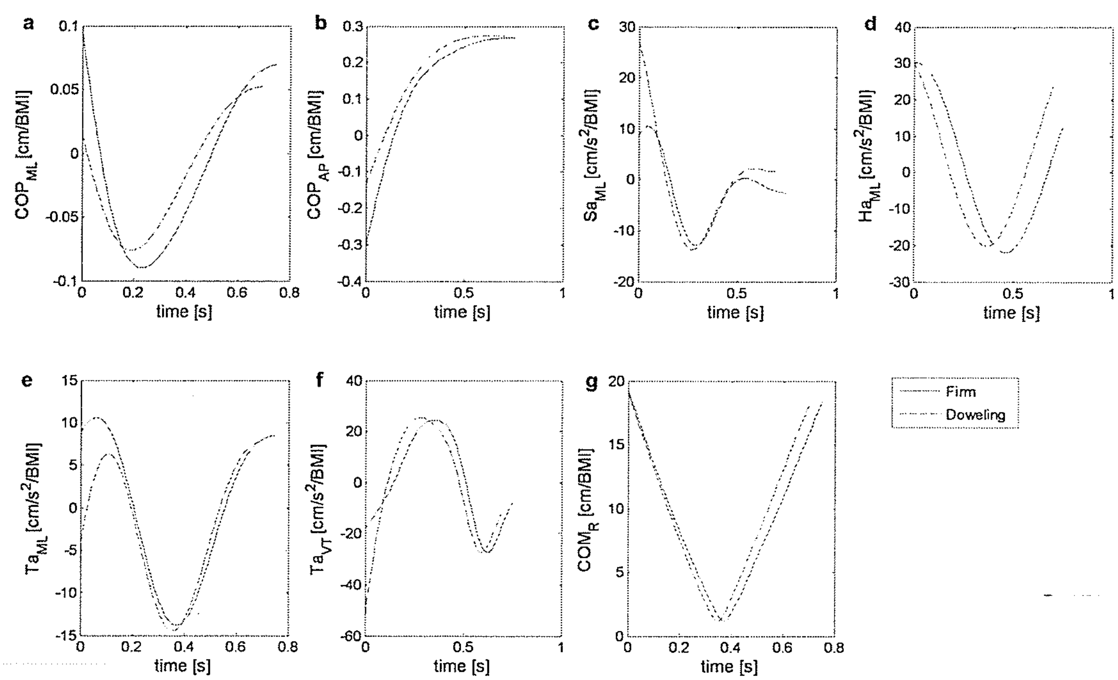

Fig. 4.4. Model inputs (a)-(f) and actual resultant COM (g), for the firm (-) and doweling (--) surfaces: a) medial-lateral (ML) center of foot pressure (COP); b) anterior-posterior (AP) COP; c) ML swing leg acceleration (Sa); d) ML hip acceleration (Ha); e) ML trunk acceleration (Ta); f) vertical (VT) Ta; and g) resultant (R) center of mass (COM). Note that all signals are normalized by the subject's BMI [kg/m^2].

occur quickly (within 300 ms). However, for the Ha_{ML} (Fig. 4.4d) the plots remain different for the entire step duration. A phase lag is also observed for all signals (Fig. 4.4a-g). The results for one step of each subject, for each surface, are shown in Fig. 4.5.

4.4.1 MODEL PERFORMANCE

In all parts of the COM estimation/prediction, the correlation coefficients with respect to the actual COM were quite high (≥ 0.997). Table 4.1 presents the summary, as well as the error defined in equation (5) and standard deviation (SD). The results show a very high correlation between the actual and estimated COM. While the error was higher for the doweling surface than that of the firm surface (as expected), the overall the COM followed the actual COM quite well (Fig. 4.5). In terms of absolute difference in centimeters, the errors were 3.62 ± 2.69 cm and 4.74 ± 3.01 cm for the firm and doweling surfaces respectively.

4.4.2 STABILITY OUTCOMES

The mean and SD for each outcome measure (amplitude mean, peak-to-peak amplitudes, phase lags, and correlation coefficient) for $\mathbf{COM}_{ML}$, $\hat{\mathbf{COM}}_{ML}$, $\mathbf{COM}_{VT}$, $\hat{\mathbf{COM}}_{VT}$,

Table 4.1. Comparison of model derived $\hat{\mathbf{COM}}_R$ and recorded $\mathbf{COM}_R$: correlation coefficient, coefficient of determination, and the error.

Surface	Correlation Coefficient (mean ± SD)	Coefficient of Determination (mean ± SD)	Error (%) (mean ± SD)	Absolute Error [cm] (mean ± SD)
Firm	0.9993±0.0010	0.9985±0.0020	16.06±11.11	3.62±2.69
Doweling	0.9988±0.0019	0.9977±0.0039	21.41±12.70	4.74±3.01

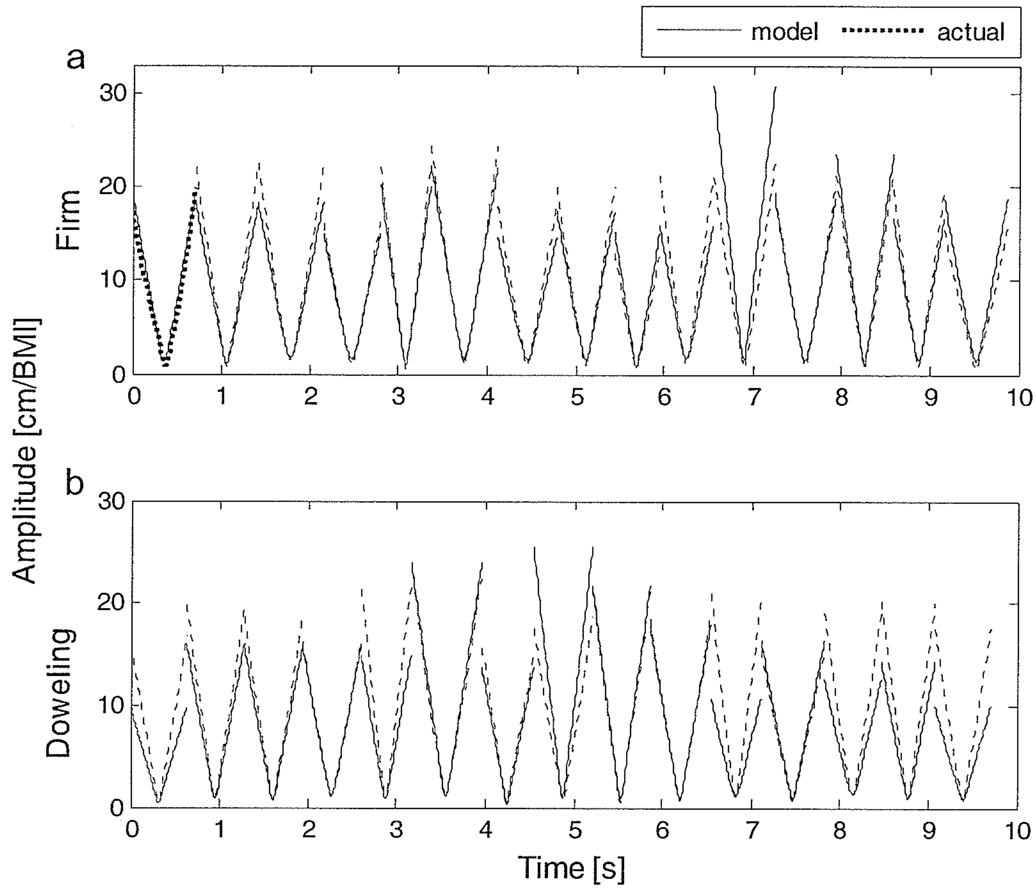

Fig. 4.5. Model results: modeled (-) and true (--) normalized resultant COM trajectory for: (**a**) the firm surface; and (**b**) the doweling surface. One step is shown for each of the 15 subjects. All signals are normalized by the subject's BMI [kg/m^2].Each step is shown as a function of time, however note that the time is not continuous across steps.

$\mathbf{COM}_R$, and $\hat{\mathbf{COM}}_R$, for each subject and surface, are given in Table 4.2. The amplitude ranges for the model inputs and outputs are given in Table 4.3. Congruent with the results presented in Table 4.1, the estimated COM resulted in stability measures similar to those calculated from the actual COM except in a few conditions. Those few exceptions were the significant differences between the mean peak-to-peak amplitude of the $\mathbf{COM}_{ML}$ and $\hat{\mathbf{COM}}_{ML}$ for the doweling surface, significant differences in the correlation coefficient values between $\mathbf{COM}_{ML}$ and $\hat{\mathbf{COM}}_{ML}$ for both surfaces and those of $\mathbf{COM}_{VT}$ and $\hat{\mathbf{COM}}_{VT}$ for the firm surface.

4.5 DISCUSSION

Subjects walked to the end of a 7 m walkway, with forward progression being maintained. The relationship between the COP and acceleration signals and the resultant COM was modeled, with the $\hat{COM}_{ML}$, $\hat{COM}_{AP}$, and $\hat{COM}_{VT}$ components existing as hidden outputs. The resultant COM trajectory accounts for the overall COM motion and

Table 4.2. Outcome measures (across all subjects) for $\hat{\mathbf{COM}}_R$ and $\mathbf{COM}_R$, $\hat{\mathbf{COM}}_{ML}$ and $\mathbf{COM}_{ML}$, and $\hat{\mathbf{COM}}_{VT}$ and $\mathbf{COM}_{VT}$: mean amplitude, peak-to-peak amplitude, phase lag, and correlation coefficient for the given signal across all subjects. Significant differences are denoted by *. Shading represents the pairs of outcomes for the estimated and actual trajectories.

Surface	Mean Amplitude [cm/BMI] (mean ± SD)					
	$\hat{COM}_R$	COM_R	$\hat{COM}_{ML}$	COM_{ML}	$\hat{COM}_{VT}$	COM_{VT}
Firm	9.73 ± 1.98	9.74 ± 1.25	0.00 ± 0.00	0.00 ± 0.00	0.00 ± 0.00	0.00 ± 0.00
Doweling	8.82 ± 2.07	9.65 ± 1.33	0.00 ± 0.00	0.00 ± 0.00	0.00 ± 0.00	0.00 ± 0.00
	Peak-to-peak Amplitude [cm/BMI] (mean ± SD)					
	$\hat{COM}_R$	COM_R	$\hat{COM}_{ML}$	COM_{ML}	$\hat{COM}_{VT}$	COM_{VT}
Firm	17.97 ± 3.99	18.80 ± 2.63	1.48 ± 0.07	1.28 ± 0.37	2.30 ± 0.15	2.38 ± 0.29
Doweling	16.17 ± 4.13	18.58 ± 2.71	1.60 ± 0.19*	1.23 ± 0.41	2.22 ± 0.08	2.34 ± 0.21
	Phase Lag [ms] (mean ± SD)					
	$\hat{COM}_R$	COM_R	$\hat{COM}_{ML}$	COM_{ML}	$\hat{COM}_{VT}$	COM_{VT}
Firm	67.5 ± 5.8	67.5 ± 5.8	67.7 ± 5.9	68.7 ± 6.2	67.4 ± 6.2	66.0 ± 6.0
Doweling	64.4 ± 6.1	62.5 ± 2.4	64.4 ± 6.7	66.7 ± 6.7	67.4 ± 6.2	66.3 ± 6.3
	Correlation Coefficient (mean ± SD)					
	$\hat{COM}_R$	COM_R	$\hat{COM}_{ML}$	COM_{ML}	$\hat{COM}_{VT}$	COM_{VT}
Firm	0.95 ± 0.07	0.95 ± 0.07	0.84 ± 0.36*	0.73 ± 0.33	0.84 ± 0.14*	0.90 ± 0.12
Doweling	0.95 ± 0.06	0.95 ± 0.05	0.92 ± 0.12*	0.68 ± 0.29	0.84 ± 0.19	0.87 ± 0.19

Table 4.3. Mean peak-peak amplitude ranges for model inputs (pressure and accelerations) across all subjects.

Parameter	Surface	
	Firm	Doweling
ML COP [cm/BMI] (mean ± SD)	0.16 ± 0.07	0.14 ± 0.06
AP COP [cm/BMI] (mean ± SD)	0.48 ± 0.14	0.44 ± 0.12
Swing ML Acceleration [cm/s²/BMI] (mean ± SD)	49.19 ± 29.16	52.75 ± 27.82
Hip ML Acceleration [cm/s²/BMI] (mean ± SD)	46.45 ± 15.73	41.29 ± 16.48
Trunk AP Acceleration [cm/s²/BMI] (mean ± SD)	28.85 ± 10.68	26.67 ± 11.40
Trunk VT Acceleration [cm/s²/BMI] (mean ± SD)	67.53 ± 17.51	66.65 ± 19.34

the coupling that occurs between each movement direction. The model accurately accounted for the variability in the $\mathbf{COM}_R$ signal, with coefficients of determination values of 0.9985 ± 0.0020 and 0.9977 ± 0.0039, for the firm and doweling surfaces, respectively. This is not surprising if we look at the low variability of the model inputs across all subjects (Table 4.4). In general, there is low variability in the body accelerations and COP (model inputs) across subjects after normalization by BMI. Hence, the model (with inputs from miniature accelerometers attached to a few select body locations) could predict COM without the need for calibration for every subject or the need for coordination of data from all body segments.

The AP, VT and resultant COM trajectories have low variability across subjects. This is important as the COM_{AP} component has the largest contribution to the resultant COM trajectory. However, there was higher variability in the COM_{ML} component [Menz et al., 2003; Oddsson et al., 2004]. In [Kaya et al., 1998] it was found that during locomotion, elderly subjects had a difficult time controlling ML momentum. This

Table 4.4. Correlation coefficients (variability) of the data signals across all subjects.

Signal	Firm Surface (mean ± SD)	Irregular Surface (mean ± SD)
ML COP	0.84 ± 0.36	0.93 ± 0.09
AP COP	0.98 ± 0.16	0.93 ± 0.29
ML Swing Acceleration	0.74 ± 0.30	0.70 ± 0.28
ML Hip Acceleration	0.86 ± 0.26	0.57 ± 0.42
AP Trunk Acceleration	0.84 ± 0.18	0.79 ± 0.24
VT Trunk Acceleration	0.84 ± 0.14	0.84 ± 0.19
ML COM	0.73 ± 0.33	0.68 ± 0.29
AP COM	1.00 ± 0.00	1.00 ± 0.00
VT COM	0.90 ± 0.12	0.87 ± 0.19
Resultant COM	0.95 ± 0.07	0.95 ± 0.05

resulted in ML accelerations being more phasic; hence in single support stance, ML COM acceleration and deceleration occurred within a short time period (100-200 ms). Similarly, [Menz et al., 2003] found that healthy subjects exhibited irregular acceleration patterns in the ML pelvis trajectory. Thus, ML movements are the hardest to perceive and predict. While there was low variability in the AP and VT directions across subjects, there is higher variability in the ML direction. This is expected as corrective movements take place in the ML direction about a narrow base of support, which require rapid short duration movements of a more random nature that do not follow a consistent pattern. This was reflected in the significant difference across subjects in the variability of the estimated and actual COM_{ML}, particularly when on the doweling surface. It should be noted that the model did not account for visual information, which would include information about these corrective movements.

Partial validation of the model was done through the percentage error difference between the actual and estimated COM values. The error represents how well the model can predict the COM; the model is expected to have some error as it is blind of other useful spatial information, such as visual and vestibular sensory information.

At steady state gait, the COM trajectory during walking on a fixed and firm surface is generally rhythmic. When challenging surface conditions are encountered, in which foot contact information and ground reaction forces are distorted and unpredictable, the gait rhythm becomes disrupted [Oddsson et al., 2004]. In this study, the doweling surface caused disturbing and temporarily unopposed ground reaction forces, which in turn produced errors in the movement/balance system's predictions and caused a disruption to the gait rhythm. This was reflected in the increased error between the estimated and actual resultant COM for the doweling surface ($21.41 \pm 12.70\%$) when compared to that for the firm surface ($16.06 \pm 11.11\%$). Also note the signal differences between the plots (Fig. 4.4) for the firm and doweling surfaces; in the early time periods there is quite a large difference, however, the mechanical disturbance and balance loss is momentary, with quick restoration of stability (200-300 ms).

Although the overall output of our model (Fig. 4.3) was $\mathbf{C\hat{O}M}_R$, as hidden outputs the model also has $\mathbf{C\hat{O}M}_{ML}$, $\mathbf{C\hat{O}M}_{AP}$, and $\mathbf{C\hat{O}M}_{VT}$. Since the ML and VT components of COM are also used in stability analysis [Chou et al., 2003; Patla, 2003], the stability outcomes were additionally calculated for these components. For the firm surface, no significant difference ($p < 0.05$) was observed in the amplitude mean, peak-to-peak amplitude, or the phase lag between $\mathbf{C\hat{O}M}_R$ and $\mathbf{COM}_R$, $\mathbf{C\hat{O}M}_{ML}$ and $\mathbf{COM}_{ML}$, or $\mathbf{C\hat{O}M}_{VT}$ and $\mathbf{COM}_{VT}$. Thus, the derived relationship for the firm surface could accurately represent the amplitude excursions and phase information for the $\mathbf{COM}_{ML}$, $\mathbf{COM}_{VT}$, and $\mathbf{COM}_R$. For the doweling surface, a significant difference was observed in the peak-to-peak amplitude between $\hat{\mathbf{COM}}_{ML}$ and $\mathbf{COM}_{ML}$. In response to the irregular doweling surface, compensatory reactions were reflected in greater variability of the COM_{ML} signal (Table 4.4). In addition, the COP_{ML} and hip acceleration trajectories for the doweling surface had peak-to-peak amplitudes that were significantly different than those for the firm surface ($p < 0.05$). Specifically, the mean peak-to-peak amplitude for the COP_{ML} and hip accelerations on the doweling surface was smaller than those for the firm surface. In effect, this means that the doweling surface had a damping effect on the gait rhythm; if the brain has difficulty with accurately predicting or determining the surface properties (ground reaction forces), then the motion of the swing leg (transition from double to single support phases) will be delayed until the muscles of the single support stance leg can absorb and compensate for the disturbing effects of the irregular surface.

The variability of $\hat{\mathbf{COM}}_{ML}$, $\hat{\mathbf{COM}}_{VT}$, and $\hat{\mathbf{COM}}_{R}$ across all subjects were also investigated. Significant differences existed in the variability between $\hat{\mathbf{COM}}_{ML}$ and $\mathbf{COM}_{ML}$ and between $\hat{\mathbf{COM}}_{VT}$ and $\mathbf{COM}_{VT}$ for the firm surface. On the doweling surface, significant difference was observed in the variability between $\hat{\mathbf{COM}}_{ML}$ and $\mathbf{COM}_{ML}$. In these cases, the variability was less for the estimated signals. This is a result of the input signals generally having a lower variability than the COM_{ML}. The variability in $\hat{\mathbf{COM}}_{R}$ was not significantly different than $\mathbf{COM}_{R}$ for either surface.

4.6 SUMMARY

The center of body mass, center of foot pressure, and body segment acceleration signals are commonly used to indicate movement performance and stability during standing activities and walking. The COM is generally calculated using a video-based motion analysis system, via motion trajectories of markers placed on each body segment and anthropometric models. The resultant three-dimensional COM trajectory provides us with a picture of the overall movement and the coupling that exists between the movements in different planes. During standing activities and walking, important stability outcomes for the COM trajectory are:

- The amplitude mean [Marigold and Patla, 2005],
- The peak-peak amplitude [Marigold and Patla, 2005],
- The phase lag, relative to step initiation, representing the time to peak amplitude across each step [MacLellan and Patla, 2006],
- The signal variability, due to differences in biomechanical configurations and balance strategies/gait patterns, from step to step and across subjects [Moe-Nilssen and Helbostad, 2006].

For balance maintenance and restoration, the human brain is capable of estimating and predicting the COM even in the absence of visual or vestibular information. Thus, we hypothesized that the COM may be acquired through the processing of proprioceptive somatosensory information, represented by body segment accelerations, and an external spatial reference, the ground support, represented by the COP. To investigate this hypothesis, we modeled the relationships that exist between the COP and accelerometer data with the 3D COM trajectory, while healthy participants walked on firm and irregular surfaces.

The model was validated by calculating:

1. The correlation coefficient and coefficient of determination between the

actual ($\mathbf{COM}_R$) and estimated ($\hat{\mathbf{COM}}_R$) resultant COM trajectories, representing the amount of variability the model accounts for. The models accounted for $99.85 \pm 0.20\%$ and $99.77 \pm 0.39\%$ of the resultant COM trajectory's variability for the firm and irregular surfaces, respectively.

2. The percentage error and actual error between $\mathbf{COM}_R$ and $\hat{\mathbf{COM}}_R$. The percentage error for the model was 16.06 ± 11.11 % and 21.41 ± 12.70 % for the firm and doweling surfaces, respectively. In turn, this translated into an absolute error between the true and actual resultant COM of 3.62 ± 2.69 cm and 4.74 ± 3.01 cm for the firm and doweling surfaces respectively.

In terms of clinical outcomes, the aforementioned stability outcomes derived from the actual and estimated COM were compared. The estimated COM resulted in stability measures similar to those calculated from the actual COM.

The derived model is novel in that it does not require any calibration and provides a reasonably accurate estimation of the COM, which can be compared to the brain's balance performance. Hence, this model could be used instead of the cumbersome method of video motion analysis for COM calculation.

5 CONCLUSIONS AND FUTURE RESEARCH

5.1 SUMMARY AND CONCLUSION

In this dissertation, the novelty of the research is threefold. Together, the three different studies done in this thesis give an integrated and coherent understanding of the human balance system. The primary results of this research can be combined together to achieve an interactive tool to provide therapy to the patient, to assess their ability to learn required recalibrations of internal to external spatial reference frames, and to evaluate their stability and control through a COM estimation method that is independent of any calibration. This research concerned the development of objective clinical tools for treating balance disorders and evaluating the performance of the main tasks related to community ambulation and instrumental activities of daily living. The outcomes are important as balance and mobility are necessary in order to independently perform acts of daily living and to avoid falls causing injuries and/or hospitalization. The ability to effectively treat and assess balance disorders will play a role in maintaining and restoring quality of life. A summary and conclusion of current research is provided, followed by a summary of future research.

5.1.1 INTERACTIVE REHABILITATION TOOL

An interactive virtual environment-based tool was developed, which coupled the COP movement trajectory to an on-screen cursor (computer sprite). Four different tasks were developed, which sought to improve dynamic balance control in both standing and short-sitting. The tasks Tic-Tac-Toe and Memory Match were designed to elicit quick, short movements followed by a sustained hold in the computer sprite position, in order to select a square/card. In Under Pressure, the participant had to shift their body weight to move a receptacle in line to catch a moving object. In Balloon Burst, quick and precise movements were required to move the computer sprite to intersect and pop the stationary balloon.

The benefits of the developed interactive virtual environment-based tool were evaluated via standing and short-sitting case studies, where the amount of recovery was quantified pre-exercise versus post-exercise. Six different tasks were performed on two different surfaces (Section 2.3.3.2). For the short-sitting case study, the number of falls was compared pre- and post-exercise. Post-exercise, all subjects exhibited a lower fall count and in fact, were able to complete all balance 6 balance tasks on both surfaces. For the standing case study, the number of falls, the range of COP excursion, and the COP path length outcomes were compared pre- and post-exercise. Post-exercise, the subjects exhibited a lower fall count and a decreased COP excursion limits for some tasks. In both the short-sitting and standing case studies, participants exhibited an increased practice volume and an increased attention span during training. These findings demonstrated that graded, dynamic balance exercises on different surfaces can be effectively coupled to virtual environment-based tasks. In addition to the training program being enjoyable, all subjects improved their fall rate performance after using the virtual environment-based therapy. Hence, this approach was effectively applied to:

- People who had severe balance and mobility limitations, including long standing, secondary musculo-skeletal problems.

- People with CNS lesions, hemiplegia and bilateral deficits.
- People who were actively participating in sports.

5.1.2 STRATEGIES FOR LEARNING NEW MOTOR TASKS

As we perform basic and instrumental activities of daily living, many sensory motor transformations are required to successfully map the changing relationships between one's self, the environment, and objects moving in the environment. Recent developments strongly suggest that elements of rehabilitative regimes incorporating real-time biofeedback into interactive virtual environments, which emulate egocentric and allocentric relationships, are motivational and can substantially increase volume of task-specific therapy. The combined use of physical movements and virtual psychomotor tasks, having specific goals and graded spatiotemporal accuracy, represents functional training with a wide scope of sensory motor and cognitive experiences, which can drive neuroadaptation sufficient for producing behavioral recovery. It is important to train and evaluate people under congruent visual-spatial conditions and those which systematically change the sensory motor transformation. Hence, in the virtual task Balloon Burst, a visual rotation was applied which changed the manner in which the COP trajectory (biofeedback signal) moved the computer screen sprite (perceived visually). Specifically, the computer sprite's movement was rotated counter-clockwise by 60°.

In this study, healthy participants first learned how to play Balloon Burst with a congruent mapping between the COP and computer sprite motion. Next, the rotation was applied to the mapping and five sessions were played. Finally the task was again performed with the congruent mapping. The following parameters were calculated in order to investigate the adaptation strategies:

- The displacement angle between the COP trajectory and the direct line path between the starting COP position and target position.
- The maximum perpendicular displacement between the COP trajectory and the direct line path to the balloon target.
- The preferred direction, which represents the directional sum of the muscle activity (RMS) for a given muscle, for the tibialis anterior, soleus, and peroneus longus muscles.

We observed that two different strategies emerged for adapting to this shift in the spatial mapping. The two strategies consisted of a clockwise tuning of the peroneus longus muscle along with either a clockwise or counterclockwise shift of the tibialis anterior muscles. This adaptation occurred after gaining knowledge of the computer sprite's position relative to the target under the new spatial relationship. While a decrease in perpendicular displacement was exhibited, the ability to preplan corrections (i.e. decrease displacement angle) to account for the rotational offset resulted in better learning. This is supported by the fact that no aftereffects were observed for the perpendicular displacement. The decrease in the displacement angle was optimally achieved through the counterclockwise tuning of the tibialis anterior muscles. The counterclockwise tuning of the tibialis anterior was observed after the participants originally tuned the muscle in a clockwise manner. This result is important as it shows changing the learning strategy can result in better recalibration of internal to external spatial reference frames required to learn task specific visual based rotations viewed in a virtual environment.

5.1.3 STABILITY ASSESSMENT AND CENTER OF MASS ESTIMATION DURING WALKING

The center of body mass, center of foot pressure, and body segment acceleration signals are commonly used to indicate movement performance and stability during standing activities and walking. The COM is generally calculated using a video-based motion analysis system, via motion trajectories of markers placed on each body segment and anthropometric models. The resultant three-dimensional COM trajectory provides us with a picture of the overall movement and the coupling that exists between the movements in different planes. During standing activities and walking, important stability outcomes for the COM trajectory are:

- The amplitude mean [Marigold and Patla, 2005],
- The peak-peak amplitude [Marigold and Patla, 2005],
- The phase lag, relative to step initiation, representing the time to peak amplitude across each step [MacLellan and Patla, 2006],
- The signal variability, due to differences in biomechanical configurations and balance strategies/gait patterns, from step to step and across subjects [Moe-Nilssen and Helbostad, 2006].

For balance maintenance and restoration, the human brain is capable of estimating and predicting the COM even in the absence of visual or vestibular information. Thus, we hypothesized that the COM may be acquired through the processing of proprioceptive somatosensory information, represented by body segment accelerations, and an external spatial reference, the ground support, represented by the COP. To investigate this hypothesis, we modeled the relationships that exist between the COP and accelerometer data with the 3D COM trajectory, while healthy participants walked on firm and irregular surfaces.

The model was validated by calculating:

1. The correlation coefficient and coefficient of determination between the actual ($\mathbf{COM}_R$) and estimated ($\hat{\mathbf{COM}}_R$) resultant COM trajectories, representing the amount of variability the model accounts for. The models accounted for $99.85 \pm 0.20\%$ and $99.77 \pm 0.39\%$ of the resultant COM trajectory's variability for the firm and irregular surfaces, respectively.
2. The percentage error and actual error between $\mathbf{COM}_R$ and $\hat{\mathbf{COM}}_R$. The percentage error for the model was 16.06 ± 11.11 % and 21.41 ± 12.70 % for the firm and doweling surfaces, respectively. In turn, this translated into an absolute error between the true and actual resultant COM of 3.62 ± 2.69 cm and 4.74 ± 3.01 cm for the firm and doweling surfaces respectively.

In terms of clinical outcomes, the aforementioned stability outcomes derived from the actual and estimated COM were compared. The estimated COM resulted in stability measures similar to those calculated from the actual COM.

The derived model is novel in that it does not require any calibration and provides a reasonably accurate estimation of the COM, which can be compared to the brain's balance performance in the absence of visual information. However, the model has to be tested and validated with data from different aged populations. Once the accuracy of the model estimation is confirmed on different populations, the model could be used instead of the cumbersome method of video motion analysis for COM calculation.

5.2 FUTURE RESEARCH

5.2.1 INTERACTIVE REHABILITATION TOOL

Future research includes further tool validation through a randomized proof of principal study, embedding an assessment module into the tasks, and development of the interactive tool for treadmill walking, as described in the following paragraphs.

A randomized proof of principal study could be conducted in order to assess and validate the feasibility of using the developed virtual environment-based tool to:

1. Motivate subjects and achieve a greater volume of practice.
2. Improve both static and dynamic balance control.

A total of 24 subjects would be recruited from community dwelling and ambulatory older adults who attend the Riverview Health Centre Day Hospital for treatment of balance impairment and mobility limitations. Inclusion criteria would include:

1. Age: 65-85 years.
2. Mini-Mental State Examination score greater than 24, which indicates no cognitive impairment [Folstein et al., 1975].
3. English-speaking with the ability to understand the nature of the study and provide informed consent.
4. Independent in ambulatory functions, with or without an assistive device (cane or walker).

Exclusion criteria would include any medical condition or disability that prevents participation in an exercise program.

Participants would be randomly assigned to either the *control* or the *experimental* group. There would be 16 sessions scheduled, two per week. Each session would last 45 minutes and would include rest periods as needed. The training sessions for the two groups would be scheduled at different times of the day. The control group would receive the typical rehabilitation program presently provided at the Day Hospital. The Day Hospital physiotherapy program would consist of individual strengthening and balance exercises, which would be done in standing, sitting, and lying positions. The control group would also receive an assessment of walking aides, a gait re-education program, and an unsupervised walking program. The experimental group would receive a program using dynamic balance exercises coupled to our developed virtual environment-based tasks.

The following outcome measures could be recorded pre-treatment and post-treatment:

1. Number of falls and fall injuries (via a fall log).
2. Six Minute Walk test scores [Hamilton and Haennel, 2000].
3. Berg Balance Scale scores [Berg et al., 1995].
4. Step Test scores [Hill et al., 1996].
5. The stability measurements discussed in Section 2.3.3.2.
6. Spatial-temporal gait parameters, as recorded via the GaitRite system.
7. Activities-Specific Balance Confidence Scale scores [Powell and Meyers, 1995].
8. Late Life Function and Disability Instrument scores [Jette et al., 2002].
9. Health-related quality of life scores, obtained via the SF36 self-report questionnaire of general health perceptions and mental health status [Brazier et al., 1992].

The outcomes for individual subjects and between subjects could then be compared and evaluated.

As the COP signal provides functional information to the clinician, an embedded assessment module could be incorporated into the tasks in order to increase the benefits of the interactive tool and make treatment assessment transparent. As previously mentioned, during the task Balloon Burst, the on-screen pressure and balloon position information are automatically logged. The assessment module could then use the logged information to calculate the following movement and stability parameters:

1. The velocity profile for the movement trajectory could be calculated. Of particular interest would be the peak and shape of the velocity profile. In general, movements of slow to moderate speeds will produce a bell shaped velocity; when the movements become quicker and less controlled, the shape of the curve will become skewed. Thus, the skewness of the velocity profile could be compared both within and between sessions. The kurtosis of the profile could be compared in the same manner, in order to determine the peaked shape of the curve.
2. The elapsed time between the appearance of the balloon and the popping of the balloon would give an indicator of how quickly the participant could perform the task.
3. The trajectory of the control marker from the time the balloon appears until when the balloon is popped could be examined. This would provide an account of spatial accuracy, indicating endpoint undershoot error and its absolute magnitude and endpoint overshoot error and its absolute

magnitude. In addition, the trajectory could be compared with the reference path, determined by the straight line joining the COP marker location at the start of the movement and the balloon location. The error between the reference line and the actual trajectory could then be calculated; this error would be correlated with successive movements, both intra- and inter-session.

Embedded assessment along with the ability to customize our system to each subject's needs and the system's portability, affords its use in monitored at-home-programs; this makes our therapy approach cost effective. A monitored program with follow-up visits ensures that programs are being performed properly and safely. Home based programs have been shown to be effective for rehabilitation. The continuing of training once at home should allow patients to further maintain and/or improve functional levels [Hale and Piggot, 2005; Giusti et al., 2006].

Our current system offers benefits for short-sitting and standing balance. However, [Messier et al., 2000] showed that aerobic walking also decreased postural sway. In order to apply the benefits our interactive tool to walking, larger mats could be used with the system in order to facilitate stepping. Treadmill training, used with or without body weight support, is an effective method of retraining subjects to walk, which produces substantial gains in balance [Barbeau and Visintin, 2003; Harris-Love et al., 2004; Wilson et al., 2006]. A study by Barbeau and Visintin [2003] demonstrated the importance of treadmill training with body weight support, in order to regain walking ability in the early stages of recovery. This is particularly important for subjects who are dependent on support for normal ambulation. Hence, the task could be developed for use with treadmill walking. The COP trajectory for tasks incorporating stepping and walking behaves differently than for standing, as the foot is not in contact with the surface during the swing phase. Thus, the on-screen cursors could be controlled via the load value of each pressure sensor (or groups of pressure sensors) in the mat as opposed to the COP. As an example, consider Balloon Burst for use with a treadmill. First, a sensor grid would be defined as the total number of squares and the number of sensors per square. During game play, a balloon would appear in a random or predetermined location, defined by the row and column coordinates of the square. In order to pop the balloon, the user must step on the sensors that correspond to the on-screen square location.

5.2.2 STRATEGIES FOR LEARNING NEW MOTOR TASKS

In the present research, an angle of 60° was used for the rotated visual-spatial transformation. This rotation degree presented a relatively difficult angle to adjust to. The participants who exhibited the clockwise rotation were likely unaware of the transformation mechanism, but were prepared to do quick corrections after the movement began. Conversely, the participants who exhibited the counterclockwise rotation probably recognized that the movement field was rotated and used this information to preplan their movements. Further research could be done where the number of trials required to fully adapt to a range of rotation angles would be determined. Thus the degrees of difficulty of each of the transformations could be determined. For contrast and comparison, different types of transformations could also be incorporated into the task, e.g., a scaling or translation of the mapping.

As situations that require recalibrations are encountered in every day activities, it would be beneficial to examine the effects of training with a rotation, while performing the Balloon Burst task on different compliant and irregular surfaces. Continuing both the original and additional studies with patients as the participants would further elucidate how learning occurs when specific deficits are present in the balance system.

With the current setup, the participant attempts to maintain single-link inverted pendulum movements, with both feet planted on the surface. Hence, it may be assumed that the muscles in both legs are tuning in a similar manner (note that this could be confirmed with an additional study). Many reaching studies have examined learning in one arm and have studied the transfer of the recalibration to the opposite arm [Shadmehr and Moussavi, 2000]. Studying this effect in the lower limbs would prove to be a more difficult task, as both limbs are supporting the body motion. However, an insole could be used in each shoe, where a single insole would be used to control the onscreen sprite's movements and the participants would be instructed to weight-bear with the control leg. The loading in the opposite leg would be determinable from the non-control insole. Muscle tuning would be examined for both legs, to ensure tuning could be observed for a single leg only. Next, the rotated task would be performed with the opposite leg without any training. We could then examine if transference of the recalibration occurred.

5.2.3 STABILITY ASSESSMENT AND CENTER OF MASS ESTIMATION DURING WALKING

After stepping on a compliant or irregular surface, the subject may temporarily loose their balance. Attempts to regain postural control after a perturbation will be manifested in the left footstep after the doweling surface is encountered [MacKinnon and Winter, 1993; Bauby and Kuo, 2000]. For example, Oddsson et al. (2004) determined that medial or lateral disturbances applied to the stance foot will result in either a narrowing or widening of the next step, respectively. This will be reflected in the ML COM component, which is adjusted in order to counteract ML instabilities [McIlroy and Maki 1999]. Hence, prediction of the COM trajectory over the entire walkway as opposed to signal steps is important.

Having an insole in each shoe would provide us with the COP for both feet and hence prediction could be done for every step of free-form walkways. However, studies have shown that gait speed can have a substantial effect on kinematic and kinetic variables, as different speeds alter the step lengths and consequently the acceleration patterns [Lelas et al., 2002]. Hence, control over the subjects gait would be beneficial and result in more consistent COM analysis and improved model accuracy. Walking on an instrumented treadmill would allow for speed control and data collection of a large number of steps. In addition to the COP, the instrumented treadmill would allow for calculation of spatial and temporal gait parameters. The use of these additional gait parameters could be used to improve the COM estimation model. Note that drawbacks to using an instrumented treadmill are a loss of portability and increased difficulty in using compliant or irregular surfaces.

After further improvement of the model using data from a young, healthy population, a study could be carried out using a large range of patient populations to determine the extent to which the proposed model is valid. A validated patient population generated model would have direct application in a clinical environment, as the stability parameters (Section 4.4.3) could be used to determine if the subject exhibited any gait abnormalities.

References

S. Abeele and O. Bock, "Mechanisms for sensorimotor adaptation to rotated visual input," *Exp Brain Res*, vol. 139, pp. 248–253, 2001.

P. Bach-y-Rita, S. Wood, R. Leder, O. Paredes, D. Bahr, E.W. Bach-y-Rita and N. Murillo, "Computer-assisted motivating rehabilitation (CAMR) for institutional, home, and educational late stroke programs," *Top Stroke Rehabil*, no. 8, vol. 4, pp. 1-10, 2002.

H. Barbeau and M. Visintin, "Optimal outcomes obtained with bodyweight support combined with treadmill training in stroke patients," *Arch Phys Med Rehabil*, no. 84, pp. 1458-65, 2003.

F. Barbier, P. Allard, K. Guelton, B. Colobert and A.P. Godillon-Maquinghen, "Estimation of the 3-D center of mass excursion from force-plate data during standing," *Trans Neural Sys Rehab Eng*, no. 11, vol. 1, pp.31-37, 2003.

L. Barratto, P.G. Morasso, C. Re and G. Spada, "A new look at posturographic analysis in the clinical context: sway-density vs. other parameterization techniques," *Motor Control*, vol. 6, no. 3, pp. 248-73, 2002.

E. Bauby and A.D. Kuo, "Active control of lateral balance in human walking," *J Biomech*, vol. 33, pp. 1433-1440, 2000.

K. Berg, S. Wood-Dauphinee and J.I. Williams, "The Balance Scale: reliability assessment with elderly residents and patients with an acute stroke," *Scand J Rehabil Med*, vol. 27, pp. 27-36, 1995.

A.L. Betker, Z. Moussavi and T. Szturm, "On modeling center of foot pressure distortion through a medium," *IEEE Trans Biomed Eng*, vol. 52, no. 3, pp. 345-352, 2005.

A. L. Betker, T. Szturm, Z. Moussavi, and C. Nett, "Video Game-Based Exercises for Balance Rehabilitation: A Single-subject Design," *Arch Phys Med Rehabil*, vol. 87, pp. 1141-9, 2006a.

A.L. Betker, Z. Moussavi, and T. Szturm, "Center of Mass Approximation as a Function of Acceleration," *IEEE Trans. Biomed. Eng.*, vol. 53, no. 4, pp. 686-693, 2006b.

A.L. Betker, A. Desai, C. Nett, N. Kapadia and T. Szturm, "Game-Based Exercises for Dynamic Short-sitting Balance Rehabilitation of Subjects with Chronic Spinal Cord and Traumatic Brain Injuries," *Phys Ther*, vol. 87, pp. 1389-98, 2007.

C. Boult, R.L. Kane, T.A. Louis, L. Boult and D. McCaffrey, "Chronic conditions that lead to functional limitation in the elderly," *J Gerontol*, vol. 49, pp. M28-36, 1994.

D. Bourbonnais, S. Bilodeau, Y. Lepage, N. Beaudoin, D. Gravel and I.L. Forget, "Effect of force-feedback treatments in patients with chronic motor deficits after a stroke," *Am J Phys Med Rehab*, vol. 81, no. 12, pp. 890-7, 2002.

J.E. Brazier, R. Harper, N.M. Jones, A. O'Cathain, K.J. Thomas, T. Usherwood, and L. Westlake, "Validating the SF-36 health survey questionnaire: new outcome measure for primary care," BMJ, vol. 305, pp. 160-164, 1992.

J. Broeren, M. Rydmark, and K. S. Sunnerhagen, "Virtual reality and haptics as a training device for movement rehabilitation after stroke: a single-case study," *Arch Phys Med Rehabil*, vol. 85, pp. 1247-50, 2004.

N. Burgess, "Spatial memory: how egocentric and allocentric combine," *Trends Cogn Sci*, vol. 10, pp. 551-557, 2006.

J.H. Carr and R.B. Shepherd, *Neurological rehabilitation: optimizing motor performance*. Oxford: Butterworth-Heinemann Ltd, 1998.

J.H. Carr and R.B. Shepherd, "The changing face of neurological rehabilitation," *Rev Bras Fisioter*, vol. 10, no. 2, pp. 147-156, 2006.

P.T. Cheng, S.H. Wu, M.Y. Liaw, A.M.K. Wong and F.T. Tang, "Symmetrical body-weight distribution training in stroke patients and its effect on fall prevention," *Arch Phys Med Rehab*, vol. 82, pp. 1650-4, 2001.

L.-S. Chou, K. R. Kaufman, M. E. Hahn, and R. H. Brey, "Medio-lateral motion of the center of mass during obstacle crossing distinguishes elderly individuals with imbalance," *Gait Posture*, vol. 18, pp. 125-133, 2003.

A. Cogan, J. Madey, W. Kaufman, G. Holmlund and P. Bach-y-Rita, "Pong game as a rehabilitation device," In *Proc. of the 4th Ann. Conf. on Sys. and Devices for the Disabled,* pp. 187-188, 1977.

J.C. Culham, C. Cavina-Pratesi, and A. Singhal, "The role of parietal cortex in visuomotor control: What have we learned from neuroimaging?" *Neuropsychologia*, vol. 44, pp. 2668-2684, 2006.

D. Cunningham and M. Krishack, "Virtual reality: a holistic approach to rehabilitation," *Stud Health Tech Info*, vol. 62, pp. 90-3, 1999.

C.M. Dean, C. L. Richards and F. Malouin, "Task-related circuit training improves performance of locomotor tasks in chronic stroke: a randomized, controlled pilot trial," *Arch Phys Med Rehab*, vol. 81, pp. 409-417, 2000.

W.J.G. DeWeerdt and M.A. Harrison, "The use of biofeedback in physiotherapy," *Physiotherapy*, vol. 71, pp. 9-12, 1985.

V. Dietz and S. Harkema, "Locomotor activity in spinal cordinjured persons," *Can J Appl Physiol*, vol. 96, pp. 1954–1960, 2004.

M. Dozza, L. Chiari and F. Horak, "Audio-biofeedback improves balance in patients with bilateral vestibular loss," *Arch Phys Med Rehabil*, vol. 86, pp. 1401-3, 2005.

V. Edgerton, N. Tillakaratne, A.J. Bigbee, R. De Leon, and R.R. Roy, "Plasticity of the spinal neural circuitry after injury," *Annu Rev Neurosci*, vol. 17, pp. 145–167, 2004.

J. M. Ellermann, J. D. Siegal, J. P. Strupp, T. J. Ebner, and K. Ugurbil, "Activation of visuomotor systems during visually guided movements: a functional MRI study," *J Magn Reson*, vol. 131, pp. 272-85, 1998.

V. Femery, P. Moretto, J.M. Hespel and G. Lensel, "Plantar pressure biofeedback device for foot unloading," *Proc. 5th Symp Footwear Biomech*, 2001.

J. Fernandez-Ruiz, R. Diaz, C. Hall-Haro, P. Vergara, J. Mischner, L. Nunez, R. Drucker-Colin, A. Ochoa, and M.E. Alonso, "Normal prism adaptation but reduced after-effect in basal ganglia disorders using a throwing task," *Eur J Neurosci*, vol. 18, no. 3, pp. 689-94, Aug. 2003.

L. Fogassi and G. Luppino, "Motor functions of the parietal lobe," *Curr Opin Neurobiol.*, vol. 15, pp. 626-631, 2005.

M.F. Folstein, S.E. Folstein and P.R. McHugh, "Mini-Mental State: A practical method for grading the cognitive state of patients for the clinician," *J Psychiatr Res*, vol. 12, no. 2, pp. 189–198, 1975.

F. Frassinetti, V. Angeli, F. Meneghello, S. Avanzi, and E. Làdavas, "Long-lasting amelioration of visuospatial neglect by prism adaptation," *Brain*, vol. 125, pp. 608-623, 2002.

L.P. Fried, S.J. Herdman, K.E. Huhn, G. Rubin and K. Turano, "Preclinical disability," *J Aging and Health*, vol. 3, no. 2, pp. 285-300, 1991.

R.A. Geiger, J.B. Allen, J. O'Keefe and R.I.L. Hicks, "Balance and mobility following stroke: effects of physical therapy interventions with and without biofeedback/force plate training," *Phys Ther*, vol. 81, pp. 995-1005, 2001.

J. Gill, J.H.J. Allum, M.G. Carpenter, M.H. Ziolkowska, A.L. Adkin, F. Honegger, and K. Pierchala, "Trunk sway measures of postural stability during clinical balance tests: effects of age," *J Gerontol: Biol Sci Med Sci*, vol. 56A, no. 7, pp. M438-47, 2001.

A. Giusti, A. Barone, M. Oliveri, M. Pizzonia, M. Razzano, E. Palummeri, and G. Pioli, "An Analysis of the Feasibility of Home Rehabilitation Among Elderly People With Proximal Femoral Fractures," *Arch Phys Med Rehabil, vol. 87, no. 6, pp. 826-831, 2006.*

M. Glanz, S. Klawansky and T. Chalmers, "Biofeedback therapy in stroke rehabilitation: a review," *J Royal Soc Med*, vol. 90, pp. 33-9, 1997.

E. M. Gutierrez-Farewik, Å. Bartonek, and H. Saraste, "Comparison and evaluation of two common methods to measure center of mass displacement in three dimensions during gait," *Hum Mov Sci*, vol. 25, no. 2, pp. 238-256, 2006.

M. E. Hahn, and L-S. Chou, "Can motion of individual body segments identify dynamic instability in the elderly?" *Clin Biomech*, vol. 18, no. 8, pp. 737-744, 2003.

L. A. Hale and J. Piggot, "Exploring the Content of Physiotherapeutic Home-Based Stroke Rehabilitation in New Zealand," *Arch Phys Med Rehabil, vol. 86, no. 10, pp. 1933-1940, 2005.*

D.M. Hamilton and R.G. Haennel, "Validity and reliability of the 6-minute walk test in a cardiac rehabilitation population," *J Cardiopulm Rehabil*, vol. 20, pp. 156-164, 2000.

J.E. Harris, J.J. Eng, D.S. Marigold, C.D. Tokuno, and C.L. Louis, "Relationship of balance and mobility to fall incidence in people with chronic stroke," *Phys Ther*, vol. 85, pp. 150-8, 2005.

M.L. Harris-Love, R.F. Macko, J. Whitall, and L. W. Forrester, "Improved hemiparetic muscle activation in treadmill versus overground walking," *Neurorehabil Neural Repair*, vol. 18, no. 3, pp. 154-60, 2004.

M. Henriksen, H. Lund, R. Moe-Nilssen, H. Bliddal, and B. Danneskiod-Samsoe, "Test-retest reliability of trunk accelerometric gait analysis," *Gait Posture*, vol. 19, no.3 , pp. 288-297, 2004.

S. M. Henry, J. Fung, and F. B. Horak, "Effect of Stance Width on Multidirectional Postural Responses," *J Neurophysiol*, vol. 85, pp. 559-570, 2001.

K.D. Hill, J. Bernhardt, A.M. McGann, D. Maltese and D. Berkovits, "A new test of dynamic standing balance for stroke patients: Reliability, validity and comparison with healthy elderly," *Physiother Canada*, Fall, pp. 257- 262, 1996.

H. Huang, S.L. Wolf and J. He, "Recent developments in biofeedback for neuromotor rehabilitation," *J NeuroEng Rehabil*, vol. 3, no. 11, 2006.

S. Jansson and A. Soderlund, "A new treatment programme to improve balance in elderly people - an evaluation of an individually tailored home-based exercise programme in five elderly women with a feeling of unsteadiness," *Disabil Rehabil*, vol. 26, no. 24, pp. 1431–1443, 2004.

A.M. Jette, S.M. Haley, W.J. Coster, J.T. Kooyoomjian, S. Levenson, T. Heeren, and J. Ashba, "Late life function and disability instrument: I. Development and evaluation of the disability component," *J Gerontol A Biol Sci Med Sci*, vol. 57, pp. M209-216, 2002.

Y. Jian, D. A. Winter, M. G. Ishac, and L. Gilchrist, "Trajectory of the body COG and COP during initiation and termination of gait," *Gait Posture*, vol. 1, pp. 9-22, 1993

H. Johansen-Berg, H. Dawes, C. Guy, S. M. Smith, D. T. Wade, and P. M. Matthews, "Correlation between motor improvements and altered fMRI activity after rehabilitative therapy," *Brain*, vol. 125, pp. 2731-42, 2002.

K. E. Jones, J. Wessberg, and A. Vallbo, "Proprioceptive feedback is reduced during adaptation to a visuomotor transformation: preliminary findings," *Neuroreport*, vol. 12, pp. 4029-33, 2001.

M. P. Kabada, H. K. Ramakrishnan and M. E. Wooten, "Measurement of lower extremity kinematics during level walking," *J. Orthop. Rel. Res.*, vol. 8, pp. 383-392, 1990.

B. K. Kaya, D. E. Krebs and P. O. Riley, "Dynamic stability in elders: momentum control in locomotor ADL," *J. Gerontology Series A: Biol. Sci. and Med. Sci.*, vol. 53, no. 2, pp. M126-M134, 1998.

J. W. Krakauer, M. R. Ghilardi, and C. Ghez, "Independent learning of internal models for kinematic and dynamic control of reaching," *Nat Neurosci*, vol. 2, pp. 1026-31, 1999.

J. W. Krakauer, Z. M. Pine, M-F. Ghilardi, and C. Ghez, "Learning of Visuomotor Transformations for Vectorial Planning of Reaching Trajectories," *J Neurosci*, vol. 20, pp. 8916–8924, 2000.

D.M.W. Kriegsman, D.J.H. Deeg, J.T.H.M. van Eijk, B.W.J.H. Penninx and A.J.P. Boeke, "The additional influence of disease-specific characteristics on mobility in patients with different chronic diseases: a study in The Netherlands," *J Epidemiol Comm Health*, vol. 51, pp, 676-85, 1997.

A.D. Kuo, "An optimal control model for analyzing human postural balance," *IEEE Trans Biomed Eng*, vol. 42, no. 1, pp. 87-101, 1995.

G. Kwakkel, "Impact of intensity of practice after stroke: Issues for consideration," *Disabil Rehabil*, vol. 28, no. 13–14, pp. 823–830, 2006.

D. Lafond, M. Duarte, and F. Prince, "Comparison of three methods to estimate the center of mass during balance assessment," *J Biomech,* vol. 37, pp. 1421-1426, 2004.

N. Lazzaro and K. Keeker, "What's my method?: a game show on games," *Ext abs Human Factors Comp Sys*, pp. 1093-4, 2004.

M.Y. Lee, M.K. Wong and F.T. Tang, "Using biofeedback for standing steadiness, weight-bearing training," *IEEE Eng Med Biol*, vol. 15, no. 6, 112-6, 1996.

J.L. Lelas, G.J. Merriman, P.O. Riley and D.C. Kerrigan, "Predicting peak kinematic and kinetic parameters from gait speed," *Gait Posture*, vol. 17, pp. 106-12, 2003.

F. E. Leon-Sarmiento, W. Bara-Jimenez, and E. M. Wassermann, "Visual deprivation effects on human motor cortex excitability," *Neurosci Lett*, vol. 389, pp. 17-20, 2005.

R. A. Liston and B. J. Brouwer, "Reliability and validity of measures obtained from stroke patients using the Balance Master," *Arch Phys Med Rehabil*, vol. 77, pp. 425-30, 1996.

C.D. MacKinnon and D.A. Winter, "Control of whole body balance in the frontal plane during human walking," *Biomech*, vol. 26, no. 6, pp. 633-644, 1993.

M. J. MacLellan, and A. E. Patla, "Adaptations of walking pattern on a compliant surface to regulate dynamic stability," *Exp. Brain. Res.*, vol. 73, pp. 521–530, 2006.

B. E. Maki, and W. E. McIlroy, "The control of foot placement during compensatory stepping reactions: Does speed of response take precedence over stability?" *IEEE Trans Rehab Eng*, vol. 7, no. 1, pp. 80-90, 1999.

T.W. Malone, "Heuristics for designing enjoyable user interfaces: Lessons from computer games," *Proc Human Factors Comp Sys*, pp. 63-8, 1982.

D. S. Marigold and A. E. Patla, "Adapting Locomotion to Different Surface Compliances: Neuromuscular Responses and Changes in Movement Dynamics," *J Neurophysiol*, vol. 94, pp. 1733–1750, 2005.

I. Marklund and M. Klassbo, "Effects of lower limb intensive mass practice in poststroke patients: single-subject experimental design with long-term follow-up," *Clin Rehabil*, vol. 20, pp. 568-576, 2006.

J.J. Marotta, G.P. Keith, and J.D. Crawford, "Task-specific sensorimotor adaptation to reversing prisms," *J Neurophysiol*, vol. 93, pp. 1104-10, 2005.

W.E. McIlroy, B.E. Maki, "The control of lateral stability during rapid stepping reactions evoked by antero-posterior perturbation" does anticipatory control play a role?" *Gait Posture*, vol. 9, pp. 190-8, 1999.

H. B. Menz, S. R. Lord, and R. C. Fitzpatrick, "Acceleration patterns of the head and pelvis when walking on level and irregular surfaces," *Gait Posture*, vol. 18, no. 1, pp. 35-46, 2003.

S.P. Messier, T.D. Royer, T.E. Craven, M.L. O'Toole, R. Burns and. W.H. Ettinger Jr., "Long-term exercise and its effect on balance in older, osteoarthritic adults: results from the Fitness, Arthritis, and Seniors Trial (FAST)," *J Am Geriatr Soc*, vol. 48, no. 2, pp. 131-8, 2000.

R. Moe-Nilssen, and J. L. Helbostad, "Estimation of gait cycle characteristics by trunk accelerometry," *J Biomech*, vol. 37, pp. 121–126, 2004.

E. G. Moore, M. W. Rosenberg and S. H. Fitzgibbon, "Activity limitation and chronic conditions in Canada's elderly, 1986-2011," *Dis and Rehab*, vol. 21, pp. 196-210, 1999.

W. Mou, T. P. McNamara, B. Rump, and C. Xiao, "Roles of Egocentric and Allocentric Spatial Representations in Locomotion and Reorientation," *J Exp Psychol Learn Mem Cogn*, vol. 32, pp. 1274-90, 2006.

F. A. Mussa-Ivaldi, "Modular features of motor control and learning," *Curr Opin Neurobiol*, vol. 9, pp. 713-717, 1999.

T.J. O'Connor, R.A. Cooper, S.G. Fitzgerald, M. J. Dvorznak, M. L. Boninger, D. P. VanSickle, and L. Glass, "Evaluation of a manual wheelchair interface to computer games," *Neurorehabil Neural Repair*, vol. 14, no. 1, pp. 21-31, 2000.

L. I. E. Oddsson, C. Wall, M. D. McPartland, D. E. Krebs, and C. A. Tucker, "Recovery from perturbations during paced walking," *Gait Posture*, vol. 19, pp. 24-34, 2004.

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

A. E. Patla, "Strategies for dynamic stability during adaptive human locomotion," *IEEE Eng Med Biol Mag*, vol. 22, no. 2, pp. 48-52, Mar-Apr 2003.

J. Patton and F. A. Mussa-Ivaldi, "Robot-assisted adaptive training: Custom force fields for teaching movement reaching patterns," *IEEE Trans Biomed Eng*, vol. 51, pp. 636–646, 2004.

M. A. Perez, B. K. Lungholt, K. Nyborg, and J. B. Nielsen, "Motor skill training induces changes in the excitability of the leg cortical area in healthy humans," *Exp Brain Res*, vol. 159, pp. 197-205, 2004.

R.J. Peterka, "Sensorimotor integration in human postural control", *J. Neurophysiol.*, vol. 88, pp. 1097-1118, 2002.

L.E. Powell and A.M. Myers, "The Activities-specific Balance Confidence (ABC) Scale," *J Gerontol A Biol Sci Med Sci*, vol. 50A, pp. M28-34, 1995.

M.S. Remple, R.M. Bruneau, P.M. VandenBerg, C. Goertzen and J.A. Kleim, "Sensitivity of cortical movement representations to motor experience: evidence that skill learning but not strength training induces cortical reorganization," *Behav Br Res*, vol. 123, pp.133-41, 2001.

C.L. Richards, F. Malouin, G. Bravo, F. Dumas and S. Wood-Dauphinee, "The role of technology in task-oriented training in persons with subacute stroke: a randomized controlled trial," *Neurorehabil Neural Repair*, vol. 18, no. 4, pp. 199-211, 2004.

R.L. Roig, G.M. Worsowicz, D.G. Stewart and D.X. Cifu, "Geriatric rehabilitation. 3. physical medicine and rehabilitation interventions for common disabling disorders," *Arch Phys Med Rehabil*; no. 85, suppl. 3, pp. S12-7, 2004.

Y. Rossetti, K. Koga, and T. Mano, "Prismatic displacement of vision induces transient changes in the timing of eye-hand coordination," *Percept Psychophys*, vol. 54, no. 3, pp. 355-64, Sep. 1993.

Y. Rossetti, "Implicit short-lived motor representations of space in brain damaged and healthy subjects," *Consciousness and Cognition*, vol. 7, pp. 520-558, 1998.

S. Rossignol, "Locomotion and its recovery after spinal injury," *Curr Opin Neurobiol*, vol. 10, no. 6, pp. 708-16, 2000.

P. Rougier, "Optimising the visual feedback technique for improving upright stance maintenance by delaying its display: behavioural effects on healthy adults," *Gait Posture*, vol. 19, pp. 154–163, 2004.

J.D. Schaechter, "Motor rehabilitation and brain plasticity after hemiparetic stroke," *Prog Neurobiol*, vol. 73, pp. 61-72, 2004.

V. Segers, P. Aerts, M. Lenoir, and D. De Clerq, "Dynamics of the body centre of mass during actual acceleration across transition speed," *J. Exp. Biol.*, vol. 210, pp. 578-585, Feb. 2007.

R. Shadmehr and F. A. Mussa-Ivaldi, "Adaptive representation of dynamics during learning of a motor task," *J Neurosci*, vol. 14, pp. 3208-3224, 1994.

R. Shadmehr and Z. M. K. Moussavi, "Spatial Generalization from Learning Dynamics of Reaching Movements," *J Neuroscience*, vol. 20, pp. 7807–7815, 2000.

N.T. Shepard and S.A. Telian, "Excerpt - balance system function," *Amer. J. of Audiology*, vol. 1, no. 4, pp. 45-50, 1992.

A. Shumway–Cook and F.B. Horak, "Assessing the influence of sensory interaction on balance," *Phys Ther*, vol. 66, pp. 548–1550, 1986.

A. Shumway-Cook, D. Anson and S. Hailer, "Postural sway biofeedback: its effects on reestablishing stance stability in hemipleglc patients," *Arch Phys Med Rehab*, vol. 69, pp. 395-400, 1988.

K. Singh and S. H. Scott, "A motor learning strategy reflects neural circuitry for limb control," *Nat Neurosci*, vol. 6, pp. 399-403, 2003.

S.E. Sihvonen, S. Sipila and P.A. Era, "Changes in postural balance in frail elderly women during a 4-week visual feedback training: a randomized controlled trial," *Gerontology*, vol. 50, no. 2, pp. 87-95, 2004.

G. L. Smidt, J. S. Arora, and R. C. Johnston, "Accelerographic analysis of several types of walking," *Am J Phys Med*, vol. 50, no. 6, pp. 285-300, 1971.

H. Sveistrup, "Motor rehabilitation using virtual reality," *J NeuroEng Rehabil*, vol. 1, no. 10, 2004.

T. Szturm and B. Fallang, "Effects of varying acceleration of platform translation and toes-up rotations on the pattern and magnitude of balance reactions in humans," *J Vestib Res*, vol. 8, pp. 381-397, 1998.

E. Taub, G. Uswatte and R. Pidikiti, "Constraint-induced movement therapy: a new family of techniques with broad application to physical rehabilitation - a clinical review," *J Rehabil Res Dev*, vol. 36, no. 3, pp. 237-251, 1999.

B. P. Tripp, W. E. McIlroy, and B. E. Maki, "Online mutability of step direction during rapid stepping reactions evoked by postural perturbation", *IEEE Trans Neural Sys Rehabil Eng*, vol. 12, no. 1, pp. 140-152, 2004.

W.W. Tsang and C.W. Hui-Chan, "Effects of exercise on joint sense and balance in elderly men: Tai Chi versus golf," *Med Sci Sports Exerc*, vol. 36, no. 4, pp. 658-67, 2004.

O. Urhan and H. Dincer, "Gait trainer for children with spastic cerebral palsy," *Proceedings 23rd IEEE Eng Med Biol Soc Conf (EMBC'01)*, vol. 2, 2001, pp. 25-28.

A. N. Vasvada, B. W. Peterson, and S. L. Delp, "Three-dimensional spatial tuning of neck muscle activation in humans," *Exp Brain Res*, vol. 147, pp. 437-448, 2002.

A. Viau, A. G. Feldman, B. J. McFadyen, and M. F. Levin, "Reaching in reality and virtual reality: a comparison of movement kinematics in healthy subjects and in adults with hemiparesis," *J Neuroengineering Rehabil*, vol. 1, pp. 1-11, 2004.

J.S. Webster, P.T. McFarland, L.J. Rapport, B. Morrill, L. A. Roades, and P. S. Abadee, "Computer-assisted training for improving wheelchair mobility in unilateral neglect patients," *Arch Phys Med Rehabil*, vol. 82, no. 6, pp. 769-775, 2001.

R. Whipple, L. Wolfson, C. Derby, D. Singh and J. Tobin, "Altered sensory function and balance in older persons," *J Gerontol*, vol. 48, pp. 71-6, 1999.

D.J. Wilson, M. Powell, J.L. Gorham JL, and M. K. Childers, "Ambulation training with and without partial weightbearing after traumatic brain injury: Results of a randomized, controlled trial," *Am J Phys Med Rehabil*, vol. 85, pp. 68–74, 2006.

D. A. Winter, "Camera speeds for normal and pathological gait analyses," *Med Biol Eng Comput*, vol. 20, pp. 408-412, 1982.

D. Zambarbieri, M. Schmid, M. Magnaghi, G. Verni, V. Macellari and A. Fadda, "Biofeedback techniques for rehabilitation of the lower limb prosthetic subjects," *Proceedings VIII Mediterranean Conf Med Biol Eng Comp (MEDICON 98),* 1998.