A multiple segment model of human balance for objective assessment of clinical scales

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NANYANG
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**A MULTIPLE SEGMENT MODEL OF HUMAN
BALANCE FOR OBJECTIVE ASSESSMENT OF
CLINICAL SCALES.**

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Contents

Contents	ii
Abstract	v
List of Figures	vi
List of Tables	xi
1 Introduction	1
1.1 Background	1
1.1.1 Importance of Studying Balance	1
1.1.2 Human Balance	3
1.1.3 Balance Rehabilitation	6
1.1.4 Limitations of Clinical Scales	8
1.2 Objectives	9
1.3 Scope and Contributions	10
1.4 Chapter Outline	13
2 Literature Review	15
2.1 Clinical Aspects of Balance	15
2.1.1 Components of the human balance system	15
2.1.2 Normal Postural Control	19
2.1.3 Balance Disorders	21
2.2 Neuromotor Rehabilitation	22
2.2.1 Brain plasticity	23
2.2.2 Stroke	23
2.2.3 Stroke Recovery	24
2.3 Balance Rehabilitation	25
2.3.1 Clinical Assessment of Balance	25
2.3.2 Balance Training	28
2.4 Technology in Balance Rehabilitation	29
2.4.1 Posturography	30

2.5	Biomechanics of Balance	32
2.5.1	Human Balance Models	33
2.6	Summary	37
3	Clinical Scales for Objective Evaluation	39
3.1	Analysis of Clinical Assessment Scales	39
3.1.1	Commonly Used Scales	40
3.1.2	Comparison of Clinical Scales	41
3.1.3	Quantifying Parameters	42
3.2	Automated Assessment	44
3.2.1	Posturography	44
3.2.2	Motion Capture	44
3.3	3D Multiple Segment Model for Human Balance	45
3.3.1	OpenSim Implementation	47
3.4	Summary	50
4	Experimental Setup	51
4.1	Instrumentation	51
4.2	Experimental Protocol	52
4.3	System Configuration and Calibration	55
4.4	OpenSim modelling platform	57
4.4.1	Capabilities	58
4.4.2	Gait Extraction Toolbox	59
4.5	Model Scaling	60
4.6	Data Analysis	62
5	Results	65
5.1	Model Verification	65
5.2	Passive Tests	66
5.2.1	Static Standing	67
5.2.2	Standing with eyes closed	70
5.2.3	Standing with feet together	71
5.2.4	Standing with one foot in front	73
5.2.5	Standing on one leg	73
5.3	Active Tests	75
5.3.1	Alternate Stair Touch	75

5.3.2	Sit to Stand to Sit	78
5.3.3	Functional Reach	79
5.3.4	Pick up from floor	81
5.3.5	Look Over Shoulder	82
5.3.6	Rise to Toes	83
5.3.7	In place response	84
5.3.8	Compensatory Step	87
5.4	Walking Tests	89
5.4.1	Walking with Head Turns	90
5.4.2	Walking over Level Surface	92
5.4.3	Stepping over Obstacle	94
5.5	Discussion	95
6	Conclusions	98
6.1	Significance and Contributions	101
6.2	Future Work	102
6.2.1	Clinical Trial	103
6.3	Summary	104
	Bibliography	105
A	Appendix	117
A.1	Categorization of Tests within Clinical Scales	118
A.2	Passive Tests	119
A.3	Active Tests	120
A.4	Walking Tests	121
B	Clinical Scales	122

Abstract

An increasingly aging population has led to a growth in subjects with balance disorders. For an elderly person, falling is a risk that has to be challenged on a day to day basis and must be monitored efficiently. For a stroke patient, the inability to maintain upright posture is a crippling limitation that affects the performance of abilities of daily living and often requires a long process of rehabilitation.

A review of the state of the art in balance recovery and rehabilitation has revealed gaps in the process of assessing and monitoring balance performance in patients at risk. The currently accepted practice is to use Clinical Assessment Scales to evaluate the performance of the subjects. Although simple and easy to use, these methods rely on visual analysis, and the results are given on a qualitative numerical scale. The outcomes are highly subjective and can often lead to confusion for both the patient and the therapist in interpreting the results. This can lead to loss of patient motivation and lack of confidence in the tests themselves. More tools are required to provide accurate and reliable information regarding balance assessment. Up to now, research in the assessment of human balance has almost exclusively been done using simple two dimensional kinematic models, which has been shown to not fully incorporate significant parameters needed for advanced analyses.

The main objective of this research is to develop a new assessment method that provides quantitative information of a subject's performance. To develop this system, the currently used clinical scales have been analyzed, to identify the key parameters that are evaluated and propose an objective assessment tool that focuses on measurable data instead of visual observations. For this purpose, each individual test on the most commonly used scales has been classified into three categories according to the type of motion. Based on the analysis of the clinical scales we have selected two parameters, the center of pressure and the center of mass as the main focus of study.

We propose a hybrid posturography/motion capture system to extract both kinematic and ground reaction force information from test subjects. This data is used with a 12-segment, 31- degree-of-freedom articulated linkage model that is implemented in the OpenSim dynamic simulation software. 19 tests were performed to a cohort of 20 healthy subjects, and the relationship of the CoM and CoP was analyzed in the context of static and dynamic stability. The results show a notable increase in the CoP-CoM distance as the intensity of motion increases. This variation is more significant at the transition phases between different stable poses. These results are consistent with current theories regarding the behavior of the CoM and CoP under static and dynamic conditions. The data reveals that the velocity of the CoM is also an important factor to consider when analyzing stability in dynamic conditions.

The proposed system has shown to be an accurate method to determine the CoP and CoM of a subject while performing clinical evaluation tests. We have determined that our model is significantly more accurate than existing models in the literature. This alternative method has the potential to overcome the limitations of using only posturography as a method for determining stability limits. In the future it can be used to evaluate the performance and evolution of the patient and subsequently evaluate the effectiveness of the rehabilitation process.

List of Figures

1.1	National estimates of the leading causes of nonfatal injuries treated in hospital emergency departments highlighting unintentional falls, United States, 2008; and National estimates of the leading causes of injuries leading to death highlighting ranks of unintentional falls, United States, 2007	2
1.2	Injury types by mechanism in admittances to a Singapore emergency department.	2
1.3	Underlying medical causes for falls at home in admittances to a Singapore emergency department.	3
1.4	Location of the center of pressure and center of mass.	4
1.5	Integration of sensorimotor control systems required for balance.	4
1.6	Iterative process of clinical assessment and determination/modification of training in balance rehabilitation.	7
2.1	Displacement of the otolithic membrane when forces area applied on the head.	16
2.2	Functional organization of the semicircular canals	16
2.3	Muscle Spindle	18
2.4	Integration of the various components of the human balance system. The cerebellum plays a key role in integrating the sensory information and producing an output signal to the motor cortex.	19
2.5	Movement strategies in response to disturbances.	20
2.6	Planes used to describe body motion.	21
2.7	Disorders that affect balance. Different disorders affect distinct components of the human balance system.	22
2.8	Differences between ischemic and hemorrhagic stroke.	24
2.9	Influence of clinical assessment in the formulation of a training plan. . . .	26
2.10	The Balance master by Neurocom.	31
2.11	Different models used in postural analysis.	34

2.12	Location of the 21 markers used in the 14 segment model to estimate the CoM.	37
3.1	Multi segment kinematic model used.	46
3.2	Articulated linkage with 12-segments, 31-degrees-of-freedom implemented in the OpenSim environment.	47
3.3	Location of the lower body segmental reference frames.	48
3.4	Modeling of the ankle, subtalar and metatarsophalangeal joints as frictionless revolute joints.	48
3.5	Schematic of the human knee joint.	49
4.1	Diagram of instrumentation procedure, including force platform posturography and motion tracking.	51
4.2	Location of the Markers on a subject.	54
4.3	Motion Analysis Laboratory	56
4.4	Calibration wand and L-Frame used to define the global origin and calibrate the motion capture system.	56
4.5	Setup of the motion capture lab with force platforms and motion capture cameras. 8 optical infrared cameras captures marker information and 2 force plates measure ground reaction force and center of pressure. A centralized PC is used to combine the information.	57
4.6	OpenSim graphic user interface and Inverse Kinematics tool.	58
4.7	Location of the global, model and force plate coordinate systems. The origins of these reference frames are the calibration point, the pelvis, and the center of the force platform respectively.	59
4.8	Scaled model (right), and original (left).	62
4.9	Overview of marker tracking and model coordinate matching.	63
5.1	Comparison of CoP - CoM estimation error in standing test for three distinct models. Our OpenSim model has a significantly lower estimation error than both Winter's segmented mass distribution model and Kudoh's Gaussian distribution model.	66
5.2	Example of raw marker data (upper panels) and OpenSim model (lower panels) for tests in the passive category. Tests are in order: Quiet standing, standing with feet close together, standing with one foot in front, and standing on one leg respectively.	67

5.3	Measured CoP and Calculated CoM of one subject in a Static Standing test. Comparison between the two parameters for each of the three trials in the sagittal and frontal planes. The bottom two graphs display the associated error for each trial. Even though there are variations in the CoP (solid lines), the CoM (dotted lines) varies accordingly.	68
5.4	Measured CoP and Calculated CoM error of one subject in a Static Standing test. The error is constant throughout the trial.	69
5.5	Absolute error for Static standing for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. The error remains constant throughout the trial.	70
5.6	Absolute error for Standing with Eyes Closed for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. The reduction in sensory feedback causes the CoP-CoM error to increase on average, and also to fluctuate more significantly.	71
5.7	Absolute error for Standing with Feet Together for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a noticeable variation in the CoP-CoM error associated with an increase in sway.	72
5.8	Absolute error for Standing with one foot in front for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a significant increase in error at the transition phase at the start of the trial, which stabilizes afterwards. . .	73
5.9	Absolute error for Standing on One Leg for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. Data displayed for left leg up (L) and right leg up (R). . .	74
5.10	Different stages of the Alternate Stair Touch, seen in the upper panels as raw marker data and the lower panels after implementation in OpenSim. .	75
5.11	Measured CoP and Calculated CoM of one subject in an Alternate Stair Touch test: Comparison between the two parameters for each of the three trials in the sagittal and frontal planes.	76
5.12	Measured CoP and Calculated CoM of one subject in an Alternate Stair Touch test: CoP-CoM error for each trial.	77

5.13 Absolute Error for Alternate Stair Touch for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	78
5.14 Absolute Error for Sit to Stand to Sit for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a noticeably higher error at the transitional phase at the beginning and end of the test, corresponding to the standing and sitting component. The error is significantly reduced during the upright stance phase.	79
5.15 Absolute Error for Functional Reach Forward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	80
5.16 Absolute Error for Functional Reach Lateral for all 20 subjects, averaged for all trials. Data shown for reaching Left and Right. Comparison of the average error of the population and the average error of the subject. There is a steady increase during the lunge phase and a noticeable drop at maximum extension, followed by a constant slower error during recovery phase.	81
5.17 Absolute Error for Pick up from floor for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	82
5.18 Absolute Error for Look Behind Shoulder for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	83
5.19 Absolute Error for Rise to Toes for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	84
5.20 Absolute Error for In Place Response - Forward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	85
5.21 Absolute Error for In Place Response - Backward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	86

5.22 Absolute Error for Forward Compensatory Step for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	87
5.23 Absolute Error for Backward Compensatory Step for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	88
5.24 Gait cycle studied in walking tests. Events are measured in the following order: left foot Toe Off, left foot Heel Strike, right foot Toe Off, right foot Heel Strike.	89
5.25 Measured CoP and Calculated CoM of one subject in a Walking with Head Turns test. Comparison between the two parameters for each of the three trials in the sagittal and frontal planes.	90
5.26 Measured CoP and Calculated CoM of one subject in a Walking with Head Turns test. Error for each trial.	91
5.27 Absolute Error for Walking with Head Turns test for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	92
5.28 Absolute Error for Walking over Level Surface for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	93
5.29 Absolute Error for Step over Obstacle for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.	94
5.30 Average error for all tests shown, separated by categories: in orange passive; in yellow active; and in green walking. In general, the walking and active categories have a significantly increased error when compared to the passive category.	96

List of Tables

3.1	Characteristics of popular clinical assessment scales including Standard Error of Measurement (SEM), Minimum Detectable Change (MDC) and confidence intervals for Test-retest and Inter-rater/Intra-rater reliability. . .	41
3.2	Categories used to differentiate individual tests.	41
3.3	Example of identification of parameters for measurement in the passive category. Extracted from Appendix A.2.	42
3.4	Example of identification of parameters for measurement in the Active category. Extracted from Appendix A.3.	43
3.5	Example of identification of parameters for measurement in the Walking category. Extracted from Appendix A.4.	43
3.6	Chosen parameters and related measurements. The Body Center of Mass cannot be measured directly.	43
3.7	Segment mass distribution of the body from the Winter model.	46
3.8	Standard mass distribution of the body from the Kudoh model.	46
3.9	Inertial parameters for the body segments included in the model	50
4.1	Detailed description of body segment marker locations. Note that both right (R) and left (L) side of the body have markers.	55
5.1	Average mean error, standard deviation and standard error of the mean (SEM) for all tests shown, separated by categories: passive, active, walking. 96	

1 Introduction

This introductory chapter explains the context and relevance of the research. We start by identifying the problem and motivation behind the work. The key knowledge gaps are identified and the objectives and scope of the project are defined.

1.1 Background

Studies on human balance have received increased attention for both clinical and academic reasons. Roboticists study human balance to attempt to reproduce millions of years of Nature's evolution, identify the mechanisms involved and generate advanced models and control algorithms [1–3]. Additionally, clinicians are interested in the study of balance to gain a better understanding of the mechanisms behind the complex human sensor integrations required for upright standing and gait generation [4]. This in turn will allow better comprehension of deficiencies in any of the underlying systems, leading to better tools for assessment and treatment [5]. However, there is no current consensus on what model or methods are more adequate to study the mechanisms of human balance.

1.1.1 Importance of Studying Balance

As bipedal beings, humans are permanently at a risk of falling. This is due to the fact that a significant percentage of the body mass is located high above the ground, making us inherently unstable. The interlacing feedback mechanisms required for maintaining balance can be perturbed due to injury, disease and is particularly increased as people age. Studies have shown that up to 30% of people in America aged 65 years and older living in a community and more than 50% of those living in residential care or a nursing home fall every year [6]. Statistical data from the United States (Figure 1.1), shows the importance of falling as a health hazard. Regardless of age, unintentional falling has been labeled as the leading cause of injury. This problem intensifies in the case of the elderly, as an increase in age is highly correlated with a risk of death due to injury by falling.

In Singapore, falls are also a serious risk for the elderly, as a study reported 17.2% of subjects aged 60 and above reported falling [9]. It is also the the leading mechanism for

Figure 1.1: National estimates of the leading causes of nonfatal injuries treated in hospital emergency departments highlighting unintentional falls, United States, 2008 [7]; and National estimates of the leading causes of injuries leading to death highlighting ranks of unintentional falls, United States, 2007 [8]. Falling is the highest ranked cause of injury for most age groups, and the risk of death due to injury after falling is significantly increased as age progresses.

elderly patients sustaining trauma. In a study of admissions to a Singapore emergency department by Yeo et. al. [10], injuries at home were found to be the most common (Figure 1.2), and falling was found to be the leading cause of all injures (85.3%). Figure 1.3 shows that the main reasons for falling at home are due to loss of balance and/or dizziness.

Figure 1.2: Injury types by mechanism in a Singapore emergency department [10]. The majority of injuries occur at home.

Elderly populations are not the only ones at risk. Patients suffering from Traumatic Brain Injury, Stroke, Parkinson's Disease or Diabetes among others are also at a greater risk of falling [11–14]. In all these cases, the risk of falling imposes a severe financial and social burden on the patients and their families.

Figure 1.3: Underlying medical causes for falls at home in admittances to a Singapore emergency department [10]. The most common cause of injury occurs due to loss of balance.

1.1.2 Human Balance

Human beings are naturally unstable due to the fact that most of the body mass is located in the trunk and head, supported over a narrow base of support. Balance is a generic term that has been defined many different and sometimes contradictory ways in the literature, often confused with Posture. Here we will use the following definitions, depicted in figure 1.4:

- *Posture* is the term that describes the orientation of body segments with relation to gravity.
- *Balance*, also named *Stability*, is the body's dynamic ability to control posture in order to prevent falling.
- *Center of Mass* (CoM) or *Center of Gravity* (CoG) is the point that corresponds to the total body mass in the global reference system. It is calculated as a weighted average of each individual body segment. The term *Sway* is the variation of the CoM due to compensatory action not directly related to voluntary motion.
- *Center of Pressure* (CoP) is the point location of the Ground Reaction Force (GRF) vector, which is independent of the CoM.

The human balance system is the foundation of posture and stability instrumental for performing activities under different conditions. It allows us to stay upright when standing and know where we are in relation to gravity. Our balance system also helps us move and adjust speed and orientation without falling.

Figure 1.4: Location of the center of pressure and center of mass.

To achieve balance, the body relies on a number of sensor and control systems that work together to interpret information from the environment and determine the necessary adjustments. These systems can broadly be grouped into sensory input, integration of the sensory input and muscle motor control. Figure 1.5 shows the integration of these elements.

Figure 1.5: Integration of sensorimotor control systems required for balance [15].

Sensory Input

Information from the surrounding environment is received by the central nervous system through three peripheral sources: Visual, proprioceptive and vestibular. In the eyes, sensory receptors called rods and cones send impulses to the brain when stimulated by light. These visual cues provide information on the orientation of the person with regards to surrounding objects. Proprioception is the sense of the relative position of parts of the body such as skin, muscles and joints due to pressure or stretching of the surrounding tissues. Of great importance are the sensors contained in the neck, which give information on head orientation, and ankles, which give information on the quality of the supporting surface and body sway. The vestibular organs are responsible for providing information regarding motion (linear and rotational), spatial orientation and balance. The utricle and saccule detect gravity (vertical orientation) and linear movement while the semicircular canals detect rotational movement [16] . When the vestibular organs on both sides of the head are functioning properly, they send symmetrical impulses to the brain.

Integration of Sensory Input

The visual, proprioceptive and vestibular cues are sent to the brain stem, where information is sorted. The brain stem is located in a sensory feedback loop with the cerebral cortex and the cerebellum. The cortex is responsible for thinking and memory, and is responsible for generating voluntary motion based on previous experiences and memory. The coordination center of the brain is the cerebellum, and is responsible for generating automatic reactions that have been acquired through repeated exposure (training) [17].

This process of integration of multiple sources of information is essential for maintaining balance in the event of eventual conflicting input. In those situations, higher level thinking and memory can help in deciding the appropriate postural response.

Motor Output

After all information has been received and processed, the brain stem will transmit nerve impulses to muscles controlling the eyes as well as other muscles and joints throughout the body. This coordinated control of different muscle groups is essential for not only maintaining balance while in motion but being able to see clearly.

For vision, the vestibular system uses an automatic function called the vestibulo-ocular reflex to send motor control signals via the nervous system to the muscles of the eyes [18]. Since there are two sets of vestibular organs, the motion of the head is detected by differences in impulses from the left and right ears. These differences in impulses control eye motion and stabilize gaze.

The transmission of impulses from the brain to the muscles of the body is developed from early childhood. Through repetition, nerve pathways are strengthened and over time the brain identifies "normal" sensory input for a given task and complex movements become automatic. This process is called facilitation, and occurs throughout a person's entire lifetime, as a response to changing environments and different sensory inputs.

If one or more of these systems is debilitated due to aging, trauma or illness, the risk of falls can be increased, and balance rehabilitation may be required.

1.1.3 Balance Rehabilitation

The main objectives of balance rehabilitation have been identified as restoring/improving lost or degraded function due to age or illness; reducing the risk of falling and subsequent injury, which leads to an decreased financial and social burden for the patients and their families/communities; and finally the restoration of the ability to perform Activities of Daily Living (ADL), which can have a tremendous impact on user independence.

The complexity of the human balance system creates challenges in diagnosing and treating the underlying cause of imbalance. Currently, the common procedure for balance rehabilitation therapy is an iterative cycle consisting of assessment, determination of training protocol, training session and post assessment with possible modification of training (Figure 1.6).

Assessment

Clinical assessment of balance is a critical process in each step of balance rehabilitation therapy, due to the fact that correct identification of the patient's impairment will dictate the type, intensity and duration of training. At the end of the examination, it is possible to obtain a quantitative measurement of the patient's impairment. These values serve as a baseline for further measurements.

The most common method of assessing balance is the use of standardized clinical scales. These allow classification and comparison of different patients or groups [19]. Examples of commonly used clinical assessment scales are:

Figure 1.6: Iterative process of clinical assessment and determination/modification of training in balance rehabilitation.

- Tinetti Performance Oriented Mobility Assessment (POMA) Test [20]
- Berg Balance Scale (BBS) [21]
- Clinical Test for Sensory Integration in Balance (CTSIB) [22]
- Postural Assessment Scale for Stroke Patients (PASS) [23]
- Balance Evaluation Systems Test (BESTest) [24]

The principle of a clinical scale is to have the patient perform a series of tests while the therapist evaluates and grades performance according to defined guidelines. These tests include evaluation of standing position, resistance to disturbances, and observation of gait. The sum total of marks for all tests in a scale gives an indication of the level of impairment, while evolution of the patient is measured by a marked variation in overall score.

Training

Training protocols for balance rehabilitation are varied and depend on the patient's age and disability, the therapist and resources available. Current trends for rehabilitation methods include perturbation based methods, body weight support [25] and visual or verbal feedback [26, 27].

Perturbation based methods study the response of the patient to a given perturbation. There are two distinct types of responses to any given balance disturbance.

Anticipatory mechanisms are based on feed-forward motion resulting from well learned situations, where reactive mechanisms rely on sensory motor feedback in the presence of an unknown or unexpected situation [28]. While reactive responses reduce the likelihood that a balance loss will lead to a fall, proactive responses can prevent the loss of balance altogether. Researchers and clinicians are divided on which of these mechanisms is more important in the prevention of falls. While some believe repeatable predicted perturbations are more effective training methods [29], others believe that the response to unpredictable displacements is more relevant training [30].

It is common for implemented systems to incorporate different methods for training such as the use of movable force platforms combined with support harnesses and some sort of visual stimulation [31].

Post-Assessment

The post assessment is another important part of the balance rehabilitation program, as it gives both the patient and the therapist information regarding effectiveness of the treatment, and evolution of the patient over time. The methods used are usually the same as the ones used for the initial assessment.

Follow-Up

The last step in a rehabilitation treatment process is a follow-up evaluation, which is usually performed after a period of three months after training. The purpose of this evaluation is to analyze the lasting effects of training after the patient is no longer receiving treatment.

As can be seen from the sections above, balance assessment is a fundamental step throughout the process of balance rehabilitation. Without an accurate, objective evaluation, the patient's recovery can be delayed, the therapist will not have access to the correct information and thus an effective treatment methodology cannot be achieved.

1.1.4 Limitations of Clinical Scales

The widespread use of clinical scales for balance assessment can be attributed to their ease of use, minimal equipment and preparation required to execute and flexibility in evaluation. Additionally, clinical scales attempt to provide some correlation with situations that would occur in activities of daily living.

Despite these clear advantages, therapists are also faced with a number of issues when using these scales. Firstly, clinical scales are mostly observation based, making them highly subjective and susceptible to intra- and inter-rater variability. Two patients with similar performance may be given different scores by the same examiner, and the same patient can be given two distinct scores by two different doctors. The scoring system is also discrete, suffering from flooring and ceiling effects, and the evolution is highly nonlinear. This means that there is no clear distinction between a patient who has just begun treatment and is scored as a “1”, and that same patient several weeks later who has improved however is unable to reach the score of “2” and as such is still marked as “1”. Also, two patients that are scored at the minimum (flooring) and maximum (ceiling) can have completely different performances can sometimes be scored similarly just because the scales have so few intermediate levels. This can have a great effect on patient motivation as the time and effort spent in treatment may not be reflected in the overall score during a post assessment evaluation. Additionally, many classification tests are non-exclusive, and several tests overlap, which can lead to confusion in terms of scoring. Lastly, clinical scales lack the ability to identify the underlying pathophysiology in individual patients [32].

1.2 Objectives

Based on the conclusions given above, it is clear that there are several limitations in the current balance assessment schemes referenced above. These limitations include high subjectivity in the visual assessment performed by doctors and the lack of specificity of the ordinal scales. This work will focus on mitigating some of the issues described with the following objectives:

- Apply a 3-Dimensional model of human balance that accurately captures the dynamics of balance.

As will be detailed further in the next sections, there is a clear discrepancy between the biomechanical models used for clinical studies and those used in engineering studies. It has been shown that the choice of model used can greatly affect the accuracy of the parameters measured. A full body model is necessary to ensure that we minimize the measurement error.

- Develop a new methodology that serves as a precursor for objective assessment for balance that can track minor changes in patient performance.

The developed method will set the foundation for further studies into balance assessment, including correlating measurable parameters with the performance of subjects when evaluated by a therapist. Using more sensing tools than previously used, we can achieve an increase in sensitivity of the evaluation due to a wider range of available data. To achieve these goals, the following steps are required:

- Analyze commonly used clinical assessment scales, identify their similarities and differences.
- Identify relevant measurable parameters used in these tests, set up an appropriate measurement system.
- Implement a human balance model that incorporates the key parameters.
- Analyze the performance of subjects performing clinical scales from the perspective of the key identified parameters.

1.3 Scope and Contributions

In this work we propose to use motion tracking and force platform posturography to study the correlations between dynamic and kinematic parameters and the subjective assessment given by clinical scales. The target clinical scales are the ones that are related to stroke recovery. This population has been selected because after stroke there is an abrupt loss of function, where in the case of elderly the degradation occurs gradually over the years. Stroke patients often are more motivated to pursue rehabilitation because the feeling of loss is more noticeable. Additionally, there is less risk of reduced cognitive function, when compared to other degenerative diseases. The nature of the trauma also allows for faster recovery rates.

In an extensive review by Cretual[33] of over 850 articles related to balance, it has been revealed that in studies related to postural evaluation using clinical scales, no biomechanical model is used. Additionally, the author hints at bioengineering journals being more prone to publishing model based studies than clinical ones. Although no statistical study was made to confirm this point, the data could show a discrepancy between development of new evaluation methods and their current clinical applications.

To address this issue, we have implemented a full body model of the human to simulate the motion performed by subjects while they carry out tests on the clinical scales. The use of an appropriate model is fundamental to ensure the validity of the results. We

have validated this model with a trial in static standing conditions. Afterwards, we have conducted a limited trial with 20 healthy subjects that performed 19 of the tests belonging to some of the most commonly used clinical scales, which have been categorized according to type of motion involved.

Although a full scale clinical trial to establish correlations between clinical evaluations and measurements obtained with this method has been deemed outside the scope of this work, the methodology described in this thesis lays the foundation from which such a trial could be implemented.

- **Claim of Novelty**

We propose to analyze COP-COM relations for analysis of tests in the clinical scales. The identification of significant measurable parameters has the potential to change the way objective balance assessment is performed.

To the authors knowledge, no other study has attempted to find correlations between multiple kinematic and dynamic parameters and the respective performance score in clinical scales. While many other studies have studied the relationship between postural sway and commonly used clinical tests [34–38]. All of these have found little or no correlation between COP excursion and the test chosen. However, the only parameter tested is the center of pressure. Contrastingly, in our results we have shown that there is a significant correlation between the type of test performed and the CoP-CoM distance.

In this work, the author expects the main contributions to be: Firstly, a thorough analysis of clinical scales, similarities and differences as well as their strengths and limitations, will provide insight into what doctors look for in terms of balance assessment. This will shed light on what tests are more relevant, and why therapists still use them regularly despite their numerous limitations. Secondly, the development of a quantitative measurement system that eliminates many of the limitations of clinical scales will improve assessment of balance. This system is independent of the interpretation of the therapist, and provides detailed tracking of the center of pressure and center of mass of the subject. The development of this system lays the foundation for a future clinical trial to correlate measurable parameters with the performance of subjects when evaluated by a therapist.

Lastly, an additional contribution is the implementation of a three dimensional kinematic model of the human body accurately captures the dynamics of balance. This model has been implemented to attempt to overcome the limitations of kinematic models currently used in balance studies. The required complexity of kinematic models

used to describe human balance is still a topic of much discussion. Although there is reticence in using more advanced models in clinical setting due to the added complexity in measurement, the overall trend is to exponentially increase the amount of quantifiable data at the clinician's disposal. This study expects to shed light in that direction, which can be important when attempting to bridge the gap between clinicians and engineers in enhancing the process of balance rehabilitation.

1.4 Chapter Outline

Chapter 1 has been an introduction to the theme of human balance, the conditions that affect it and the required steps that are taken in balance rehabilitation therapy. The main limitations of the current practices have been identified, based on which the objectives and scope of the work have been defined.

Chapter 2 starts with an in-depth literature review on the human balance system, its components and specific functions, as well as the most common balance disorders. Afterwards, the chapter provides details regarding stroke, factors that affect brain plasticity, and requirements of stroke rehabilitation. Also, a deeper understanding of the commonly used assessment methods and training procedures for balance rehabilitation is given. The chapter follows up with a review of current advances in introduction of technology in balance rehabilitation, its advantages and limitations. These will focus mainly on the use of posturography as an assessment and training tool for balance. Having identified the limitations in currently available systems, we analyze the biomechanics of balance, more specifically the relationship between the center of pressure and the center of mass. Lastly, we compare different kinematic models of the human body used in balance.

Chapter 3 contains a thorough analysis of the most commonly used clinical scales. These are classified based on measured parameters, similarities and applicability. A set of "key" quantifying parameters are also identified. Afterwards, a system of automated assessment is proposed, consisting of a force platform and a motion capture system. Through the use of posturography and motion tracking the key parameters defined previously will be measured. A multi-segmented full body human balance model is proposed to study the assessment parameters of clinical balance scales.

Chapter 4 describes the experimental setup used for testing. We begin by describing the motion capture system, its configuration, calibration and process of acquiring motion capture and force plate data. Subsequently we describe the tests that will be performed, as well as the protocol for marker placement. We then introduce the OpenSim modelling software, its applications and how it was used to implement the model. Lastly, the process of determining the center of mass and center of pressure from measured data is explained.

Chapter 5 displays the results obtained from data acquisition of 20 test subjects performing 19 different tests from 3 categories. For each category, we have presented one subject and one as representative of that category, and analyzed in depth the relationship of the CoP and CoM on a per trial basis. We also compare each test of that subject with the entire cohort to show that it is indeed representative of the entire population. Lastly, we perform a comparative analysis of the three categories, and the implications on the relationship between CoP and CoM.

Chapter 6 discusses how the results are interpreted within the context of the scope of the work and contributions, as well as providing an overview of the overall project. We explore the future direction of the work, and propose a number of avenues of research that will enhance the current method.

In the Appendix we have included the distribution of each test in the chosen clinical scales into one of three categories, as well as the comparison between clinical evaluation and measurable parameters. This comparison is the basis of the identification of key parameters discussed in Chapter 3. We also include the clinical scales analyzed in that chapter. These documents include detailed instructions on how to perform the tests, and what are the evaluation criteria.

2 Literature Review

This chapter details the current state of the art in balance rehabilitation and assessment. The chapter starts with a thorough description of the different components of the human balance system, and how healthy individuals are able to use them to maintain balance. The most common balance disorders are classified according to the systems they affect the most. As was mentioned before, this project focuses on stroke. Stroke rehabilitation is promoted via neural plasticity, a process still not well understood which includes repetitive motions to re-teach the brain. The clinical assessment methods are then analyzed and the advantages as well as the limitations of clinical scales are described. Lastly, the current advances in technology in rehabilitation are characterized, with focus on posturography. Finally the biomechanics of balance are detailed, with emphasis on the currently used kinematic models, their applications and limitations.

2.1 Clinical Aspects of Balance

The Human balance system is a complex coordination of sensory integration and continuous motor control systems that function automatically. This section will provide insight on each individual system and how they work together to allow humans to maintain posture.

2.1.1 Components of the human balance system

Vestibular System

The role of the vestibular system is to provide information regarding balance and body position and orientation in space. A properly functioning vestibular system allows the body to generate compensatory reactions when subject to internal or external forces [16]. Two distinct sets of organs are responsible for this sensory input.

The two Otolith organs (Utricle and Sacculus) detect linear displacements and accelerations. When the head is tilted, the displacement of otolithic membrane due to gravity excites the hair cells of the sensory macula. The relation between excitation/inhibition of hair cells provides direction of the tilting (Figures 2.1b and c). Similar excitation can also be achieved when the head is subject to linear acceleration (Figures 2.1d and e).

Figure 2.1: Displacement of the otolithic membrane when forces are applied to the head [39]. a) Normal upright position; b) Backward head tilt due to leaning; c) Forward head tilt due to leaning; d) Effect of linear acceleration; e) Effect of linear deceleration.

The semicircular canals are responsible for detecting angular displacement and accelerations. Inside each canal is sensory mass called the cupula, which contains highly sensitive hair cells and blocks the flow of endolymph (liquid that fills the inner ear) [40]. Head rotations cause the endolymph to displace the cupula and excite the sensory cells (Figures 2.2A and B). The semicircular canals are arranged in pairs, with each canal located on the opposite side of the head as its counterpart. This way, when the head rotates, the displacement of the cupula will generate opposing signals for each of the two partners [39] (Figures 2.2C).

Figure 2.2: Functional organization of the semicircular canals [39]. A) Unperturbed. B) Displacement of the cupula due to angular acceleration. C) Location of the semicircular canals on the head.

Vision

Vision plays a very important role in the maintenance of balance, posture and locomotion. [41] The exact mechanisms by which the eyes detect self motion and motion of the environment are unknown, and different theories have arisen among the years. Of note are the "two mode theory of vision" [42] and the "theory of ecological optics" [43]. In more recent years researchers have developed the concept of afferent and efferent motion perception as the two distinct mechanisms for visual stabilization and posture.

Afferent motion perception refers to the motion of objects in the environment. This is achieved by combining two systems: Central and Peripheral. Central vision is responsible for recognizing objects and their respective motions, while peripheral vision is sensitive to motion of the entire environment. The latter is thought to be extremely important for postural control. Spontaneous body motion causes a phenomenon called retinal slip, which the central nervous system uses as a means of feedback to compensate sway [44]. Efferent motion perception is related to the motion of the body, head or eyes. Two hypotheses are currently considered as viable explanations to the ability to maintain a stable view of the world despite ocular motion. The "inflow theory" states that proprioceptive receptors in the extra-ocular muscles provide the exact position of the eyeballs inside the orbit. The "outflow theory" proposes that visual consistency is achieved by information contained in neural outflow such as a "corollary discharge" or an "efference copy" [45]

Although it has been shown that vision is the central sensory component in balance, it is still possible to maintain upright posture when subject to dark a environment. However, spontaneous lateral body oscillations are largely reduced when objects that are in a darkened environment project a small LED[46].

Proprioception

Proprioception is the sense by which humans are aware of the position of the respective body parts in space without having to look at them. To achieve this, the body uses several sensors. Skin receptors respond to touch and pressure. When muscles contract, the amount of pressure on the skin in the area of movement changes and activate nerve fibers that conduct to the cerebellum. Within the muscles themselves, the muscle spindle provides information regarding changes in muscle length, while the Golgi tendon organ reports changes in muscle tension.

Muscle spindles are small sensory capsules located within the muscles. These are

found throughout the body in parallel to other muscle fibers (extrafusal fibers), and contain specialized fibers called intrafusal fibers. Intrafusal fibers are characterized by extremities containing contractile proteins, and a non contractile central region which is wrapped by the muscle spindle afferent. When the muscle stretches, this activates ion channels which trigger action potentials in the muscle spindle afferent [47]. Figure 2.3 displays a muscle spindle and its main components.

Figure 2.3: Muscle spindle capsule and its main components [48].

Golgi tendon organs are afferent nerve endings that are located in series with the extrafusal muscle fibers, at the junction of the muscle and tendon. If the muscle is stretched, the higher elasticity of the muscle fibers when compared to the tendon does not stimulate the sensory dendrites. However, when there is an active contraction of the muscle the resulting force acts directly on the tendon, and the Golgi tendon afferent is activated. [39]

Central Nervous System

The central nervous system (CNS) is the basis of voluntary motor control and reflexes. It allows us to maintain normal quiet, stance and to safely accomplish the majority of activities of daily living. The CNS is responsible for processing and integrating all the input from various sensors and generating coordinated signals for the muscle fibers.

Within the CNS, the Cerebellum is the regulator of all motor output. It receives information from other areas of the brain that dictate voluntary and programmed movements, while also receiving sensory feedback from receptors such as visual, vestibular and proprioceptors regarding the movements as they occur. After the information is processed, a new signal is generated to the motor cortex to modulate the output (Figure 2.4). As this post processing is done outside of the cognitive area of the brain, balance is maintained subconsciously while other tasks are performed.

Figure 2.4: Integration of the various components of the human balance system. The cerebellum plays a key role in integrating the sensory information and producing an output signal to the motor cortex.

2.1.2 Normal Postural Control

When standing upright, even healthy individuals do not stand absolutely still, as the body has a tendency to sway. Most of this motion occurs in the sagittal plane (see figure 2.6, and subsequently the majority of early research and treatment has been focused on anteroposterior (front/back) stability. More recently researchers have begun to focus on stability mechanisms in other directions [49].

To recover from postural instability due to displacements of the base of support the body uses a different number of movement strategies [50], which are used in both feed-forward and feedback control loops. The feedback loop refers to posture changes determined by information from sensory organs such as the vestibular system, proprioceptors and vision. These can include response to external perturbation, such as a displacement of the support surface, as well as a change in the gait cycle caused by tripping or slipping. The feedforward loop refers to postural changes that are of anticipatory nature, as a response to a voluntary motion. These can include prevention of a disturbance or changing of the position of center of mass (COM). The three main motion strategies for recovery in the sagittal plane are the ankle, hip and stepping strategies (Figure 2.5). Closely related to these strategies are characteristic patterns of muscle activity, called muscle synergies. A muscle synergy is a coupling of a group of muscles which causes them to act as a unit.

Ankle Strategy

As the name implies, the ankle strategy restores stability using the ankle joint. When leaning forward, the role of the gastrocnemius is to reverse the forward motion, while the hamstrings and paraspinal muscles extend the hips and knees. When leaning back-

Figure 2.5: Movement strategies to restore balance in presence of disturbances: a) ankle strategy, b) hip strategy, and c) stepping response [51].

wards, it is the role of the anterior tibialis to reverse the motion, followed by activation of the quadriceps and the abdominal muscles. This type of strategy appears to be most commonly used when both the perturbation is small and the support surface is stable [51].

Hip Strategy

The hip strategy generates large body motion at the hip, with antiphase rotation of the ankle. Unlike the ankle strategy, the principal muscles activated are the abdominal/quadriceps and paraspinal/hamstrings muscles, with little contribution from the anterior tibialis or gastrocnemius. It has been suggested that the hip strategy is used in response to perturbations of larger magnitude, or in cases when the base of support is unstable [51].

Stepping Strategy

If the two above strategies are insufficient to restore postural stability, a step is required to realign the base of support under the COM. Early research indicated that a stepping response occurred only when the perturbation was great enough to displace the COM outside the base of support. Nowadays, it has been shown that stepping can occur even when the COM is well within the limits of the base of support [51].

Although these strategies have been described independently, in reality there is a wide range of intermediate responses between pure ankle and pure hip, as well as stepping responses. As an example, when slipping occurs, there is a continuous motion of the ankle and hip throughout the recovery.

Mediolateral Stability

Mediolateral stability refers to all motion that occurs within the body's frontal plane (see figure 2.6). The alignment of the body joints and respective muscles allows very little movement in the mediolateral direction for the ankle. For this reason, recovery of stability occurs mainly at the hip and trunk. The primary mediolateral motion is the lateral movement of the pelvis, caused by abduction of one leg and adduction of the other [52].

Figure 2.6: Planes used to describe body motion.

Maintaining postural control in a dynamic environment requires us to modify the movement strategy used. We are constantly adapting the type and amplitude of the postural response to the changes in surrounding conditions. Additionally, with repeated exposure to a given task, the body optimizes its response and becomes more efficient [53].

2.1.3 Balance Disorders

Balance disorders are conditions that affect one or more of the sensorimotor systems that control upright stance. Although many of the following disorders affect multiple systems simultaneously, they can be divided according to which system is most affected (figure 2.7):

Disorders that affect the Brain

Parkinson's is a progressive degenerative disorder of the CNS that causes tremors, bradykinesia (slow movement), rigidity and loss of postural reflexes causing instability.

Figure 2.7: Disorders that affect balance. Different disorders affect distinct components of the human balance system.

Both Stroke patients and patients with Traumatic Brain Injury (TBI) often suffer from loss of brain function, which can lead to the inability to move one or more limbs on one side of the body.

Disorders that affect proprioception

Diabetes is a disease that is caused either by insufficient production of insulin from the pancreas or if the cells are unable to respond to the production of insulin. This causes high blood sugar which leads to vascular problems and in extreme cases can lead to a condition called diabetic neuropathy, which causes numbness and pain in the feet, as well as muscle fatigue and weakness. Patients with complete or partial SCI can have limited or no control of lower sections of the body.

Disorders that affect the vestibular system

Benign Paroxysmal Positional Vertigo (BPPV) and Meniere's Disease are conditions that commonly cause loss of balance and vertigo.

In disorders that are related to the brain, recovery is caused by a poorly understood phenomenon called neural plasticity.

2.2 Neuromotor Rehabilitation

Neuromotor rehabilitation is a topic related to the recovery of sensorimotor impairment caused by illness or trauma. This section will shed light on the process of brain plasticity, as well as its importance in stroke recovery. The main causes of stroke and its influence on the level of impairment are also described, and their relation to balance rehabilitation.

2.2.1 Brain plasticity

The concept of brain plasticity refers to adaptive nature of the central nervous system. Currently, there is strong evidence to support that neurons and other brain cells are able to alter both their structure and function as a response to internal stimuli [54].

One of the first to study the phenomenon of brain plasticity was Donald O. Hebb, who was interested in identifying the role of neurons in learning. According to Hebb, when axons of adjacent nerve cells fire together, the efficiency of the interconnections between those cells gets strengthened [55]. This process of brain reorganization begins at the earliest stages of human life, and occurs continuously through adulthood, whenever the person is subject to a different task, or in the event of brain injury.

Some of the most important factors that influence brain plasticity after brain injury can be described as:

- Repetition of a task is necessary to provide lasting neural changes [56].
- Intensity of stimulation and training can greatly affect the restructuring of the brain. Too little intensity can lead to weakening of synaptic responses [57], while excessive intensity can lead to overuse and degradation of function.
- Time also plays a crucial role in post injury plasticity. Not only is the whole process of neural reorganization a gradual process, but also the elapsed time between the occurrence of brain injury and the learning process. Studies have shown that training before at an early stage of injury is more effective [58].
- Age leads to decline in neuroplastic responses of the brain [59].

In this work, we will focus on one specific type of brain injury, the cerebrovascular accident or stroke.

2.2.2 Stroke

A stroke is classified as the permanent loss of brain function caused by a disruption of the supply of blood to the brain. Without constant supply of oxygen and nutrients, brain cells begin to die. The consequences of the loss of brain cells can be varied, ranging from inability to formulate words, loss of hearing, reduced vision and motor dysfunctions [60]. Two distinct types of stroke exist: ischemic and hemorrhagic. Figure 2.8 shows the differences in each type.

Figure 2.8: Differences between ischemic and hemorrhagic stroke [61].

Ischemic stroke is the most common type of stroke, representing over 80% of all strokes [62, 63], and occurs when a blood clot obstructs one of the blood vessels supplying blood to the brain. This causes the brain cells to deteriorate due to lack of oxygen and nutrients, leading to reduced function of the affected area. According to the location of the clot, ischemic strokes can further be characterized into thrombotic strokes, if the clot is located in the brain, or embolic if it is located elsewhere in the body. Hemorrhagic stroke is caused by a rupture of one of the blood vessels in the brain, leading to blood leakage into the brain (intracerebral hemorrhage) or the surrounding tissues (subarachnoid hemorrhage). The accumulation of blood increases pressure and compresses the surrounding brain tissues. There is also a third type of stroke, referred to as Transient Ischemic Attack TIA or mini stroke. this occurs when one of the arteries leading to the brain is temporarily blocked, which normally leaves no permanent damage in the brain [60, 64].

2.2.3 Stroke Recovery

The process of brain restructuring after a stroke is still mostly unknown in the medical community. While some patients experience significant early recoveries, others are impaired for life and require extensive treatment. The level of impairment is dependent on the location and severity of the stroke, as well as how fast emergency medical treatment is found.

According to the National Stroke Association of America [65]:

- 10% of patients recover almost completely.

- 25% recover with minor impairments.
- 40% experience moderate to severe impairments requiring special care.
- 10% require care in a nursing home or other long-term care facility.
- 15% die shortly after the stroke.

Immediate medical treatment following stroke is focused on maintaining essential vital signs, and preventing life threatening conditions such as arrhythmia [66]. After the patient is stable, the focus shifts towards rehabilitation, which should begin as quickly as possible. Evidence shows that it is crucial for rehabilitation to start early, as the first 12 weeks after a stroke are when recovery is most evident. [67].

Most stroke victims survive with a number of motor, sensory and/or cognitive disabilities which limit the ability to perform basic activities of daily living (ADL) such as eating, dressing, walking, toileting and basic hygiene [68]. Consequently, the goal in stroke rehabilitation is to improve function of the patient to enable independence in performing ADL. Rehabilitation of a stroke victim involves a interdisciplinary group of people including a psychologist, speech therapist, physical therapist and occupational therapist.

One of the most detrimental aftermaths of stroke is impaired postural control, due to the severe limitations it has on ability to perform independent ADL [11]. Thus, it is essential to focus rehabilitation of the stroke patient in balance control.

2.3 Balance Rehabilitation

As was mentioned in chapter 1, balance rehabilitation follows an iterative process which is highly dependent on the patient, therapist and available resources.

2.3.1 Clinical Assessment of Balance

Clinical examination of balance occurs on three distinct levels: Evaluation of the ability to perform functions that require postural control; the different movement and sensory strategies required to execute said functions, and the fundamental sensory and/or motor impairments that limit them. This information is then used to formulate a training plan, short- and long-term goals and promote functional recovery of the patient.

Figure 2.9: Influence of clinical assessment in the formulation of a training plan.

Impairment

When assessing a patient, it is important to examine impairments that have the most impact on postural control. Examiners should focus on the ankle muscles that play a crucial role in the maintenance of upright stance, as well as the range of motion at the ankle and hips. Lastly, it is necessary to evaluate the integrity of the visual system, as well as somatosensory input at the feet and ankles.

Functional Assessment

The analysis of the functional capabilities of the patient are the most complex to perform. The initial evaluation should contain a self report such as the Activities-specific Balance Confidence (ABC) Scale [69]. In this report the patient rates his/hers own ability to perform certain tasks. The results guide the therapist into specific training in those tasks where the confidence levels are the lowest.

Performance based tests for functional balance compare the level of the patient with pre-established norms, serving as a baseline for further measurements and directing treatment according to the patients evolution. Some of these tests include: the Timed Up and Go (TUG) test [70], which measures the ability to stand up, walk a predetermined distance and return; the Functional Reach Test [71], which measures the how far a patient is able to reach forward while still maintaining balance; or The Multi-Directional Reach Test [72] examines stability not only in the sagittal plane but also in mediolateral directions.

Sensorimotor Strategies

Assessment of the sensorimotor strategies used to maintain stability in voluntary motion and in case of perturbations is a fundamental component of balance assessment. This examination begins by verifying the alignment of the patient with regards to gravity. Measuring the movement strategies in response to perturbations requires a series of tests. Self initiated body movements are examined while the patient voluntarily shifts weight forwards, backwards and side to side. Additionally, anticipation of possibly destabilizing movements can be tested by asking patients to place one foot on a stool or lifting a heavy object [49]. To test the reactive response the patient is given the “nudge test” [20]: a small perturbation in the forwards or backwards direction, and the recovery process is evaluated. To examine the maintenance of stability under different environments, differences in sensory information are tested. The most simple of these tests is Romberg’s test [73], where the patient’s vertical sway is observed with feet together, eyes open, followed by eyes closed. The sharpened or tandem Romberg test is a variation of the original test, where the patient has to place his feet in heel-to-toe position, with one foot directly in front of the other. The Clinical Test for Sensory Integration in Balance (CTSIB)[22] uses six distinct sensory conditions to ascertain the ability to maintain balance when up to two out of three (visual, vestibular or proprioceptive) sensory systems are impaired.

Clinical Scales

In light of all the different available tests for various tasks, motor and sensory conditions, some researchers have attempted to develop assessment tools that incorporate components from all the above mentioned methods. These Clinical Scales are commonly used as a standardized and simple examination that provides important information for balance recovery. Examples of these Clinical Scales are the Tinetti Performance Oriented Mobility Assessment (POMA) Test [20], the Berg Balance Scale (BBS) [21], Postural Assessment Scale for Stroke Patients (PASS) [23]; and the Balance Evaluation Systems Test (BESTest) [24].

These scales are widely applied due to their simplicity, flexibility and ease of use.

Clinical scales are restricted by two types of limitations:

- **Method based limitations** - Inherent to the procedure, derived from the fact that each test is a simplification of real world scenarios thus not all parameters can be taken into account. They are built around a limited set of environmental conditions,

and thus cannot predict what the actual performance of the patient will be when subject to complex environments. Also, most scales do not incorporate tests for all three aspects of postural control: steady state, reactive and anticipatory. Lastly, clinical scales lack the ability to identify which subsystems are responsible for a decline in performance. [32, 49].

- **Evaluation based limitations** - Dependent on the observer, and therefore are subject to inter- and intra-rater variability. They consist of discrete values with non-linear progression, and suffer from ceiling and flooring effects.

2.3.2 Balance Training

With regards to training protocols, the same three levels are present: Impairment, movement/sensory strategies and functional tasks.

Impairment

The first step is to attempt correct any impairment that the patient may have, such as pain in the lower extremities or loss of sensitivity. This allows the patient to resume strategies previously used. Studies have shown that patients stability improves with strength training, aerobic conditioning, sensory organization training and balance training such as Tai Chi [74]. In cases where the impairment is permanent, it is necessary to develop new strategies that will enable completion of tasks.

Sensorimotor recovery

The recovery of steady-state, reactive and anticipatory control of balance is a key process in the rehabilitation process. Motor rehabilitation includes training in maintenance of alignment, controlling the movement of the COM relative to a constant base of support, and changes in the base of support to maintain stability.

The goal of alignment retraining is to allow the patient to be able to have a steady position that is appropriate for the task, is efficient in respect to gravity, and maintains the COM over the limits of stability. Approaches for alignment training include verbal and manual cues to assist the patient in locating vertical posture, use of visual feedback or queues such as mirrors or flashlights, or somatosensory queues from leaning against a wall or chair.

Retraining movement strategies is a key aspect in the development of effective movements for postural control. This retraining occurs within the context of anticipatory and

reactive control. Although there is no consensus over which mechanism is more important for the prevention of falls, both can be used effectively for balance retraining.

Reactive balance control training is based on exposing the patient to unpredictable displacements with varying frequency, directions and intensities. Patients that undergo perturbation based training have reduced risks of falling due to improved ability to respond to perturbations [75]. In anticipatory balance control training, patients learn to control movements of the COM over an increasingly wide area. The premise is that repetition of perturbations in a predictable and gradually incremental fashion causes long-term effects on postural stability for prevention of loss of balance and falls [76].

Sensory training is also fundamental for the patient to learn how to organize and select the most appropriate sensory input for postural control. Patients who are reliant on vision are requested to perform a number of tasks while blindfolded, keeping their eyes closed or with restricted vision. Patients who rely on somatosensory inputs are requested to stand on compliant surfaces such as foam or moving surfaces. Lastly, to encourage the improvement of the vestibular senses, patients are asked to not stand on compliant surfaces and have their visual cues removed or limited. It is common practice to provide a support harness for patients undergoing such tests.

Functional Tasks

Training at a functional task level is applying and adapting the different sensorimotor strategies to changing environments, such as home or work, and specific tasks. It is crucial to practice postural control responses in a wide variety of conditions, which require adapting the sensorimotor strategies used. Additionally, the difficulty of the tasks should be increased gradually over time, to stimulate the optimization of the postural response. Lastly, it is important to notice that the training of impairments, sensorimotor strategies and functional tasks is not in sequential order, but must be run in parallel.

2.4 Technology in Balance Rehabilitation

As a complement to clinical examination, researchers have implemented instrumental analysis as a means of providing more exact quantitative measurements and evaluate therapeutic efficiency [77]. The use of technology is a natural step in the rehabilitation process, as it provides more tools for both researchers that are developing new assessment and training methods and clinicians that are implementing them.

Examples of these implementations include: Use of wearable accelerometers recording over long periods of time [78]; shoe mounted sensors such as the Mobifit or Gaitshoe [79, 80]; video recording systems for gait kinematic and kinetic analysis [81]; and electromyographic (EMG) measurements [82].

One of the most researched tools for use in the clinical environment has been computerized posturography.

2.4.1 Posturography

Posturography is a technique that examines active and passive postural control regulation by analyzing the trajectory of the center of pressure (CoP) given by a force plate. It is divided into static and dynamic posturography.

Static Posturography

Static posturography examines postural control while patients are in an unperturbed state on a fixed surface, where the center of gravity can be approximately extracted directly from CoP measurements with a 97% certainty [83]. The variations of the position of the CoP during testing are used to determine quantitative measures such as the velocity of the CoP on the anteroposterior or mediolateral axes, the path of COP, and the 95% confidence ellipse area of COP [84]. These measurements enable the determination of the energy expenditure required to maintain balance [85].

The advantages of static posturography are the ability to provide a linear, quantitative and objective measurement of static balance [86]. However, it is limited due to its simplistic nature, which does not reflect real life conditions and activities of daily living. Another crucial limitation in the applicability of static posturography to the clinical environment is the wide variety of equipment and procedures, which makes standardization practically impossible. Examples of key parameters that should be considered standard are feet position, test duration, velocity and path of the CoP [87]. These parameters define the width of the base of support, the amount of time taken which can affect fatigue, and the path taken which determines the total displacement of the CoP.

Examples of commercial uses of static posturography based systems include Pro Balance by Lab-Rehab [88], the Balance SD by Biodex [89] and the Nintendo Wii Balance Board [90, 91].

Dynamic Posturography

Dynamic posturography focuses on the study of the effect of induced perturbations, which can be directed at upper body segments, movable support base, moving the visual surround, or voluntary weight shifting [32].

Figure 2.10: The Balance master by Neurocom.

Dynamic posturography focuses on the study of the effect of induced perturbations, which can be directed at upper body segments, movable support base, moving the visual surround, or voluntary weight shifting [92]. The main advantage of this method is that the assessment may provide detailed data on direct sensory contribution to balance control, since each of the sensory inputs can be manipulated independently [93]. Additionally, the variety of different stimuli and operating conditions can be more closely related to certain activities of daily living (ADL) [32].

The Balance Master by Neurocom [94] uses dynamic posturography via a movable force platform (Figure 2.10). The system provides not only CoP data but also an Equilibrium Score (ES), which is determined as the angular difference between the subject's calculated antero-posterior CoP displacement and a theoretical maximum. In the Equitest system also by Neurocom [95], the subject stands on a force platform surrounded by a visual enclosure, which is able to provide a sway referenced environment, and introduces the Postural Stability Index (PSI), defined as the percentage ratio of the total stabilizing ankle torque and the destabilizing torque due to gravity. Other systems such as the Balance Quest-MultiTest by Framiral [96] provide a percentage of stability calculated from deviations both in the antero-posterior and mediolateral directions.

Despite the clear advantages, posturography has yet to be incorporated into routine clinical evaluations, which can be due to a number of factors. Firstly, the requirements

in terms of hardware and associated financial cost can be high (a force platform is estimated at 6,500. USD, while more advanced systems can go up to 100,000. USD [94]). Additionally, posturography assessment requires individual calibration, takes longer to administer and may not correlate well with certain ADL. Furthermore, a critical limitation in posturography is that the body movement analysis is restricted to the examination of the CoP excursion, from which the center of gravity trajectory is extracted. [19]. This leads to a great dependence on the kinematic model used for determination of the center of mass.

In a 2011 meeting of the European Society for Clinical Evaluation of Balance Disorders (ESCEBD), "Consensus was that a force platform cannot be considered as a technique which is sufficient on its own to perform a clinically relevant test for the assessment of neuro-otological and musculo-skeletal conditions, evaluation of compensation or treatment (rehabilitation) or prediction of falls" [97].

Although the use of posturography shows great promise as a tool for assessing balance, it is insufficient as a standalone tool to provide all the relevant information required for an accurate evaluation of the subject condition.

2.5 Biomechanics of Balance

From a mechanical standpoint, the study of balance appears simple, as the only forces in action are weight, applied to the center of mass (CoM), and the ground reaction force, applied to the center of pressure (CoP). The weight is a constant, so only the measurement of the ground reaction forces (GRF) is required to determine the balance of forces. However, any horizontal displacement of the CoM, either due to voluntary motion or perturbation compensation, will generate a torque that further increases the displacement and can lead to falling if no counteracting forces are applied. This mechanical instability requires continuous control of both CoP and CoM positions.

The most simple and commonly used method for determination of stability is the zero moment point (ZMP) criterion proposed by Vukobratovic et al. [98], initially developed for control of bipedal robots. A zero moment point is defined as a point where the sum of the moments of the interaction forces between the feet and the base of support is zero. Stability is ensured if the position of the ZMP is located inside the convex hull constructed by the feet in contact with the ground, assuming non slip conditions. The distance between the ZMP and the edge of the support surface indicates the level of stability. In the condition of flat foot, the ZMP and the CoP coincide.

However, It has been argued that the sole measurement of the CoP is insufficient in analyzing the factors that affect stability in dynamic posture. In [99], it is hypothesized that the regulation of the angular momentum about the center of mass of the body is highly regulated in human locomotion and can be used as a stability measure. In [100], the author states that not only the CoP excursion, but also the velocity of the center of mass (CoM) is a key factor in balance control. Subsequently, the author proposes using the extrapolated center of mass, a vector constructed by adding the CoM position and its velocity multiplied by a factor, as a measure of stability to be used as a complement to the CoP excursion. Other works have also proposed the joint analysis of both the CoP and CoM [101–105]. If the distance between the CoP and the CoM increases, the amount of postural control energy required to counteract the resultant moment also increases. In a two part paper Hsue et. al [106, 107], demonstrated that the CoP-CoM distance, as well as the instantaneous CoP velocity, to be important to classify stability in patients with cerebral palsy.

It is clear that a complete study of the biomechanics of balance needs to consider both CoP and CoM. Since neither parameter is an anatomical mark, indirect measurements and computational estimates are required. The position of the center of pressure can be determined through pressure mapping sensors, or via measurement of ground reaction forces and torques via a force platform, such as in posturography. Determining the position of the CoM is a more complex problem, since it must take into account the relative positions and inertia of each body segment. Currently, there is no real time measurement system to accurately determine the CoM, so naturally a simplified model is required.

2.5.1 Human Balance Models

Biomechanical models of human balance with varying degrees of complexity have been used extensively for studies on human posture and movement [108, 109]. The majority of these studies that use anatomical models consider two approximations. Firstly, the body is composed of a set of rigid segments, and secondly, these segments are articulated by mass-less, frictionless joints [110]. With this hypothesis the center of mass of each segment remains stationary with respect to its extremities, and the global CoM can be determined as the weighted sum of the segmental centers of mass.

Each study will apply specific models in accordance to individual experimental conditions. Identifying the minimum number of kinematic variables required for accurate

description of the dynamics of balance and gait is a key step in choosing a model. In the literature, six distinct categories of models were found, as depicted in figure 2.11. Within each category the model may (single plane), or may not (biplane) be restricted to an anatomical plane. This is relevant to define antero-posterior (AP) movement in the sagittal plane, and mediolateral (ML) movement in the frontal plane. Movement in the horizontal plane is usually restricted to CoP models. Most planar segmental models study movement on the sagittal plane, since this is the plane where most of the balance recovery motion occurs [111]. These models are often based on direct feedback mechanisms [112].

Figure 2.11: Different models used in postural analysis: a) Center of Pressure alone (CoP); b) Center of Mass alone (CoM); c) Center of Pressure and Center of Mass (CoP/CoM); d) Simple inverted pendulum; e) Double inverted pendulum; f) Multi-segmented. Modified from [33]

• CoP

The simplest possible model considers only the measurement of the center of pressure alone. This is by far the most commonly used model in biomechanical postural analysis, and movement is considered on the horizontal plane. As explained previously, several studies have shown that measuring CoP alone is not enough to determine stability. The choice of this model may often be due to technological constraints rather than methodological reasons [33].

• CoM

Studies that consider only the Center of Mass are less common, due to the difficulty in its exact determination. Studies that do so are usually interested in following motion across a wider area, where access to force platform data is more limited. The CoM is often approximated to the third lumbar vertebra [113].

- **CoM/CoP**

Models that consider both the CoP and CoM combine the properties of both models described above, while focusing on the relationship between these two parameters. To note that often studies that use this model often use only one parameter, or derive one from the other.

- **Simple inverted pendulum**

The inverted pendulum model is the simplest and most widely segmental model of human balance [3, 5, 114, 115]. It consists of a Center of Mass (CoM) which rests on a single, mass-less rod, which in turn is attached with a joint (ankle) to a triangular foot. To control the inverted pendulum, torque is applied at the ankle.

The main advantage of this model is that it reduces the system to a single degree of freedom, as the destabilizing effect of gravity is the greatest. This type of model can be used to identify postural sway during quiet standing, using static posturography, where the ankle strategy is the predominant movement strategy applied. However, complex motions that require coordination of multiple joints cannot be identified by this model. More recently, many studies have shown that the one-link inverted pendulum model is unable to accurately capture the dynamics of postural sway [116, 117].

- **Double inverted pendulum**

As was described above, during rapid or high intensity perturbations, the body responds using the so called hip strategy. This implies that the human balance model should be multi-segmental. The double inverted pendulum model introduces another degree of freedom by including the hip joint. In this model the human body is represented by two segments (legs and trunk), which are controlled by two actuators (ankle and hip). A variation of the double inverted pendulum is the triple inverted pendulum (figure 2.12a), which considers the knee joint as well. Although not commonly used, Boonstra et al.[118] have shown that the combined contributions of the hip, ankle and knee joints are relevant in analyzing balance, while Pinter et al. reported the impact

of non rigidity of both the hip and knee joints was non negligible [116]. When studying human postural control in the mediolateral plane. double inverted pendulum models have been implemented (ankle and hip), since the knee does not bend in the mediolateral direction (Figure 2.12b). In these models, pelvic motion is the main source of CoM displacement.

• Multisegmented

More complex three dimensional models have also been used to analyze dynamics of walking, specially when considering the effect of the hip on both mediolateral and sagittal plane stability [120, 121]. Nakamura et al. [122] use a seven link, six joint model as a basis for a assistive walking controller. Robert et al. [123] used a 34 degree of freedom (DOF) multi-segmented model to calculate the inverse dynamics of the body to determine individual joint load in response to sudden perturbations. In a more clinical application, Esrafilian et al. applied a multi-segmented model to study the energy consumption of patients suffering from knee osteoarthritis while walking. However, when calculating stability, only the CoP is considered. In his reference book [124], D. A. Winter proposes a standard 14-segment model (Figure 2.12), with independent mass fractions for each individual segment. This model can be further simplified to fit the necessary experimental constraints. When modeling human joints, it is standard practice to consider the neck as a ball-and-socket joint (3DOF), the shoulders as ball and socket joints, the elbows as universal joints (2DOF), the pelvis as a ball-and-socket joint, the knees as hinge joints (1DOF) and ankles as either ball-and-socket or universal joints.

Figure 2.12: Location of the 21 markers used in the 14 segment model to estimate the CoM. Four trunk segments are considered to track internal mass shifts in the thoracic and lumbar regions [104].

2.6 Summary

The human balance system is a highly complex interaction of muscular, sensory and cognitive processes that work seamlessly to help us maintain stability when interacting with the environment. If one or many of the systems responsible for maintaining equilibrium is impaired by age, disease or trauma, the risk of falling is increased. In this work we have focused mainly on the victims of stroke, since the cause of impairment is neurological and not musculoskeletal. In many cases, a balance rehabilitation program can help the patients improve their ability to stand upright and perform activities of daily living.

One of the key aspects of balance rehabilitation is the process of assessing the patients condition. However, The many different interacting systems make the task of identifying the nature of the patients condition difficult. It is common clinical practice to refer to standardized clinical scales to establish a baseline for the level of the patient. These clinical scales have had widespread use due to the simplicity in performing and evaluating the tests contained therein. Albeit easy to use, clinical scales suffer from a number of limitations. Of the two types of limitations described, those that are related to the method of evaluation show the most promise for improvement.

The biomechanical approach focuses on studying the physical parameters that define balance, but often require additional measuring tools. Determination of the center

of pressure and center of mass of the subject has been shown as a key component for accurate description of postural control. Computerized posturography is a method that was developed to help researchers and clinicians identify and study the mechanisms controlling postural sway and resistance to perturbations. Despite its advantages in obtaining more detailed information on the contribution of the individual sensory systems' impact on posture, they have yet to be widely incorporated in the clinical environment. The high costs of equipment, increased set-up time and most importantly lack of standardization in the measurements has limited the applicability of these methods. It has also been shown that posturography alone, relying solely on the measurement of the CoP, is unable to determine the underlying pathology of the patient, and as such cannot be used as a standalone test in balance assessment.

The review made in this chapter has highlighted the need to develop new tools for assessing balance. The current available methods are unable to overcome the needs of patients and therapists, and research in posturography as a means to quantify the performance have as of yet been unsuccessfully introduced. Additionally, there is a current gap in the modelling tools used for clinical assessment of balance. Even though complex models have been shown to be fundamental for accurately capturing the dynamics of balance, they have up to now been overlooked in favor of overly simplistic models.

The following chapter describes a novel approach to assess balance based on evaluating the commonly used tests in the clinical scales. These tests will provide clues as to what parameters are more relevant for determination of balance performance. Afterwards, a multi-segmented model is implemented, and used to study the relationship between the CoP and CoM of subjects performing these tests.

3 Clinical Scales for Objective Evaluation

It is clear that no ideal solution for assessment of balance can be identified. Clinical scales lack the objectivity and scientific rigor required to accurately evaluate patients, and current applications of posturography are unable to claim a permanent place in the rehabilitation clinic. More research into balance assessment is required to provide therapists with effective tools to improve evaluation and subsequent treatment of patients.

The key to filling this knowledge gap lies in understanding why therapists trust the results given by clinical scales despite all the problems associated with them. If the real phenomena that are being assessed can be identified, it is possible to modify the evaluation phase of the clinical scale and develop a more objective assessment. In this manner, although the intrinsic conditions of the scales cannot be removed (Method based limitations), it is possible to reduce the subjective nature of the evaluation, making it more universal and more horizontally transferable between doctors, researchers, hospitals, etc. If the score a subject achieves in a test can be attributed to a measurable, quantifiable parameter such as the center of pressure, a standard assessment method can be developed which relies on these measurements.

This chapter presents an analysis of clinical scales. Each test in the most commonly used scales has been evaluated and categorized according to the type of motion, key parameters have been identified, and afterwards linked with engineering concepts. An instrumentation method to measure these parameters has been detailed. To accurately determine said parameters, we introduce a full body model of the human body, as well as how it is implemented in the OpenSim simulation platform. This model will be used to match experimental measurements with dynamic parameters such as the center of mass and center of pressure.

3.1 Analysis of Clinical Assessment Scales

There are many clinical scales available to the balance rehabilitation therapist. With a thorough analysis of the most commonly used clinical scales, similarities and differences in the tests can be identified. Afterwards, the most commonly evaluated parameters can be correlated with measurable physical variables such as CoM/CoP, upper/lower body kinematics, and ankle/hip torque.

3.1.1 Commonly Used Scales

There is no clear consensus in which evaluation method is more valid for clinical balance assessment. As such the choice of scale used will depend on the experience of the therapist. Here we describe four of the most commonly used scales that are used for balance assessment post stroke.

Berg Balance Scale

The Berg Balance Scale (BBS) [21] was developed to measure balance among older people with impairment in balance function by assessing the performance of functional tasks. It consists of 14 items evaluated through a five-point scale (see Appendix B), ranging from 0-4, where 0 indicates the lowest level of function and 4 the highest level of function. The maximum score is 56, and a change in function is revealed when there is a difference of at least 8 points between two distinct assessments.

Tinetti Performance Oriented Mobility Assessment (POMA) Test

The Tinetti Performance Oriented Mobility Assessment Test [20] is an assessment test designed to evaluate both static and dynamic conditions for older adults. It is divided into a balance section with 9 items, and a gait section with 7 items, which are sometimes administered as individual tests (see Appendix B). Each item is scaled from 0-2, where the highest score indicates independence within each test item.

Functional Gait Assessment

The Functional Gait Assessment [125] was developed to assess the postural stability of patients while performing various tasks. It consists of 10 items (see Appendix B) scored from 0-3 where “0” represents severe impairment and “3” represents normal ambulation.

Balance Evaluation Systems Test (BESTest)

The BESTest was developed as an assessment that targets 6 different control systems relevant to balance, in an attempt to provide specific training practices for individual balance deficits [24]. It is an extensive test and comprehensive, consisting of 36 items grouped into 6 categories: biomechanical constraints; stability limits/verticality; anticipatory postural adjustments, postural responses, sensory orientation and stability in gait

(see Appendix B). Each test is evaluated on an ordinal scale from 0-3, and the total score is afterwards calculated into a percentage score.

Table 3.1 shows an overview of the different clinical scales as well as their performance measures.

Table 3.1: Characteristics of popular clinical assessment scales including Standard Error of Measurement (SEM), Minimum Detectable Change (MDC) and confidence intervals for Test-retest and Inter-rater/Intra-rater reliability.

Scale	Tests	Time (m)	Points Scale	SEM	MDC	Test-retest	Inter/Intra-rater
BBS	14	15-20	0-4	1.79	8	0.72-0.95	0.95
BESTest	30	30-40	0-3	2.8	7.81	0.88	0.87-0.98
POMA	16	10-15	0-2	N/A	N/A	0.97	N/A
FGA	10	5-10	0-3	1.52	4.2	0.95	0.94

3.1.2 Comparison of Clinical Scales

Looking at the four scales described above it is easy to see that there are minimal differences within some scales, as some items are repeated across the scales. The question arises as to whether these items are the ones actually measuring properties that have a relevant impact in the accuracy of the assessment. To test this premise, each item has been grouped into one of four categories, as seen in table 3.2.

Category	Description
Pre-existing	Tests that assess the physical impairments such as pain or range of motion.
Passive	Tests where the patient maintains a constant posture with no voluntary motion.
Active	Tests where the patient is subject requested to perform small movements, or is subject to external perturbations.
Walking	Tests where the patient is required to walk a number of steps.

Table 3.2: Categories used to differentiate individual tests.

The rationale behind this division is to categorize each action from the simplest (passive) to the most complex (walking), in terms of the demand to the sensorimotor systems. Pre-existing conditions such as joint pain or stiffness causing loss of range of motion have been placed in a separate category. Although they are important for maintenance of stability in , they can be identified and accounted for outside of the traditional testing apparatus. The “passive” category includes all activities that consist of maintaining steady state posture over a constant and firm base of support. The “active” category includes the tests that require change in base of support, or that incorporate external perturbations or voluntary displacements of the CoM. Lastly, the “walking” category in-

cludes all components that require multiple steps, or where the gait pattern is the subject of evaluation. Appendix A.1 shows the categorization of all tests in the chosen clinical scales.

3.1.3 Quantifying Parameters

After each category has been selected, the next step is to determine what measurable parameters can be extracted for each motion. Naturally, within each category there may be an overlap of parameters, which in turn may indicate that such a parameter plays a key role in the evaluation of the task at hand. For each category, a table has been generated identifying the clinical objectives of the test, the possible forces and motions measurable, and potential instrumentation required.

Passive

Table 3.3 shows an example of the process of analyzing the clinical aspects of the tests grouped under the 'passive' category. As this category encompasses scenarios where the base of support remains unchanged, and no perturbations are present, the relative motion is expected to be low, and a force platform is likely to provide sufficient information.

Passive				
Clinical Perspective		Eng Perspective		
Name of Test	Clinical Measurements	Forces Measured	Motions Measured	Instrumentation
CoM alignment	Antero-posterior, mediolateral CoM alignment and segmental postural alignment	Centre of pressure - extract COM position	Static posture (motion of body in static balance)	Force platform Image capturing for segment kinematics?
Verticality and lateral lean	Patient returns to vertical smoothly without overshooting or undershooting.	Centre of pressure - extract COM variations	Lean - variation of CoM from center to max and return	Force platform Image capturing
Functional reach forward	Maximum horizontal distance reached	Centre of pressure - extract COM variation min to max	Horizontal distance forward	Force platform Image capturing

Table 3.3: Example of identification of parameters for measurement in the passive category. Extracted from Appendix A.2.

Active

Table 3.4 shows an example of the process of analyzing the clinical aspects of the tests grouped under the 'Active' category. These tests incorporate a wider range of motion, and although a force plate may provide useful information, it becomes necessary to use additional sensors to detect the kinematics of the body.

Active				
Clinical Perspective		Eng Perspective		
Name of Test	Clinical Measurements	Forces Measured	Motions Measured	Instrumentation
rise to toes	stability and height, compare force used when holding hands	force used with hands for support, variation of CoP	Ankle elevation	Force sensor on support bar. Motion capture
alternate stair touching	number of successful touches and time required to complete 8 touches	pressure on supporting foot. Extract varying COM for each step	leg kinematics	force platform on supporting foot, force sensor on step. Motion capture
standing arm raise	stability, use of step to maintain balance, ability to lift quickly, maximum attainable angle	COP - COM	trunk / arm kinematics	force platform. motion capture

Table 3.4: Example of identification of parameters for measurement in the Active category. Extracted from Appendix A.3.

Walking

The tests grouped under the walking category are exemplified in table 3.5. These tests require the subject to ambulate a number of steps, and thus identification of segmental and joint kinematics is of greater importance. While force platforms may be able to identify the variation of the CoP during the gait cycle, motion capture has a more important role in analyzing the subject's posture during the evaluation.

Walking				
Clinical Perspective		Eng Perspective		
Name of Test	Measurements	Forces Measured	Motions Measured	Instrumentation
gait - level surface	ability to walk distance, time required	foot pressure	body kinematics	pressure pads on walkway/treadmill, visual system
change in speed	balance during transitions	foot pressure	body kinematics, COM acceleration	pressure pads on walkway/treadmill, visual system
step over obstacle	change of speed, maintenance of balance - hesitation, short steps or obstacle touch	foot pressure	body kinematics, COM acceleration	pressure pads on walkway, visual system

Table 3.5: Example of identification of parameters for measurement in the Walking category. Extracted from Appendix A.4.

Relevant Parameters

Parameters	Measurements
Center of Pressure	Ground Reaction Force
Joint Kinematics	Angle and Angular Velocity
Segment Kinematics	Position and Velocity
Body Center of Mass	-

Table 3.6: Chosen parameters and related measurements. The Body Center of Mass cannot be measured directly.

Based on the tables described previously, the most relevant parameters to be measured have been determined, and are displayed in table 3.6.

Appendix B includes transcripts of the four clinical scales analyzed: the Berg Balance

Scale, the Balance Evaluation Systems Test, Tinetti Balance Assessment Tool and the Functional Gait Assessment. These include the tests that should be performed, as well as details on how to guide the subject and instructions on how to score based on performance criteria. It is clear that although many of the tests overlap between methods, there seems to be no overall consensus on the type and number of tests, as well as differences in scoring criteria.

In Appendix A.1 we display the complete categorization of all tests in all four clinical scales studied. It is clearly seen that there is significant overlap in regards to the type of tests performed. Appendixes A.2, A.3 and A.4 are a comparative analysis of observable versus measurable parameters that are used in evaluating the performance.

3.2 Automated Assessment

There are many methods to measure the parameters described in the previous sections. Some of the most common parameters found are the position of the CoM, and the kinematics of the ankle and hip joints. Tables 3.3, 3.4 and 3.5 already propose sensors that might enable the detection of such parameters.

3.2.1 Posturography

The clinical uses of posturography have already been discussed in the previous chapter. A force-plate will typically consist of three or four force sensors that measure ground reaction forces. The location of the resulting ground reaction force is called the center of pressure (CoP) and can be easily determined from the individual contributions. In the literature, the standard parameters used are the range of CoP in both directions, the CoP standard deviation in both directions, and the CoP mean velocity [87]. The parameter that is desired is the body center of mass (CoM), and in quasi-stationary situations it can be extracted directly from the CoP, since it is a vertical projection. In dynamic situations, the body's inertia becomes an important factor which influences the deviation of the CoP [126].

3.2.2 Motion Capture

In order to capture the kinematics of joint motion, a motion capture system is required. The type of motion tracking system determines key factors in the system design, such as the amount of space required, whether or not the patient can ambulate, the

costs of equipment and the ease of use. Examples of commonly used sensors are inertial measuring units, electromagnetic sensors and optoelectronic systems [119]. Inertial measuring units generate data based on the fusion of gyroscopes, magnetometers and accelerometers. They are cheap and easily accessible but have limited accuracy. Electromagnetic sensors relay the position of receivers to an external transmitter. Since the data transmission is electromagnetic, there is no need for line of sight, however they are highly affected by signal noise and interference. Optoelectronic systems are much less sensitive to noise, however they require line of sight to the target, as well as multiple markers/cameras. The main advantage to these systems is that they have been used extensively in balance and gait analysis, and thus a wide range of computer algorithms is available.

3.3 3D Multiple Segment Model for Human Balance

As was described in the literature review, there is no current consensus regarding which model is more adequate to describe the intricacies of human balance. Up to now, researchers studying clinical scales have used over simplistic models, or in many cases forgoing models altogether, due to technical constraints. Recent studies have shown that these models are inadequate for describing important characteristics of the postural response mechanisms.

With that in mind, the author proposes the use of a three dimensional multiple segmented model for use in this study [127]. The human body has been modeled as a 12-segment, 31-degree-of-freedom articulated linkage (Figure 3.1). The head and torso are lumped together as a single rigid body that articulates around the pelvis via a ball-and-socket joint. Each shoulder is a ball-and-socket joint, and each elbow is a universal joint. Each hip is a ball-and-socket joint, each knee is a translating hinge joint [128], each ankle is a universal joint comprised of two non-intersecting hinge joints.

Determination of the segmental center of mass will depend on the inertia of each body segment. The information extracted from the force platform and the cameras cannot be used to determine this parameter exactly, however there are a number of ways to calculate the approximate value. In a nine year longitudinal study, Jensen [129] studied the changes in segment mass, the coordinates of the mass centre and the principal moments of inertia the segments and the body of a sample of males aged 4-20 years old. Winter [124] proposes a segmental mass distribution based on anthropomorphic data shown in table 3.7. Other works such as Kudoh [130] use simpler method by approxi-

Figure 3.1: Multi segment kinematic model used.

imating the body masses to a standard mass distribution (Table 3.8).

Segment	Mass Fraction
Head	0.081
Thorax	0.216
Abdomen	0.139
Pelvis	0.142
Thighs	0.100
Legs	0.0465
Feet	0.0145
Upper arms	0.028
Fore arms	0.022
Total	1.000

Table 3.7: Segment mass distribution of the body. Modified from Winter 2009 [124].

Element	Mass (%)	Total
Upper Arm	3.6 x2	7.2
Fore Arm	2.9 x2	5.8
Torso	50	50
Thigh	1.4 x2	22.8
Shank	5.3 x2	10.6
Foot	1.8 x2	3.6

Table 3.8: Standard mass distribution of the body. Modified from Kudoh 2004 [130].

As can be seen by analyzing both tables, there are important differences in the mass distributions for each individual segments. Also of note is that in Winter's model, the position of the center of mass of each segment is explicitly defined in relation to bony landmarks. Kudoh's model approximates each segment to a cylinder, thus the position of the segmental center of mass is located at its geometric center.

To determine the kinematic and dynamic parameters of the segments, the model has been implemented in OpenSim software [131], a freely available application for building and analyzing computer models of the musculoskeletal system and performing dynamic

simulations.

3.3.1 OpenSim Implementation

The model used in this work is a variation of the *Gait2392* model included in the OpenSim standard distribution. The default unscaled properties are for a subject measuring 1.8m and weighing 75.16Kg, based on measurements from [132]. For this model, the surface data for the bones in the pelvis and thigh was obtained by marking surfaces with a polygon mesh, and using a three dimensional digitizer. A visual representation of the model in the OpenSim environment is shown in Figure 3.2.

Figure 3.2: Articulated linkage with 12-segments, 31-degrees-of-freedom implemented in the OpenSim environment.

Kinematics

The upper extremities consist of the torso, humerus, radius/ulna and hand. The torso segment connects to the humerus segment at the acromial-clavicular joint. The elbow joint is located at the midpoint between the lateral and medial condyles of the ulna. The

wrist joint is fixed at the midpoint between the ulnar head and the styloid process of the radius.

The lower extremities contain six rigid body segments: the pelvis, femur, tibia/fibula, talus and foot. The foot includes the calcaneus, navicular, cuboid, cuneiforms and metatarsals. The pelvic reference frame is fixed at the midpoint of the line connecting the two anterior iliac spines (ASIS), where it connects to the torso. The femoral frame is fixed at the center of the femoral head. The tibial frame is located at the midpoint of the line between the medial and lateral femoral epicondyles. The talar frame is located at the midpoint of the line between between the apices of the medial and lateral malleoli. The toe frame is located at the base of the second metatarsal (Figures 3.3 and 3.4).

Figure 3.3: Location of the lower body segmental reference frames [133].

Figure 3.4: The ankle, subtalar and metatarsophalangeal joints are modeled as frictionless revolute joints with axes oriented as shown [133].

The knee joint is a particularly complex structure, due to the non-circular aspect of the femoral condyles, the joint does not operate as a simple pin [128]. Here, the

knee is modeled based on a simplified single degree of freedom model first developed by Yamaguchi et al. [134], which accounts for the kinematics of both the tibiofemoral joint and the patellafemoral joint in the sagittal plane as well as the patellar levering mechanism. This was further explored by Delp et al. [133], who represented the femoral condyles as ellipses, who roll and slide on the tibia plateau, which in turn is represented by a line segment (Figure 3.5). Seth et al. [128] later adapted the model by removing the patella to avoid kinematic constraints.

Figure 3.5: Schematic of the human knee joint [128]. In this model, the tibia has one rotational degree of freedom, θ , but translates in the rotation plane (x, y) with respect to the femur.

Inertial properties

The inertial parameters for the lower body segments are adapted from the 10-segment Anderson and Pandy model [135]. In that study, the mass and inertial properties were based on anthropomorphic data averaged from five subjects. The dimensions of the segments are taken from [133]. The upper body parameters are adapted from [127]. Table 3.9 displays the mass and principal moments of inertia for each segment used.

Table 3.9: Inertial parameters for the body segments included in the model

Body Segment	Mass (Kg)	Moments of Inertia (Kg.m ²)		
		xx	yy	zz
Torso	26.8266	1.4745	0.7555	1.4314
Humerus	2.0325	0.011946	0.004121	0.013409
Ulna	0.6075	0.002962	0.00618	0.003213
Radius	0.6075	0.002962	0.00618	0.003213
Hand	0.4575	0.000892	0.000547	0.00134
Pelvis	11.777	0.1028	0.0871	0.0579
Femur	9.3014	0.1339	0.0351	0.1412
Tibia	3.7075	0.0504	0.0051	0.0511
Talus	0.1000	0.0010	0.0010	0.0010
Calcaneus	1.250	0.0014	0.0039	0.0041
Toe	0.2166	0.0001	0.0002	0.0010

3.4 Summary

This chapter has presented a thorough analysis of existing commonly used clinical scales, and categorized them according to the type of motion involved and presence of external perturbations. These categories are active, passive and walking. We have identified that the center of pressure and center of mass are key parameters in describing the dynamics of balance within the context of clinical scales.

The analysis of the literature has identified that there is a significant discrepancy between the complexity of the balance model used and the environment of the study. Studies that focus on clinical evaluations of balance performance use very simple models, however these has been shown to not fully capture the dynamics of balance. Here we have presented a multi-segmented model of the human body based on anthropomorphic data from experimental measurements. This model is of enough complexity to capture the dynamics of postural sway, and will be used to determine the center of pressure and center of mass of subjects performing tests in clinical scales representative of the three categories.

4 Experimental Setup

This chapter details the process of data acquisition and analysis. The sensors used are described in detail, as well as the overall system setup. The method for capturing motion data and force platform is described, as well as how the information must be formatted to accommodate the implementation into OpenSim.

4.1 Instrumentation

The following experimental setup has been implemented to analyze the parameters described above, shown in Figure 4.1. Force platforms detect the patient's response by measuring the ground reaction forces and subsequent CoP. Additionally, an optical marker based motion capture system will be used to determine the movements of body landmarks. This data is fed into our human model in OpenSim, where information regarding center of mass and joint kinematics is calculated.

Figure 4.1: Diagram of instrumentation procedure, including force platform posturography and motion tracking.

4.2 Experimental Protocol

Twenty subjects volunteered to participate in the study. All subjects were healthy adults and were not suffering from any musculoskeletal injury likely to affect their balance at the time of testing. All experiments were conducted at the Motion Analysis Laboratory, Dept of Mechanical Engineering, Nanyang Technological University. Participants height, weight, age, and gender were collected at the start of each session. Participants were requested to wear sandals, shorts and a tight-fitting sleeveless top. Participants were requested to do 19 tests, repeating 3 times per test. Forces and motions are captured for the duration of all tests. Adequate rest time between attempts was given to prevent fatigue. The following is a description of the tests performed, which have been grouped into three categories according to the selection criteria described in the previous chapter.

Static Trial

The first test performed is the Static Trial. In this test, the subject stands with feet shoulder width apart and with arms straight slightly separated from the body and holds the position for about 5s. This test is required to capture the default marker position for that specific subject, and is used to perform the scaling of the generic model as described below.

Passive Tests

- **Standing Unsupported:** The subject is asked to stand with both feet on one platform, arms at the sides and look forward, holding position for 30 seconds.
- **Standing with eyes closed:** Subject is asked to stand with both feet on one platform, arms at the sides, close eyes and hold for 15 seconds.
- **Standing with feet together:** Subject is asked to stand on one force platform with both feet together touching at the ankles, and hold for 15 seconds.
- **Standing with one foot in front:** Subject is asked to place one foot in front of the other, ankle to toes, and hold for 20 seconds.

Active Tests

- **Sit to Stand to Sit:** Subject begins test seated on a chair with feet on force platforms. Upon request, the subject will rise up until upright, and after a brief delay

sit down again.

- **Functional Reach Forward:** Subject is asked to stand with both feet on one platform, and reach forward as far as possible without lifting the heels from the ground, then return to straight.
- **Functional Reach Lateral:** Subject is asked to stand with both feet on one platform, and reach to the one side as far as possible without lifting the heels from the ground, then return to straight. Repeat for the other side.
- **Pick up from floor:** Subject is asked to pick up an item placed on the floor between their feet and return to upright.
- **Look behind shoulder:** Subject is asked to stand on one platform and look over the left shoulder at a sign setup 10 meters behind, returning to face forward afterwards. After a brief pause repeat looking over the other shoulder.
- **Alternate Stair Touch:** Subject is asked to stand on one platform with both feet, alternating touching a box placed on another platform 8 times per foot.
- **Rise to toes:** Subject is asked to stand with both feet on one platform. Upon request, the subject is to rise onto his toes and hold for up to 3 seconds.
- **In place response: Forward:** A research assistant stands in front of the subject, placing one hand on each shoulder and lightly pushing the patient backward until the anterior ankle muscles contract, then suddenly releases.
- **In place response: Backward:** A research assistant stands behind the subject, placing one hand on each scapula and isometrically resisting the backwards push from the patient. As soon as the heels are about to be lifted the assistant releases the pressure.
- **Compensatory step: Forward:** A research assistant stands in front and to the side of the subject with one hand on each shoulder and asks them to push forward. Once the subject has leaned until their shoulders and hips are in front of their toes, the support is suddenly removed. The test must elicit a step.
- **Compensatory step: Backward:** A research assistant stands at the back and to the side of the subject with one hand on each scapula and asks them to lean backward. Once the subject has leaned until shoulders and hips are behind the heels the support is suddenly removed. This test must elicit a step.

Walking Tests

- **Walk over level surface:** The subject is asked to perform 8 steps ensuring that the feet will contact the force plates independently. Multiple trials are allowed until the subject is comfortable with the steps without noticeable change in gait.
- **Walk with head turns:** The subject is asked to perform 8 steps along the walkway, while simultaneously turning their head from side to side.
- **Step Over Obstacle:** The subject is asked to perform 8 steps, stepping over two stacked boxes of 23cm in height.

Marker derived data was retrieved using a three dimensional video motion capture system (Motion Analysis Eagle system, Motion Analysis Corp, Santa Rosa, CA, USA) Small reflective markers (14mm in diameter) were mounted over specific landmarks on the trunk, pelvis, legs and arms (Table 4.1 and Figure 4.2)[136].

Figure 4.2: Location of the Markers on a subject.

Table 4.1: Detailed description of body segment marker locations. Note that both right (R) and left (L) side of the body have markers.

Segment	Marker	Position
Trunk	R(LSH)	Marker Placed on the tip of the shoulder [acromio-clavicular (AC) joint]
	C7	Marker placed over on the spinous process of 7th cervical vertebra
Pelvis	R(L)ASI	Marker placed over anterior superior iliac spine (ASIS)
	SACR	Marker placed at the midpoint between left and right posterior superior iliac spines (PSIS)
Thigh	R(L)THAP	Marker located at the proximal anterior aspect of the thigh
	R(L)THAD	Marker located at the distal anterior aspect of the thigh
	R(L)THLP	Marker located at the proximal lateral aspect of the thigh
	R(L)THLD	Marker located at the distal lateral aspect of the thigh
	R(L)THLP	Marker located at the proximal lateral aspect of the thigh
	R(L)LEPI	Marker over lateral epicondyle of femur
	R(L)MEPI	Marker over medial epicondyle of femur
Shank	R(L)TIAP	Marker located on the proximal 1/3 of the anterior shaft of the tibia
	R(L)TIAD	Marker located on the distal 1/3 of the anterior shaft of the tibia
	R(L)TILAT	Marker located on the mid-lateral aspect of the tibia
	R(L)LMAL	Marker located over the lateral malleolus
Foot	R(L)MMAL	Marker located over the medial malleolus
	R(L)HEEL	Marker on distal aspect of bisection of posterior calcaneum
	R(L)MFS	Marker on superior medial mid-foot
	R(L)MFL	Marker on lateral mid-foot
	R(L)P1MT	Marker on medial aspect of 1st metatarsophalangeal (MTP) joint
	R(L)P5MT	Marker on lateral aspect of 5th MTP joint
Arm	R(L)TOE	Marker on distal end of first toe of foot
	R(L)ELB	Marker located over the lateral epicondyle of the humerus
	R(L)WR	Marker over the dorsal aspect of wrist

4.3 System Configuration and Calibration

Marker trajectories were recorded using 8 optical infrared cameras with a sampling rate of 120Hz (Motion Analysis Eagle system, Motion Analysis Corp, Santa Rosa, CA, USA) using the Cortex motion capture software. Each camera has a specific IP address and connects to a central hub via TCP. The Master camera collects all the information and connects to the central processing PC via A/D connector for synchronization with force platform data. Ground reaction force and center of pressure data were measured using 2 AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) sampling at 1080Hz. Each force platform feeds a signal amplifier before connecting to the PC via A/D connector. To remove high frequency background noise, ground reaction forces were also filtered at 50Hz using a fourth order Butterworth filter. This system is implemented in the Motion Analysis Laboratory at NTU shown in Figure 4.3.

Before any data capture can be made the laboratory configuration must be recorded and the motion capture system must be calibrated. To calibrate the motion capture system, we used the “Calibrate” tool included in the Cortex software. Firstly, the laboratory reference origin is defined by the position of an L-frame. This calibration frame has 4 markers attached at a fixed distance of 400mm and 800mm. For all experiments performed in this work, the location of the global origin chosen was the same. The next step

Figure 4.3: Motion Analysis Laboratory

is the calibration of the motion capture system, which is performed using a calibration wand with three markers distanced by 500mm. Figure 4.4 shows the calibration wand and L-frame.

Figure 4.4: Calibration wand and L-Frame used to define the global origin and calibrate the motion capture system.

The wand is then waved around the entire capture area making sure that the 8 cameras have a clear sight of all three markers. This enables the system to sync up the information from all cameras and provide one clear signal. Once the motion capture system is calibrated, 4 reflective markers are placed at each of the corners of the force platforms, and their position is recorded. This step is essential to be able to synchronize

the force and motion data in post processing.

Figure 4.5: Setup of the motion capture lab with force platforms and motion capture cameras. 8 optical infrared cameras captures marker information and 2 force plates measure ground reaction force and center of pressure. A centralized PC is used to combine the information.

A schematic of the laboratory setup is shown in figure 4.5. The motion capture lab was calibrated before each session, and force platforms were reset digitally to eliminate hysteresis after each trial. Data was saved into C3D format, and a Gait-Extract toolbox [137] was used to process the kinematic and force data before importing into OpenSim.

4.4 OpenSim modelling platform

As has been mentioned throughout this work, the OpenSim modelling software has been used to implement the model and calculate the center of pressure and center of mass. OpenSim is a freely available, community driven and user extensible software system developed in Stanford University. OpenSim is built on the fast and robust Simbody dynamics engine. Simbody is free and open source and includes its own set of documentation and examples.

OpenSim lets users develop models of musculoskeletal structures and create dynamic simulations of movement, while visualization of models and simulation analysis is shown via a graphical user interface. The open source and extensible software also includes an application programming interface (API) that developers can use to extend the software.

4.4.1 Capabilities

OpenSim is a very flexible tool that allows the user to construct models of a wide variety of mechanisms and musculoskeletal models, which can include rigid bodies, joints, physical constraints, springs, dampers, contact models, muscles, external forces and actuators. Muscle models include the Hill-type which include both active and passive muscle properties based on tried and true dynamic models from the literature.

One of the key advantages of OpenSim is the extensive validated musculoskeletal model library of both the upper and lower extremities. Additionally, the scale tool enables the creation of subject specific models based on experimentally measured data. As a community driven platform, there are a number of user generated toolboxes to import experimental model data from common motion capture systems. One of these toolboxes is the Gait Extraction Toolbox developed by Tim Dorn [137].

Once the model has been defined, there are a number of tools that can be used for static and dynamic analyses, namely inverse kinematics, dynamics and forward simulations of movement. The Computed Muscle Control tool can generate muscle-driven forward simulations of movement, while the she static optimization tool solves the muscle redundancy problem based on algorithms in the literature. Figure 4.6 shows the GUI and Inverse Kinematics tool.

Figure 4.6: OpenSim graphic user interface and Inverse Kinematics tool.

OpenSim also enables visualization of data by either plotting trajectories, or using its imaging or video capture tools. It also supports importing geometry created in other software platforms via .stl and .obj files. The full OpenSim library of models and tools

is available through Matlab and GUI scripting. This allows Matlab programmers and GUI users to set up batch processing routines or extend OpenSims functionality, without programming in C++. Lastly, Opensim contains an Extensible Application Programming Interface (API), providing access to the underlying algorithms. This enables creating new plug-ins and custom made analyses for a variety of different needs.

4.4.2 Gait Extraction Toolbox

As mentioned above, extracting raw motion capture data and preparing for OpenSim implementation was done with the C3D Gait Extraction Toolbox [137]. This step is essential as there are three distinct coordinate systems which need to be taken into account when extracting kinematic data from the C3D file. Analog data from the force plates (FP) and the motion capture data from the laboratory (LAB) are each on their respective local coordinate system. However, before loading into OpenSim, all data must be converted to the model's local reference frame (MODEL). Figure 4.7 shows each of these coordinate systems in their respective origins.

Figure 4.7: Location of the global, model and force plate coordinate systems. The origins of these reference frames are the calibration point, the pelvis, and the center of the force platform respectively.

The gait extract toolbox was developed to extract and process experimental gait data for use with OpenSim. This toolbox requires the Motion Labs C3D Server. While origi-

nally implemented with the VICON Nexus motion capture software, it has been modified in this work to allow using our own configuration. This was achieved by modifying the coordinate systems, force plate names and marker sets in the *loadLabels.m* as described in the C3D toolbox manual. Since there is no EMG data captured in this work, those sections were ignored.

A key aspect in using the Gait Extract toolbox is defining events. Events are specific frames of interest within a given trial. The most common events are heel strikes and foot offs used when studying gait. However, when studying an event where both feet remain still it is still necessary to define at least a “start” event and “end” event. For each of these events, it is required to define which foot is in contact with which force plate.

When performing a static trial, the toolbox can be used to generate a *Scale_Setup* file, which compares the experimentally the measured markers to the generic model markers and generates scale factors used in model scaling.

4.5 Model Scaling

The model parameters described previously were relative to a male approximately 1.79m of height and weighing around 75Kg. In order to be able to apply this model to the different types of subjects, it is necessary to apply a scaling factor. All scaling in this work was done using OpenSim’s *Scale* tool.

Scaling is a comparison of the distances between the virtual markers placed at anatomical locations in the model, and the same marker data obtained experimentally via motion capture. The dimensions, mass and inertial properties of each body segment in the model are then modified to fit the experimental data based on pre-determined scale factors.

Computing these scale factors can be achieved by comparing the locations of specific marker pairs that define body segments, such as the length of the femur. The experimental distance for all measured frames is then compared with the model distance, and the overall scale factor is calculated based on the average of the scale factors computed due to all the specified marker pairs. The scale factors are then used to scale both the sizes of the body segments as well as their masses, so that the total mass of the body matches the subject’s mass.

In this work, mass scaling was obtained by preserving mass distribution, ensuring the subject’s model segments maintain the same proportion as the generic model. The individual segments also have their inertial tensors updated. An alternative method

would be to scale the individual segment masses according to their specific scale factor. However, this can lead to errors since the summed mass of the bodies will most likely not equal the total mass of the subject.

The final step of the scaling process is to match the model markers to experimental markers obtained in a static pose. Which is computed by solving an inverse kinematics (IK) problem. If certain coordinates have well known trajectories, they do not need to be computed, and are called *prescribed* coordinates. *Unprescribed* coordinates are the ones that need to be determined by inverse kinematics. This is achieved by determining the best model pose that matches experimental marker and coordinate values, by minimizing the following weighted least squares equation:

$$\min_q \left[\sum_{i \in \text{markers}} w_i \|\mathbf{x}_i^{exp} - \mathbf{x}_i(\mathbf{q})\|^2 + \sum_{j \in \text{unprescribed coords}} \omega_j (q_j^{exp} - q_j)^2 \right] \quad (4.1)$$

$\mathbf{q}_j = \mathbf{q}_j^{exp}$ for all prescribed coordinates j

Where $\mathbf{q}$ is the vector of generalized coordinates being solved for; $\mathbf{x}_i^{exp}$ is the experimental position of marker i ; $\mathbf{x}_i(\mathbf{q})$ is the position of the corresponding marker on the model; q_j^{exp} is the experimental value for coordinate j ; w_i are the marker weights and ω_j are the coordinate weights, specified by the user. The least squares problem is solved using a general quadratic programming solver, with a convergence criterion of 0.0001 and a limit of 1000 iterations.

Figure 4.8: Scaled model (right), and original (left).

Figure 4.8 shows the original model and a scaled version for a subject 1.9m of height and weighing 85Kg.

4.6 Data Analysis

Once the scaled model has been obtained for each specific subject, kinematics and dynamics can be determined for all trials. The process of determining Inverse Kinematics is similar to the one described in the scaling section: in this case the marker data is no longer in a static pose, so each step will correspond to a different pose and calculate the best match according to equation 4.1. The experimental markers are matched by varying the joint angles in the model for each step. Figure 4.9

After the kinematics are obtained, it is possible to determine the generalized forces and torques at each joint via Inverse Dynamics (ID). Having access to the states of motion (motion capture) as well as external load data (force platform), it is possible to solve the following classical equation of motion:

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{C}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{G}(\mathbf{q}) = \boldsymbol{\tau} \quad (4.2)$$

where N is the number of degrees of freedom, $\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}} \in \mathbf{R}^N$ are the vectors of generalized positions, velocities and accelerations; $\mathbf{M}(\mathbf{q}) \in \mathbf{R}^{N \times N}$ is the mass matrix of the

Figure 4.9: Overview of marker tracking and model coordinate matching.

system; $\mathbf{C}(\mathbf{q}, \dot{\mathbf{q}}) \in \mathbf{R}^N$ is the vector of Coriolis and centrifugal forces; $\mathbf{G}(\mathbf{q}) \in \mathbf{R}^N$ is the gravitational vector and $\boldsymbol{\tau} \in \mathbf{R}^N$ is the vector of generalized forces.

The final step is determining the trajectories of the body center of mass, which can be obtained by weighted sums of each of the individual segments.

$$\begin{aligned}\mathbf{CM} &= \frac{1}{M} \sum_{i=1}^n m_i \mathbf{cm}_i \\ \dot{\mathbf{CM}} &= \frac{1}{M} \sum_{i=1}^n m_i \dot{\mathbf{cm}}_i \\ \ddot{\mathbf{CM}} &= \frac{1}{M} \sum_{i=1}^n m_i \ddot{\mathbf{cm}}_i\end{aligned}\tag{4.3}$$

where $\mathbf{CM}, \dot{\mathbf{CM}}, \ddot{\mathbf{CM}}$ are the vectors of position, velocity and acceleration of the global body center of mass; M is the total mass of the body; m_i are the individual segment masses, and $\mathbf{cm}, \dot{\mathbf{cm}}, \ddot{\mathbf{cm}}$ are the vectors of position, velocity and acceleration of the segment center of mass.

Determining Center of Pressure

Since the center of mass has been determined in model coordinates, it is now necessary to also convert the individual force plate CoP into a global CoP in model coordinates. The C3D extract toolbox already converts each of the two force plates' data into model coordinates, however we still need to calculate the overall CoP. This is done by combining the individual center of pressures factored in by their respective total vertical force contributions according to the following [138]:

$$CoP(x, z) = CoP_{right} \times \left(\frac{GRF_{yright}}{GRF_{yright} + GRF_{yleft}} \right) + CoP_{left} \times \left(\frac{GRF_{yleft}}{GRF_{yright} + GRF_{yleft}} \right)\tag{4.4}$$

In case of both feet being on the same platform, $CoP_{right} = CoP_{left}$ and $GRF_{right} = GRF_{left}$.

5 Results

This chapter displays the results for a validation test on the model accuracy, as well as results obtained from performing 19 tests on the subjects. All data was generated based on a scaled model of each subject, which was obtained following the procedure detailed in the previous chapter. The Center of Pressure results are taken from the force plate data and converted to the model reference frame. The C3D Gait Extraction Toolbox was used to generate motion files, which were inserted into OpenSim. Subsequently, the Inverse Kinematics, Inverse Dynamics and Analyze (Body Kinematics) tools were used to calculate the Center of Mass. Post processing of data for plot generation was done using Matlab and Excel.

5.1 Model Verification

The first step is to verify that the model used can accurately determine the position of the center of mass. To test this, we have taken data from a subject standing in static pose, and determined the CoP using force platform data as explained above. Afterwards we compute the CoM of the subject in the quiet standing using both Kudohs Gaussian mass distribution model and Winters segmented mass distribution model. It should be noted that both these models were not implemented in OpenSim, and thus the determination of the CoM followed the methods detailed in those publications. Since equilibrium equations dictate that in static standing, the vertical projection of the CoM corresponds to the CoP. By comparing the difference between the vertical projection of the CoM and the CoP of the subject calculated by all three methods, we can determine the error in CoM estimation for the three distinct models.

Figure 5.1 shows the comparison of the error in CoM estimation for the models tested. To analyze these results, a two tailed parametric t-test was performed, with a 95% significance interval. Results show that the OpenSim model has a significantly lower ($p < 0.05$) estimation error than both the Winter and Kudoh approaches. Additionally, Winter's model has a slightly lower estimation error than Kudoh's model.

Figure 5.1: Comparison of CoP - CoM estimation error in standing test for three distinct models. Our OpenSim model has a significantly lower estimation error than both Winter's segmented mass distribution model and Kudoh's Gaussian distribution model.

5.2 Passive Tests

As was described above tests classified under the “passive” category are when the subject is under no external forces, and no voluntary motion is undertaken. All motion is the direct result of compensatory actions. For these reasons, it is expected that the error between the ground component of the center of mass and the center of pressure be very small.

Figure 5.2 shows the process of data acquisition, from raw marker data into the OpenSim model implementation, for a few of the tests in the Passive category. The top panels show the motion capture system detecting the subject's markers, as well as the ground reaction forces for each movement. The bottom panels show the same tests after the data has been input to the scaled model in OpenSim. It can be seen that the model accurately portrays the posture of the subject for wide range of motions.

Figure 5.2: Example of raw marker data (upper panels) and OpenSim model (lower panels) for tests in the passive category. Tests are in order: Quiet standing, standing with feet close together, standing with one foot in front, and standing on one leg respectively.

5.2.1 Static Standing

Figure 5.3 shows a comparison of the measured center of pressure (CoP) and the planar component of center of mass (CoM) calculated using the OpenSim modelling for one subject in the quiet standing test. For both the sagittal plane and the frontal plane these two parameters match for all trials. Also, the displacement of the CoP and CoM is also much higher on the sagittal plane for the case of quiet standing, as this is the preferred direction of sway.

Figure 5.3: Measured CoP and Calculated CoM of one subject in a Static Standing test. Comparison between the two parameters for each of the three trials in the sagittal and frontal planes. The bottom two graphs display the associated error for each trial. Even though there are variations in the CoP (solid lines), the CoM (dotted lines) varies accordingly.

However, the calculated error between CoP and CoM remains similar between both planes, as is shown in Figure 5.4. It is relevant to note that the CoP-CoM error remains relatively stable throughout the entire trial, even though there is noticeable CoP and CoM displacement in the sagittal plane.

Figure 5.4: Measured CoP and Calculated CoM error of one subject in a Static Standing test. The error is constant throughout the trial.

In order to verify if the chosen subject is a valid representative of the population, the two planar components were combined and the absolute error is calculated and shown in Figure 5.5. As can be seen, the subject displayed has an average error that is representative of the entire cohort. Additionally, the average error for the entire population remains stable at around 3 millimeters.

Figure 5.5: Absolute error for Static standing for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. The error remains constant throughout the trial.

For the remainder of the tests in the passive category, only the average error for the population is displayed.

5.2.2 Standing with eyes closed

In figure 5.6 we can see that closing the eyes leads to a sensory deprivation which increases the overall error and variability. In this test, the population variability is also more noticeable.

Figure 5.6: Absolute error for Standing with Eyes Closed for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. The reduction in sensory feedback causes the CoP-CoM error to increase on average, and also to fluctuate more significantly.

5.2.3 Standing with feet together

Figures 5.7 and 5.8 show the effect of reduction in the base of support by having both feet together or one foot in front of the other. There is a noticeable increase in the variability of the error, even though on average it does not increase significantly. In the case of the standing with one foot in front, it is worth noticing that the large spike in error at the start of the trial corresponds to the act of placing one foot in front of the other. The error then stabilizes as the subject holds the new position. This transitional phase will be explored further.

Figure 5.7: Absolute error for Standing with Feet Together for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a noticeable variation in the CoP-CoM error associated with an increase in sway.

5.2.4 Standing with one foot in front

Figure 5.8: Absolute error for Standing with one foot in front for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a significant increase in error at the transition phase at the start of the trial, which stabilizes afterwards.

5.2.5 Standing on one leg

An extreme case of base of support reduction is the case of standing on one leg, shown in Figure 5.9. Once again there is a significant increase in error at the beginning and end of the trial, associated with lifting and dropping the leg. This error however stabilizes for the leg up period. It is interesting to note that when comparing data from the left and right legs, even though there is a more noticeable sway when the right leg is up, this does not translate in a significant increase in the error. This small difference could be explained by the fact that the subjects are right handed, and may be more noticeable with subjects that have a preferred or injured side.

Figure 5.9: Absolute error for Standing on One Leg for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. Data displayed for left leg up (L) and right leg up (R).

5.3 Active Tests

Tests in the *active* category include those where there is voluntary motion, or when motion is caused by response to externally applied forces. Based on what was seen in the passive category, it is expected that the error in the tests that include motion will be higher during the transition phases. Figure 5.10 shows the process of data acquisition for the alternate stair touch test. The upper panels display raw marker data and ground reaction forces for different stages of the test. The lower panels show the visualization of the OpenSim model for the equivalent time frames.

5.3.1 Alternate Stair Touch

Figure 5.10: Different stages of the Alternate Stair Touch, seen in the upper panels as raw marker data and the lower panels after implementation in OpenSim.

Figure 5.11: Measured CoP and Calculated CoM of one subject in an Alternate Stair Touch test: Comparison between the two parameters for each of the three trials in the sagittal and frontal planes. .

Figure 5.11 shows a comparison of the measured center of pressure (CoP) and the planar component of center of mass (CoM) calculated using the OpenSim modelling for one subject in the Alternate Stair Touch test. As can be seen for both sagittal plane and the frontal plane there is a significant gap in between these two parameters every time there is a weight shift. While the CoM transition is smooth, the CoP has a more abrupt variation. However, in the stages between steps the error is greatly reduced. This effect is particularly noticeable in the sagittal plane which also corresponds to the main plane of movement (Figure 5.12).

Figure 5.12: Measured CoP and Calculated CoM of one subject in an Alternate Stair Touch test: CoP-CoM error for each trial.

The two planar components of CoM and CoP are then combined for an absolute error which is representative of the total population in Figure 5.13. It is clearly seen that the error starts off small, however as soon as movement is initiated there is a significant increase in error that remains until movement stops at the end of the trial.

Figure 5.13: Absolute Error for Alternate Stair Touch for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.3.2 Sit to Stand to Sit

Figure 5.14 displays the average error for all subjects in the sit to stand to sit trial. Note that in this test, the tests were normalized to the frame in which the body is no longer in contact with the chair. This is due to the fact that there is no force platform beneath the chair, thus the CoP can only be calculated correctly once all the weight of the subject is on the force platforms. As expected, the CoP-CoM error is significantly higher at the start and end of the trial, which correspond to the transitional period of standing and sitting respectively. However, the error assumes a lower value during the stable “upright” stance.

Figure 5.14: Absolute Error for Sit to Stand to Sit for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject. There is a noticeably higher error at the transitional phase at the beginning and end of the test, corresponding to the standing and sitting component. The error is significantly reduced during the upright stance phase.

5.3.3 Functional Reach

The functional reach forward test reveals the same pattern. In Figure 5.15, we can see that as soon as the movement begins the error increases, and remains until the subject has returned to a stable upright position, at which point it is reduced. Curiously, the lateral reach test has a slightly different behavior, shown in Figure 5.16. Although there is a noticeable increase during the early stages of the test, corresponding to the lunging component, there is a noticeable drop in error once maximum distance is achieved. This error remains smaller than the original error for the remainder of the test. This can be explained due to a slower recovery from maximum extension by the subject when compared to a lunge phase. Additionally, no noticeable difference was found between

the lunging left or right, however injury or loss of function in one side would affect these results.

Figure 5.15: Absolute Error for Functional Reach Forward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

Figure 5.16: Absolute Error for Functional Reach Lateral for all 20 subjects, averaged for all trials. Data shown for reaching Left and Right. Comparison of the average error of the population and the average error of the subject. There is a steady increase during the lunge phase and a noticeable drop at maximum extension, followed by a constant slower error during recovery phase.

5.3.4 Pick up from floor

In the case of picking up an object from the floor, shown in Figure 5.17 this departure from stability leads to a rapid increase in error during the beginning of the drop phase, followed by a smaller error during the drop and another peak just before the rise. Once the subject is ascending, the error steadily reduces until the stable upright position is reached.

Figure 5.17: Absolute Error for Pick up from floor for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.3.5 Look Over Shoulder

For the look over shoulder test, shown in Figure 5.18, there are two peaks which correspond to the motion over each shoulder. These rapid increases correspond to the weight shift required to rotate the torso. The valley in between is due to the brief pause that the subject performs in the centered position.

Figure 5.18: Absolute Error for Look Behind Shoulder for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.3.6 Rise to Toes

In the case of the rise to toes, shown in figure 5.19, the results are a more extreme version of what was verified previously. The transition phase has a rapid increase in the error, however due to the unstable nature of the position the error has a large variability during the trial.

Figure 5.19: Absolute Error for Rise to Toes for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.3.7 In place response

The in place response tests and compensatory response tests (Figures 5.20 to 5.23, refer to a different type of scenario, as there is now an external perturbation that affects the results. It can clearly be seen that there is a noticeable spike in error followed by a sudden drop due to the release of pressure.

Figure 5.20: Absolute Error for In Place Response - Forward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

Figure 5.21: Absolute Error for In Place Response - Backward for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.3.8 Compensatory Step

Figure 5.22: Absolute Error for Forward Compensatory Step for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

Figure 5.23: Absolute Error for Backward Compensatory Step for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.4 Walking Tests

Tests in the *walking* category were performed to differentiate the transitional phase that was emphasized above and the stable phase present in quiet standing. Since the body is in continuous motion over the measured time frame, there is no significant acceleration of the CoM. An important note to take into account for these tests is the limitation of number of force plates. Considering that it is only possible to measure the CoP when the patient is in contact with the force plates, the amount of data captured is defined by the number of steps covered by force plates. In case of motion capture, the available work area is much larger thus enabling a greater time span to be captured. For the following tests, CoP was only determined during one gait cycle, defined by the events of left foot Toe Off, left foot Heel Strike, right foot Toe Off, right foot Heel Strike. These events are depicted in Figure 5.24 for both raw data capture and OpenSim model calculations.

Figure 5.24: Gait cycle studied in walking tests. Events are measured in the following order: left foot Toe Off, left foot Heel Strike, right foot Toe Off, right foot Heel Strike.

5.4.1 Walking with Head Turns

Figure 5.25 shows a comparison of the measured center of pressure (CoP) and the planar component of center of mass (CoM) calculated using the OpenSim modelling for one subject in the Walking with Head Turns test. In gait, the Center of mass pattern consists of motion in both planes: a cyclic sinusoidal motion in the frontal plane and a steady linear motion in the sagittal plane. Contrastingly, the CoP behaves in a more abrupt way, with rapid changes when a foot strike occurs. This is revealed in Figure 5.26, where the error linearly changes until foot strike occurs, which creates a significant variation in error. Combining components from both planes (Figure 5.27) leads to a local minimum error halfway through the swing phase and a local maximum at foot strike.

Figure 5.25: Measured CoP and Calculated CoM of one subject in a Walking with Head Turns test. Comparison between the two parameters for each of the three trials in the sagittal and frontal planes.

Figure 5.26: Measured CoP and Calculated CoM of one subject in a Walking with Head Turns test. Error for each trial.

Figure 5.27: Absolute Error for Walking with Head Turns test for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.4.2 Walking over Level Surface

Figures 5.28 and 5.29 display the average errors for the subject and the entire population for Walking in Level Surface and Stepping over obstacle. As expected, walking over level surface has a slightly lower error associated, while the act of stepping over an obstacle reduces the valley that would be created by the swing phase.

Figure 5.28: Absolute Error for Walking over Level Surface for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.4.3 Stepping over Obstacle

Figure 5.29: Absolute Error for Step over Obstacle for all 20 subjects, averaged for all trials. Comparison of the average error of the population and the average error of the subject.

5.5 Discussion

We have shown that our method is more accurate at determining the CoM than the other two methods, having a mean estimation error that is significantly lower. This can be attributed to OpenSim's ability to incorporate anatomical measurements in the determination of the base model, which is then scaled to the subjects specific proportions. As expected, Winters model also has a lower error than Kudohs model, likely due to the anatomical correlations that are present. While Kudoh uses a cylinder approximation to infer segmental center of mass position, Winter defines the location of the segmental center of mass based on the position of anatomical landmarks. Additionally, having a mass distribution based on clinical data is more accurate than an approximation based on statistical distribution.

In the previous sections we have shown that there is a clear variation in the error between the CoP and the CoM for different types of scenarios. The overall average error for each test is shown in Figure 5.30 and Table 5.1. As can be seen, there is a clear relationship between the range of motion and the error between the CoP and CoM. Tests from the passive category (yellow) have significantly lower average errors than tests in the active (orange) and walking (green) categories.

Table 5.1: Average mean error, standard deviation and standard error of the mean (SEM) for all tests shown, separated by categories: passive, active, walking.

Test	Mean	StdDev	SEM
Quiet standing	3.088E-03	5.451E-04	6.276E-06
Stand eyes closed	4.114E-03	1.284E-03	2.024E-05
Stand with feet together	3.869E-03	1.605E-03	2.519E-05
Stand one foot in front	1.399E-02	7.393E-03	8.438E-05
Stand on one leg (L)	7.642E-03	8.134E-03	1.109E-04
Stand on one leg (R)	7.293E-03	6.819E-03	9.552E-05
Sit to stand to sit	1.617E-02	1.581E-02	4.146E-04
Reach forward	1.383E-02	8.925E-03	1.708E-04
Pick up object from floor	9.608E-03	6.883E-03	1.677E-04
Look over shoulder	9.646E-03	6.662E-03	1.211E-04
Alternate stair touch	3.810E-02	1.768E-02	2.457E-04
Lateral reach	2.313E-02	9.413E-03	1.830E-04
Rise to toes	1.104E-02	8.966E-03	1.654E-04
In place response forward	1.685E-02	1.174E-02	2.354E-04
In place response backward	4.331E-02	3.440E-02	6.899E-04
Compensatory step forward	6.738E-02	4.679E-02	1.035E-03
Compensatory step backward	8.683E-02	5.718E-02	1.619E-03
Walk on level surface	1.072E-01	3.720E-02	1.953E-03
Walk with head turns	1.008E-01	3.716E-02	1.901E-03
Step over obstacle	1.074E-01	5.207E-02	2.279E-03

Figure 5.30: Average error for all tests shown, separated by categories: in orange passive; in yellow active; and in green walking. In general, the walking and active categories have a significantly increased error when compared to the passive category.

As was described in the previous sections, tests classified under the “passive” category are when the subject is under no external forces, and no voluntary motion is

undertaken. All motion is the direct result of compensatory actions. As such, since the weight is a constant, the balance of forces predicts that the ground component of CoM and the CoP should match. This is in concordance with the theory of static stability. It can be seen from the results shown that in quasi-static conditions, the CoP is a sufficiently accurate descriptor of CoM, since no significant differences were found. These results are in concordance with others found in literature where quiet standing is the pose being studied.

Tests in the “active” category include those where there is voluntary motion, or when motion is caused by response to externally applied forces. Contrary to the tests in the “passive” category, the results show a significant difference between the CoP and CoM during the transitional phases. It is important to note that after the transition phase the error gets reduced when the subject assumes a new “stable” position, even though the corresponding sway is sometimes higher than when the subject was in the original pose. Additionally, the Cop always has a larger displacement than the Com which is concurrent with the ZMP Concept of the trolley under a mass. These results indicate that there is a correlation between the velocity of the CoM and the position of the CoP.

Tests in the “walking” category were performed to differentiate the transitional phase between quiet standing (described above) and continuous motion. In these tests the CoP-CoM error was significantly higher than those in both the passive and active categories. This is also concurrent with our previous notion that the CoM velocity, which is significantly higher in the tests of this category, is directly correlated with the position of the center of pressure.

6 Conclusions

Balance disorders are a common problem in the aging society we live in. Falling has been found to be the number one reason for hospital emergency visits for elderly people in Singapore, and the associated financial and social burden is high. Elderly people are at a risk of falling in their everyday lives, and regular monitoring is required. Additionally, patients that suffer from stroke are susceptible to falling because the central nervous system is unable to interpret the signals that it receives and act fast enough. For them, the act of standing upright is an exercise that often requires months of physical rehabilitation. Identifying the nature of the disability is the first step in the process, and physicians and therapists are looking for better tools to help them assess balance problems. These tools should be simple to use and understand and most importantly the results should be clear and independent of the person doing the evaluation. The current process of evaluating balance in a clinical environment consists of widespread use of clinical scales. These tests are simple and ease of use, with quick results that are fairly easy to interpret. The problems arise when one analyzes the method of evaluation with a critical eye. The subjectivity of the evaluation, leading to intra- and inter-rater variability, as well as non-linearity in the grading system has cast doubt on the validity of these methods. Both the therapists confidence and the patients motivation can be affected, which can hamper the effectiveness of the rehabilitation plan.

In the literature, researchers have attempted to tackle the problem of balance assessment by one of two ways: a clinical approach where focus is on developing new clinical scales that reduce subjectivity and provide more details about the underlying problem; and a technical approach, where researchers attempt to determine measurable parameters adequate to classify the patients risk of falling. Many systems have been developed that measure biometric properties of the subjects. Among them, force platform posturography is the most commonly used method to study the dynamics of balance and gait. Posturography measures the variation of the Center of Pressure (COP) of the subject under varied conditions, and is divided into two categories. Static posturography measures body sway when the subject is in unperturbed stance, while dynamic posturography measures the effect of external perturbations such as applied forces, change in visual surrounding or movement of the base of support. A number of com-

mercially available posturography systems exist, and all of them consider the COP excursion as the measurement of static and dynamic stability. However, many restrictions associated with these systems have made their acceptance into the clinical environment limited.

In this work we have presented an alternative methodology that provides a wider spread of data with more significant information that can be used to further improve the patient rehabilitation process. Using a combination of force plate and motion capture, we are able to extract ground reaction forces and motion capture kinematics from subjects performing tests from clinical scales. This information has been inserted into a kinematic model of the human: a multi-segmented 31-DOF articulated linkage. This model has been implemented in OpenSim, an open source modelling and simulation tool developed at Stanford University. This platform allows us to perform a number of simulations by synchronizing the motion capture data with the ground reaction forces. One of the key aspects of using this tool is the possibility of scaling the size and properties of the generic model to the experimental measurements of each subject under trial. Based on the performed motions and applied forces, we have calculated the inverse kinematics and dynamics, as well as the position, velocity and acceleration of the CoM, as well as the position of the global CoP based on multiple force plates.

While force platform posturography has been extensively used to study stability in the past, the combination of center of mass and center of pressure analysis has rarely been addressed. In those few cases, calculating the center of mass was done using very simple inverted pendulum models [101–105], or in some cases no model was used. As was shown in the literature review, a number of works have shown that simple inverted pendulum models and even double inverted pendulum models are insufficient in capturing the dynamics of postural sway, as the contributions of the ankle, hip and knee are all required for accurate description of movement [117]. The full body model used in this work allows us to more accurately calculate the dynamics of the center of mass in a wide range of motions and poses.

The static standing test provides a good verification method for the proposed model. The condition of static stability states that the ground component of the center of mass will coincide with the center of pressure when the subject is in quiet stance. Our results show that this condition is met with a relatively small error associated. To validate the approach taken, we have compared the CoP-CoM estimation error for static standing for a subject using three distinct models. Statistical analysis has shown that the method

described in this work provides a significant improvement in determining the CoM of a subject over other models used in the literature. Additionally, we would like to do a small test to validate the applicability of the method for clinical use. Unfortunately, this step was not achieved within the time constraints of this work due to limited availability of test subjects and associated clinical data.

We have focused on the relationship between the center of pressure and the center of mass, as both the literature and our preliminary results have shown to be the most relevant parameters to define stability. In this work we have discovered that the ground component of the center of mass corresponds to the center of pressure in static or quasi-static conditions, which is consistent with the static definition of balance. However, when transitions between poses are required these parameters diverge significantly. This is highlighted further when the tests are grouped into categories. It is clearly seen that as the intensity of motion increases, the error between CoP and CoM also increase. These results have shown that there is a clear relationship between the position and velocity of the CoM and the CoP. This is in concurrence with other works. The results obtained can be interpreted in simplification as mass (CoM) on an inverted pendulum under a cart (ZMP). In order to maintain stability of the CoM the ZMP will vary more significantly under voluntary motion or perturbations. This behavior confirms the theory of dynamic stability which defines that the CoP, as a direct measurement of the ZMP in flat foot conditions, is a main indicator of stability. as it has a larger variability under different conditions.

However, there are many limitations to the exclusive measurement of the CoP exclusively in the clinical environment, which have also been highlighted by this work. Firstly, the CoP can only be calculated when the subject is over a force platform. This is a critical limitation since many of the clinical tests require movement over a number of steps. This would require a large number of force plates to capture the full range of motion. Motion capture system are less affected by this, since the area of data capture is much larger. Additionally, for tests that require external perturbations, these perturbations will affect the CoP measurement. Having a system that allows determination of stability without these constraints would greatly enhance the applicability of instrumented systems in clinical settings.

6.1 Significance and Contributions

In this work, we have identified the patterns of movement associated with tests performed in clinical balance evaluation. We have categorized them as active, passive or walking, based on the type of motion required. This categorization allows us to look at the tests under the perspective of dynamic or static stability, and the implications that those conditions have on the relationship between the center of mass and center of pressure. This quantitative assessment eliminates the subjectivity of the observer. While other works have delved into the study of clinical scales, they have all focused on improving the inter- and intra-rater variability. This work is the first to analyze the tests from an engineering perspective.

We have developed a novel method for determining the center of mass and center of pressure of a subject performing clinical balance evaluation tests. This method combines force platform posturography and motion capture to generate accurate measurements. While the required complexity of kinematic models used to describe human balance is still a topic of much discussion, it has been shown that simple inverted pendulum models used in the majority of the literature [118] are unable to capture the required dynamics.

In this work, we have implemented a full body model that fully incorporates the dynamics of balance and motion. This 31-DOF model can be scaled to be subject-specific, and enables the determination of the center of mass in a wide array of motions and poses. This model can also be used in future experiments to analyze other components such as joint kinematics and dynamics. We have shown that our model and method present a significant improvement over existing models in calculating the center of mass of a subject. While other, simpler models can and have been used to study balance, we have shown that a full body model is required to accurately represent the dynamics of even simple motions. This result contributes one more step towards identifying the exact complexity and scope of models used to study and evaluate balance.

To the authors knowledge, no other study has attempted to find correlations between multiple kinematic and dynamic parameters and the respective performance score in clinical scales. While a number of studies have studied the relationship between postural sway and commonly used clinical tests [34–38]. All of these have found little or no correlation between COP excursion and the test chosen. However, the only parameter tested is the center of pressure. Contrastingly, in our results we have shown that there is

a significant correlation between the type of test performed and the CoP-CoM distance.

The method developed in this work enables future analysis of the CoP and CoM of post stroke patients performing clinical balance evaluation tests. This will allow a quantifiable measurement of these parameters which will be compared to their subjective performance in the clinical scales.

This quantitative method has the potential to revamp the clinical balance rehabilitation assessment methods currently in use. Removing the subjectivity of the assessment leads to greater confidence in the results, as well as a greater sensitivity to changes in performance. The patient is the overall benefactor, as the therapist can modify the rehabilitation program according to their specific needs.

6.2 Future Work

Although the results shown in this work show a great potential, the methodology described in the previous chapters still needs to be evaluated in an environment where loss of balance is more likely to occur. In the near future, we are looking to perform a small scale test to help validate the methodology presented. This test will include three different subjects with noticeable variability in postural control. The subjects' performance would be acquired using a force and motion capture system, with the assistance of a trained physiotherapist who would also be giving a simultaneous independent qualitative assessment. This will allow us to correlate the subjective assessment with quantitative evaluation provided by the method. The results of those tests will confirm if the approach taken here is applicable to a clinical test. The ultimate goal of this strategy is to develop a direct correspondence between performance on clinical tests and measurable parameters such as the CoP and CoM. Unfortunately this final validation was not achievable due to availability of clinical data.

With regards to the methodology, the process might be easier to apply if we developed a user interface that processed all the data and provided information directly to the user. This would imply establishing connections between the force platform and video capture system, generating a scaled OpenSim model for the subject and using it to perform the necessary calculations. OpenSim already provides support for this via its Application Program Interface. We feel that this step would be important once the clinical trial has shed light on the most relevant parameters to show. We would also attempt to determine a function that estimates the fall risk of a subject as a function of the CoP and CoM position, velocity and acceleration. While other works have done some of this

in the past, they have resorted to over simplified methods of calculating the CoM, thus it would be relevant to compare their results with the ones achieved with our method. To increase the amount of data at our disposal, we would like to increase the number of force platforms to have a wider range of data points for gait calculations. Regarding the full body modelling, it would be interesting to determine a way to calculate the center of pressure without requiring a use of force platform. Recent studies have started to delve into this area using Kinect to determine the ZMP of subjects without ground reaction forces [139].

6.2.1 Clinical Trial

In addition, in the future we hope to undertake a small scale clinical trial with subjects that are in post-stroke rehabilitation. This trial would include a cohort of 10 subjects: 5 male and 5 female. As a control, 10 non-stroke subjects with similar sex and age distribution would also be tested for a direct comparison. Each subject would be evaluated by a physiotherapist for any non stroke related medical conditions that might cause deficiency in ability to balance prior to be accepted for the study. All subjects must be able to stand unassisted for a period of 30s and be able to walk with minimal assistance for 8 meters. The objective of these selection criteria is to limit by all reasonable means the number of possible variables that may affect the measured parameters. Additionally the study will focus on subjects that have suffered a stroke no more than 12 weeks beforehand, as this has been shown to be the period of time where recovery is most evident and changes in the results would be more apparent. They would also be enrolled in a balance recovery program at their local hospital or clinic. Subjects in the control cohort would only require one session to measure the base level of performance, while subjects in the post-stroke cohort would be evaluated four times over a period of 12 weeks (measurements would be made at weeks 1, 4, 8 and 12). At all these evaluating sessions a physiotherapist would be overseeing the measurement process and patient interaction. For tests requiring external force application, the physiotherapist would administer the test directly. The objectives of this trial would be to accompany the cohort over the rehabilitation period, comparing the measured performance with the qualitative assessment provided by the therapist. We would look at relationship between the CoP and the CoM of the subjects and identify any correlations between these and the clinical scales values. The expected results for this trial would be a range of values for the variability of CoP and CoM that more accurately describe the behavior of a patient pre-

scribed with a given clinical scale score. Depending on the results, we hypothesize that with our methodology it might be possible to limit the amount of tests currently required to give an accurate and reliable estimate of the fall risk of the patient.

6.3 Summary

Several studies have begun to address the relationship between the CoM and CoP, but up to now have done so in an overly simplistic fashion. The methodology described in this work allows a more in depth analysis of these parameters for clinical studies. Our results show that there is a significant difference in the CoP-CoM distance in the presence of perturbations or voluntary motion, which is concurrent with the criterion of dynamic vs static stability. However we would like to expand on the notion by studying the relationship between the CoP and the velocity and acceleration of the CoM. The method described provides a system for determining the center of mass of a subject under motion capture. We have shown that it is significantly more accurate than other models present in the literature.

The system developed in this study has shown that there is merit in studying the relationship between the CoP and CoM in a clinical environment. We have identified that studying the CoP or CoM alone do not provide a full picture of stability even when the subjects are healthy young adults. We expect that the results would be significantly more explicit when the subjects are balance impaired. The next step is to identify the variability of these parameters with subjects at different levels of balance ability. Establishing this correlation has the potential to greatly improve the way in which balance assessment is performed.

This method has laid the foundation for objective evaluation of patient capabilities, which will enable tracking the patient progress over the course of the rehabilitation period. This will in turn reflect on patient motivation, as the improvement can be seen through measurable parameters. Getting the patient motivated and involved in the rehabilitation process is a key step in improving recovery. While the experience and knowledge of the therapist are invaluable assets that should take priority in determining the treatment program, providing them with accurate, quantifiable data will allow them to make better informed decisions.

Additional investigations would be required in order to further understand the impact of these parameters within a clinical environment, however we feel this is a productive avenue for further research.

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A Appendix

This appendix includes detailed information regarding the distribution of clinical scales in the categories described in chapter 3. Appendix A.1 displays the categorization of all tests in all four clinical scales studied. Appendixes A.2, A.3 and A.4 are a comparative analysis of observable vs measurable parameters that are used in evaluating the performance. In appendix B we include transcripts of the clinical scales: the Berg Balance Scale, the Balance Evaluation Systems Test, Tinetti Balance Assessment Tool and the Functional Gait Assessment. These include the tests that should be performed, as well as details on how to guide the subject and instructions on how to score based on performance criteria.

A.1 Categorization of Tests within Clinical Scales

BESTest	
Biomechanical constraints	1 Base of support
	2 CoM alignment
	3 Ankle strenght and range
	4 Hip/trunk strenght and range
	5 Sit on floor and stand up
Stability limits / Verticality	6 Verticality and lateral lean
	7 Functional reach forward
	8 Functional reach lateral
	9 Sit on floor and stand
Anticipatory postural adjustments	10 Rise to toes
	11 Stand on one leg
	12 Alternate stair touching
	13 Standing arm raise
Postural responses	14 In place response - forward
	15 In place response - backward
	16 Compensatory stepping correction - F
	17 Compensatory stepping correction - B
Sensory integration	18 Compensatory stepping correction - L
	Modified CTSIB:
	19 Eyes open firm surface
	20 Eyes closed firm surface
	21 Eyes open foam surface
	22 Eyes closed foam surface
Stability in gait	23 Incline - eyes closed
	24 Gait - level surface
	25 Change in speed
	26 Walk with head turns
	27 Walk with pivot turns
	28 Step over obstacle
	29 Timed get up and go test
	30 Timed get up and go test with dual task

BBS	
1	Sit to stand
2	Stand unsupported
3	Sitting with back unsupported
4	Stand to sit
5	Transfers
6	Stand unsupported eyes closed
7	Stand unsupported feet together
8	Reach forward
9	Pick object from floor while standing
10	Look over shoulder while standing
11	Turn 360
12	Alternate stepping on stool
13	Stand unsupported one foot in front
14	Stand on one leg

Functional Gait Assessment	
1	Gait level surface
2	Change in gait speed
3	Gait with horizontal head turns
4	Gait with vertical head turns
5	Gait and pivot turn
6	Step over obstacle
7	Gait with narrow base of support
8	Gait with eyes closed
9	Ambulating backwards
10	Steps

POMA	
Balance	1 Sitting balance
	2 Arise
	3 Attempts to arise
	4 Immediate standing balance
	5 Standing balance
	6 Nudge
	7 Eyes closed
	8 Turn 360
	9 Sit down
Gait	10 Initiation of gait
	11 Step length and height
	a Right swing foot
	b Left swing foot
	12 Step symmetry
	13 Step continuity
	14 Path deviation
	15 Trunk
	16 Walk stance

Color Code	
	PRE-EXISTING
	PASSIVE
	ACTIVE
	GAIT

A.2 Passive Tests

Passive				
Clinical Perspective		Eng Perspective		
Name of Test	Clinical Measurements	Forces Measured	Motions Measured	Instrumentation
CoM alignment	Antero-posterior, mediolateral CoM alignment and segmental postural alignment	Centre of pressure - extract COM position	Static posture (motion of body in static balance)	Force platform Image capturing for segment kinematics?
Verticality and lateral lean	Patient returns to vertical smoothly without overshooting or undershooting.	Centre of pressure - extract COM variations	Lean - variation of CoM from center to max and return	Force platform Image capturing
Functional reach forward	Maximum horizontal distance reached	Centre of pressure - extract COM variation min to max	Horizontal distance forward	Force platform Image capturing
Functional reach lateral	Maximum horizontal distance reached	Centre of pressure - extract COM variation min to max	Horizontal distance lateral	Force platform Image capturing
Stand eyes open - firm surface	Degree of sway	Centre of pressure - extract COM variations, compare to trial with feet at shoulder width	Variation of CoM	Force Platform Image capturing
Stand eyes closed - firm surface	Degree of sway	Centre of pressure - extract COM variations	Variation of CoM	Force Platform
Stand eyes open - foam surface	Degree of sway	Centre of pressure - extract COM variations	Variation of CoM	Force Platform
Stand eyes closed - foam surface	Degree of sway	Centre of pressure - extract COM variations	Variation of CoM	Force Platform
Stand eyes closed - incline	Compare sway to standing on level ground, notice alignment to vertical	Centre of pressure - extract COM variations	Vertical alignment	Force Platform imaging for vertical alignment
Stand - firm surface with visual conflict dome or sway-referenced visual surround	Degree of sway	Centre of pressure - extract COM variations	Relative motion to visual surround	Force platform, rotating surrounding
Stand - compliant surface w visual conflict dome or sway-referenced visual surround	Degree of sway	Centre of pressure - extract COM variations	Relative motion to visual surround	Force platform, rotating surrounding
Sitting with back unsupported	Measure level of supervision required and time	Force used by legs/thighs/ on 2 bases of support: gluteus and feet	Motion of trunk/head	Two distinct pressure sensors for each BoS. Video analysis to analyze segmental variations over time.
Stand on one leg	Ability to lift independently Measure time	Pressure on foot - COM	Motion of trunk	Force Platform
Look over shoulder while standing	Ability to weight shift for trunk rotation	Pressure on each foot	-	Force Platform, Image capturing
Stand unsupported one foot in front	Ability to place foot ahead and hold	Pressure on each foot - COM variations based on reduced base of support	-	Force Platform

A.3 Active Tests

Active				
Clinical Perspective		Eng Perspective		
Name of Test	Clinical Measurements	Forces Measured	Motions Measured	Instrumentation
rise to toes	stability and height, compare force used when holding hands	force used with hands for support, variation of CoP	Ankle elevation	Force sensor on support bar. Motion capture
alternate stair touching	number of successful touches and time required to complete 8 touches	pressure on supporting foot. Extract varying COM for each step	leg kinematics	force platform on supporting foot, force sensor on step. Motion capture
standing arm raise	stability, use of step to maintain balance, ability to lift quickly, maximum attainable angle	COP - COM	trunk / arm kinematics	force platform. motion capture
in place response - forward	stability with ankles, arms or hips. Use of stepping to recover	pressure on feet - variation at release pressure exerted by patient	COM acceleration upon release, ankle and upper body kinematics	force platform. segmental body imaging.
in place response - backward	stability with ankles, arms or hips. Use of stepping to recover	pressure on feet - variation at release pressure exerted by patient	COM acceleration upon release, ankle and body kinematics	force platform segmental body imaging
compensatory stepping correction - F	independent recovery, length and number of steps to maintain balance	maximum CoP variation from vertical, response to release (stabilization)	number and length of steps, COM acceleration and lower body kinematics	force sensors to detect stepping, motion capture
compensatory stepping correction - B	independent recovery, length and number of steps to maintain balance	maximum CoP variation from vertical, response to release (stabilization)	number and length of steps, COM acceleration and lower body kinematics	force sensors to detect stepping, motion capture
compensatory stepping correction - L	independent recovery, number of steps to maintain balance, length/width of steps	maximum CoP variation from vertical, response to release (stabilization)	number and length of steps, width of steps, COM acceleration and lower body kinematics	force sensors to detect stepping, motion capture
sit to stand	independent stabilization, use of hands	variation of CoP with change of BoS	body kinematics, arm motion	force platform. motion capture for trunk/leg motion
stand to sit	controlled descent, use of hands	variation of CoP with change of BoS	body kinematics, arm motion COM acceleration	force platform. motion capture for trunk/leg motion
transfers	ability to transfer with minimal use of hands	variation of CoP with change of BoS	COM velocity/acceleration arm motion	force platform. motion capture for trunk/leg motion
pick object from floor while standing	ability to reach object and return to standing position	CoP	variation of COM along motion, stabilization at end	force platform, visual analysis of maximum reach
turn 360	stability while turning, required time to complete	CoP	body sway while turning	force platform, motion capture

A.4 Walking Tests

Walking				
Clinical Perspective		Eng Perspective		
Name of Test	Measurements	Forces Measured	Motions Measured	Instrumentation
gait - level surface	ability to walk distance, time required	foot pressure	body kinematics	pressure pads on walkway/treadmill, visual system
change in speed	balance during transitions	foot pressure	body kinematics, COM acceleration	pressure pads on walkway/treadmill, visual system
step over obstacle	change of speed, maintenance of balance - hesitation, short steps or obstacle touch	foot pressure	body kinematics, COM acceleration	pressure pads on walkway, visual system
timed get up and go test	speed and evidence of imbalance	foot pressure, COM	body kinematics	pressure pads on walkway, visual system
timed get up and go test with dual task	differences in rate/accuracy of counting backwards from sitting and standing, change of speed	foot pressure, COM	body kinematics	pressure pads on walkway, visual system
initiation of gait	hesitancy or multiple attempts to start	COM	initial acceleration	pressure pads on walkway, visual system
step length and height (right and left)	passes stance foot with step, swing foot clears floor completely	foot pressure	height and length of step	pressure pads on walkway/treadmill, visual system
step symmetry	right and left step length are equal	foot pressure	height and length of step	pressure pads on walkway/treadmill, visual system
step continuity	stopping or discontinuity between steps	foot pressure		pressure pads on walkway/treadmill
path deviation	deviation from path, use of walking aid	foot pressure		pressure pads on walkway/treadmill
trunk	sway, use of walking aid, flexion of knees or arms spread	foot pressure	body kinematics	pressure pads on walkway/treadmill, visual system
walk stance	distance between heels while walking	foot pressure	lower leg kinematics	pressure pads on walkway/treadmill, visual system
gait with horizontal head turns	head turns smoothly, change in gait, deviation from path	foot pressure	upper body kinematics, head rotation	pressure pads on walkway, visual system
gait with vertical head turns	head turns smoothly, change in gait, deviation from path	foot pressure	upper body kinematics, head rotation	pressure pads on walkway, visual system
gait and pivot turn	speed of turn, loss of balance	foot pressure, CoM variation during rotation	upper body kinematics, head rotation	pressure pads on walkway, visual system
gait with narrow base of support	able to ambulate 10 steps, staggering	foot pressure, CoM variation	upper body kinematics	pressure pads on walkway, visual system
gait with eyes closed	use of assistive device, speed, gait pattern, path deviation	foot pressure, CoM variation	upper body kinematics	pressure pads on walkway, visual system
ambulating backwards	use of assistive device, speed, gait pattern, path deviation	foot pressure, CoM variation	upper body kinematics	pressure pads on walkway, visual system
steps	alternating feet, use of rail	foot pressure, CoM variation, acceleration	upper and lower body kinematics	pressure pad in each step, visual system

B Clinical Scales

The following are direct transcripts of the following Clinical Scales:

- Berg Balance Scale (BBS). Extracted from [21]
- Balance Evaluation Systems Test (BESTest). Extracted from [24]
- Tinetti's Performance Oriented Mobility Assessment (POMA). Extracted from [20]
- Functional Gait Assessment (BESTest). Extracted from [125]

Berg Balance Scale

The Berg Balance Scale (BBS) was developed to measure balance among older people with impairment in balance function by assessing the performance of functional tasks. It is a valid instrument used for evaluation of the effectiveness of interventions and for quantitative descriptions of function in clinical practice and research. The BBS has been evaluated in several reliability studies. *A recent study of the BBS, which was completed in Finland, indicates that a change of eight (8) BBS points is required to reveal a genuine change in function between two assessments among older people who are dependent in ADL and living in residential care facilities.*

Description:

14-item scale designed to measure balance of the older adult in a clinical setting.

Equipment needed: Ruler, two standard chairs (one with arm rests, one without), footstool or step, stopwatch or wristwatch, 15 ft walkway

Completion:

Time: 15-20 minutes

Scoring: A five-point scale, ranging from 0-4. "0" indicates the lowest level of function and "4" the highest level of function. Total Score = 56

Interpretation:

41-56 = low fall risk
21-40 = medium fall risk
0-20 = high fall risk

A change of 8 points is required to reveal a genuine change in function between 2 assessments.

Berg Balance Scale

Name: _____

Date: _____

Location: _____

Rater: _____

ITEM DESCRIPTION

SCORE (0-4)

Sitting to standing	_____
Standing unsupported	_____
Sitting unsupported	_____
Standing to sitting	_____
Transfers	_____
Standing with eyes closed	_____
Standing with feet together	_____
Reaching forward with outstretched arm	_____
Retrieving object from floor	_____
Turning to look behind	_____
Turning 360 degrees	_____
Placing alternate foot on stool	_____
Standing with one foot in front	_____
Standing on one foot	_____

Total _____

GENERAL INSTRUCTIONS

Please document each task and/or give instructions as written. When scoring, please record the lowest response category that applies for each item.

In most items, the subject is asked to maintain a given position for a specific time. Progressively more points are deducted if:

- the time or distance requirements are not met
- the subject's performance warrants supervision
- the subject touches an external support or receives assistance from the examiner

Subject should understand that they must maintain their balance while attempting the tasks. The choices of which leg to stand on or how far to reach are left to the subject. Poor judgment will adversely influence the performance and the scoring.

Equipment required for testing is a stopwatch or watch with a second hand, and a ruler or other indicator of 2, 5, and 10 inches. Chairs used during testing should be a reasonable height. Either a step or a stool of average step height may be used for item # 12.

Berg Balance Scale

SITTING TO STANDING

INSTRUCTIONS: Please stand up. Try not to use your hand for support.

- () 4 able to stand without using hands and stabilize independently
- () 3 able to stand independently using hands
- () 2 able to stand using hands after several tries
- () 1 needs minimal aid to stand or stabilize
- () 0 needs moderate or maximal assist to stand

STANDING UNSUPPORTED

INSTRUCTIONS: Please stand for two minutes without holding on.

- () 4 able to stand safely for 2 minutes
- () 3 able to stand 2 minutes with supervision
- () 2 able to stand 30 seconds unsupported
- () 1 needs several tries to stand 30 seconds unsupported
- () 0 unable to stand 30 seconds unsupported

If a subject is able to stand 2 minutes unsupported, score full points for sitting unsupported. Proceed to item #4.

SITTING WITH BACK UNSUPPORTED BUT FEET SUPPORTED ON FLOOR OR ON A STOOL

INSTRUCTIONS: Please sit with arms folded for 2 minutes.

- () 4 able to sit safely and securely for 2 minutes
- () 3 able to sit 2 minutes under supervision
- () 2 able to sit 30 seconds
- () 1 able to sit 10 seconds
- () 0 unable to sit without support 10 seconds

STANDING TO SITTING

INSTRUCTIONS: Please sit down.

- () 4 sits safely with minimal use of hands
- () 3 controls descent by using hands
- () 2 uses back of legs against chair to control descent
- () 1 sits independently but has uncontrolled descent
- () 0 needs assist to sit

TRANSFERS

INSTRUCTIONS: Arrange chair(s) for pivot transfer. Ask subject to transfer one way toward a seat with armrests and one way toward a seat without armrests. You may use two chairs (one with and one without armrests) or a bed and a chair.

- () 4 able to transfer safely with minor use of hands
- () 3 able to transfer safely definite need of hands
- () 2 able to transfer with verbal cuing and/or supervision
- () 1 needs one person to assist
- () 0 needs two people to assist or supervise to be safe

STANDING UNSUPPORTED WITH EYES CLOSED

INSTRUCTIONS: Please close your eyes and stand still for 10 seconds.

- () 4 able to stand 10 seconds safely
- () 3 able to stand 10 seconds with supervision
- () 2 able to stand 3 seconds
- () 1 unable to keep eyes closed 3 seconds but stays safely
- () 0 needs help to keep from falling

STANDING UNSUPPORTED WITH FEET TOGETHER

INSTRUCTIONS: Place your feet together and stand without holding on.

- () 4 able to place feet together independently and stand 1 minute safely
- () 3 able to place feet together independently and stand 1 minute with supervision
- () 2 able to place feet together independently but unable to hold for 30 seconds
- () 1 needs help to attain position but able to stand 15 seconds feet together
- () 0 needs help to attain position and unable to hold for 15 seconds

Berg Balance Scale continued...

REACHING FORWARD WITH OUTSTRETCHED ARM WHILE STANDING

INSTRUCTIONS: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. (Examiner places a ruler at the end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the fingers reach while the subject is in the most forward lean position. When possible, ask subject to use both arms when reaching to avoid rotation of the trunk.)

- ☐ 4 can reach forward confidently 25 cm (10 inches)
- ☐ 3 can reach forward 12 cm (5 inches)
- ☐ 2 can reach forward 5 cm (2 inches)
- ☐ 1 reaches forward but needs supervision
- ☐ 0 loses balance while trying/requires external support

PICK UP OBJECT FROM THE FLOOR FROM A STANDING POSITION

INSTRUCTIONS: Pick up the shoe/slipper, which is in front of your feet.

- ☐ 4 able to pick up slipper safely and easily
- ☐ 3 able to pick up slipper but needs supervision
- ☐ 2 unable to pick up but reaches 2-5 cm (1-2 inches) from slipper and keeps balance independently
- ☐ 1 unable to pick up and needs supervision while trying
- ☐ 0 unable to try/needs assist to keep from losing balance or falling

TURNING TO LOOK BEHIND OVER LEFT AND RIGHT SHOULDERS WHILE STANDING

INSTRUCTIONS: Turn to look directly behind you over toward the left shoulder. Repeat to the right. (Examiner may pick an object to look at directly behind the subject to encourage a better twist turn.)

- ☐ 4 looks behind from both sides and weight shifts well
- ☐ 3 looks behind one side only other side shows less weight shift
- ☐ 2 turns sideways only but maintains balance
- ☐ 1 needs supervision when turning
- ☐ 0 needs assist to keep from losing balance or falling

TURN 360 DEGREES

INSTRUCTIONS: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.

- ☐ 4 able to turn 360 degrees safely in 4 seconds or less
- ☐ 3 able to turn 360 degrees safely one side only 4 seconds or less
- ☐ 2 able to turn 360 degrees safely but slowly
- ☐ 1 needs close supervision or verbal cuing
- ☐ 0 needs assistance while turning

PLACE ALTERNATE FOOT ON STEP OR STOOL WHILE STANDING UNSUPPORTED

INSTRUCTIONS: Place each foot alternately on the step/stool. Continue until each foot has touched the step/stool four times.

- ☐ 4 able to stand independently and safely and complete 8 steps in 20 seconds
- ☐ 3 able to stand independently and complete 8 steps in > 20 seconds
- ☐ 2 able to complete 4 steps without aid with supervision
- ☐ 1 able to complete > 2 steps needs minimal assist
- ☐ 0 needs assistance to keep from falling/unable to try

STANDING UNSUPPORTED ONE FOOT IN FRONT

INSTRUCTIONS: (DEMONSTRATE TO SUBJECT) Place one foot directly in front of the other. If you feel that you cannot place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the subject's normal stride width.)

- ☐ 4 able to place foot tandem independently and hold 30 seconds
- ☐ 3 able to place foot ahead independently and hold 30 seconds
- ☐ 2 able to take small step independently and hold 30 seconds
- ☐ 1 needs help to step but can hold 15 seconds
- ☐ 0 loses balance while stepping or standing

STANDING ON ONE LEG

INSTRUCTIONS: Stand on one leg as long as you can without holding on.

- ☐ 4 able to lift leg independently and hold > 10 seconds
- ☐ 3 able to lift leg independently and hold 5-10 seconds
- ☐ 2 able to lift leg independently and hold ≥ 3 seconds
- ☐ 1 tries to lift leg unable to hold 3 seconds but remains standing independently.
- ☐ 0 unable to try or needs assist to prevent fall

TOTAL SCORE (Maximum = 56)

eAppendix.

Balance Evaluation Systems Test (BESTest)^a

Test Number/Subject Code _____ Date _____

Examiner Name _____

Examiner Instructions for BESTest

1. Subjects should be tested with flat-heeled shoes or with shoes and socks off.
2. If subject must use an assistive device for an item, score that item one category lower.

Tools Required

- Stopwatch
- Measuring tape mounted on wall for Functional Reach Test
- Approximately 60 × 60 cm (2 × 2 ft) block of 4-inch, medium-density, Tempur® foam^b
- 10-degree incline ramp (at least 2 × 2 ft) to stand on
- Stair step, 15 cm (6 in) in height for alternate stair tap
- 2 stacked shoe boxes for obstacle during gait
- 2.5-kg (5-lb) free weight for rapid arm raise
- Firm chair with arms with 3 m in front marked with tape for Timed “Get Up & Go” Test
- Masking tape to mark 3-m and 6-m lengths on the floor for Timed “Get Up & Go” Test

Summary of Performance: Calculate Percent Score

Section I: _____/15 × 100 = _____ Biomechanical Constraints

Section II: _____/21 × 100 = _____ Stability Limits/Verticality

Section III: _____/18 × 100 = _____ Anticipatory Postural Adjustments

Section IV: _____/18 × 100 = _____ Postural Responses

Section V: _____/15 × 100 = _____ Sensory Orientation

Section VI: _____/21 × 100 = _____ Stability in Gait

TOTAL: _____/108 points = _____ Percent Total Score

(Continued)

Balance Evaluation Systems Test (BESTest)

eAppendix.

Continued

Balance Evaluation Systems Test (BESTest) Interrater Reliability

Subjects should be tested with flat-heeled shoes or shoes and socks off. If subject must use an assistive device for an item, score that item one category lower. If subject requires physical assistance to perform an item, score the lowest category (0) for that item.

I. Biomechanical Constraints

Section I: _____/15 Points

1. Base of support

- (3) Normal: Both feet have normal base of support with no deformities or pain
- (2) One foot has deformities and/or pain
- (1) Both feet have deformities *OR* pain
- (0) Both feet have deformities *AND* pain

2. CoM alignment

- (3) Normal AP and ML CoM alignment and normal segmental postural alignment
- (2) Abnormal AP *OR* ML CoM alignment *OR* abnormal segmental postural alignment
- (1) Abnormal AP *OR* ML CoM alignment *AND* abnormal segmental postural alignment
- (0) Abnormal AP *AND* ML CoM alignment

3. Ankle strength and range

- (3) Normal: Able to stand on toes with maximal height and to stand on heels with front of feet up
- (2) Impairment in either foot of either ankle flexors or extensors (ie, less than maximum height)
- (1) Impairment in two ankle groups (eg, bilateral flexors or both ankle flexors and extensors in one foot)
- (0) Both flexors and extensors in both left and right ankles impaired (ie, less than maximum height)

4. Hip/trunk lateral strength

- (3) Normal: Abducts both hips to lift the foot off the floor for 10 s while keeping trunk vertical
- (2) Mild: Abducts both hips to lift the foot off the floor for 10 s but without keeping trunk vertical
- (1) Moderate: Abducts only one hip off the floor for 10 s with vertical trunk
- (0) Severe: Cannot abduct either hip to lift a foot off the floor for 10 s with trunk vertical or without trunk vertical

5. Sit on floor and stand up

Time _____ seconds

- (3) Normal: Independently sits on the floor and stands up
- (2) Mild: Uses a chair to sit on floor *OR* to stand up
- (1) Moderate: Uses a chair to sit on floor *AND* to stand up
- (0) Severe: Cannot sit on floor or stand up, even with a chair, or refuses

II. Stability Limits/Verticality

Section II: _____/21 Points

6. Sitting verticality and lateral lean

<u>Lean</u>			<u>Verticality</u>		
<u>Left</u>	<u>Right</u>		<u>Left</u>	<u>Right</u>	
(3)	(3)	Maximum lean, subject moves upper shoulders beyond body midline, very stable	(3)	(3)	Realigns to vertical with very small or no overshoot
(2)	(2)	Moderate lean, subject's upper shoulder approaches body midline or some instability	(2)	(2)	Significantly overshoots or undershoots but eventually realigns to vertical
(1)	(1)	Very little lean, or significant instability	(1)	(1)	Failure to realign to vertical
(0)	(0)	No lean or falls (exceeds limits)	(0)	(0)	Falls with the eyes closed

(Continued)

eAppendix.

Continued

7. Functional reach forward *Distance reached: _____ cm OR _____ inches*

- (3) Maximum to limits: >32 cm (12.5 in)
- (2) Moderate: 16.5–32 cm (6.5–12.5 in)
- (1) Poor: <16.5 cm (6.5 in)
- (0) No measurable lean—or must be caught

8. Functional reach lateral *Distance reached: **Left** _____ cm (_____ in) **Right** _____ cm (_____ in)*

<u>Left</u>	<u>Right</u>	
(3)	(3)	Maximum to limit: >25.5 cm (10 in)
(2)	(2)	Moderate: 10–25.5 cm (4–10 in)
(1)	(1)	Poor: <10 cm (4 in)
(0)	(0)	No measurable lean, or must be caught

III. Anticipatory Postural Adjustments**Section III: _____/18 Points****9. Sit to stand**

- (3) Normal: Comes to stand without the use of hands and stabilizes independently
- (2) Comes to stand on the first attempt with the use of hands
- (1) Comes to stand after several attempts or requires minimal assist to stand or stabilize or requires touch of back of leg or chair
- (0) Requires moderate or maximal assist to stand

10. Rise to toes

- (3) Normal: Stable for 3 s with good height
- (2) Heels up, but not full range (smaller than when holding hands so no balance requirement) OR slight instability and holds for 3 s
- (1) Holds for less than 3 s
- (0) Unable

11. Stand on one leg

<u>Left</u>	<i>Time in seconds</i> _____	<u>Right</u>	<i>Time in seconds</i> _____
(3)	Normal: Stable for >20 s	(3)	Normal: Stable for >20 s
(2)	Trunk motion, OR 10–20 s	(2)	Trunk motion, OR 10–20 s
(1)	Stands 2–10 s	(1)	Stands 2–10 s
(0)	Unable	(0)	Unable

12. Alternate stair touching *# of successful steps: _____ Time in seconds: _____*

- (3) Normal: Stands independently and safely and completes 8 steps in <10 s
- (2) Completes 8 steps in <10 seconds, but shows instability such as inconsistent foot placement, excessive trunk motion, hesitation, or arrhythmic stepping
- (1) Completes <8 steps—without assistance (ie, assistive device) OR >10 s for 8 steps
- (0) Completes <8 steps in 10 s, even with assistive device

13. Standing arm raise

- (3) Normal: Remains stable
- (2) Visible sway
- (1) Steps to regain equilibrium/unable to move quickly without losing balance
- (0) Unable, or needs assistance for stability

(Continued)

Balance Evaluation Systems Test (BESTest)

eAppendix.

Continued

IV. Postural Responses**Section IV: _____/18 Points****14. In-place response—forward**

- (3) Recovers stability with ankles, no added arm or hip motion
- (2) Recovers stability with arm or hip motion
- (1) Takes a step to recover stability
- (0) Would fall if not caught *OR* requires assist *OR* will not attempt

15. In-place response—backward

- (3) Recovers stability at ankles, no added arm/hip motion
- (2) Recovers stability with some arm or hip motion
- (1) Takes a step to recover stability
- (0) Would fall if not caught *OR* requires assistance *OR* will not attempt

16. Compensatory stepping correction—forward

- (3) Recovers independently with a single, large step (second realignment step is allowed)
- (2) More than one step used to recover equilibrium, but recovers stability independently *OR* one step with imbalance
- (1) Takes multiple steps to recover equilibrium, or needs minimum assistance to prevent a fall
- (0) No step *OR* would fall if not caught *OR* falls spontaneously

17. Compensatory stepping correction—backward

- (3) Recovers independently with a single, large step
- (2) More than one step used, but stable and recovers independently *OR* one step with imbalance
- (1) Takes several steps to recover equilibrium or needs minimum assistance
- (0) No step *OR* would fall if not caught *OR* falls spontaneously

18. Compensatory stepping correction—lateralLeft

- (3) Recovers independently with one step of normal length/width (crossover or lateral OK)
- (2) Several steps used, but recovers independently
- (1) Steps, but needs to be assisted to prevent a fall
- (0) Falls, or cannot step

Right

- (3) Recovers independently with one step of normal length/width (crossover or lateral OK)
- (2) Several steps used, but recovers independently
- (1) Steps, but needs to be assisted to prevent a fall
- (0) Falls, or cannot step

V. Sensory Orientation**Section V: _____/15 Points****19. Sensory integration for balance (modified CTSIB)**A—Eyes open,
firm surface

- Trial 1 _____ s
Trial 2 _____ s
(3) 30 s stable
(2) 30 s unstable
(1) <30 s
(0) Unable

B—Eyes closed,
firm surface

- Trial 1 _____ s
Trial 2 _____ s
(3) 30 s stable
(2) 30 s unstable
(1) <30 s
(0) Unable

C—Eyes open,
foam surface

- Trial 1 _____ s
Trial 2 _____ s
(3) 30 s stable
(2) 30 s unstable
(1) <30 s
(0) Unable

D—Eyes closed,
foam surface

- Trial 1 _____ s
Trial 2 _____ s
(3) 30 s stable
(2) 30 s unstable
(1) <30 s
(0) Unable

20. Incline—eyes closedToes Up

- (3) Stands independently, steady without excessive sway, holds 30 s, and aligns with gravity
- (2) Stands independently 30 s with greater sway than in item 19B *OR* aligns with surface
- (1) Requires touch assist *OR* stands without assist for 10–20 s
- (0) Unable to stand >10 s *OR* will not attempt independent stance

(Continued)

eAppendix.

Continued

VI. Stability in Gait

Section V: ____/21 points

21. Gait-level surface

Time ____ seconds

- (3) Normal: Walks 20 ft, good speed (≤ 5.5 s), no evidence of imbalance
- (2) Mild: Walks 20 ft, slower speed (> 5.5 s), no evidence of imbalance
- (1) Moderate: Walks 20 ft, evidence of imbalance (wide base, lateral trunk motion, inconsistent step path)—at any preferred speed
- (0) Severe: Cannot walk 20 ft without assistance or severe gait deviations *OR* severe imbalance

22. Change in gait speed

- (3) Normal: Significantly changes walking speed without imbalance
- (2) Mild: Unable to change walking speed without imbalance
- (1) Moderate: Changes walking speed but with signs of imbalance
- (0) Severe: Unable to achieve significant change in speed *AND* signs of imbalance

23. Walk with head turns—horizontal

- (3) Normal: Performs head turns with no change in gait speed and good balance
- (2) Mild: Performs head turns smoothly with reduction in gait speed
- (1) Moderate: Performs head turns with imbalance
- (0) Severe: Performs head turns with reduced speed *AND* imbalance *AND/OR* will not move head within available range while walking

24. Walk with pivot turns

- (3) Normal: Turns with feet close, fast (≤ 3 steps) with good balance
- (2) Mild: Turns with feet close, slow (≥ 4 steps) with good balance
- (1) Moderate: Turns with feet close at any speed with mild signs of imbalance
- (0) Severe: Cannot turn with feet close at any speed and significant imbalance

25. Step over obstacles

Time ____ seconds

- (3) Normal: able to step over 2 stacked shoe boxes without changing speed and with good balance
- (2) Mild: steps over 2 stacked shoe boxes but slows down, with good balance
- (1) Moderate: steps over shoe boxes with imbalance or touches box.
- (0) Severe: cannot step over shoe boxes *AND* slows down with imbalance or cannot perform with assistance.

26. Timed "Get Up & Go" Test

Get Up & Go: Time ____ seconds

- (3) Normal: Fast (< 11 s) with good balance
- (2) Mild: Slow (> 11 s) with good balance
- (1) Moderate: Fast (< 11 s) with imbalance
- (0) Severe: Slow (> 11 s) *AND* imbalance

27. Timed "Get Up & Go" Test With Dual Task

Dual Task: Time ____ seconds

- (3) Normal: No noticeable change between sitting and standing in the rate or accuracy of backward counting and no change in gait speed
- (2) Mild: Noticeable slowing, hesitation, or errors in counting backwards *OR* slow walking (10%) in dual task
- (1) Moderate: Affects on *BOTH* the cognitive task *AND* slow walking ($> 10\%$) in dual task
- (0) Severe: Cannot count backward while walking or stops walking while talking

(Continued)

TINETTI BALANCE ASSESSMENT TOOL

Tinetti ME, Williams TF, Mayewski R, Fall Risk Index for elderly patients based on number of chronic disabilities. Am J Med 1986;80:429-434

PATIENTS NAME _____ D.o.b. _____ Ward _____

BALANCE SECTION

Patient is seated in hard, armless chair;

		Date		
Sitting Balance	Leans or slides in chair Steady, safe	= 0 = 1		
Rises from chair	Unable to without help Able, uses arms to help Able without use of arms	= 0 = 1 = 2		
Attempts to rise	Unable to without help Able, requires > 1 attempt Able to rise, 1 attempt	= 0 = 1 = 2		
Immediate standing Balance (first 5 seconds)	Unsteady (staggers, moves feet, trunk sway) Steady but uses walker or other support Steady without walker or other support	= 0 = 1 = 2		
Standing balance	Unsteady Steady but wide stance and uses support Narrow stance without support	= 0 = 1 = 2		
Nudged	Begins to fall Staggers, grabs, catches self Steady	= 0 = 1 = 2		
Eyes closed	Unsteady Steady	= 0 = 1		
Turning 360 degrees	Discontinuous steps Continuous	= 0 = 1		
	Unsteady (grabs, staggers) Steady	= 0 = 1		
Sitting down	Unsafe (misjudged distance, falls into chair) Uses arms or not a smooth motion Safe, smooth motion	= 0 = 1 = 2		
	Balance score		/16	/16

P.T.O.

TINETTI BALANCE ASSESSMENT TOOL

GAIT SECTION

Patient stands with therapist, walks across room (+/- aids), first at usual pace, then at rapid pace.

	Date		
Indication of gait (Immediately after told to 'go'.)	Any hesitancy or multiple attempts = 0 No hesitancy = 1		
Step length and height	Step to = 0 Step through R = 1 Step through L = 1		
Foot clearance	Foot drop = 0 L foot clears floor = 1 R foot clears floor = 1		
Step symmetry	Right and left step length not equal = 0 Right and left step length appear equal = 1		
Step continuity	Stopping or discontinuity between steps = 0 Steps appear continuous = 1		
Path	Marked deviation = 0 Mild/moderate deviation or uses w. aid = 1 Straight without w. aid = 2		
Trunk	Marked sway or uses w. aid = 0 No sway but flex. knees or back or uses arms for stability = 1 No sway, flex., use of arms or w. aid = 2		
Walking time	Heels apart = 0 Heels almost touching while walking = 1		
	Gait score	/12	/12
Balance score carried forward		/16	/16
Total Score = Balance + Gait score		/28	/28

Risk Indicators:

Tinetti Tool Score

≤18

19-23

≥24

Risk of Falls

High

Moderate

Low

Appendix.

Functional Gait Assessment^a

Requirements: A marked 6-m (20-ft) walkway that is marked with a 30.48-cm (12-in) width.

1. GAIT LEVEL SURFACE

Instructions: *Walk at your normal speed from here to the next mark (6 m [20 ft]).*

Grading: Mark the highest category that applies.

- (3) Normal—Walks 6 m (20 ft) in less than 5.5 seconds, no assistive devices, good speed, no evidence for imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside of the 30.48-cm (12-in) walkway width.
- (2) Mild impairment—Walks 6 m (20 ft) in less than 7 seconds but greater than 5.5 seconds, uses assistive device, slower speed, mild gait deviations, or deviates 15.24–25.4 cm (6–10 in) outside of the 30.48-cm (12-in) walkway width.
- (1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, or deviates 25.4–38.1 cm (10–15 in) outside of the 30.48-cm (12-in) walkway width. Requires more than 7 seconds to ambulate 6 m (20 ft).
- (0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside of the 30.48-cm (12-in) walkway width or reaches and touches the wall.

2. CHANGE IN GAIT SPEED

Instructions: *Begin walking at your normal pace (for 1.5 m [5 ft]). When I tell you "go," walk as fast as you can (for 1.5 m [5 ft]). When I tell you "slow," walk as slowly as you can (for 1.5 m [5 ft]).*

Grading: Mark the highest category that applies.

- (3) Normal—Able to smoothly change walking speed without loss of balance or gait deviation. Shows a significant difference in walking speeds between normal, fast, and slow speeds. Deviates no more than 15.24 cm (6 in) outside of the 30.48-cm (12-in) walkway width.
- (2) Mild impairment—Is able to change speed but demonstrates mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside of the 30.48-cm (12-in) walkway width, or no gait deviations but unable to achieve a significant change in velocity, or uses an assistive device.
- (1) Moderate impairment—Makes only minor adjustments to walking speed, or accomplishes a change in speed with significant gait deviations, deviates 25.4–38.1 cm (10–15 in) outside the 30.48-cm (12-in) walkway width, or changes speed but loses balance but is able to recover and continue walking.
- (0) Severe impairment—Cannot change speeds, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width, or loses balance and has to reach for wall or be caught.

3. GAIT WITH HORIZONTAL HEAD TURNS

Instructions: *Walk from here to the next mark 6 m (20 ft) away. Begin walking at your normal pace. Keep walking straight; after 3 steps, turn your head to the right and keep walking straight while looking to the right. After 3 more steps, turn your head to the left and keep walking straight while looking left. Continue alternating looking right and left every 3 steps until you have completed 2 repetitions in each direction.*

Grading: Mark the highest category that applies.

- (3) Normal—Performs head turns smoothly with no change in gait. Deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.
- (2) Mild impairment—Performs head turns smoothly with slight change in gait velocity (eg, minor disruption to smooth gait path), deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width, or uses an assistive device.

- (1) Moderate impairment—Performs head turns with moderate change in gait velocity, slows down, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width but recovers, can continue to walk.
- (0) Severe impairment—Performs task with severe disruption of gait (eg, staggers 38.1 cm [15 in] outside 30.48-cm (12-in) walkway width, loses balance, stops, or reaches for wall).

4. GAIT WITH VERTICAL HEAD TURNS

Instructions: *Walk from here to the next mark (6 m [20 ft]). Begin walking at your normal pace. Keep walking straight; after 3 steps, tip your head up and keep walking straight while looking up. After 3 more steps, tip your head down, keep walking straight while looking down. Continue alternating looking up and down every 3 steps until you have completed 2 repetitions in each direction.*

Grading: Mark the highest category that applies.

- (3) Normal—Performs head turns with no change in gait. Deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.
- (2) Mild impairment—Performs task with slight change in gait velocity (eg, minor disruption to smooth gait path), deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width or uses assistive device.
- (1) Moderate impairment—Performs task with moderate change in gait velocity, slows down, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width but recovers, can continue to walk.
- (0) Severe impairment—Performs task with severe disruption of gait (eg, staggers 38.1 cm [15 in] outside 30.48-cm (12-in) walkway width, loses balance, stops, reaches for wall).

5. GAIT AND PIVOT TURN

Instructions: *Begin with walking at your normal pace. When I tell you, "turn and stop," turn as quickly as you can to face the opposite direction and stop.*

Grading: Mark the highest category that applies.

- (3) Normal—Pivot turns safely within 3 seconds and stops quickly with no loss of balance.
- (2) Mild impairment—Pivot turns safely in >3 seconds and stops with no loss of balance, or pivot turns safely within 3 seconds and stops with mild imbalance, requires small steps to catch balance.
- (1) Moderate impairment—Turns slowly, requires verbal cueing, or requires several small steps to catch balance following turn and stop.
- (0) Severe impairment—Cannot turn safely, requires assistance to turn and stop.

6. STEP OVER OBSTACLE

Instructions: *Begin walking at your normal speed. When you come to the shoe box, step over it, not around it, and keep walking.*

Grading: Mark the highest category that applies.

- (3) Normal—Is able to step over 2 stacked shoe boxes taped together (22.86 cm [9 in] total height) without changing gait speed; no evidence of imbalance.
- (2) Mild impairment—Is able to step over one shoe box (11.43 cm [4.5 in] total height) without changing gait speed; no evidence of imbalance.
- (1) Moderate impairment—Is able to step over one shoe box (11.43 cm [4.5 in] total height) but must slow down and adjust steps to clear box safely. May require verbal cueing.
- (0) Severe impairment—Cannot perform without assistance.

(Continued)

Appendix.

Continued

7. GAIT WITH NARROW BASE OF SUPPORT

Instructions: *Walk on the floor with arms folded across the chest, feet aligned heel to toe in tandem for a distance of 3.6 m [12 ft]. The number of steps taken in a straight line are counted for a maximum of 10 steps.*
Grading: Mark the highest category that applies.

- (3) Normal—Is able to ambulate for 10 steps heel to toe with no staggering.
- (2) Mild impairment—Ambulates 7–9 steps.
- (1) Moderate impairment—Ambulates 4–7 steps.
- (0) Severe impairment—Ambulates less than 4 steps heel to toe or cannot perform without assistance.

8. GAIT WITH EYES CLOSED

Instructions: *Walk at your normal speed from here to the next mark (6 m [20 ft]) with your eyes closed.*

Grading: Mark the highest category that applies.

- (3) Normal—Walks 6 m (20 ft), no assistive devices, good speed, no evidence of imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width. Ambulates 6 m (20 ft) in less than 7 seconds.
- (2) Mild impairment—Walks 6 m (20 ft), uses assistive device, slower speed, mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width. Ambulates 6 m (20 ft) in less than 9 seconds but greater than 7 seconds.
- (1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width. Requires more than 9 seconds to ambulate 6 m (20 ft).
- (0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width or will not attempt task.

9. AMBULATING BACKWARDS

Instructions: *Walk backwards until I tell you to stop.*

Grading: Mark the highest category that applies.

- (3) Normal—Walks 6 m (20 ft), no assistive devices, good speed, no evidence for imbalance, normal gait pattern, deviates no more than 15.24 cm (6 in) outside 30.48-cm (12-in) walkway width.
- (2) Mild impairment—Walks 6 m (20 ft), uses assistive device, slower speed, mild gait deviations, deviates 15.24–25.4 cm (6–10 in) outside 30.48-cm (12-in) walkway width.
- (1) Moderate impairment—Walks 6 m (20 ft), slow speed, abnormal gait pattern, evidence for imbalance, deviates 25.4–38.1 cm (10–15 in) outside 30.48-cm (12-in) walkway width.
- (0) Severe impairment—Cannot walk 6 m (20 ft) without assistance, severe gait deviations or imbalance, deviates greater than 38.1 cm (15 in) outside 30.48-cm (12-in) walkway width or will not attempt task.

10. STEPS

Instructions: *Walk up these stairs as you would at home (ie, using the rail if necessary). At the top turn around and walk down.*

Grading: Mark the highest category that applies.

- (3) Normal—Alternating feet, no rail.
- (2) Mild impairment—Alternating feet, must use rail.
- (1) Moderate impairment—Two feet to a stair; must use rail.
- (0) Severe impairment—Cannot do safely.

TOTAL SCORE: _____ MAXIMUM SCORE 30

^a Adapted from Dynamic Gait Index.¹ Modified and reprinted with permission of authors and Lippincott Williams & Wilkins (<http://lww.com>).