Shibaura Institute of Technology

Investigation of Human Balance Ability and Development of a New Sensory Oriented Posture Control Model Based on Joint Stiffness Characteristic

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ABSTRACT

This research thesis introduces a new biomechanics approaches to determine balance disability and to enhance human postural control model in pursuance of developing a better balance ability assessment tool which allows for timely and targeted therapeutic intervention and rehabilitation. The concept of this research is to combine both functional and physiological assessment principal in order to find the key factor or parameter which can represent the resource of human control component. This research proposed a measurement of joint stiffness during normal and perturbed stance. Degree of posture movement during external perturbation applied much depended on effective joint stiffness. Three phase of investigation were performed. The first phase involved an interview session with physiotherapist about recent needs and effective assessment in evaluating balance ability. The second phase consists of clinical study. In this phase, individual were asked to undergo experiment to record posture movement and muscle activation when external platform perturbation is applied. In order to determine the effect of sensory information toward posture movement, three sensory input condition were applied (i.e., vision, vestibular sense, and somatosensory (from light fingertip touch). Response of the ankle and hip joint stiffness at all conditions were measured and compared with a conventional balance assessment score (Functional Reach Test) with the aim of determining its correlation with balance ability. Joint stiffness profile then were used to develop a new sensory oriented posture control model based on joint stiffness characteristic. The joint stiffness profile were later included in the close loop double inverted pendulum model to represent a neuromuscular control. The experiment results have shown that, joint stiffness value is able to discriminate different intensity of perturbation and sensory conditions where good balance condition generated less joint stiffness especially at ankle. Besides, it is also able to show a negative correlation ($r>-0.5$) with Centre of Mass (COM) where high stiffness is required to keep the body sway in a small range. The simulated result from the model have shown it able to generate existence of ankle and hip strategy at particular condition and simulated joint motion displacement almost followed the experiment data with absolute percentage difference less than 20 percent (considered close and near miss). This study concludes that assessment of balance ability via joint stiffness characteristic during perturbed stance provides a means for contributing to the development of a better assessment technique.

Keywords: Balance ability, joint stiffness, sensory manipulation, and perturbed stance**.**

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CHAPTER 1

INTRODUCTION

Balance for human body can be defined as an ability to keep in equilibrium state by adjusting the centre of mass (COM) and centre of pressure (COP) which frequently changes due to changes in positions and movements of the body segment [1]. Furthermore, balance state can be achieved by adjusting postural structure and joint motion based on integration of vestibular, visual, proprioceptive and tactile information. This strategy also known as posture-control strategy. Balance disorder and body instability can be caused by certain health conditions, medications, injuries, disease such as Parkinson disease and stroke or even ageing. Balance disorder can affect daily life activities where someone may feel dizziness or vertigo, falling, faintness, floating sensation, blur vision, confusion or disorientation and most importantly it will effect standing ability [2, 3]. However, according to D A Winter (1995), the death rate among elderly due to falls was quite higher which around 185.6 per 100 000 [4]. Degeneration of balance control system has become the main reason of high risk of falls among older people.

In order to understand about posture-control system, there are many research have been done on risk of falls among healthy individual, elderly and individual with specific disease. This area of research provides information regarding to effect of decreasing in visual acuity, allocation of attention, lack of muscle strength and cognitive demand towards postural task. Typically, a treatment will be provided based on the cause of disorder. A disorder which due to vestibular problem is normally requires an implant treatment. Meanwhile balance disorder affected by neurological problem will undergo rehabilitation treatment. Thus, a reliable assessment method is required to monitor the progression of disease and the treatment efficiency. In recent years, there are many research have been done in order to understand the important characteristic of posture control system and thus, provides a reliable assessment method. The literature of this thesis will reviews all the current finding on posture control system strategy due several conditions such as external perturbation and weakened of body function in term of joint motion and stiffness, and control nervous system (CNS) response through muscle activation. Then, comparisons between those characteristics will be made to evaluate its significant trend with the aims to clarify the current and ongoing research in analysis of posture control strategy and development of balance assessment system.

1.1 Problem Statement

As mentioned before, postural control is the act of maintaining, achieving or restoring a state of balance during any posture or activity according to response creating by the centre nervous system (CNS) [1]. Like before, balance control was described by inverted pendulum model and was assumed as a set of reflexes to maintain equilibrium based on visual, somatosensory and vestibular responses. Unfortunately, these assumptions are quite limited and thus, limit the assessment focus to assess the risks of falling accurately. Actually, postural control system consists of many subcomponent which plays an important role in determination of individual balance ability [5]. However, the existence balance ability assessment only focus or specifies on only one or two of them.

Generally, functional assessments which commonly use in clinical field are using a scoring system completion of each task and focus on performance and quality of movement which obviously only determine physical constraints. The total score given will determine the risk of falls. It provide information about the ability of the patient to function independently but unfortunately further information is required especially to monitor small changes in patient ability to balance especially in term of physiological aspects. Besides, it is important to consider dynamic condition or environment that mimic a real environment, thus a reliable test for balance assessment can be produced. This situation rises a demand for an accurate and comprehensive assessment for measurement of balance ability. This might be can be archived by combining the essential principle of both functional and physiological assessment and may give a consistent evaluation to the patient. This research thesis focuses on the investigation of posture control strategy based on joint motion in correlation of balance ability. Their response towards different sensory input and environment condition will be analysed. Thus, a significant trend of posture changes towards instability can be define. Therefore, an experimental data of posture strategy trends should be considered in evaluating balance ability and the development of simulation model.

1.2 Research Objective

In order to fulfil the needs, three main objectives have been determined as listed below;

i. To investigate the characteristic of human postural movement especially joint stiffness response to maintain balance position during both normal and perturbed stance in order to gather a deep understanding on relationship between posture motion and balance ability.
ii. To compare the characteristic of joint stiffness in different sensory inputs condition and different frequency of repetitive platform perturbation with regard to both balance and cognitive ability.
iii. To develop a simulation module based on joint stiffness profile that will simulate human postural control strategy in dynamic environment and also different sensory input condition.

1.3 New Finding Knowledge

This research will result in new potential towards development of balance ability assessment system that combined both functional and physiological theory. The novel knowledge acquired from this research will lead to a new exploration as follow;

i. Exploration of human posture modulation pattern under various conditions as an essential investigation to improve activity performance and ergonometric aspects.
ii. Development of active fall prevention system or treatment based on postural orientation changes.
iii. Development of simple and non –invasive balance assessment system based on posture image capture.

iv. The therapist perception in using the computerized system and perturbed stance approach to evaluate risk of fall.

v. Exploration of posture control system adaptation due to loss of other extremities.

1.4 Significance of Research

Early detection of disease bring a huge possibility of recovery. However, not many knows that lack of balance ability is actually a sign of various serious health problems. Instability not only can cause injuries but also death. Due to the need of evaluation of balance capability of an individual, over a decades, many assessment methods and approaches have been developed and used. An increases of demands for a better assessment have encouraged researchers and physiotherapists to come out with many idea to improve evaluation method to fit in the clinical setting and environment. However, there is an argument that the existence assessment system unable to provide meaningful evaluation. It is reported to have less correlation between quality of posture movement and physiological change and have poor discriminative ability between healthy and disable subject. Furthermore, a complex attachment on patient's body to gather physiological information such as electromyography to analyse muscle activity will cause discomfort and less ecological validity which can leads to a wrong diagnosis of balance ability.

This research will result in investigating the possible characteristic of human biomechanics measures especially joint stiffness which can define both posture performance and physiological aspects. The significant trend between existence conventional balance assessment score and joint stiffness would proof the possibility of joint stiffness measurement as a promising method to evaluate balance ability. The change of joint stiffness under different sensory condition (including vision, vestibular sense and somatosensory) can provided a better knowledge on its response due to lack and improvement of body function. This allows a future develop of reliable and simple assessment with only based on biomechanics measures.

1.5 State of Art

The aims of this research is to investigate the possible measure to determine balance ability, thus predict risk of fall. In general, balance ability assessment is centred to both centre of mass (COM) and centre of pressure (COP) deterioration. The available assessment methods and computerized system have been used widely in clinical environment. However, there are still room for improvement especially the need of neuromuscular response properties representation and reliability of the assessment. According to the literature review, analysis of joint stiffness properties have shown a positive characteristic which not only can described constraint of movement, but also muscular response. The main focus of this research is to observe changes of stiffness's magnitude or gain due to different sensory condition and external perturbation. Also, its correlation with balance ability. The development of simulation model is required to estimate the posture control model and the effectiveness of joint stiffness measurement to predict balance impairment.

1.6 Scope and limitation

i. The experiment conducted using healthy young male subjects between aged of 24.24 ± 2.19 years old.

ii. All measurement and experiment set up will be based on determination of posture modulation (referring to ankle, hip and neck joint motion) and joint stiffness changes.

iii. All sensory manipulations will be based on physical manipulations without involving invasive stimulation.

iv. The measurement and simulation program are coded in MATLAB language and SIMULINK.

1.7 Outline of Thesis

The research title is "Investigation of Human Balance Ability and Development of a New Sensory Orientated Posture Control Model Based on Joint Stiffness Characteristic". This section briefly described the content of this research thesis which consists of six chapters including introduction, literature review, methodology, results, discussion and conclusion chapter.

Chapter 1: The first chapter provides a general introduction and background of the whole research including problem statement, research objective, scope and limitation and outline of the thesis.

Chapter 2: The second chapter is the literature review section which explained the previous study that related to this research. This chapter is begin with an introduction of balance and stability concept. Then, it continue with the description of human postural control system. Comparison of available assessment method and general knowledge of joint stiffness measurement is also discussed. Finally, representation of human posture as inverted pendulum model and it description are also included.

Chapter 3: The third chapter provides information on how the research is conducted. This chapter begin with the description of research framework and then, research design which includes of three difference phase; instigation phase, clinical study phase and modelling phase. In the investigation phase, it is includes survey and discussion with physiotherapist regarding to their opinion on the existence balance ability evaluation method. For clinical study phase, it is include details on the experiment setup, equipment, participant, recording parameter and setting, and data analysis methods. Finally, for the modelling phase, the simulation model of posture control development methods are discussed including the parameter setting and equation used.

Chapter 4: This chapter is where the results gathered from the clinical experiment and simulation model were discussed. At the beginning of the chapter, the physical data of the participant were described. Then, it continue with comparison between joint stiffness and score from conventional balance assessment. Analysis of joint stiffness in response to balance ability and limited vision and vestibular sense input during dynamic perturbation is deliberated. After that, analysis of human posture strategy scheme with the existence of additional somatosensory input from fingertip is discussed. At the end of this chapter, the output gathered from the simulation model of human posture control is analysed.

Chapter 5: The fifth chapter discussed issues including stability and joint stiffness response, weakness in vestibular and vision sensory in response to perception of posture response. Then it proceed with discussion on motor learning ability, effect of light touch to stability and posture modulation and role of wrist and finger to body motion. Lastly, discussion continue with simulation of human posture control using joint stiffness profile.

Chapter 6: The sixth chapter which is the last chapter of this thesis explained the conclusion of the whole research finding and recommendation for upcoming development of the new assessment system.

CHAPTER 2

LITERATURE REVIEW

2.1 Balance and Stability

Balance can be defined by using Newton's First Law where it is the condition where the resultant forces or load actions acting upon an object are zero.

$$\sum F = 0$$

In general, the ability of an object to stay in balance actually depends on several aspects such as base of support (BoS) or commonly known as centre of pressure (COP) and centre of mass (COM) or also known as centre as gravity (COG). Position of centre of mass (COM) and centre of pressure (COP) play an important role to ensure an object to remain balance especially in static condition. For example, if COM located within the COP range, then the object is balanced. However, stability is connected to movement. When an object is moving, the COM will also move away from COP, thus become unbalance and instable. Stability can be archived with a large COP so that the COM will always lie within COP therefore will create better stability. Figure 1 below shows the relationship of COM and COP in determining the stability.

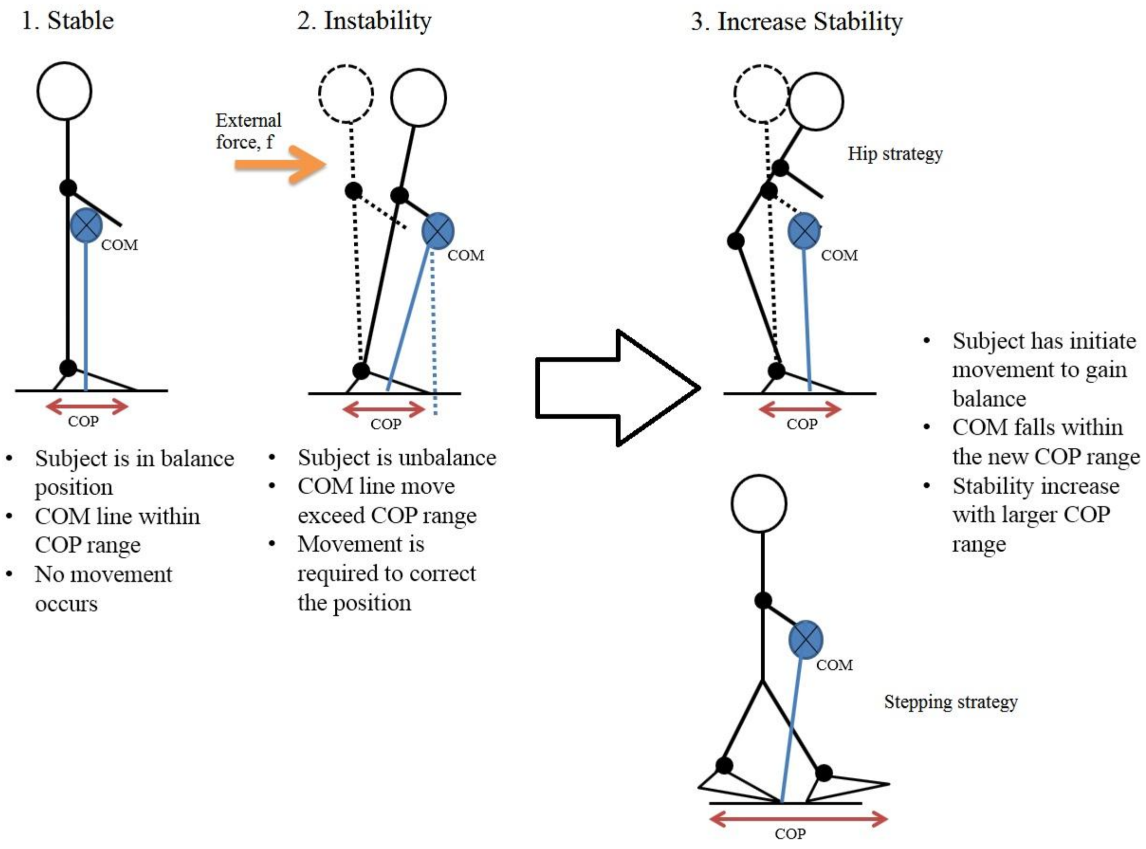


Figure 1 The relationship between centre of gravity (COM) and centre of pressure (COP) in defining balance and stability ([1]).

Stability and balance principle mentioned above are applicable to human. The unique of human posture structure has allow various combination movement from joints to make sure both COP and COM in equilibrium state [6]. In addition, human has the natural ability to sense the threat to stability and to use muscular activity to counteract the force of gravity in order to prevent falling. This response always refer to posture control system which control by centre nervous system (CNS).

2.2 Human Postural Control System

Postural control system responsible to initiate any movement as the act of maintaining, achieving or restoring a state of balance during any posture or activity [1].This postural control has been considered as reflex-like responses produced by a sensory stimulus. In term of balance, the body is depending on many variables in the centre nerves system (CNS) to be maintained. Strategies of postural control therefore vary depending on individual's goals and environment condition. Based on figure below, when a human facing disturbance which will affect balance, a sensory system which normally related to proprioceptive information will give response to central nerves system. The CNS will translate the response and provide an information regarding reaction towards threat, posture maintenance and movement strategy required. Those strategies will be transmitted as signals to motor system and thus creating movement of joint, limb and support to preserve balance.

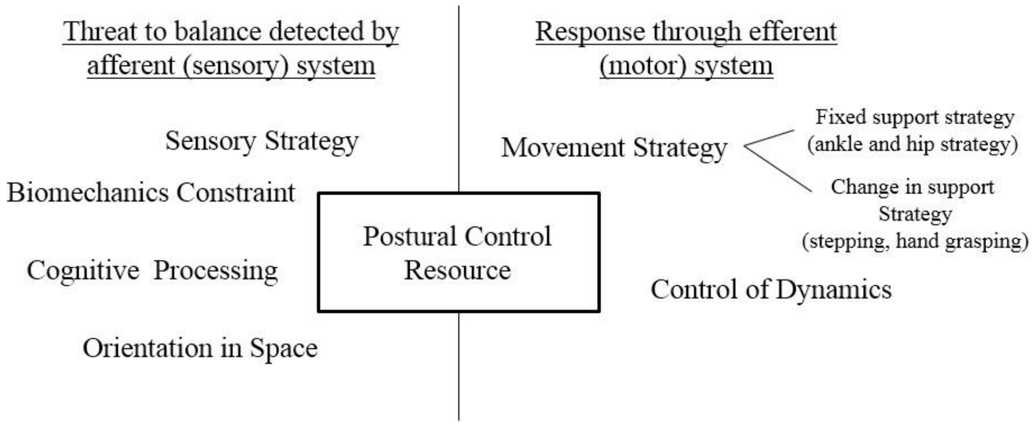


Figure 2 Summary of human postural control strategies [1, 7].

Balancing process occurs through compensation of feed forward and feedback control. According to Kandel (2000), human body movement reaction towards to external perturbation or any condition can be voluntary (desired response) or involuntary (reflective) depend on the situation faces [8]. Then, the nervous central system (CNS) learns to correct for such external perturbation in two way as mentioned before;

a. Feedback control

The CNS monitors sensory signals and uses this information to act directly on the limb itself. This control mechanism occurs to provide reflective movement. For example, to maintain desire position. Figure below illustrated the feedback control;

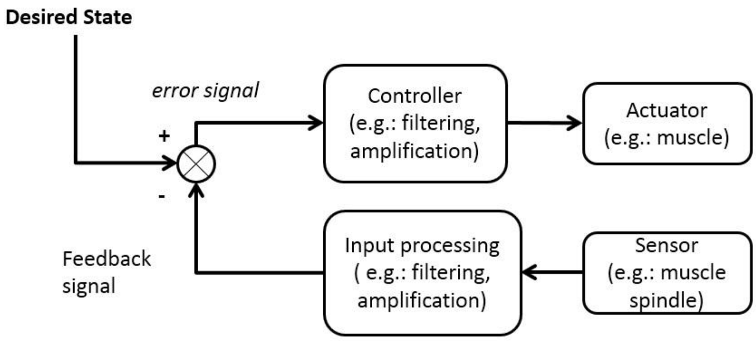


Figure 3 Feedback control [8].

b. Feedforward control

The CNS used input from the same or different senses such as vision, hearing and touch to detect any possible or forthcoming perturbation and initiate proactive strategy based on that information and experiment. Activities such as kicking a ball is one example of feedforward mechanism. This mechanism also can used to describe the voluntary response.

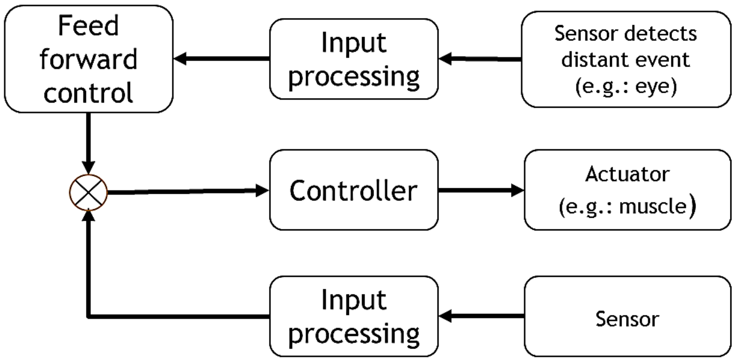


Figure 4 Feedforward control [8]

Applicability of either both control mechanism or each one of them depend on the sensory information and cognitive system. By involving both memory and motor learning ability which part of cognitive system, the CNS observed able to reduce the magnitude of action required (i.e.; muscle activation and movement) [9, 10]. There are three major senses involved in posture balance control such as vision (planning motion and avoiding obstacles), vestibular system (detect linear and angular acceleration) and somatosensory system (sense the position, velocity, orientation of gravity and contact with external object) [4]. Sufficient feedback information from the senses make the balancing process effective.

Visual information plays an importance role in stabilizing balance. According to Buchanan (1999), visual information helps in reducing the variability of head's position and the position of the centre of mass (COM) within the support surface defined by the feet. Furthermore, visual system and vestibular system works closely to maintain balance. Normally when the head is moving, fluid located inside semi-circular canal (a part of labyrinth in ears) will also moving. This movement will activate other part of inner ear and finally creates a signal to the brain to indicated movement and position of the head. Figure 5 below describes the mechanism of vestibular nuclei that trigger vestibular spinal tracts which plays an important role in movement and posture [11, 12].

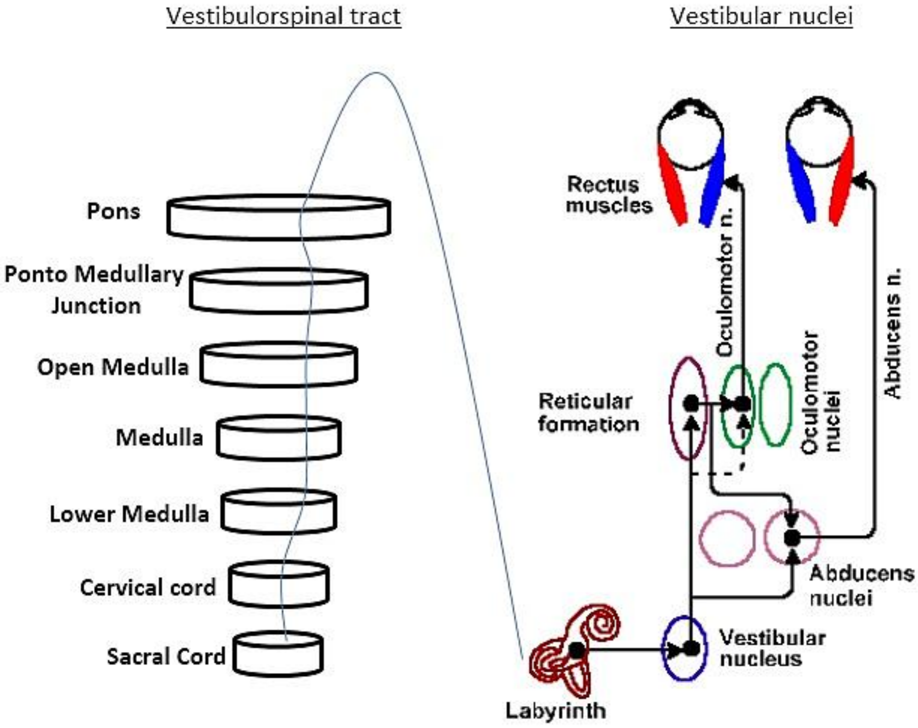


Figure 5 Brainstem pathways for control of eye movement and posture by semi-circular canal.

The pathway of interaction between afferent and efferent of human nervous system is too complex. However, it is enough to understand the pathway of vestibular spinal tract. Based on the figure above, it is understand that with any movement experiences by head due to external influence will trigger changes in labyrinth where it is not only causes eyes move conjugate with head movement to maintain gaze on desire object but at the same time , transferring information for posture and movement adjustment.

Degeneration of posture control system function can lead to balance disorder. Balance disorder can affect daily life activities where someone may feel dizziness or vertigo, falling, faintness, floating sensation, blur vision, confusion or disorientation and most importantly it will effect standing ability [2]. Balance disorder will lead to fall risk. According to Winter (1995), the death rate among elderly due to falls was quite higher which around 185.6 per 100 000 [4]. Balance disorder and body instability can be caused by certain health conditions, medications, injuries, disease such as Parkinson disease and stroke or even ageing. Weakened of function of above mechanism explained in Figure 2 has become the main reason of high risk of falls among older people. According to Hill and Schwarz (2004), factors that contribute to risk of falls among elderly can be divided into two; intrinsic factor and extrinsic factor. Intrinsic factor includes demographic (age and history of falls), lower extremities impairment (muscle weakness, walking problem and etc.), sensory and neuromuscular decreases (Parkinson disease, stroke, impaired cognitive and etc.) and medication intake (psychoactive medication, polypharmacy); meanwhile extrinsic factors include quality of life, residential care and hospital setting [13]. Moreover, the reason of balance impairment among healthy elderly was agreed to be caused by the lack of cognitive system which can be observed by measuring joint stiffness [14].

These researches have provided information regarding the effect of demography [15], visual acuity [16], allocation of attention, lack of muscle strength and cognitive demand towards postural task [14, 16]. Typically, a treatment will be provided based on the cause of disorder. A disorder which due to vestibular problem is normally requires an implant treatment. Meanwhile, balance disorder affected by neurological problem will undergo rehabilitation treatment such as balance and vestibular physical therapy. Thus, a reliable assessment method is required to monitor the progression of disease and the treatment efficiency. In recent years, there are many research have been done in order to understand the important characteristic of posture control system and thus, provides a reliable assessment method.

2.3 Functional and Physiological Assessment

Assessment of balance disorder has an important role especially as an aid to clinical diagnosis and the assessment of treatment efficacy, as an aid to identify elderly people with history of falls and it risk, and lastly as an aid to understand how the postural control system works. In general, there are two type of balance assessment method available which are functional assessment and physiological assessment. Functional assessment generally used in clinical practice by doctors and physiotherapist due to fast monitoring and not required expensive equipment. There are method which really specific towards balance disorder assessment such as Berg Balance Scale (BBS), Functional Reach Test (FRT) and Tinetti Assessment Tool (TAT), meanwhile others are more likely to have some component that related to balance assessment. For example, Fugl-Meyer Assessment which the main objective is to evaluate motor and sensory impairment faced by both lower and upper extremities but somehow the result will help to predict fall risks. The BBS and Tinetti assessment evaluate the patient with specific task such as sitting to standing, standing one foot, standing unsupported and many more. Then, the physiotherapist will give a score based on completion of the task. This is different with FRS where a quantitative data will be gathered. In FRT, measurement of maximal distance that one can reach beyond arm’s length while maintaining balance will be taken. However, these tasks were applied to trigger individual postural control as a whole. As mentioned, postural control is the act of maintaining, achieving or restoring a state of balance during any posture or activity [1]. For example, when a patient was asked to standing from sitting position or standing with one support, actually the central nerves system (CNS) will create a response and provide information regarding reaction towards the movement, posture maintenance and voluntary movement required. Those strategies will be transmitted as signal to motor system and thus creating movement of joint, limb, head and support to preserve balance. However, these postural strategies cannot be evaluated by using functional assessment methods. Most of the methods are using a scoring system completion of each task and focus on performance and quality of movement. The total score given will determine the risk of falls. Nevertheless, Berg Balance Scale was reported to have a good reliability in predicting balance impairment, meanwhile Tinetti Assessment Tool and Functional Reach Test was informed to have acceptable characteristic to be apply as assessment tool for balance disorder [17-20]. Even though, it have a good reliability still these method was reported to have low sensitivity and responsiveness. For example, the patient who assessed by using the TAT, still there are patients who fell even they scored high score in the assessment (high score predicts less fall risk) [18].

These assessments explained above were reported to be exposed with human error and being suggested to include with other method to support the results [18, 19, 21]. Optoelectronic system was reported to be used together with the FRT method. It was reported to help provide useful information and improve quality of assessment [22]. On the other hand, most of the method required continues assessment to determine the problem. Some required more than a month to predict the balance disorder [23, 24].

On the other hand, physiological assessment evaluates balance by measuring patient's physiological changes for example by calculating both centre of pressure (COP) and movement of centre of gravity (COM) [25]. Physiological analysis tells more about the capabilities and limitation of sensory system. Generally, it measures kinetic, kinematics and electromyography aspects of the patients during posture changes [26]. This assessment method required a specific system to trigger patient's movement and create required perturbation in evaluating balance. Some approaches involved sophisticated device such as wearable sensor like gyroscope to record particular changes during gait to detect balance ability [27-29]. The most favourable test is by using moveable platform where a postural control strategy under dynamic condition can be defined. The effects from inclined or declined surface; and forward or backward acceleration produced by moveable platform will influence the whole body including COM, COP, proprioceptive system and joints movement [30].

By following the concept of physiological assessment, computerized dynamic posturography (CDP) system has been developed and commercialized. In early 1980s, posturography has been reported used in clinical application which is the combination of devices such as dynamic force platform; body movement recording system and electromyography (EMG) were used with horizontal and vertical direction of surface perturbation [31, 32]. Besides, visual inputs were applied to this system to monitor the vestibular and proprioceptive system functions. This computerized system can be assumed as a complete system to monitor a physiological change in human body regarding to movement and postural strategy when perturbation occurs. It allowed the clinician to introduce several manipulations such as sensory and cognitive manipulation in order to address the physiological changes. However, dynamic posturography is less preferred by physiotherapist (PT) due to several barriers.

One of the main reason is it is difficult for all PT to implicate the result of this application. PT must require appropriate or additional knowledge to interpret the data. Besides, this equipment requires a high cost to purchase and for long term maintenance. But, it still cannot be denied that such advance technology system like posturography system was reported to be have a moderate correlation with two common functional test and able to discriminate the effects of ageing and disease to human posture control system [25]. Figure below described the available posturograph available in market. Currently there are two type of posturograph device which produced by NeuroCom® and Biodex Medical System.

(a) (b)

Figure 6 Computerized balance assessment available such as (a) Computed Dynamic Posturograph (CDP) by NeuroCom® and (b) Balance System SD by Biodex Medical System.

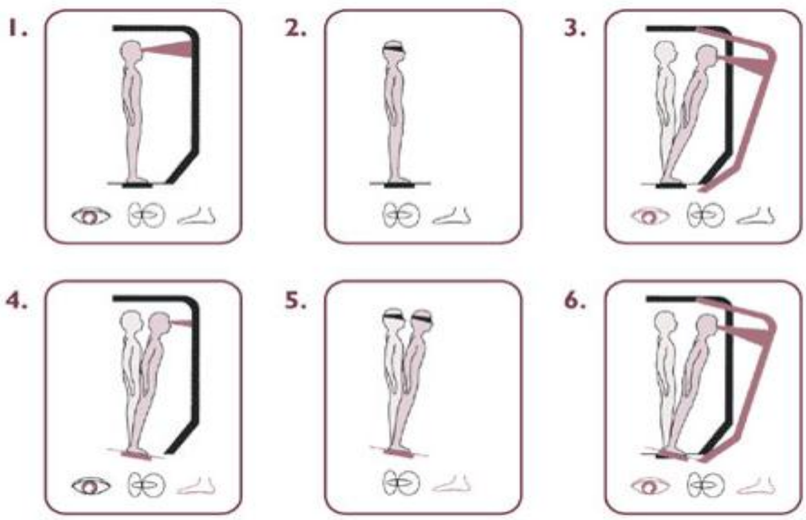


Figure 7 Illustration on the sensory test scheme available in the CDP system.

Table 1 below shown the comparison between 6 different functional assessment methods which commonly used to monitor balance disorder. This comparison includes the advantages and limitations of each method. Both assessments were seen produced different type of feedback for patients and physicians. Furthermore, each of assessment required different duration to predict the risk of fall and detect balance disorder disease. Some demand continues assessment to produce a reliable result meanwhile some can predict the impairment on the spot. However, there is no specific standard or guideline provided for physiotherapist about the appropriate assessment to assess balance impairment. Selection of an assessment normally based on arbitrary selection but not founded by relevant assessment required by individual [1]. Thus, there are still needs and space for a new assessment method that is not just simple, but reliable and efficient.

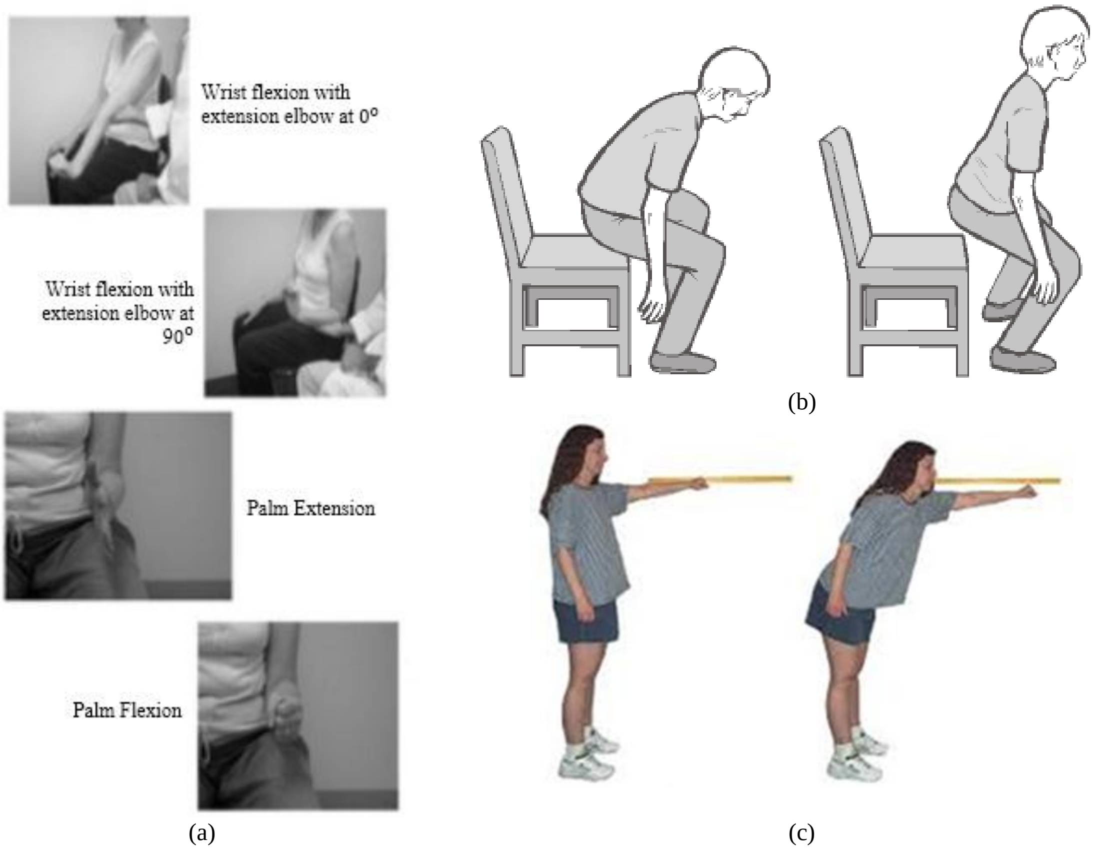


Figure 8 Example of functional assessment available; (a) Example of Fugl-Mayer Hand and Wrist assessment, (b) Tinetti Balance Test (Get up and Go assessment) and (c) Functional Reach Test (FRT)

Table 1 Balance ability assessment method available.

Assessment	Advantage	Limitation	Reference
1) Functional Assessment			
BBS	-no special equipment -good internal consistency reliability -Interrater reliability	-high possibility of misjudge -demonstrated only 53% sensitivity -low level of responsiveness	[21],[17],[23]
Tinetti	-fit with clinical environment - acceptable to be used as a screening test for fall	-rapid drop in sensitivity -was recommended to be improved	[20],[18]
Barthel	-useful as daily record for patient	-very brief -cannot evaluate balance ability accurately -more towards independency evaluation	[33]
Fall risk index	-useful and reliable as screening tool	- required continuous assessment	[24]
Fugl-Meyer	-only for half impairment patient	-longer time -need a quiet place -low level of responsiveness	[23]
Functional Reach Test	-useful in prediction of postural stressor	-need to use additional tool to improve data collection - exposed to human error.	[34],[19]
2) Physiological Assessment			
Dynamic Posturography	-able to apply sensory manipulations -useful and able to detect disease due to weakened of sensory and cognitive function	- poor discriminative ability -complex data analysis and data collection -complex attachment on patient body - unable to detect defect on posture performance	[26]
	a. CDP -functional evaluation for BVL patient, and dizziness.	-only reported sensitive to disease related to vestibular system.	[35, 36]
	b. **Balance System SD** -measures stability from COP displacement at medial-lateral and anterior- posterior direction	-author found small contribution from medial-lateral and anterior-posterior stability index to overall stability index.	[37]

2.4 Joint Stiffness measured Balance Ability

As we can see from available assessment as mentioned in the section before, quiet standing approach is the simplest approach to analyse balance impairment in a patient. Under different external perturbation, changes of centre of mass (COM) and centre of pressure (COP) can give a hint on weakness of posture control system. Furthermore, dominance of hip or ankle strategy can be observed through this approach. The elderly was reported to have higher COP component than young subject during quiet standing and relied more on hip strategy to keep balance [4]. However, there is still limited information regarding the central of nervous system (CNS) adaptation to maintain balance under different intensities of perturbation and changes due to ageing.

As early as 1970s, investigation on joints stiffness has begun where research was focus on biomechanics properties of single joint such as viscous elasticity of ankle joint stiffness [38]. After that the focus has become wider, as the researcher started to investigate the reflexive response [39, 40], voluntary response [39, 41], and reactive control [6], muscle activation during stiffness control [40, 42]. Besides, in posture control system study, joint stiffness is contributes to body sway. Body sway defines the reaction of body to maintain the position of both centre of mass (COM) and centre of pressure (COP). The joint stiffness control would act to correct the COP to move in the same direction as the COM to maintain in balance position [6]. Furthermore, research by Fitzpatrick et.al (1992) concluded that posture sway confine when reflex response was higher [39].

In analysis of movement and gait performance, joint stiffness also was observed among knee and hip osteoarthritis's patient [43, 44]. It was suggested that, defected joint is more stiff than others. Moreover, analysis of torque and dynamic stiffness of joint especially at ankle among elderly have given a clue that altered posture control strategy applied by elderly during stepping down. And it suggested that any rehabilitation and exercise strategy of elderly should be more focus on development and maintain both ankle and foot function [45].

Our previous study on joint stiffness characteristic over difference frequency intensity of translation perturbation have shown ankle and hip joint applied different stiffness characteristic over high intensity of perturbation [46]. As the frequency of perturbation increase, stiffness at ankle was observed to be increased meanwhile hip become less stiff and sway more to maintain balance. Under increment of intensities of perturbation, stiffness pattern able to explain the posture strategy shifting from ankle to hip strategy. Furthermore, theoretical study done by Edwards (2007), based on database of average people physical measurement agreed with the study where stiffness at every joint is counteract to each other and it was measured to be decreased at one joint as it increased at other joints [47]. Besides, our previous experiment also have shown that the adaptation of CNS over repeated perturbation can be seen through stiffness characteristic [46, 48]. The stiffness response under repeated perturbation under period of time is able to describe the CNS response in term of motor learning ability. However, how balance ability related to pattern of joint stiffness are still remains unclear. Table 2 is summary of previous research which explained related parameter measured in correlation with joint stiffness.

Table 2 Parameter measured in correlation with joint stiffness.

Parameter measure	**Type of perturbation**	**Intensity**	**Reference**
Reflex response	- Stand still and ease - plantar flexion and dorsiflexion	- torque of 10 N/m	[39], [49]
Variability of COP and COM	- quiet standing	- different stance width	[6],[40],[50]
Muscle properties and activation	- quiet standing - Gaussian torque -goal directed movement	 - 0-50 Hz band-limited - high speed	[40],[8],[50], [51, 52] [53]
Intrinsic properties	 -tilting perturbation -Step-like disturbances	 -0.78,1◦ - amplitude for 150 ms duration	[49],[54],[55]
Stability	- theoretical study	-based on average physical measurement	[47]
Ageing	-perturbed stance	- Rotated pseudo peak-to-peak amplitude of 4◦ and a cycle duration of 48.4 s	[56], [57], [58]
CNS adaptation	-repeated perturbation	- longer duration (more than 60s)	[46, 59, 60]
Postural Strategy	- forward and backward translation	- different frequency and velocity	[46, 61]
Gait performance	-hip anthroplasty patient -knee antroplasty patient	- normal walking	[44] [43]
Voluntary sway	-different stance condition	-wide, narrow and tandem position	[41]
Active and passive mechanisms	- perturbed stance (rotational)	- Rotated pseudo randomly with peak-to-peak amplitude of 4◦ and a cycle duration of 48.4 s -at different frequencies (0.02–2.80 Hz).	[57],[62],[63]

In addition, Figure 9 below has summarized that measurement of joint stiffness is able to describe and differentiate balance impairment based on five different response and characteristic. And those characteristic have shown that joint stiffness is capable to be one of the simple and reliable measurements to detect and measure balance ability and finally, predict the risk of fall. Due to this positive finding on joint stiffness characteristic which normally model as inverted pendulum, further research will be pursue in order to develop a new algorithm and alternative in measuring human balance ability and impairment.

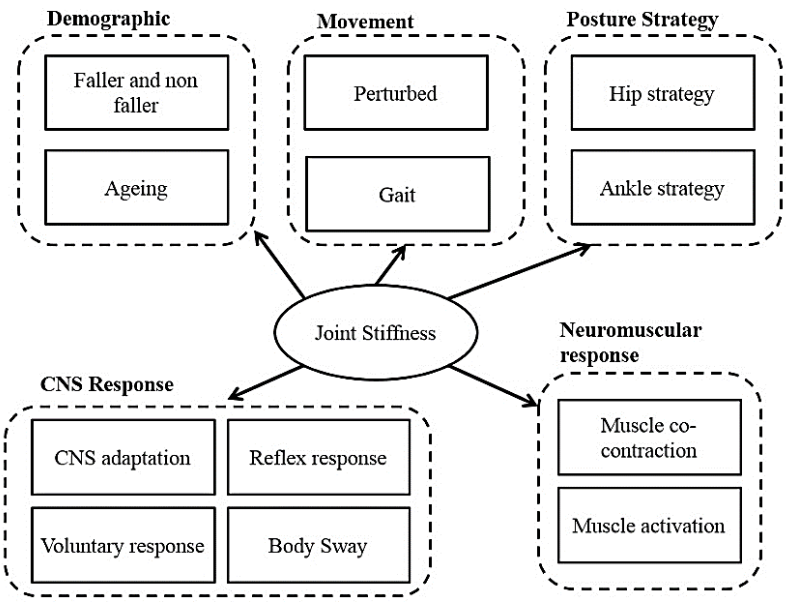


Figure 9 Five main characteristic of human posture control that can be measured from joint stiffness parameter.

2.5 Human Posture as Inverted Pendulum Model

Human balance characteristic or human posture strategy usually observed as an inverted pendulum model [63-67] . Inverted pendulum model has been beneficial to describe postural sway and it is used widely in analysis of posture control system.

Stiffness at ankle joint describes as below in a component of sway angle;

$$K_a = \frac{\tau}{\theta_a} \tag{1}$$

where τ_a is torque of ankle and θ is angle of ankle sway. The single link inverted model is describes as below*;*

$$I\ddot{\theta}_a + B_a\dot{\theta}_a + K_a\theta_a = m_1 gl \, sin\theta_a \tag{2}$$

where I is moment of inertia, B_a is damping at ankle joint, $\ddot{\theta}$ is angular acceleration, $\dot{\theta}$ is angular velocity, m is mass, g is gravitational acceleration and l is a distance of COM from ankle joint. Based on Figure 10, at equilibrium state, total of ankle joint torque should follow the equation below;

$$m_1 gx + F_v y \approx 0 \tag{3}$$

where Fv is a vertical component of ground reaction force, **x** is distance from COM to ankle joint and y is distance from Fv to ankle joint. In common experiment set up where Fv is measured using force plate, it can be also be defined or assumed as the COP [68]. From eq. (1) and (2), angle of joint sway is important to determine stiffness and damping parameter. The change of angle over time will determine the way of joint counteract against the oscillation of perturbation which is may results from evoking the intrinsic mechanical properties of joint, and muscle contractile elements which triggered by the

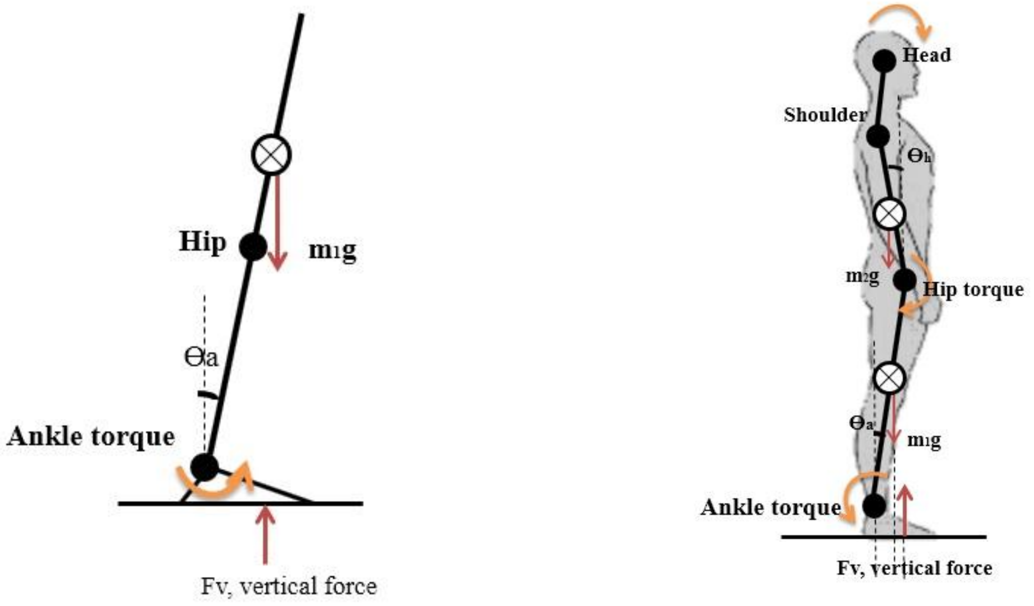
CNS. Damping coefficient may describe the ability of joint to absorb shock. However, it has a very limited physiological meaning [69]. It was observed triggered at early stage of perturbation and after that remains low and unchanged over time [46]. Due to this phenomenon, damping coefficient and characteristic can be neglected. Stiffness value varied according to condition, however posture stiffness was reported at average of 500-600Nm/rad [70]

Meanwhile, under more difficult situation, Buchanan et al, (1999) suggested that this single link inverted pendulum will split into multi-link model. And thru this situation, we will able to understand how posture control system shifting it strategy from ankle strategy to hip strategy [7]. Thus, a simplest way to describe shifting of strategy as mentioned before is by using double link inverted pendulum as shown in Figure 10 (b). Measurement of stiffness at both ankle and hip joint are still can be done by using eq. (1). However, mathematical model of double-link inverted pendulum to represent human posture control strategy can be described as below;

$$\tau_1 - \tau_2 = -[m_1 g l_1 \, sin\theta_a - m_2 g l_2 + K_h\theta_h - K_a\theta_a - B_a\dot{\theta}_a - I\ddot{\theta}_a] \qquad (4)$$

Where τ_1 and τ_2 is torque acting at both ankle and hip joint. Meanwhile, K_h and θ_h are stiffness and angle of sway at hip joint. Besides, the role of other segment and joint such as knee , head and arm segment are believe not to be a main part in posture control system but as additional part that will improved posture control response in preserving balance condition [47, 71].

Even though joint stiffness may able to describe restriction in movement in related to CNS's adaptation to preserve balance condition, it is important to relate it with muscle activities and effect of sensory limitation towards joint stiffness characteristic.



(a) (b)

Figure 10 Inverted pendulum model to represent human posture control system for both (a) single link and (b) multiple-link inverted pendulum.

In same case, stiffness indicates disadvantages due to constraint to movement, but in case of balancing ability, stiff at certain joint is a must. Still, too much stiff at joints will not give an efficient strategy to keep balance and may increase a risk of fall. Because of that, further investigation to determine the efficient and appropriate joint stiffness value that enough to keep body in balance is needed. Through this investigation, we believe that a strong understanding related to human balance, performance of movement and the CNS response can be achieved. If previously, researchers are more focus on evaluate the properties of COM and COP to describe balance ability, we suggest to look into more at a specific area which is joint stiffness as we already known that this characteristic is the essential part lead to pattern of COM and COP.

Many research have been done in order to model human balance process. One of the earliest research done was done by Fitzpatrick et al,(1992), where they tried to model human posture modulation at waist level [39]. However, there were some weakness where it failed to provide direct estimation of the ankle motion characteristic responses because the experiment done and these afford then continued by unaccounted descending motor command. Similar approaches was done by Jiang et, al (2005) where upright body was simulated with two-link inverted pendulum [72]. In this study, ankle and lumbosacral joint are in reverse phase but in the same amplitude and it's were reported quite consistence with the experiment data [72]. However, these results are contradict with previous study which focus on examine posture changes during perturbed stance where ankle and hip joint move opposite each other when individual faces high velocity of perturbation (≥ 0.5 Hz) [7, 73].

On the other hand, system dynamics of ankle-foot measured from ground contact response and external applied force impulses have been developed by Granata et al, (2004). This model established based on two components which a linear and second-order parametric model in order to model the behaviour of the effective stiffness and a non-liner feedback response [74]. However, this model only considered parametric comparison where based on the fix condition and only focus on ankle movement. Further study is still warranted to quantify musculoskeletal dynamics and others joint response in variable environment in order to complete the postural control system modelling.

In the same year, Maurer et, al (2005) have developed a simple model of human posture control at spontaneous sway based on a single link inverted pendulum where output is restricted to sagittal plane [64]. This model able to simulate both the COM and COP traces. The result have shown that this simple feedback model able to reproduce realistic sway behaviour and produced high correlation with experimental results. Unfortunately, this model did not interpreted the motion of joint which play an important role in determination of posture control strategy.

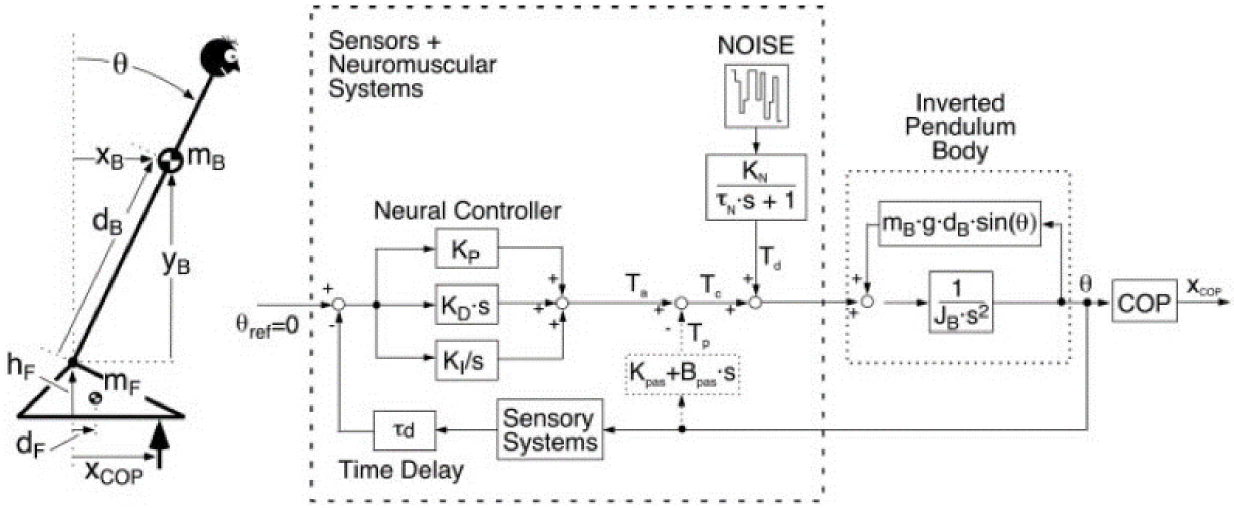


Figure 11 Human postural control model [64].

Even though relationship between neurological response such as muscle co-contraction and joint stiffness is still not clear, we suggest a total analysis of muscle activation under dynamic condition to be done. Muscle have different action at individual joints. According Kandel (2000), muscle force may be used to create stiffness at joints [8]. For example, in case of individual trying to stand on the deck of a small boat pitching back and forth in the water, large force must be applied in order to pull the centre of mass back from any direction. By contracting muscle especially at ankle before these perturbation, that person will increase the stiffness at the joint. Previous research have shown that muscle co-contracted to improve accuracy of movement [12, 53, 75]. Those finding have gave information about positive correlation between joint stiffness and muscle contraction.

Dynamic condition will almost trigger a natural response of posture control system and is expected to give a reliable correlation between neurological and kinematic response. Thus, further investigation on this area is warranted, especially under dynamic condition, so that a clear understanding on posture control system can be achieved.

CHAPTER 3

METHODOLOGY

3.1 Research Framework

In order to determine the significant trend on posture changes due to internal or external perturbation, a detail research design should be prepared so that the important measures will not overlook. Besides, opinions from stakeholder from rehabilitation field such as physiotherapist, researcher and hospital management personal should be taken in count. For this study, research framework as in Figure 12 was referred. Briefly, as mentioned before, there were two type of assessment method to evaluate human balance ability and those assessment have been characterized based on its focus. According to Figure 12, each of the assessment has their own focus which in average accounted not more than two of the resource of human control. Thus, this can be concluded that, combination between both physiological and functional assessment concept can led to introduction of new system concept which will cover more aspect of human control resource.

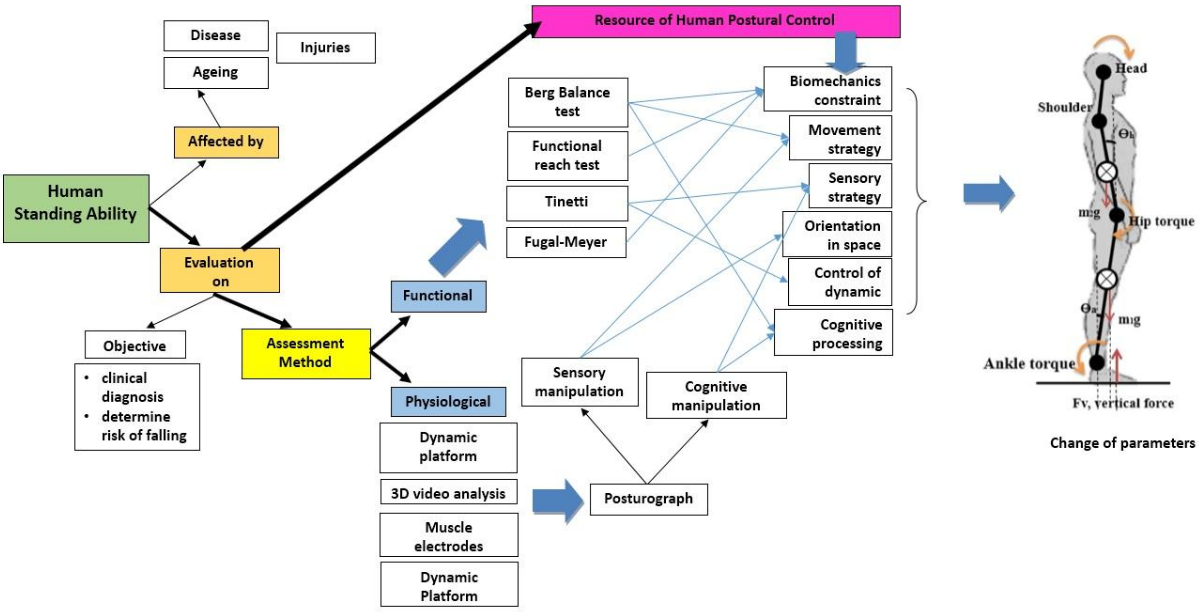


Figure 12 Research Framework

According to information gather in pervious chapter, it suggested that the change of human posture control resource can be seen through changes joint motion orientation when facing any external perturbation. Therefore, a proper research design is warranted in order to determine reliable characteristic. Recently, this concept have gather interest from other research group in order to provide a better solution for the existence method [76].

3.2 Research Design

In this section, a detail explanation on the research design will be discussed. This research design is created by considering all important aspect in determine the relationship between posture modulation and balance ability. Not to forget, the opinion from stakeholders about the existence concept of assessment that have been widely used in clinical field.

For this study, research process can characterized in three phase such as investigation, clinical study and modelling phase. Each phase was created with its own objective as mentioned detail in section below;

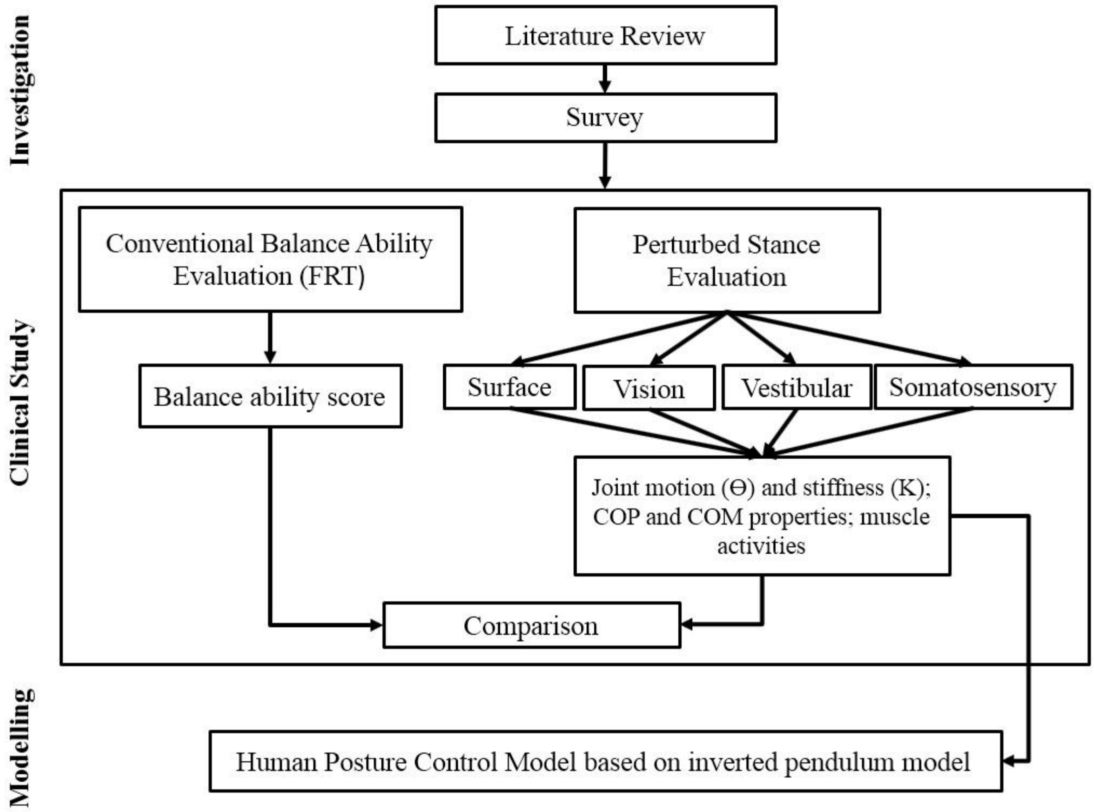


Figure 13 Flowchart of research design

3.2.1 Investigation Phase

In the investigation phase, literature from previous studies was discussed and compared in order to establish a theoretical framework about the study as mentioned in previous section. Here, all important parameter, key terms and terminology was identified. Then, a survey was conducted in order to hear the opinion on physiotherapist.

Here, a survey form regarding their opinion on preferred assessment to asses balance ability and the use of computerized system in clinical field was distributed to physiotherapist. Feedback from them were then analysed and adapted for initial evaluation of balance ability.

3.2.2 Clinical Study Phase

After that, it continued with a clinical study. In this phase, physical and physiological data were recorded. It is understood that in order to observe any change in human balance ability, there must be an external force whether voluntary or involuntary; or any specific task that required to trigger response of human posture control. According to pervious research, lower extremities plays an important role in balance. Ascending sensory pathway from sole then regulates muscle activation to initiate ankle motion and then, activated hip motion and finally the upper body including trunk and head [77] . Without a sufficient activation or response from lower extremities, it is difficult to the nervous system to provide a feedback to generate an efficient strategy and then preventing from falling. Therefore, it is believed that perturbation at surface level is sufficient enough to challenge posture control system. This approaches also have been adapted in dynamic posturography assessment as mentioned in previous chapter (Table 1). Furthermore, bipedal standing position also have widely apply as mentioned in Chapter 2. This perturbed stance experiment also conducted with three sensory manipulation which are vision, vestibular sense and somatosensory sense. These approaches were done in order to observe changes in human body motion towards limited sensory input. Besides, this method was applied in order to mimic condition experience by elderly people. Pattern of posture motion parameter such as joint sway angle, joint stiffness, both COP and COM properties together with muscle activation were gathered and compared with the balance ability score from Functional Reach Test (FRT).

For the experiment equipment; a 6-axis movable platform (MB-150, Cosmate, Japan) was used to introduce a surface type perturbation to the subject. This platform includes six electric cylinder that can expands and contracts that allow the user to create six degree of freedom (DOF) of movement. Furthermore, both platform motion displacement and frequency also can be regulated. On the other hands, a motion capture system by seven high-precision infrared cameras (HWK-200RT camera, Motion Analysis, USA) at a sampling frequency of 200 Hz, force plate (9286A, Kistler, Japan), a electromyography (EMG) and 3-axis force sensor with a built-in amplifier (MFS20-010, Liniax, Japan) were used for recording physiology and posture motion changes during the experiment. Each device were connected to A/D converter.

3.2.2.1 Participant

In this study, only healthy young male subjects aged 25.24 ± 2.19 years old participated. These is because, posture modulation pattern of normal individual is essential task of this study. With the existence of data from normal individual, further comparison between normal and individual with health problem can be done. Each subject was fully briefed regarding the possible risk and each provided written confirmation of informed consent prior to participation, in order to comply with the Declaration of Helsinki. On the other hand, subject details such as age, height, weight, and vision’s quality, history of falls and health history were recorded for references.

3.2.2.2 Subject Preparation and Data Recording

There are a few setting were implemented for subject preparation. These included number of reflective marker used for body motion recording using motion capture system. In this study a fix of 19 reflective markers were attached on subject's joints. These marker location details as described in Table below;

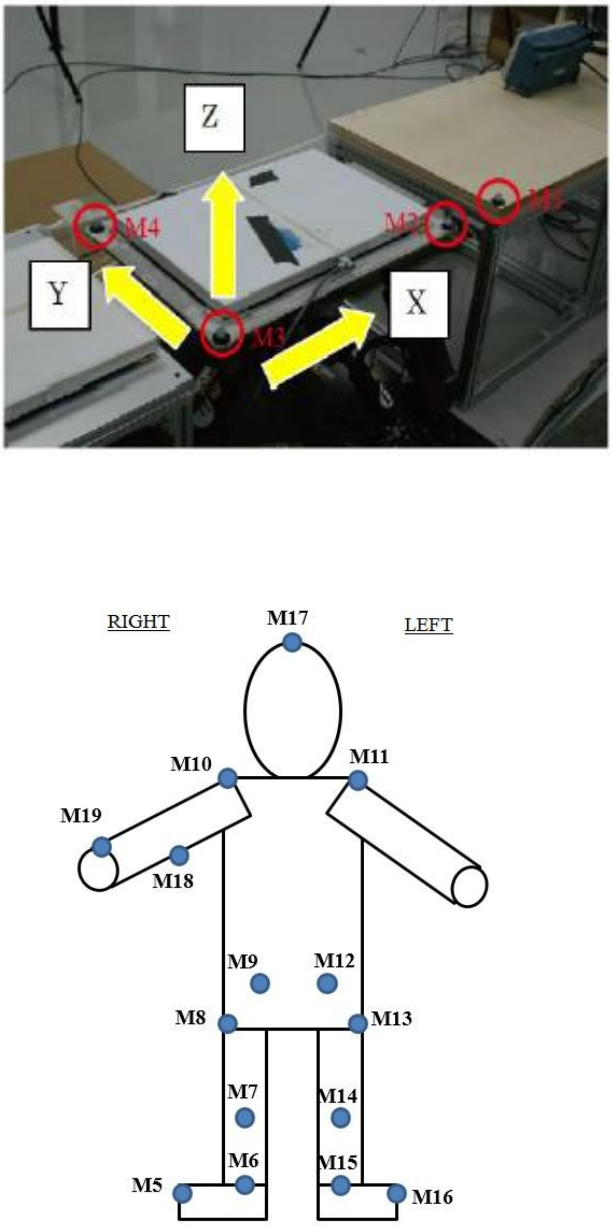


Figure 14 Reflective marker location (left) on moveable platform and (right) subject's body.

Besides, during all experiment trials, both right and left knee joints locked using wood splints to prevent bias movement at the knees. This approach is implemented so that a clear ankle or hip strategy can be obtained. Figure below described how the splint was implemented.

Table 3 Reflective marker location

Marker Name	**Location**
M1, M2,M3,M4	On moveable platform
M5, M16	3rd metatarsal
M6, M15	Lateral malleolus (ankle)
M7, M14	Lateral condyle
M8, M13	Trochanter of femur (hip)
M9, M12	Iliac crest (pelvic)
M10, M11	Acromion of scapula (shoulder)
M17	Top of head
M18	Olecranon bursa (elbow)
M19	Ulna (wrist)

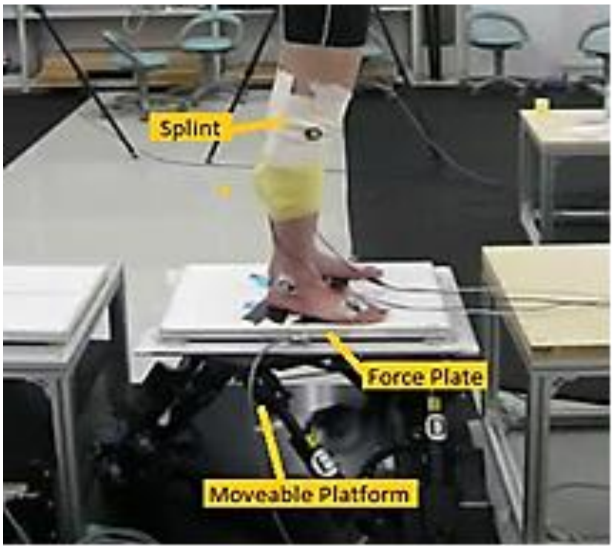


Figure 15 The used of splint to avoid knee movement

Furthermore, subject body also was attached with electromyography (EMG) electrodes to capture muscle activities while subject tried to maintain their position on the moving platform. The electrodes was positioned at five different muscles at lower extremities which are bicep femoris (BF), rectus femoris (RF), tibia anterior (TA), and medial gastrocnemius (MGAS) were recorded at a sampling frequency of 1 kHz. These muscles play an important role in initiating or limiting movement at joints. Figure 16 below illustrates the muscle location.

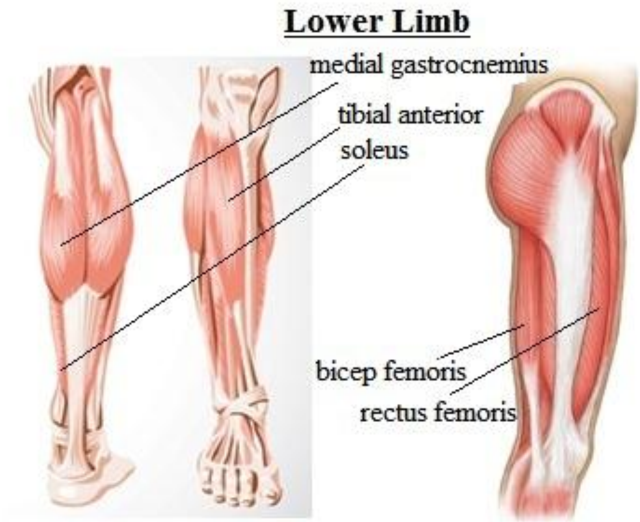


Figure 16 EMG electrode's location on subject's lower limb.

3.2.2.3 External Perturbation

Subjects were asked to stand on the experiment platform where external surface perturbation applied. With both hands were crossing, they required to maintain their balance position for a specific period. During the experiment session, posture motion and muscle activities were recorded simultaneously. However, experiment trial will be re-record if subject was almost felt, or initiated stepping. Table 4 below describes setting for external perturbation used and Figure 17 shows how the external perturbation works;

Table 4 Perturbation Setting

Perturbation	Setting
Direction	x-direction (anterior- posterior) (translation, T)
	z-direction (superior-inferior) (tilt up-tilt down, TT)
Displacement	70 mm for T
	6 degree for TT
Frequency	Four different frequencies (0.2,0.4, 0.6, and 0.8 Hz)

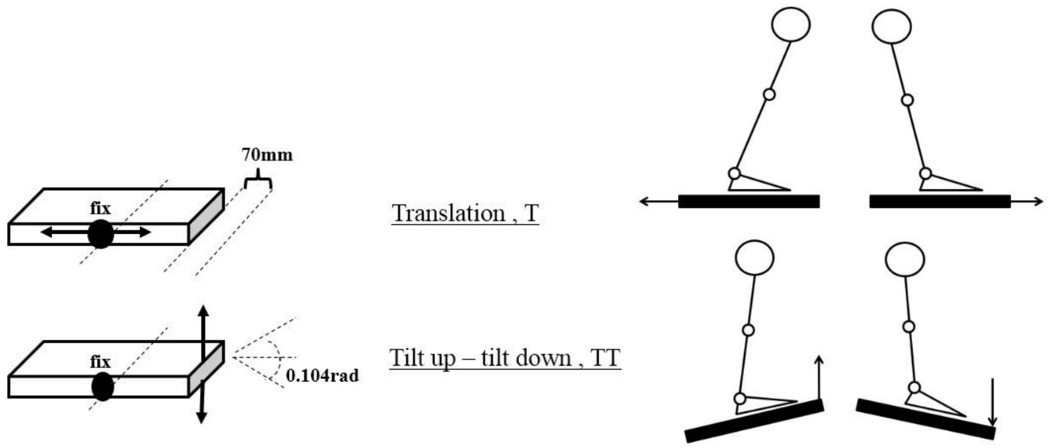


Figure 17 Platform movement for both type of perturbation and illustration of body movement during perturbation applied.

3.2.2.4 Sensory perturbation

Other than facing external surface perturbation, subject was also facing some sensory manipulation condition in order to determine effect of specific sensory information to the posture control system. This research tried to understand the effect of main sensory input towards the change in posture modulation scheme and muscle strength. The sensory system includes vision sensory, vestibular sensory and somatosensory. Sensory perturbation was applied non-invasively without any insertion of external device inside the body.

For vision sensory manipulation, the subject were requested to stand while eyes-opened (EO) and eyes-closed (EC). During EO, the subject were asked to fix their glance at ‘X’ mark on the wall located one metre in front of the subject. By doing this, the effect of vision existence in influencing balance ability can be determined.

Besides, for vestibular sense manipulation, according to previous research, the vestibular system input can be manipulated by constricting the movement of the neck and head. The use of neck collar (ADFIT collar, ADVAN FIT) was implemented during the experiment in order to created disturbance to vestibular sense. This supported by previous research that this step interfered with the function of proprioceptors in the neck muscle, thus leading to vestibular malfunction [11].

Figure 18 The use of neck collar on subject during vestibular sense manipulation experiment.

Current interest on the effect of additional sensory information from hand have been increased. This have encourages investigation on effect of additional somatosensory from hand touch towards change in posture modulation changes. In this research, additional set up improvement have been added in order to determine the changes. A pole at level of waist position is used to provide a place to touch. This pole then was imbedded with force sensor (MFS20-010, Liniax, Japan) on the top to detect force created by fingertip while facing external surface perturbation as shown in figure below;

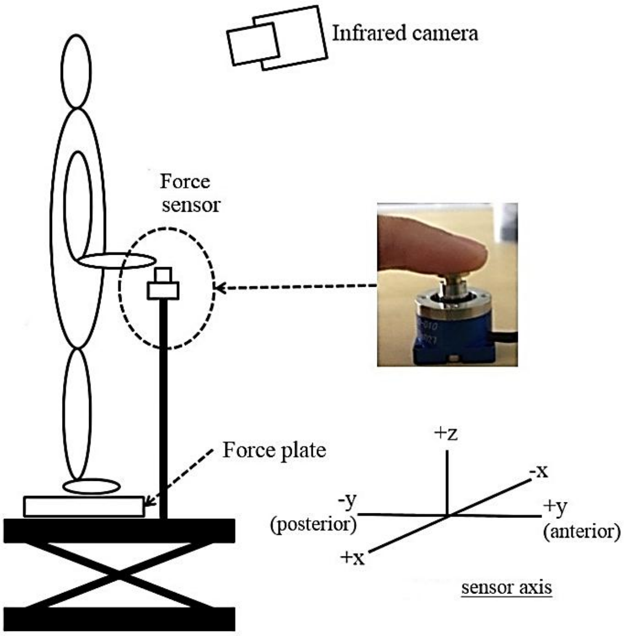


Figure 19 Illustration of experiment setup (inset indicates details of force sensor)

3.2.2.5 Data processing

Raw data from each of recording devices especially during clinical study phase need to be processed in order to eliminate noise that might occur during recording period However, markers data from motion analysis system need to undergo its own post processing in order to create and gather marker's name and marker's coordinate. All disconnected frame of marker position need to be corrected and smoothed to eliminate spike. Furthermore, the other raw data from the force plate, the EMG and force sensor, each of them were then filtered with were filtered by a second-order Butterworth filter with a 60 Hz cut-off to eliminate noise especially from power line and movement. Each data were then resampled to sampling frequency of 200 Hz.

After that, each data undergo separate process to obtain amplified output value. For force plate, each value produced by each strange gauges were then summed up to obtain vertical force as equation below;

$$F1Z1\ [N] = (raw\ data) x\ range\ [V]\ x\ 2\ x\ 4096\ x\ force\ plate\ constant$$

$$F1Z2 = \ \dots\dots\dots\dots\dots\dots..$$

$$F1Z3 = \ \dots\dots\dots\dots..$$

$$F1Z4 = \cdots \dots..$$

$$FZ = F1Z1 + F1Z2 + F1Z3 + F1Z4$$

Range value was obtained from the recording setting on the device and force plate constant value can be obtained from the force plate datasheet. This measurement also applied to the EMG data as mentioned below;

$$EMG\ [mV] = (raw\ data) x\ range\ [V]\ x\ 2\ x\ 4096$$

Similar with force plat, range value was obtained from the initial recording setting. Moreover, for the force sensor (MFS20-010, Liniax, Japan), calibration data from manufacturer. The voltage analogue data obtained then calculated using the equation below to determine the amount of force recorded.

$$Fz\ [N] = \frac{(F - 2.5V)}{0.4\ V/N}\ ,\ Fy\ [N] = \frac{M - 2.5V}{0.1\frac{V}{Ncm}} x\ 3cm\ \ ,\ Fx[N] = \frac{M - 2.5V}{0.1\frac{V}{Ncm}} x\ 3cm$$

Where *Fz* is vertical force, *Fy* is horizontal force at y-direction and *Fx* is horizontal force at x-direction.

3.2.2.6 Data analysis

Several analysis method were applied in this study especially to determine the posture modulation pattern. In general, a common statistical analysis (mean and standard deviation) was used to describe subject population results. Each result were then compared using One way ANOVA with Turkey post hoc test at a significant level of $p<0.05$ to determine a significant different between data produced.

This study focused on measuring the amount of stiffness at both the ankle and hip joints where it were obtained by using the equation below. Joint stiffness was measured according to the free body diagram in Figure 7, Chapter 2 and equation below. The ground reaction force (*Fv*), the horizontal component in y-direction direction force (*Fy*) and force plate moment at the x-axis (*Fx*), recorded from force plate as shown in Figure 17. Joint movement coordinate (x, y, z) obtained from motion analysis system were used to measure joint sway angle (θ) and body segment length (*h*) for segmental centre of mass locations. The average joint stiffness was measured from the torque of the ankle joint (*Ka*), torque of the hip (*Kh*) along with the period of recording time (*t*). The COP displacement was determined from (3) below where *dz* was the distance from the surface to the platform origin. The COM was obtained from the total segment torque as mention in (3). Besides, COP and COM properties such as displacement, velocity and range can be obtained from reading from force plate following equation below;

Joint torque:

$$\tau_{ankle} = m_1 g h_A \sin\theta_a, \tau_{hip} = m_2 g h_H \sin\theta_h \quad (1)$$

Assume that ;

$$m_n g \approx \beta_{seg} F_v$$

where β_{seg} is percentage from Plagenhoef's Body Segment Weight data

Joint stiffness at ankle (Ka) and hip (Kh):

$$Ka(Nm/rad) = \frac{\tau_{ankle}}{\Delta\theta_a}, Kh(Nm/rad) = \frac{\tau_{hip}}{\Delta\theta_h} \quad (2)$$

COP and COM displacement in anterior – posterior direction:

$$M_x = a\,(f_{z1} + f_{z2} + f_{z3} + f_{z4}),$$

$$COP_{A-P}(mm) = \frac{M_x - (F_y \,.\, d_z)}{F_v}, COM_{A-P}(mm) = \frac{\sum(F_v.\ \beta_{seg}).\ h_{seg}}{F_v} \quad (3)$$

Where a= sensor offset value, d_z =thickness parameter of force plate

COP and COM velocity:

$$COP_v(mm/s) = \frac{\sum_{n=1}^{N} COP_{A-P}(n+1) - COP_{A-P}(n)}{t},$$

$$COM_v(mm/s) = \frac{\sum_{n=1}^{N} COM_{A-P}(n+1) - COM_{A-P}(n)}{t} \quad (4)$$

COP and COM range:

$$COP_r\,(mm) = COP_{A-P}(max) - COP_{A-P}(min), \quad (5)$$

$$COM_r(mm) = COM_{A-P}(max) - COM_{A-P}(min)$$

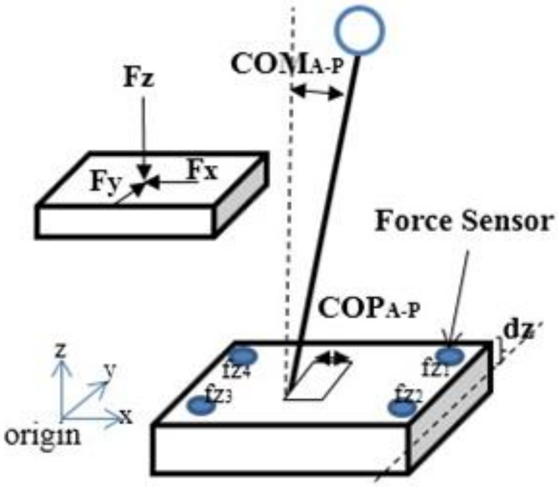


Figure 20 Force distribution on force plate

It is expected that along the repetitive external perturbation, a normal individual may shows some changes especially at the amplitude of sway angle and stiffness amount. Therefore, this kind of variation or adaptation also need to take into account. The adaptation strategy of CNS towards the joint stiffening response was determined by measuring the area under graph (AUG) using the trapezoidal rule, applying Eq. (6) and Eq. (7) as shown in Figure 19. K (t) was joint stiffness along the perturbation period, where *t* was the time for one cycle of perturbation and *i* was the number of cycles as shown in Fig. 3 (Right).

Area under graph (AUG):

$$AUG = \int_0^t K(t)\ dt \tag{6}$$

Adaptation percentage (%):

$$Adaptation\ (\%) = \frac{AUG_i - AUG_{(i+1)}}{AUG_i} x\ 100\% \tag{7}$$

$$i = 1,2,3 \ldots.$$

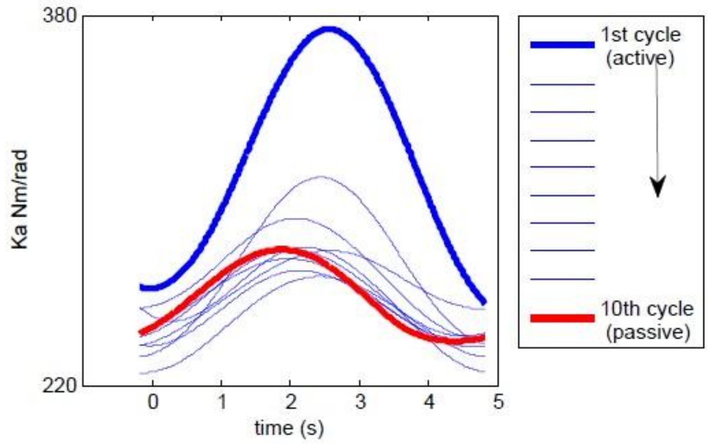


Figure 21 Example of stiffness pattern at ankle joint during 0.2 Hz of translation for Subject 1 and this windowing technique have been used to measure adaptive percentage to detect motor learning ability where first cycle was define as active mechanism meanwhile the rest as passive.

Relationship between FRT's score and body sway parameter such as joint stiffness was computed using linear regression fit. Correlation analysis was done using the Pearson function. Moreover, to determine the stability region estimation, density plots of ratio which is the comparison between amount of stiffness during perturbed and unperturbed using density plot. All measurements of stiffness, COM-COP properties, estimation of stability region, adaptation analysis and statistical analysis were completed using the MATLAB software.

3.2.3 Modelling phase

Parameter and pattern gathered from the clinical study were then simplified and describe in equation form in order to develop the mathematical model. Equation of double pendulum model was referred as main equation as mentioned in Chapter 2. Stiffness and torque component were altered according to clinical data. Comparison then were done between simulation results and actual data from experiment in order to determine the accuracy of the model. The developed model is hope to be able to be used for future development of 'hybrid' assessment method.

For the development of simulation model of double inverted pendulum, equation of motion for the system were derived. The x and y component of joint displacement were derived as below;

$$x_1 = l_1 \sin\theta_1$$

$$y_1 = l_1 \cos\theta_1$$

$$x_2 = l_2 \sin\theta_2 + l_1 \sin\theta_1$$

$$y_2 = l_2 \cos\theta_2 + l_1 \cos\theta_1$$

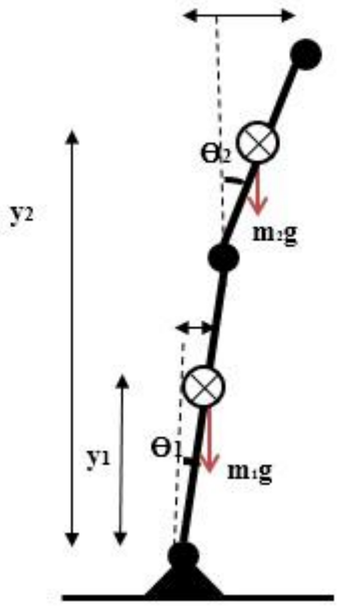


Figure 22 Double inverted pendulum diagram

Potential and kinetic energy of the system;

a. Potential energy, $PE = mgh$

$$= m_1 g\, l_1 \cos\theta_1 + \ m_2 g\ (\,l_2 \cos\theta_2 + l_1 \cos\theta_1)$$

$$= (m_1 + m_2) g\, l_1 \cos\theta_1 + \ m_2 g\ \ l_2 \cos\theta_2$$

b. Kinetic energy, $KE = \frac{1}{2} m v^2$

$$v^2 = \dot{x}^2 + \dot{y}^2$$

$$v^2{}_1 = \ l_1{}^2 \cos^2 \dot{\theta_1}^2 + \ l_1{}^2 \sin^2 \theta_1\, \dot{\theta_1}^2$$

$$v^2{}_1 = \ l_1{}^2 \dot{\theta_1}^2\ (\cos^2\theta_1 + \ \sin^2\theta_1)$$

$$v^2{}_1 = \ l_1{}^2 \dot{\theta_1}^2$$

$$v^2{}_2 = (l_2 \cos\theta_2 + l_1 \cos\theta_1)^{\ 2}\, \dot{\theta}_2{}^2 + (\, l_2 \sin\theta_2 + l_1 \sin\theta_1\ )^2\, \dot{\theta}_2{}^2$$

$$v^2{}_2 = l_2{}^2 \dot{\theta}_2{}^2 + l_1{}^2 \dot{\theta}_1{}^2 + 2 l_2 l_1\, (\cos(\,\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 + \sin\,(\theta_1 + \theta_2\,)^2\, \dot{\theta}_2{}^2)$$

Using Langrangian concept;

$$L = KE - PE$$

$$= \frac{1}{2} m v_1{}^2 + \frac{1}{2} m v_2{}^2 - (m_1 + m_2) g h - m_2 g h$$

$$= \tfrac{1}{2} m_1 l_1{}^2 \dot{\theta}_1{}^2 + \tfrac{1}{2} m_2 [l_2{}^2 \dot{\theta}_2{}^2 + l_1{}^2 \dot{\theta}_1{}^2 + 2 l_2 l_1 \left(\cos(\,\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 + \sin(\theta_1 + \theta_2\right)^2\ \dot{\theta}_2{}^2)] - (m_1 + m_2) g l_1 \cos\theta_1 - m_2 g l_2 \cos\theta_2 \ \ldots\ldots\ldots \ (8)$$

Then, simplified with Euler-Lagrange differential equation;

$$\frac{\partial}{\partial t}\left\{\frac{dL}{d\dot{\theta}_1}\right\} - \left\{\frac{dL}{d\theta_1}\right\}$$

$$\frac{\partial}{\partial t}\left\{\frac{1}{2} m_1 l_1{}^2 \dot{\theta}_1{}^2 + \frac{1}{2} m_2 [l_2{}^2 \dot{\theta}_2{}^2 + l_1{}^2 \dot{\theta}_1{}^2 + 2 l_2 l_1 \left(\cos(\,\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 + \sin(\theta_1 + \theta_2\right)^2\ \dot{\theta}_2{}^2)]\right\} - \{(m_1 + m_2) g l_1 \cos\theta_1\}$$

$$= (m_1 + m_2)\, l_1\, \ddot{\theta}_1 + m_2 l_2 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_2 \sin(\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 + (m_1 + m_2) g l_1 \sin\theta_1 \ \ldots\ldots\ldots \ (9)$$

$$\frac{\partial}{\partial t}\left\{\frac{dL}{d\dot{\theta}_2}\right\} - \left\{\frac{dL}{d\theta_2}\right\}$$

$$\frac{\partial}{\partial t}\left\{\frac{1}{2} m_2 l_2{}^2 \dot{\theta}_2{}^2 + m_2 l_2 l_1\, \cos(\,\theta_1 + \theta_2)\, \dot{\theta}_1\, \dot{\theta}_2 + m_2 l_2 l_1 \sin(\theta_1 + \theta_2\,) \dot{\theta}_1\, \ddot{\theta}_2\,]\right\} - \{m_2 l_2 \cos\theta_2\,\}$$

$$\frac{\partial}{\partial t}\left\{\frac{1}{2} m_2 l_2{}^2 \dot{\theta}_2{}^2 + m_2 l_2 l_1 \cos(\theta_1 + \theta_2)\, \dot{\theta}_1\, \ddot{\theta}_2 + m_2 l_2 l_1 \sin(\theta_1 + \theta_2)\dot{\theta}_1\, \ddot{\theta}_2\,]\right\} - \{m_2 l_2 g \sin\theta_2\}$$

$$= m_2 l_2\, \ddot{\theta}_2 + m_2 l_1 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_1 \sin(\theta_1 + \theta_2)\dot{\theta}_1{}^2 + m_2 g \sin\theta_2 \ \text{.......} \ (10)$$

For θ_1;

$$= (m_1 + m_2)\, l_1\, \ddot{\theta}_1 + m_2 l_2 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_2 \sin(\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 + (m_1 + m_2) g l_1 \sin\theta_1$$

$$\ddot{\theta}_1 = \frac{- m_2 l_2 \sin(\theta_1 + \theta_2)\, \dot{\theta}_2{}^2 - (m_1 + m_2) g l_1 \sin\theta_1}{(m_1 + m_2)\, l_1 + m_2 l_2 \cos(\theta_1 + \theta_2)}$$

.... (11)

For θ_2

$$= m_2 l_2\, \ddot{\theta}_2 + m_2 l_1 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_1 \sin(\theta_1 + \theta_2)\dot{\theta}_1{}^2 + m_2 g \sin\theta_2$$

$$\ddot{\theta}_2 = \frac{m_2 l_1 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_1 \sin(\theta_1 + \theta_2)\dot{\theta}_1{}^2 + m_2 g \sin\theta_2}{m_2 l_2}$$

.... (12)

Constant value for $(m_1, m_{2\,,}\, l_1, l_2)$ and initial angle of joint (θ_1, θ_2) were determined based on experiment data

CHAPTER 4

RESULTS

4.1 Subject Details and Balance Ability

Overall of 26 young subject were participated in this research. Details of subject physical background were described in table below. This is included subject data gathered from previous research (n=9). All subject recorded free from any neurological deceased and do not have any pass history of falling. All subject have signed the consent letter and fully aware with the procedure and risks while underwent this experiment.

Table 5 Subjects physical details

Details	
Gender	Male (n=26), Female (n-=1)
Age	24.14 (± 2.19) years old
Height	173.57 (±6.24) cm
Weight	67.56 (±2.25) kg

Before experienced the perturbed stance, subject were asked to evaluate with FRT to determine their balance ability. All subject were able to accomplish this test perfectly. The subjects' scores were within an acceptable balance range which is >25.4 cm that indicating adequate balance ability [78]. Further discussion on balance ability will be discussed at the next section.

4.2 Joint Stiffness in relation with balance ability

Figure 23 below indicated regression analysis to observe the relationship between FRT's score and stiffness. Even though all of the subjects' scores were within an acceptable balance range (>254 mm indicating adequate balance ability according to Duncan et al, (1990), correlation made between the average stiffness value at low perturbation frequency (0.2 Hz) and the FRT scores have shown a significant trend to supported the hypothesis [78]. Subjects with high scores produced less joint stiffness compared to low scorers. Based on the results, it was suggested that patients with less balance ability have stiff joints at both the ankle and hip.

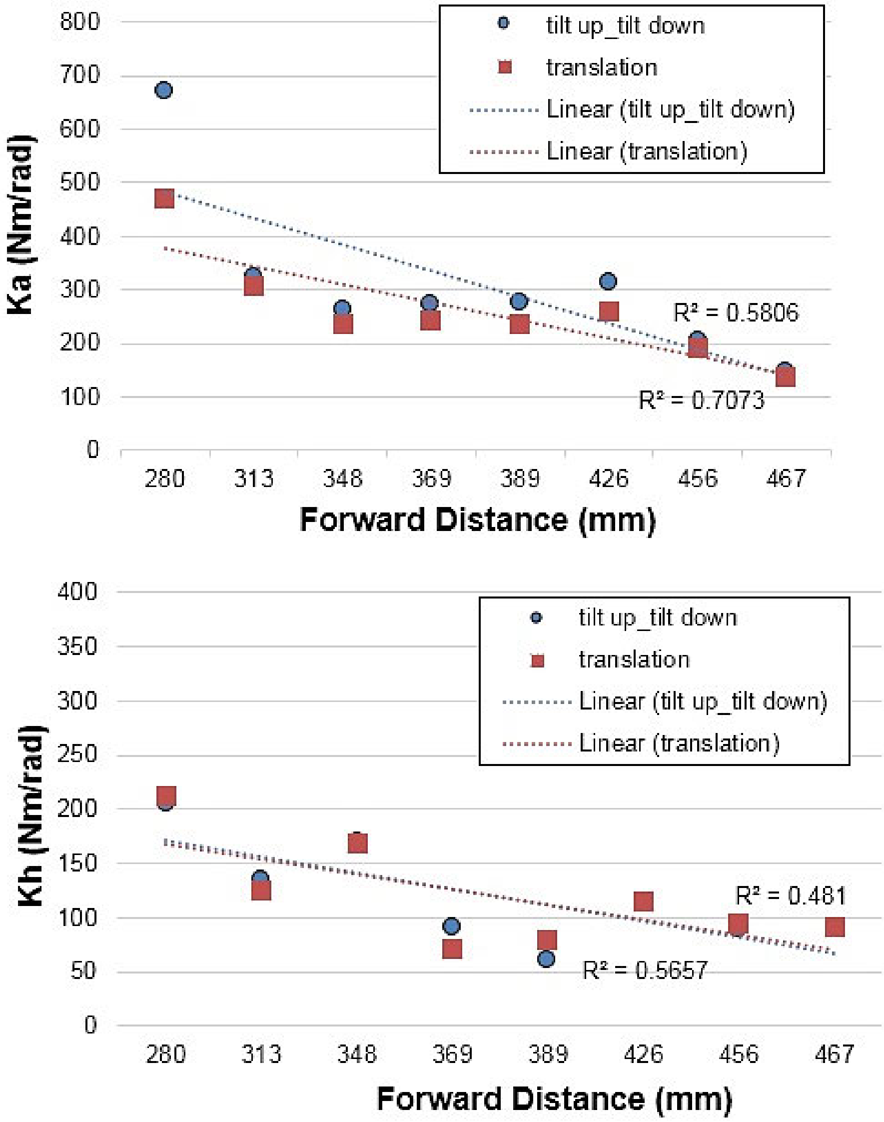


Figure 23 Joint Stiffness versus the FRT scores at ankle (*Ka*) and hip (*Kh*) joint during 0.2 Hz which is the lowest perturbation's frequency (n=8). The comparison using linear curve fitting shows the opposite relationship between balance ability and joint stiffness.

4.3 Analysis of joint stiffness of human posture in response to balance ability and limited vision and vestibular sense input during dynamic perturbation

4.3.1 Body sway under various condition

Body sway due to various condition was determined based on COP and COM properties. Figure 24 below have shown that effect of varies perturbation type, sensory condition and frequency towards COP-COM displacement range and velocity. By comparing between types of perturbation, COP range (COPr) at tilt up-tilt down was higher than the other perturbation. Meanwhile, the COM range (COMr) was greater during translation perturbation. Between different frequency, the COPr increased with frequency and a significant different between frequency was only observed during translation perturbation. For the COMr, it reduced with the increase of frequency, however, no significant different between frequency ($p>0.05$) ($F (3, 60) =1.19$, $p=0.12$). By comparing between difference sensory condition, significant different (O vs. sensory) was observed only with no vision input (*C*). Furthermore, it is observed that without vision input (*C*), a high COPr and COMr were produced compared to other sensory conditions. Meanwhile, *NO* condition does not differ much from *O*. This finding suggested that without vision, body sway more regardless type of perturbation.

On the other hand, COP-COM velocity were significantly different between different frequencies ($p<0.05$) ($F (7,160) =6.9$, $p = 0.004$) and ($F (7,160) =12.25$, $p = 0.032$) respectively. However, it were insignificant different between sensory condition. Comparison between these two types of perturbation, COP moved significantly during superior-inferior movement meanwhile during posterior-anterior movement for COM.

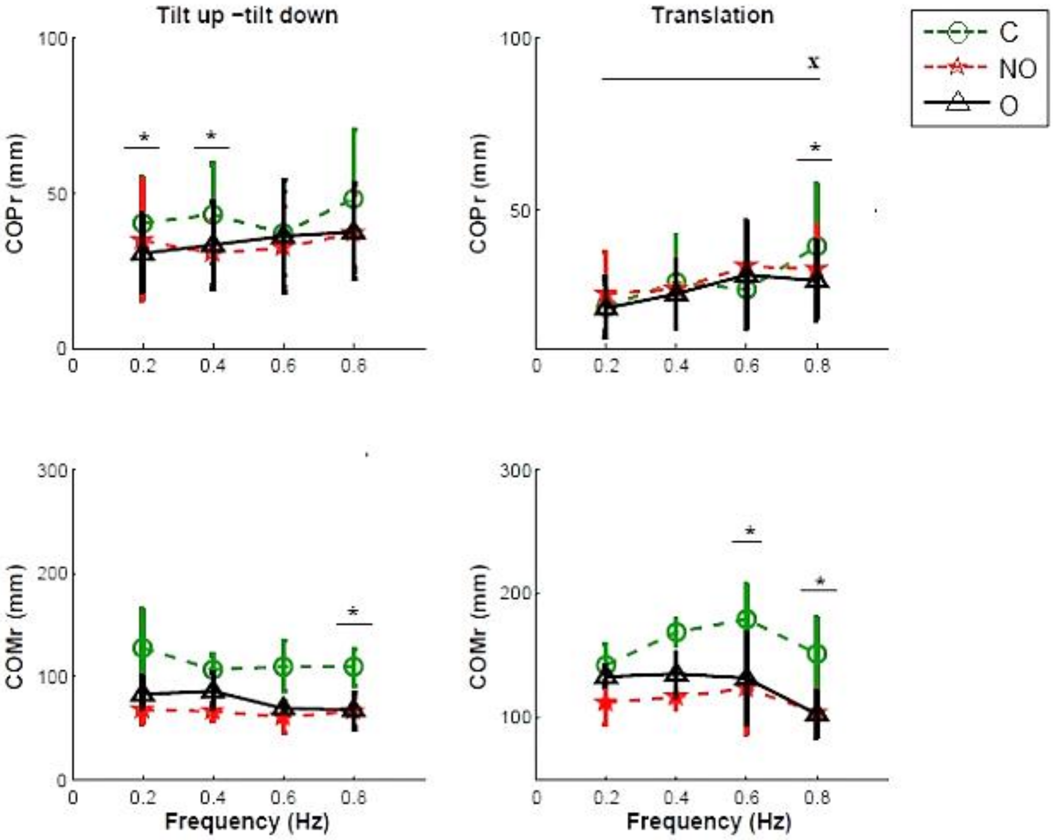


Figure 24 Both COP and COM displacement range at both type of perturbation. The (*) indicated O has significant different with C ($p<0.05$), the (^{x}) indicated significant with frequency different.

4.3.2 Joint Stiffness Pattern over different sensory condition

Based on Figure 25, in general, ankle joint stiffness (K_a) increased for about 10 percent with the increase of perturbation frequency; with correlation $R^2>0.5$ as shown in Table 1 during both perturbation. A significant different was found during translational perturbation ($p<0.05$) (F (3, 36) =6.59, $p=0.004$). Meanwhile, hip joint stiffness (K_h) shown a small decreased for about 1 percent with the increase of perturbation frequency with negative correlation for both perturbation (Table 1). At normal condition (*O*), a coaction strategy between both joint was observed as correlation become negatives (Table 1).

Analysis between sensory manipulation condition have shown that a significant different was only found at K_a during eyes closed (*C*) at certain frequency and perturbation as shown in Figure 25. Without vision sensory input (during *C*), average stiffness at both joints were observed to be higher (increased for about 20 percent) than the normal condition (*O*). The effect of limited vision input on the produced ankle joint stiffness was not different between the applied variant surface perturbations.

However, mainly no significant different found for *NO*. The use of the neck collar was observed to effectively limit the head movement as the range of head movement was smaller; about 40 percent than normal conditions ($p<0.05$) (Figure 26). Furthermore, during normal conditions (*O*), the observed head motion varied according to surface orientation. As perturbation frequency increased, head movement during the tilt up-tilt down increased but it was reduced during translation perturbation. When the neck collar were applied, the head motion was reduced with the increase of perturbation frequency during both perturbations. By analysing stiffness value produced, only the ankle joint was observed to be more stiffened during the *NO* condition; which at 0.4 and 0.6 Hz. For the hip joint, sensory input manipulation condition (both *C* and *NO*) did not showed any significant different to value of stiffness produced at *O* condition.

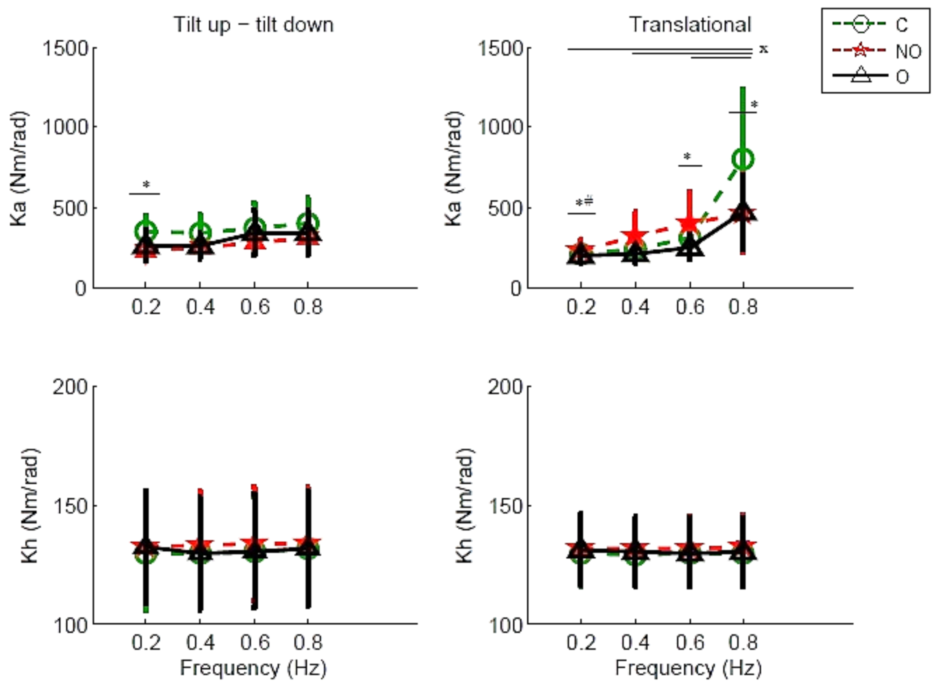


Figure 25 Comparison of the average ankle and hip joint stiffness between C and O conditions at four different frequencies (0.2, 0.4, 0.6, and 0.8 Hz). The (*) indicated O significant different with C ($p<0.05$), the ($^{\#}$) indicated O significant different with NO ($p<0.05$) and the (^{x}) indicated significant with frequency different ($p<0.05$)).

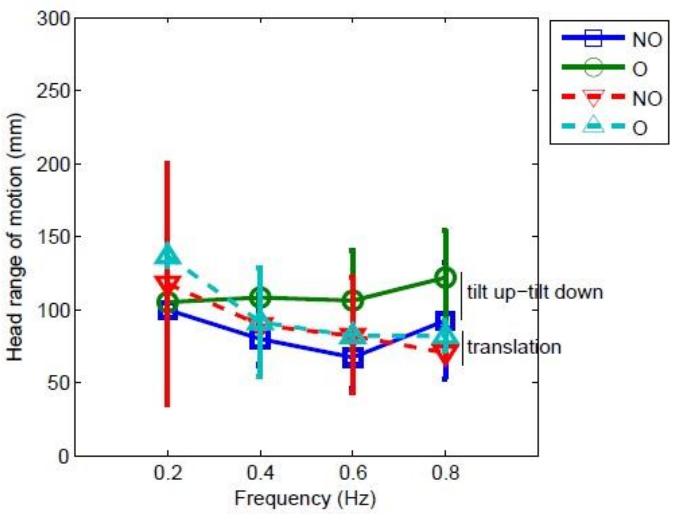


Figure 26 Range of motion for head (mm) during normal condition O and when neck collar were applied (NO).

4.3.3 Correlation between Joint Stiffness and COP-COM Properties

Both perturbations generated various COP-COM properties at the posterior-anterior plane. Joint stiffness also correlated differently with COP-COM properties. Both hip and ankle were observed to negatively correlate with COM range (COMr) at almost all conditions which meant that high stiffness was required to keep COM in a small range. The K_a was noticed to have a positive correlation with COP-COM velocity at all conditions. However, the K_h differed according to the sensory condition and surface perturbation. Stiffness at the hip was negatively correlated with COPv and COMv at *O* during both perturbations. Meanwhile, it was positive at *NO* and *C* during the tilt up-tilt down, and *NO* during translation. The results illustrate in table below suggested that high joint stiffness required to reduce body sway.

Table 6 Correlation coefficient (R^2) between frequency of perturbation, $\boldsymbol{K_a}, \boldsymbol{K_h}$, COP and COM parameter

Cond.	**Tilt up–tilt down**						**Translation**					
	$\boldsymbol{K_h}$	**Freq.**	**COMr**	**COPr**	**COMv**	**COPv**	$\boldsymbol{K_h}$	**Freq.**	**COMr**	**COPr**	**COMv**	**COPv**
O												
K_a	-0.07	0.88	-0.98	0.91	0.86	0.87	-0.32	0.86	-0.98	0.53	0.82	0.87
K_h		-0.11	-0.23	0.18	-0.13	-0.12		-0.76	0.22	-0.95	-0.80	-0.74
NO												
K_a	0.92	0.95	-0.37	0.47	0.98	0.99	0.84	0.99	-0.22	0.91	0.99	0.98
K_h		0.93	-0.68	0.29	0.94	0.93		0.86	-0.63	0.57	0.86	0.86
C												
K_a	0.95	0.92	-0.37	0.46	0.91	0.92	0.26	0.86	-0.24	0.92	0.85	0.87
K_h		0.90	-0.41	0.71	0.88	0.90		-0.03	-0.68	-0.11	-0.03	-0.02

The (-ve) value indicated negative correlation.

4.3.4 Joint Stiffness and Stability

Since the subjects who participated in this study had adequate balance ability, the estimation for both ankle and hip joint stiffness value required dynamic stability, which was determined. The stability area that established the load stiffness ratio at all conditions for the ankle and hip was shown in Figure 27 below.

Based on the Fig. 8 below, the concentrated area presented the stability area where it revealed an appropriate amount of load stiffness ratio for the perturbed condition. The load stiffness ration was determined by comparing the amount of joint stiffness during standing with perturbation and without perturbation, $K_{np} = mgh \sin\theta$ where $\sin\theta \approx 1$.Thus, joint stiffness during unperturbed stance can be assumed to be h . Concerning. Concerning the tilt up-tilt down perturbation, the ratio range was between $1.0 < K_h/mgh$<1.07 and 1.0< K_a/mgh<1.8.

Regarding the translation frequency, it was in the range of $0.9 < K_h/mgh$< 1.0 and $0.8 < K_a/mgh$<1.5. According to previous research by Suzuki (2011) which performed for quiet standing, the load stiffness ratio was $\frac{Kh}{mgh} > 0.2$ and $\frac{Ka}{mgh} > 1.0$ which was smaller compared to recent results. It may be due to the external dynamic perturbation that was applied, whereby a higher ratio was required to maintain the desired position. Thus, this result suggested that for perturbation velocity less than 0.3m/s, $0.95 < K_h/mgh$<1.035 and $0.90 < K_a/mgh$<1.65 are required for optimum stability.

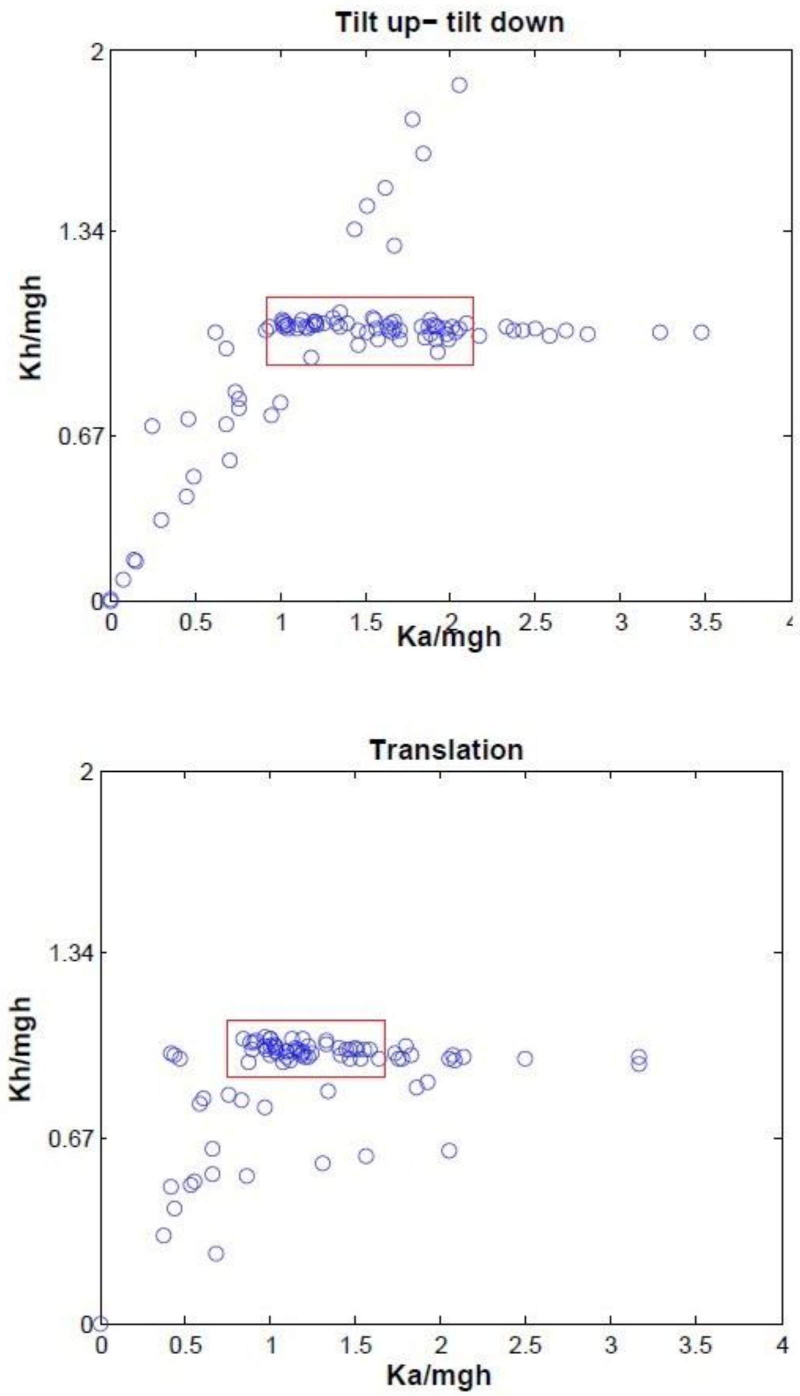


Figure 27 Density plot to determine the stability region in the (*Ka/mgh*, *Kh/mgh*) plane. Blue area (dense area) indicated the amount of stiffness where most of the subjects had applied during all conditions

4.3.5 Adaptation Ability during Repetitive Perturbation

The ability to maintain a balanced position over repetitive external perturbation can indicate the degree of motor learning ability for each individual. As previously mentioned, in our earliest studies have shown that the reduction of muscle activation amplitude and postural sway over repetitive perturbation might indicated enhanced adaptation ability [46, 79]. Analysis of moment versus angle of ankle sway of a subject at 0.2 Hz of translation perturbation have shown that the slope of linear regression at each cycle of perturbation reduced which lead to decreased of stiffness. This phenomenon have gave a clue about the existence so called adaptation strategy.

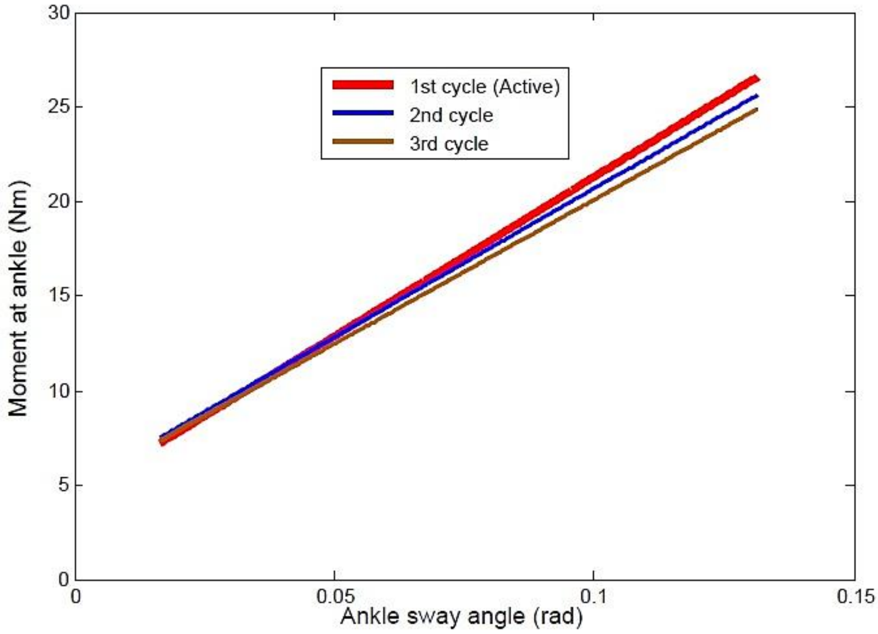


Figure 28 Graph of moment versus angle of ankle at 0.2 Hz of translation perturbation of a subject. This graph illustrates the existence of adaptation strategy response over repetitive cycle of perturbation at ankle sway.

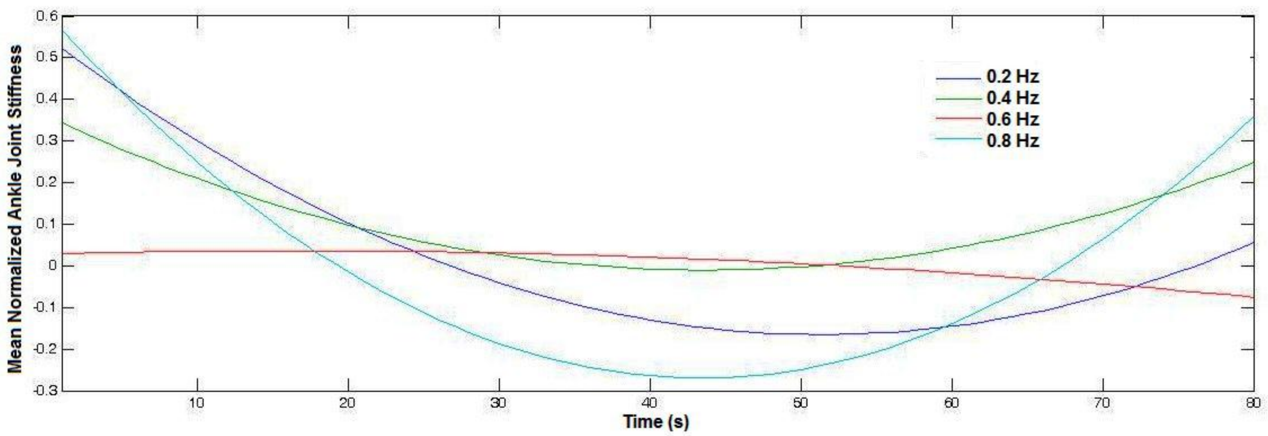


Figure 29 Estimation of ankle stiffness pattern of a subject by using polynomial curve fitting. Normalized ankle stiffness was used.

Figure 29 shows the adaptation strategy can be seen from the estimation of ankle stiffness pattern of a subject using polynomial curve fitting. However, this techniques only determine the pattern over time period. It is believed that adaptation strategy react as per movement which means that it relate closed with continuous movement. Thus, the adaptation ability was then measured by comparing the area under the graph (AUG) of each cycle of the joint stiffness response. Based on the results in Figure 30, the average adaptation percentage displayed that the normal condition (*O*) displayed better adaptation condition in comparison to the sensory manipulation condition (*NO* and *C*) as the percentage became positive. However, an insignificant difference was found. A significant difference was only discovered between various frequencies ($p<0.05$) at all perturbations (($F_{ankle,\ tt}$ (3, 36) =4.48, p=0.033), ($F_{ankle,\ t}$ (3, 36) =4.06, p=0.045), ($F_{hip,\ tt}$ (3, 36) =8.48, p=0.023), ($F_{hip,\ t}$ (3, 36) =3.39, p=0.0445)).

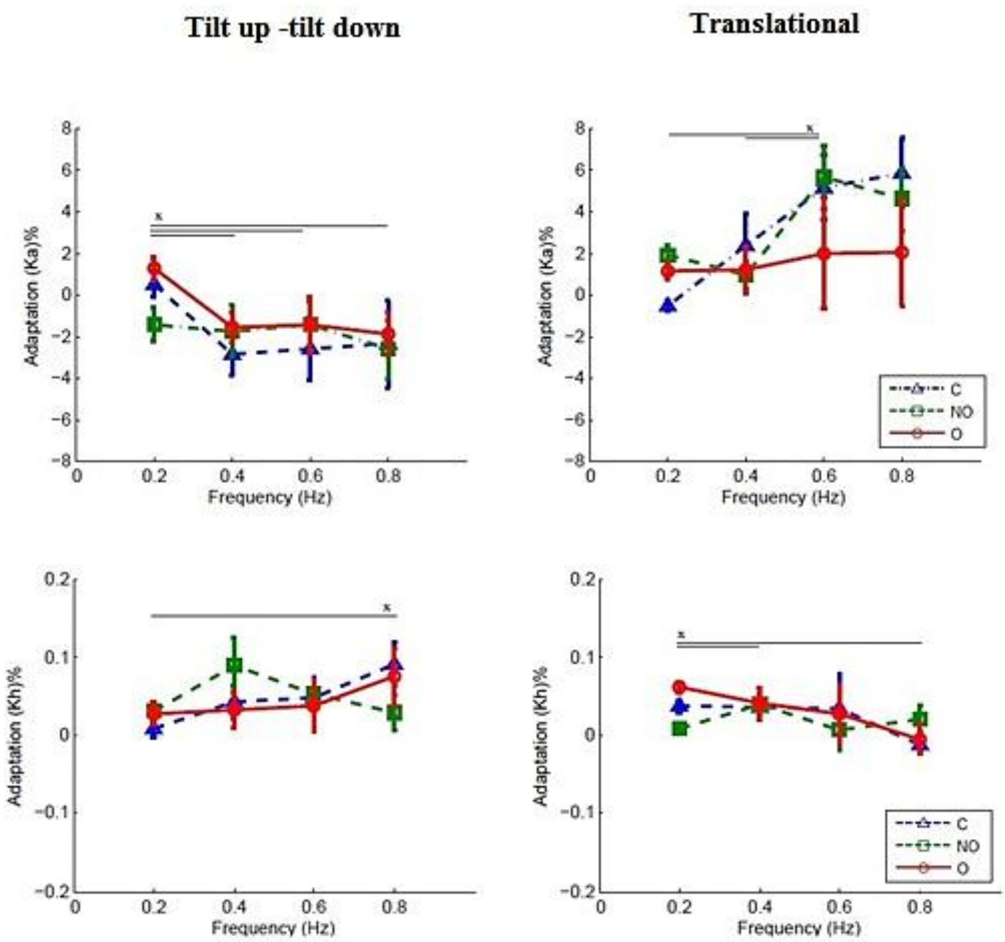


Figure 30 Adaptation percentage at different frequencies (Hz) for ankle stiffness (top row) and hip stiffness (bottom row). The (x) indicated significant with frequency different ($p<0.05$).

On average, ankle joint stiffness reduces for about 2 percent at each cycle; on the other hand, it only reduced about less than 0.1 percent for hip stiffness at the lowest frequency (0.2 Hz). This applied to both perturbations. However, with the increase of frequency, adaptation was varied; especially at the ankle joint which depended on perturbation. As the frequency increased (0.4 to 0.8 Hz), the adaptation percentage of *Ka* during the tilt up-tilt down perturbation was reduced and became more negative; however, it increased and became more positive during translational perturbation. These situations addressed the issue that not only sensory weakness affected adaptation ability, but also the frequency and type of perturbation as well.

Overall, results in this section have describes the change of stiffness value at all mentioned condition. It was observed that stiffness value change with sensory condition and over continuous movement. This particular responses information is important for development of human posture control model. Moreover, it is also observed that translation type of perturbation has triggered more posture sway than tilt up –tilt down perturbation. This can be seen in Figure 24. In order to focus more on the posture modulation scheme, analysis then will be continued with only translation perturbation.

4.3.6 Muscle activation response

As mentioned in literature review chapter, joint stiffness was reported previously to be correlated closely with muscle activation. In recent result gained from the experiment have shown that normalized value of peak muscle activation to maximum voluntary contraction (MVC) value that controlled ankle joint (TA and MGAS) increased with the increase of perturbation frequency where significant different were found between different frequency ($p<0.05$) (Figure 31). Muscle activation at hip (RF and BF) also increased with frequency. However, this response were differed with the hip stiffness characteristic where it found decreased with frequency. By comparing between different sensory conditions, similar results were obtained with stiffness response where normalized MVC improved with the improvement of sensory input. But no significant different was found ($p>0.05$).

In order, to determine whether movement at ankle and hip joint occur due to contribution of either feed forward or feedback mechanism, cross correlation analysis were done for joint movement versus muscle activation. From this analysis, time delay were obtained as shown in Figure 32. Based on the result, muscles at the ankle joint (TA and MGAS) was observed delayed the ankle motion at all frequency, thus, followed result from Finley et al, (2012) which ankle movement towards external perturbation due to feedforward control mechanism [80] . More interestingly is muscles at hip (RF and BF) changes over frequency. At frequency 0.2 and 0.4Hz of perturbation where ankle strategy applied, the RF and BF activation preceded hip motion by around 70ms. These combination of feed forward and feedback mechanism allowed hip joint to maintain its position and avoid hip to sway further. These preceding of muscle activation than joint motion or force is related to voluntary response to maintain stabilisation of joint [81]. At 0.6 Hz and above, where the shifting strategy occurred, muscle activation delayed than joint motion thus concluded the shifting strategy occurred due to involuntary response.

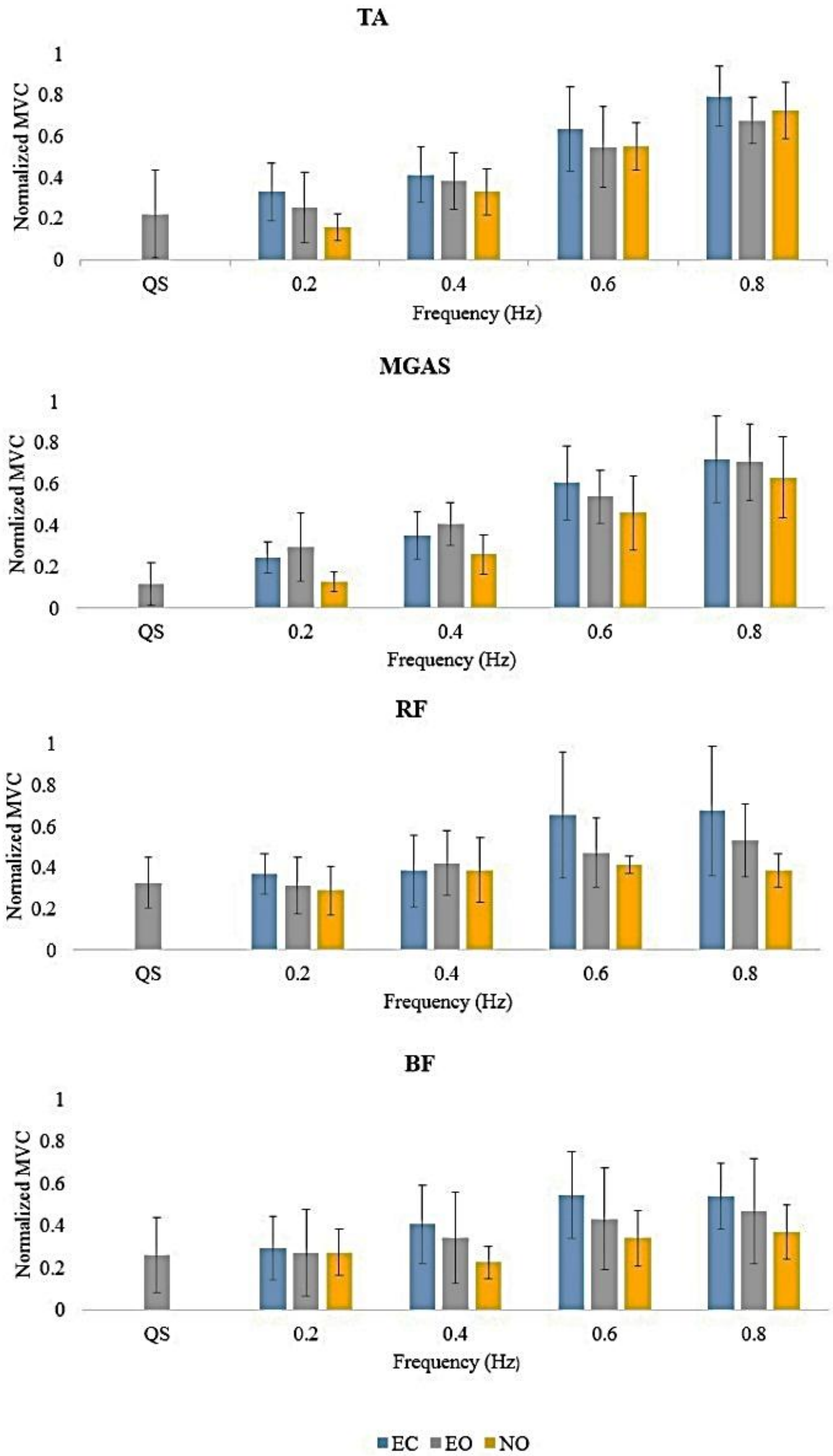


Figure 31 Normalized peak muscle activation to maximum voluntary contraction (MVC) at different perturbation frequency during translation perturbation

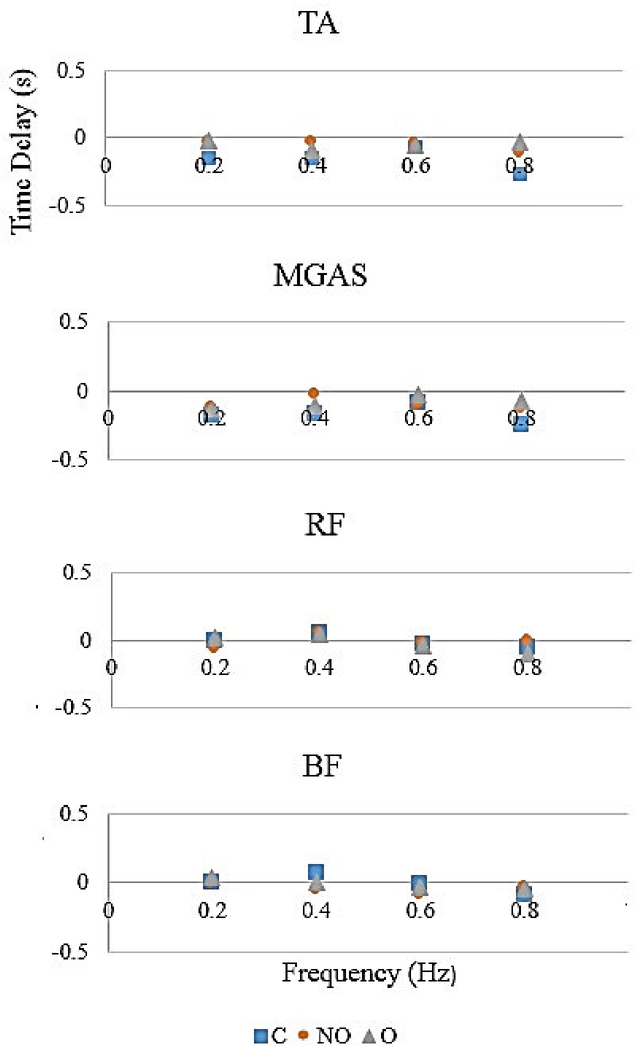


Figure 32 Time delay of joint movement vs. muscle activation. Positive value indicates joint movement precede muscle activation meanwhile negative value indicates joint movement delay muscle activation.

4.4 Analysis of human posture strategy scheme with existence of additional somatosensory input from fingertips.

In previous results, it is observed that how joint stiffness values change according to both surface and sensory perturbation. Besides, it correlation with balance ability assessment score has encouraged further analysis on posture change with existence of prominent strategy such as touch. Additional information and support, especially steadying with the hand, have a positive effect on the balance process. Handrails in buses or trains are relied upon by passengers to provide that support and prevent them from falling. Recently, interest has increased among researchers to investigate how human posture control responds to touch or grasping. Research carried out in relatively stable conditions shows these actions provide stability by reducing body sway and COP displacement [82-84]. Other studies have indicated light touch is able to provide additional spatial information to posture control systems. However, less is known about different posture responses with or without touch. In this section, a comparison of posture movement between touch and without touch; vision and without vision were analysed.

4.4.1 Body sway reduces with light fingertip touch

Fingertip touch is found to reduce joint sway and relative COM displacement. The range of motion (ROM) at ankle and hip joints is reduced with touch, as indicated in Figure 30. However, this does not occur for head motion, where there is more sway with touch. Overall the ankle and hip did not show any significant change with sensory condition [F_{ankle} (3,119) = 0.82, p = 0.82, and F_{hip} (3,119) = 0.96, p = 0.42]. However, the difference in head movement is significant in relation to sensory conditions [F_{head}

(3,119) = 9.38, p <0.001]. At higher frequencies, the ROM of ankle and hip increases, and there is a significant difference across the frequencies [Fankle (3,119) = 46.28, p <0.001 and Fhip (3,119) = 17.59, p <0.001]. However, no significant difference is found at the head [Fhead (3,119) = 1.76, p = 0.15]. Meanwhile, the relative COM displacement is lower with touch than without touch. By comparing sensory conditions (i.e., control versus sensory), a significant difference is found in the posterior direction [F (3, 95) = 9.31, p <0.01]. However, no significant difference is found in the anterior direction [F (3,127) = 1.83, p = 0.14]. At the four frequencies, there is no significant difference in the relative COM displacement in the anterior direction [F (3,103) = 2.1, p = 0.11]. However, a significant difference is found in the posterior direction [F (3,103) = 3.43, p = 0.019].

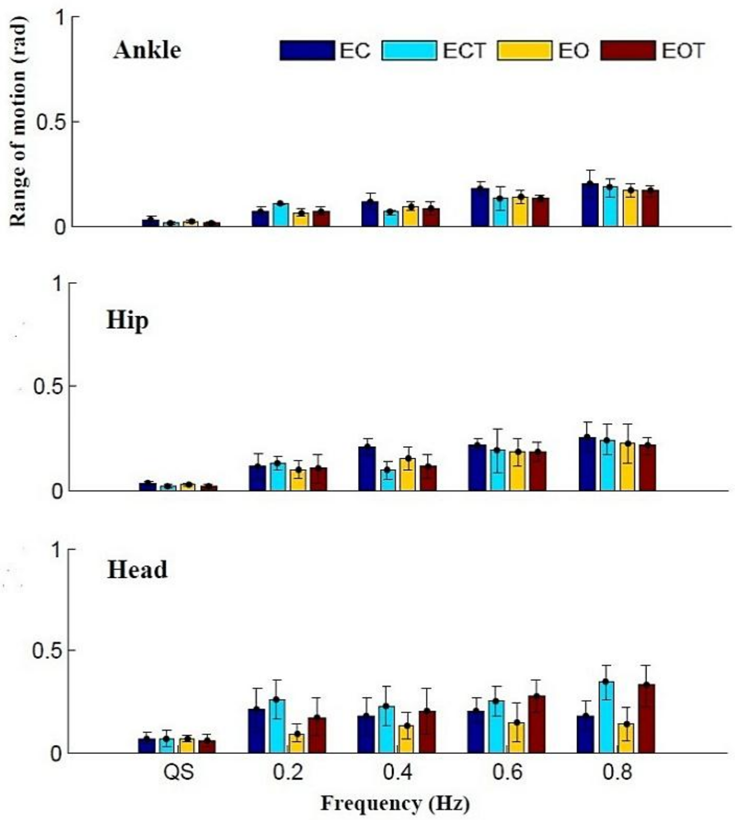


Figure 33 (Left) ROM of joints (Right) COM displacement pattern and average value (M±SE)

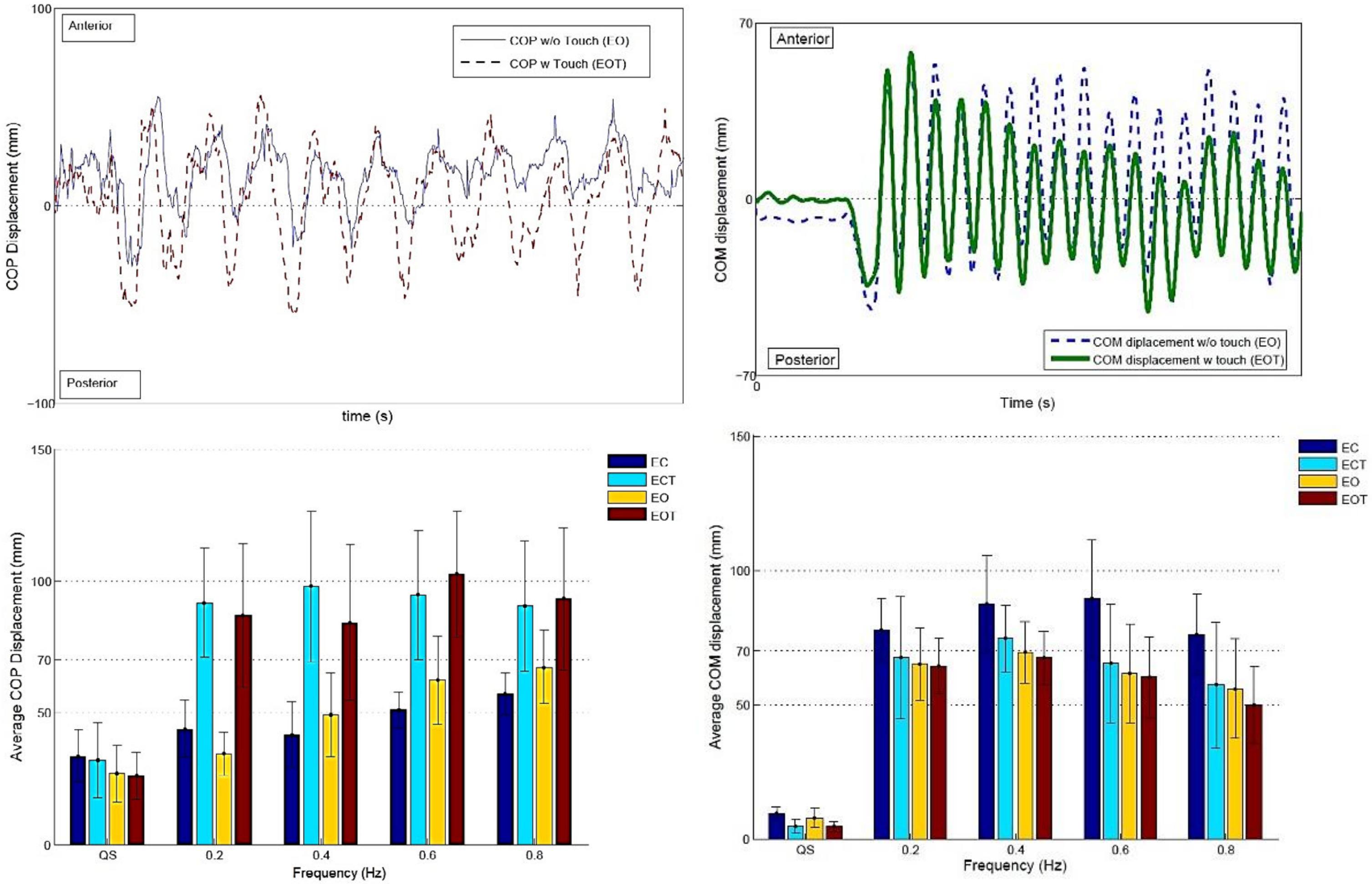


Figure 34 (Left) Comparison of COP response with and without touch. (Right) Comparison of COM response with and without touch.

The results above illustrates maximum displacement of COP at both anterior and posterior direction Based on the results in Figure 35, the COP was observed opposites response with the COM. This followed the concept of human balance described by Pollock (2000), where a better sensory inputs, individual able to provide a bigger based of support to always make sure the COM lays within the COP range [1]. By comparing between sensory condition, with existence of touch, COP displacement was increased. And, it also increased with frequency increase.

Based on statistical analysis, the average COP was observed increased with frequency with significant different found at all sensory conditions ($p < 0.05$). By comparing between sensory condition, with existence of touch, COP displacement was increased and significant different was found at all perturbation frequency ($p < 0.05$). Significant different was observed between visions and without vision; and touch and without touch. Previous research by Gatev et al, (1999) reported that, absence of vison have increase the body sway included COP [85]. Furthermore, additional sensory information like fingertip touch, was observed reduced postural sway when comparison make especially between balance and unbalance individual [82, 86, 87] . When compared with absence of vision, many research reported that COP displacement during EC higher than EO. This only can be seen at QS and 0.2 Hz of recent study. This result suggested that during perturbed stance, a larger base of support increase to provide more stability

4.4.2 Fingertips motion properties

Based on results in Figure 31, fingertip force (z-direction) significantly increases with frequency [$F(4, 79) = 5.9$, $p = 0.0007$]. From 0.2 Hz up to 0.8 Hz the force increases from 0.5 N to almost 2 N. When comparing the different vision conditions, force is higher with closed eyes compared to open eyes at all frequencies. However, no significant difference is found.

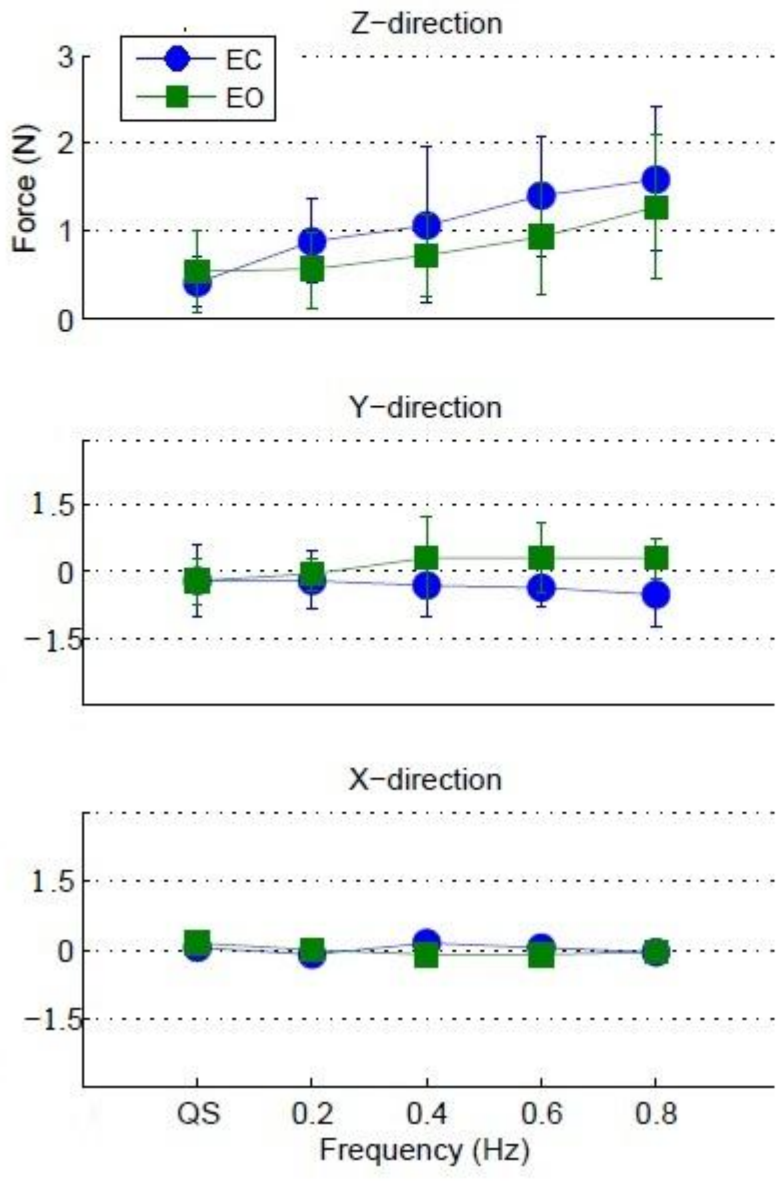


Figure 35 Average vertical force (z-direction) and horizontal force (y-direction and x-direction) produced by fingertip during EO and EC (M±SE).

Table 7 Comparison of force recorded between eyes-opened (EO) and eyes-closed (EC) using correlation coeff. (*r*) and paired t-test (*p*).

Force	**df**		**Frequency (Hz)**				
			QS	**0.2**	**0.4**	**0.6**	**0.8**
Z-direction	22	*p*	0.52	0.12	0.30	0.25	0.25
		r	0.53	0.89	0.71	0.96	0.19
Y-direction		*p*	0.91	0.58	0.14	0.25	0.01*
		r	0.94	0.25^{x}	0.05^{x}	-0.19^{x}	0.24^{x}
X-direction		*p*	0.36	0.38	0.35	0.38	0.86
		r	0.33	0.07^{x}	0.51	0.19^{x}	0.61

The (*) indicates significant difference between EO and EC with p<0.05 and (^{x}) indicates the correlation between EO and EC with r<0.3.

Furthermore, the amount of horizontal experienced at y and x directions was found to be insignificantly different between frequencies with F (4, 79) =0.18, p=0.94 and F (4, 79) =2.11, p=0.097, respectively. Almost all force in the x and y directions also found insignificant differences during difference vision conditions (e.g., EC vs. EO) except at 0.8 Hz (as shown in Table 1). More interestingly, for force at the y-direction, difference in direction was observed between EO and EC. During eyes-opened, fingertip generated more force in the anterior direction; and in the posterior direction during eyes-closed. This can be seen by the weak correlation coefficient (r<0.5) between EC and EO. These preferences are unique and give information about the possibility of different posture leanings during touch for both with and without vision.

Based on the horizontal force data, there were different in direction of force recorded between EO and EC. These have encourage for further investigation on wrist position which may contributed to this even to be happen. Figure 37 illustrates investigation of wrist movement that may lead to different horizontal force produced at the fingertips.

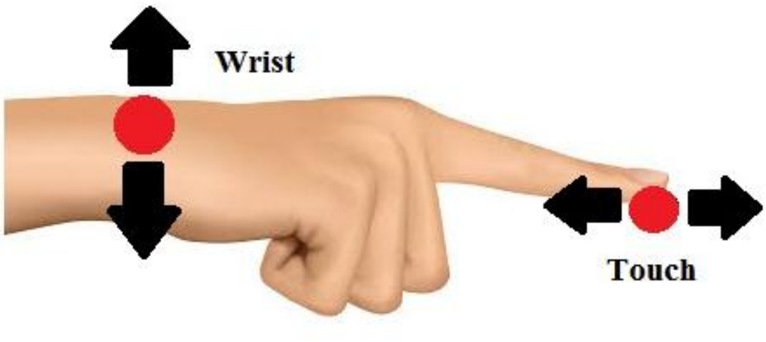


Figure 36 Wrist motion influenced fingertip movement.

According to F.-C. Su et al, (2005), motion of wrist reported influenced the fingertip motion. They have determine negative slope from regression analysis that demonstrated the so called ‘‘reciprocal’’ nature of joint motion [88]. For example, during wrist extension, passive finger joint flexion was induced and, alternatively, during wrist flexion full finger joint extension was induced. This finding have encourage the investigation on wrist motion especially in superior-anterior direction which might influenced the touch’s horizontal force direction to change.

Figure 32 above shows that, wrist at superior and inferior direction increased with the increase of frequency with significant different found only at inferior direction (p=0.03). Based on the statistical analysis of superior direction, there is no significant different found (F (1, 47) =0.05; p=0.8271). A similar results also recorded on inferior direction (F (1, 47) =0.27; p =0.6078). These insignificant results have indicated that the wrist movement might not correlated with vestibular dysfunction or more specifically vision input. These is highly recommended that the changes of force direction might cause by finger itself. In the research by Proske and Gandevia (2012), flexion of distal interphalangeal joint of finger caused by stimulation of ulnar nerve which connected to dorsal spinal cerebellar tract that control response of head, neck and upper limb [89]. It is believed due no vision inputs, the central nervous system (CNS) response by triggering ulnar nerve to cause finger flexion for better proprioceptive sense input in order to provide sufficient haptic information for position correction. In this study, flexion of finger is believed produce horizontal force towards posterior direction

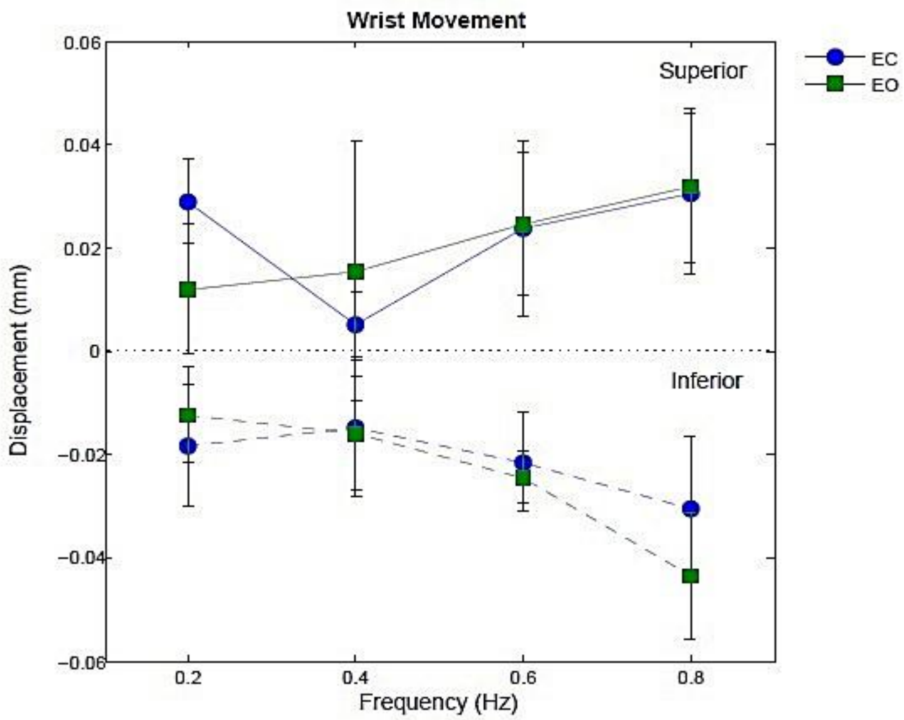
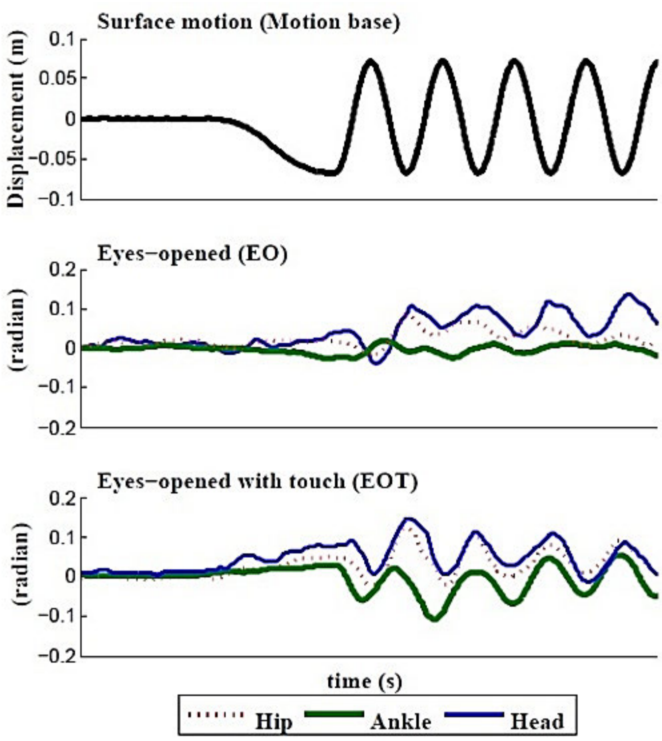


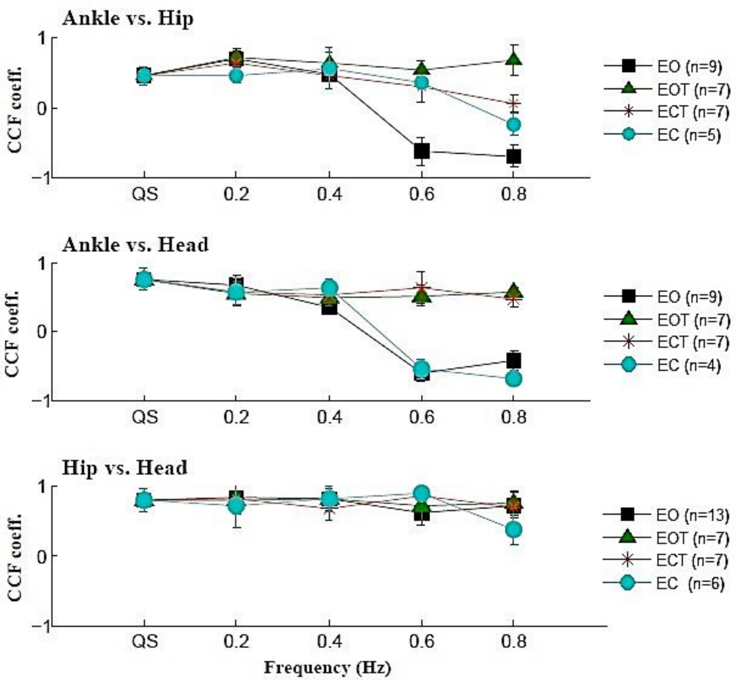
Figure 37 Wrist movement at both superior-inferior direction during EC and EO.

4.4.3 CCF coefficient between joints

This analysis is carried out to determine changes in posture strategy to maintain a balance position. According to Figure 3, results show changes from ankle to hip strategy when the perturbation frequency is 0.6 Hz. With vision the coefficient becomes negative at 0.6 Hz compared to 0.8 Hz without vision. However, no change in strategy is observed when touch is introduced. Similar responses are recorded for ankle versus head. These results indicate changes in posture modulation scheme exist even with only a small amount of support from the hand. As expected, the hip and head move at the same phase for all frequencies. There is no change with or without touch, even though greater displacement is recorded at the head with touch.



(a)



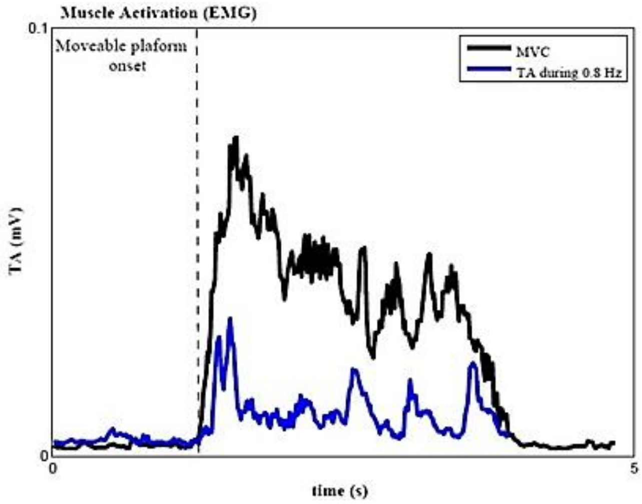
(b)

Figure 38 (*a*) Example of joint sway with and without touch at 0.6 Hz perturbation; (*b*) comparison of average CCF coefficient. The n is number of subject who produces a similar response.

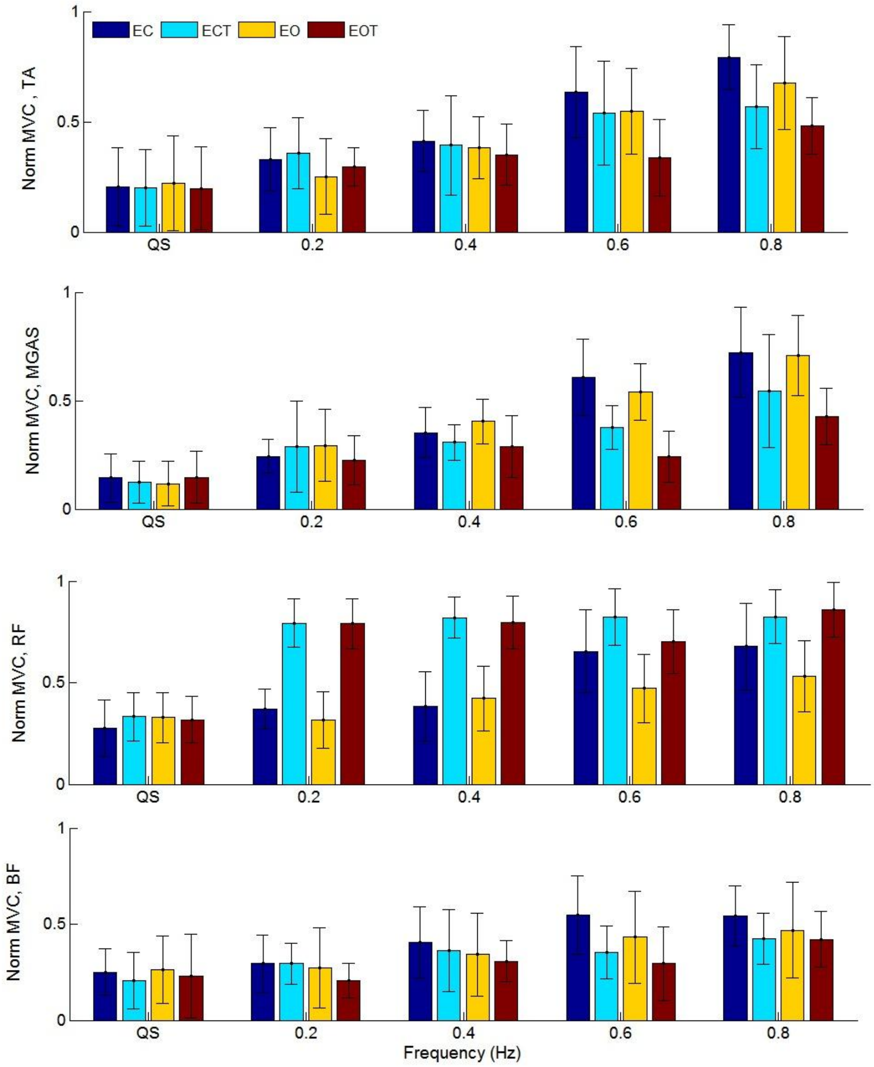
4.4.4 Touch force reduces normalised MVC of four muscles

Based on the above results, body sway reduces with touch. In our earlier hypothesis muscle activation would be higher with touch, as joint stiffness increases activation. However, the results show the opposite. Figure 33 shows that with touch, body sway and the intensity of muscle activation are reduced. These can be clearly seen at perturbation frequency above 0.6 Hz. The response is only altered at the rectus femoris (RF) where it is higher with touch. Vision and without vision indicated a consistent response at all muscles and perturbation frequencies. It is clear that with a better sensory input, the central nervous system (CNS) used an effective energy consumption by reducing the amount of muscle activation. These results followed the previous study which indicated that effective energy consumption applied by the CNS over continuous perturbations applied [53].

Table 7 shows the significant levels of normalised MVC in different conditions. By comparing the different frequency perturbations, there is a significant difference in three muscles [F_{TA} (3, 43) = 11.87, p <0.01; F_{MGAS} (3, 43) = 15.63, p <0.01; and F_{RF} (3, 39) = 3.25, p = 0.03] but no significant difference at the bicep femoris [F_{BF} (3, 41) = 1.8, p = 0.16].



(a)



(b)

Figure 39 *(*a) Normalised value gathered from MVC; (b) Average normalised MVC for muscles at different frequencies and sensory conditions

Table 8 Significant levels of normalised MVC of muscles at different conditions

			Frequency (Hz)				
Condition	**df**		**QS**	**0.2**	**0.4**	**0.6**	**0.8**
EO vs. EC	**16**	**TA**	>p	0.047*	>p	>p	>p
		MGAS	>p	0.048*	>p	>p	>p
		RF	>p	0.014*	>p	>p	>p
		BF	>p	>p	>p	>p	>p
EO vs. EOT	**17**	**TA**	>p	>p	>p	0.020*	0.010*
		MGAS	>p	0.038*	0.002*	<0.001*	<0.001*
		RF	>p	>p	>p	>p	>p
		BF	>p	0.041*	0.045*	0.009*	0.037*
EC vs. ECT	**10**	**TA**	>p	>p	>p	>p	>p
		MGAS	>p	>p	>p	>p	>p
		RF	>p	<0.001*	<0.001*	0.020*	0.026*
		BF	>p	>p	>p	>p	>p

*(The significant differences found at p <0.05 is indicated by *)*

4.4.5 Joint stiffness reduced with touch

In previous section, it was observed that touch have caused joints to sway less compared to without touch. Besides, similar response also observed in normalized muscle activation. However, those response have raise a question regarding to stiffness at each joint. Logically, when joint reduced its sway, it's indicated increase of stiffness. Increase of stiffness did not give any advantages to individual as it will increase muscle tone. But, it is found that, joint stiffness at both ankle and hip is also reduced with touch as shown in figure below. According to this figure, stiffness magnitude of both joint was observed reduced with touch at all frequency of perturbations.

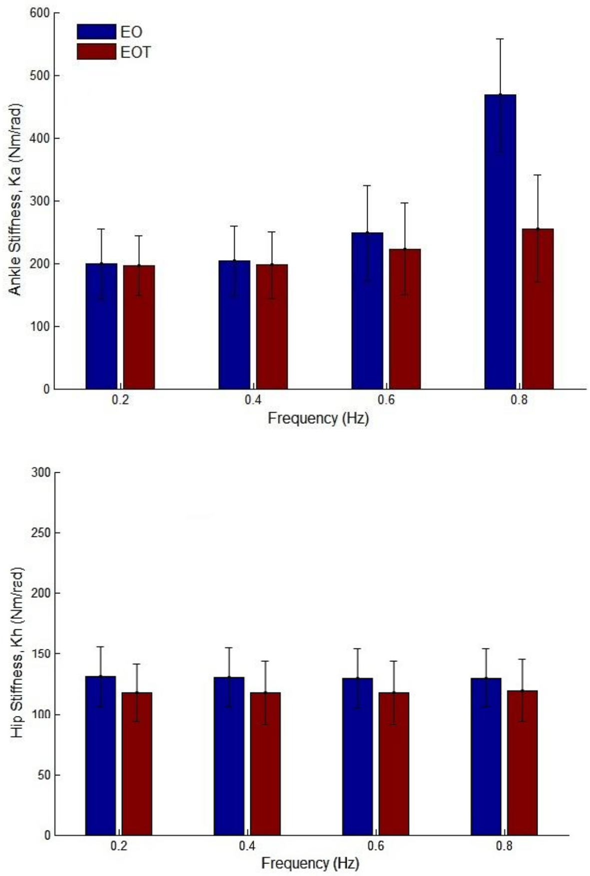


Figure 40 Comparison of joint stiffness between touch and without touch condition.

By comparing between moment and joint sway during touch and without touch, it is found that joint torque or moment experienced by the participant was higher during without touch than with the existence of touch form fingertip. This observation gave information on how touch help the whole body to resist the angular acceleration created by the introduction of surface perturbation.

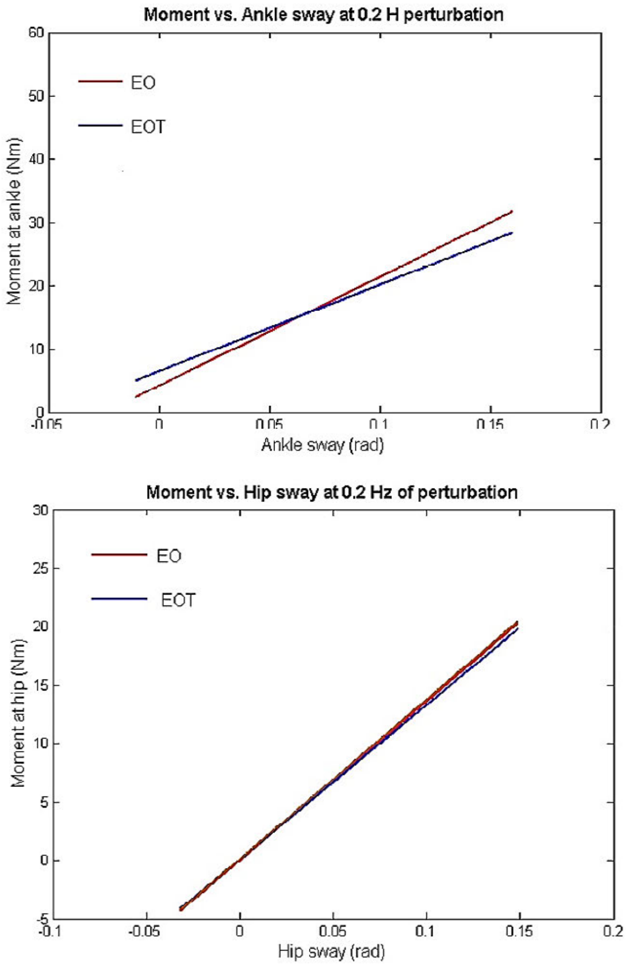


Figure 41 Moment at joint vs. joint sway angle at both touch and without touch condition

Further investigation was done to determine the significant change of vertical ground reaction force (VGF) which play an important role in creating moment at joint.

Ground reaction force to body weight ratio (N/BW) detected by force plate showed different in pattern between touch and no touch condition as shown in Figure 42 below. Without touch, the N/BW was more flattered. The research done by Scott-Pandorf et al. (2007) on peripheral arterial disease patient have found a significant flattening of the vertical force curve for the claudication patients which explained the possible pattern due to loss of sensitivity [90]. Based on the result in Table 9, it also observed that the VGF to body weight (BW) ratio range (N/BWmax – N/BWmin) increased with touch and showed a significant difference between touch conditions ($p<0.05$).

Table 9 Comparison of the ratio (N/BW) range between with and with/o touch

Comparison of VGF ratio (N/BW)		
Perturbation Frequency	**Condition**	
	With touch	**Without touch**
0.2 Hz	0.1697 ±0.061	0.1246± 0.038
0.4 Hz	0.1748± 0.027	0.1199± 0.013
0.6 Hz	0.1516± 0.032	0.1093± 0.015
0.8 Hz	0.1659± 0.063	0.1234± 0.008

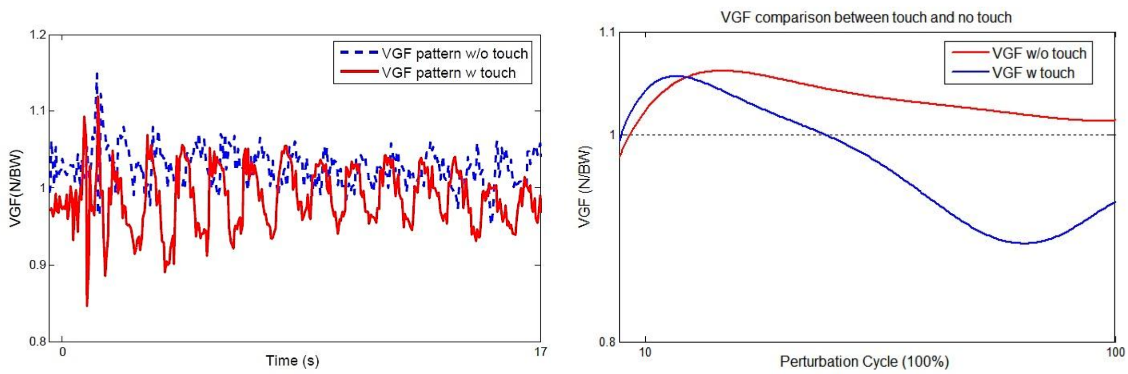


Figure 42 (Left) Raw vertical ground reaction force during both touch and no touch condition. (Right) The ground reaction force (VGF) at one perturbation cycle.

4.5 Development and analysis of human posture control simulation model.

4.5.1 Improvement of double inverted pendulum model

According on the experiment data, it was observed that changes of posture movement was due to changes of joint stiffness response caused by different sensory input. Average stiffness value at various condition were then analysis to determine ratio Y which would indicated the increase rate of stiffness from the unperturbed condition.

At condition without perturbation or $f = 0$,

$$Ka_{fo} = mgh = 516.94\ Nm/rad$$

$$Kh_{fo} = mgh = 75.67\ Nm/rad$$

For perturbed stance, which are at frequency, $f = 0.2, 0.4, 0.6, 0.8\ Hz$, additional of stiffness profile need to be included into equation (8) . Figure 37 below described the polynomial equation for ratio Y. This ratio were then included in equation below to estimate the joint stiffness value at different perturbation frequency and sensory conditions.

$$Ka_f = Ka_{fo} + y_a Ka_{fo} \text{ ……….} (13)$$

$$Kh_f = Kh_{fo} + y_h Kh_{fo} \text{ ……….} (14)$$

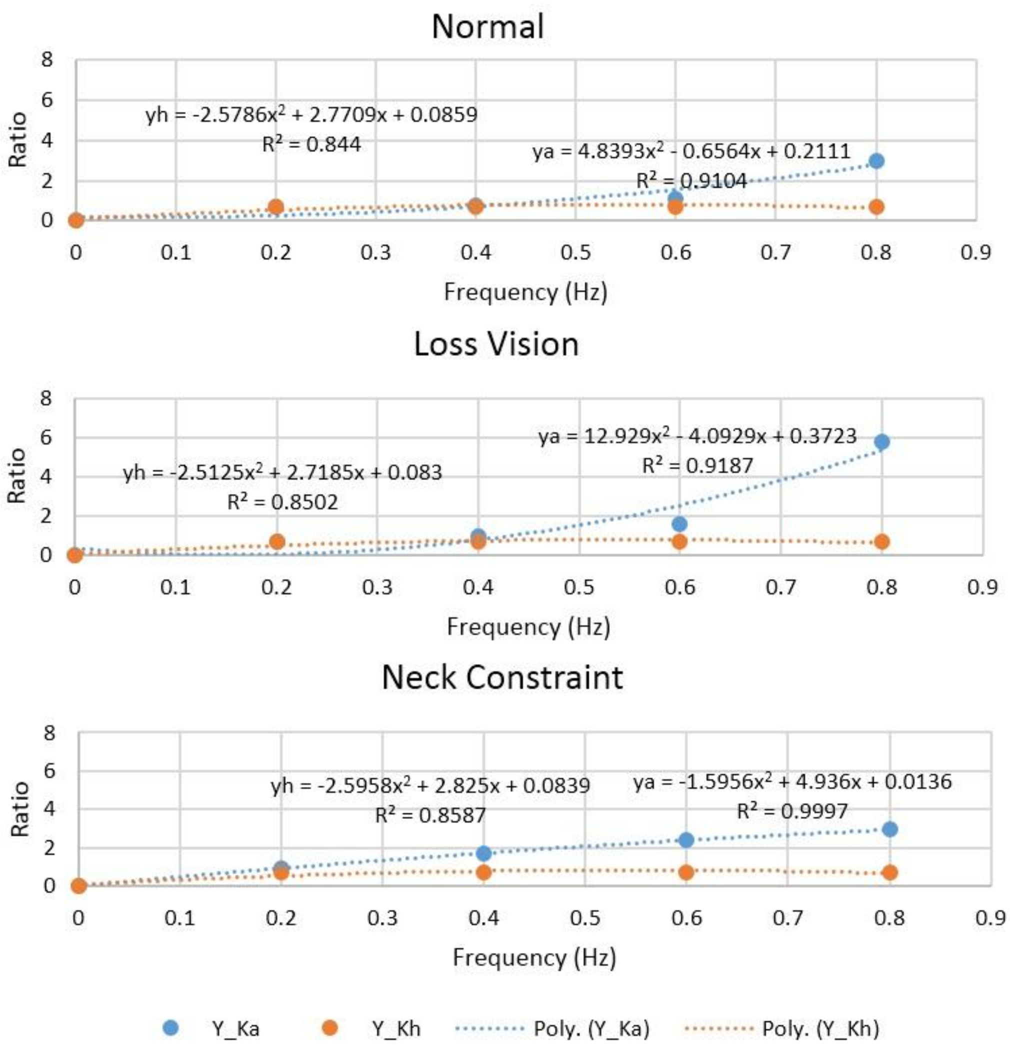


Figure 43 Ratio Y based on experiment data

However, some addition toward double pendulum model equation (8) is needed to present the neurological effect of human posture control based on joint stiffness profile. Based on the clinical study data, both joint stiffness was responded differently. Ankle stiffness was observes as involuntary effort. According to Granata et al, (2004), effective ankle stiffness was dominated by both active and reflex response [63]. Thus, in this simulation model, ankle stiffness, K_a, was categorised as kinetic energy. This also support by our result based on comparison between latency of onset EMG with joint motion has been used to predict the CNS control mechanism for posture stabilisation. Previous research stated that ankle joint applied feed forward control for stabilisation which caused muscle activation occur after joint motion for posture correction [80] . This mechanism can be considered as involuntary movement.

Based on the result, muscles at the ankle joint (TA, MGAS and SOL) was observed delayed the ankle motion at all frequency, thus, followed result from Finley et al, (2012). More interestingly is muscles at hip (RF and BF) changes over frequency. At frequency 0.2 and 0.4Hz of perturbation where ankle strategy applied, the RF and BF activation preceded hip motion by around 70ms. These feed forward feedback allowed hip joint to maintain its position and avoid hip to sway further. These preceding of muscle activation than joint motion or force is related to voluntary response to maintain stabilisation of joint [81]. At 0.6 Hz and above, where the shifting strategy occurred, muscle activation delayed than joint motion thus concluded the shifting strategy occurred due to involuntary response.

According to previous research by Gatev et al, (1999) and Fitzpatrick et al (1992), posture stabilisation is beneficial more from feed forward control mechanism than reflex feedback [39, 85]. However, possibility of reflex feedback occurrence at upper limb is still there. These change response from feed forward to reflex response might indicates inertia properties. According to Lin (2001), they observed the changes of reflex gain is significance with non-linearity of damping properties of muscle stretch [91]. Thus, with this argument, for the simulation model, hip stiffness properties might occur on term of damping, D_h.

$$D_h = K_h \theta_2 \dot{\theta}_2 \text{ ……… (15)}$$

Both ankle stiffness and damping profile were then included in equation (8);

$$= \frac{1}{2} m{v_1}^2 + \frac{1}{2} m{v_2}^2 + K_a \theta_1 - (m_1 + m_2) g h - m_2 g h - D_h \text{ ………. (16)}$$

The derivation of Euler Langrange continued and the system equation was represent in term of angular acceleration $\ddot{\theta}_1$ and $\ddot{\theta}_2$. The external perturbation which define as T was included in equation below;

For $\ddot{\theta}_1$;

$$\ddot{\theta}_1 = \frac{T - m_2 l_2 \sin(\theta_1 + \theta_2)\, {\dot{\theta}_2}^2 - (m_1 + m_2) g l_1 \sin\theta_1 + K_a \theta_1}{(m_1 + m_2)\, l_1 + m_2 l_2 \cos(\theta_1 + \theta_2)} \text{ ……….. …….. (17)}$$

For $\ddot{\theta}_2$;

$$\ddot{\theta}_2 = \frac{m_2 l_1 \cos(\theta_1 + \theta_2)\, \ddot{\theta}_1 + m_2 l_1 \sin(\theta_1 + \theta_2)\, {\dot{\theta}_1}^2 + m_2 g \sin\theta_2}{m_2 l_2 + K_h \theta_2} \text{ ………. (18)}$$

For adaptation model to represent adaptation strategy from the central nervous system (CNS), the equation below was used;

$$\theta a_{tn} = \theta a_{tn-1} + \frac{0.02}{tc}\theta a_{tn-1}\,; \qquad for\ normal\ condition\,; \text{ .... (19)}$$
$$\theta a_{tn} = \theta a_{tn-1} - \frac{0.02}{tc}\theta a_{tn-1}\,; \qquad for\ vision\ loss\ condition\,; \text{ .... (20)}$$
$$\theta h_{tn} = \theta h_{tn-1} \qquad ; \qquad for\ bot\ normal\ and\ vision\ loss\ \ condition \text{......... (21)}$$

The θa_{tn} is current angle of sway at cycle n and tc is period of one cycle (second). Furthermore, all constants parameter were defined according to experiment setup and subject physical data. The details of constant value is shown in Table 10 in the next section below;

4. 5.2 Simulation output

Simulation were then ran in SIMULINK environment. Simulation model of human posture modulation was designed using closed loop system theory and developed based on the enhancement of double inverted pendulum equation mentioned above. The close loop system was develop to represent the neuromuscular controller properties as it regulated over repeated or continuous external perturbation. Furthermore, figure below also illustrates subsystem of inverted double pendulum model and adaptation model that developed based on equation (19) to equation (21).

Table 10 Simulation constant value

Constant	Value	Note
m_1	34.45kg	Mass of lower limb, 0.51 from average weight of subjects according to Plagenhoef's Body Segment Weight
m_2	33.10kg	Mass of upper limb, 0.49 from average weight of subjects according to Plagenhoef's Body Segment Weight
l_1	0.82m	Length of lower limb from average weight of subjects according to Plagenhoef's Body Segment Weight
l_2	0.56m	Length of upperlimb from average weight of subjects according to Plagenhoef's Body Segment Weight
T	0.07m	Perturbation displacement according to experiment setting.
g	9.18 m/s^2	Gravitational acceleration
$omega$	0.2, 0.4,0.6,0.8 Hz	Perturbation setting according to experiment setting

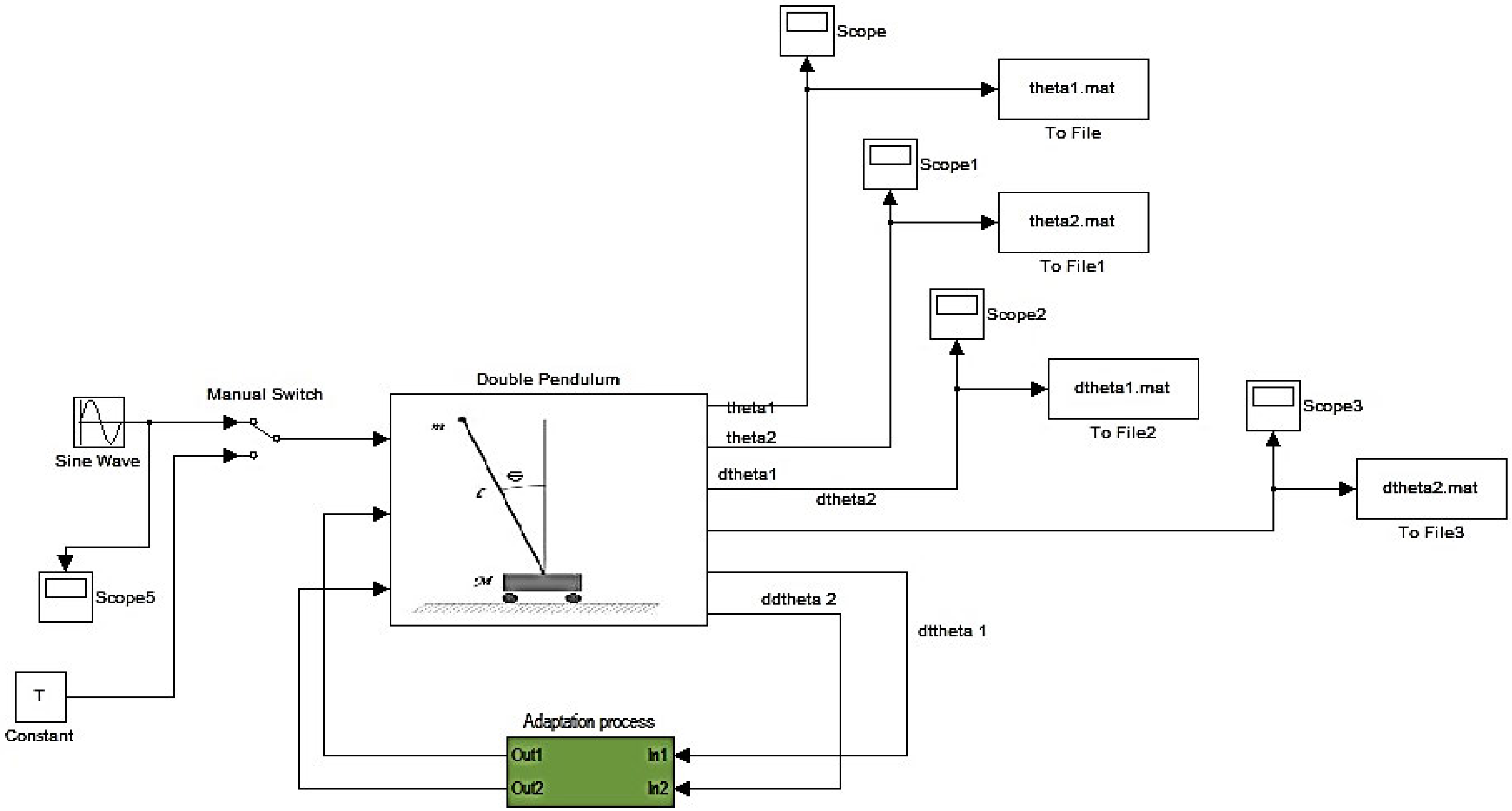


Figure 44 Simulink model system

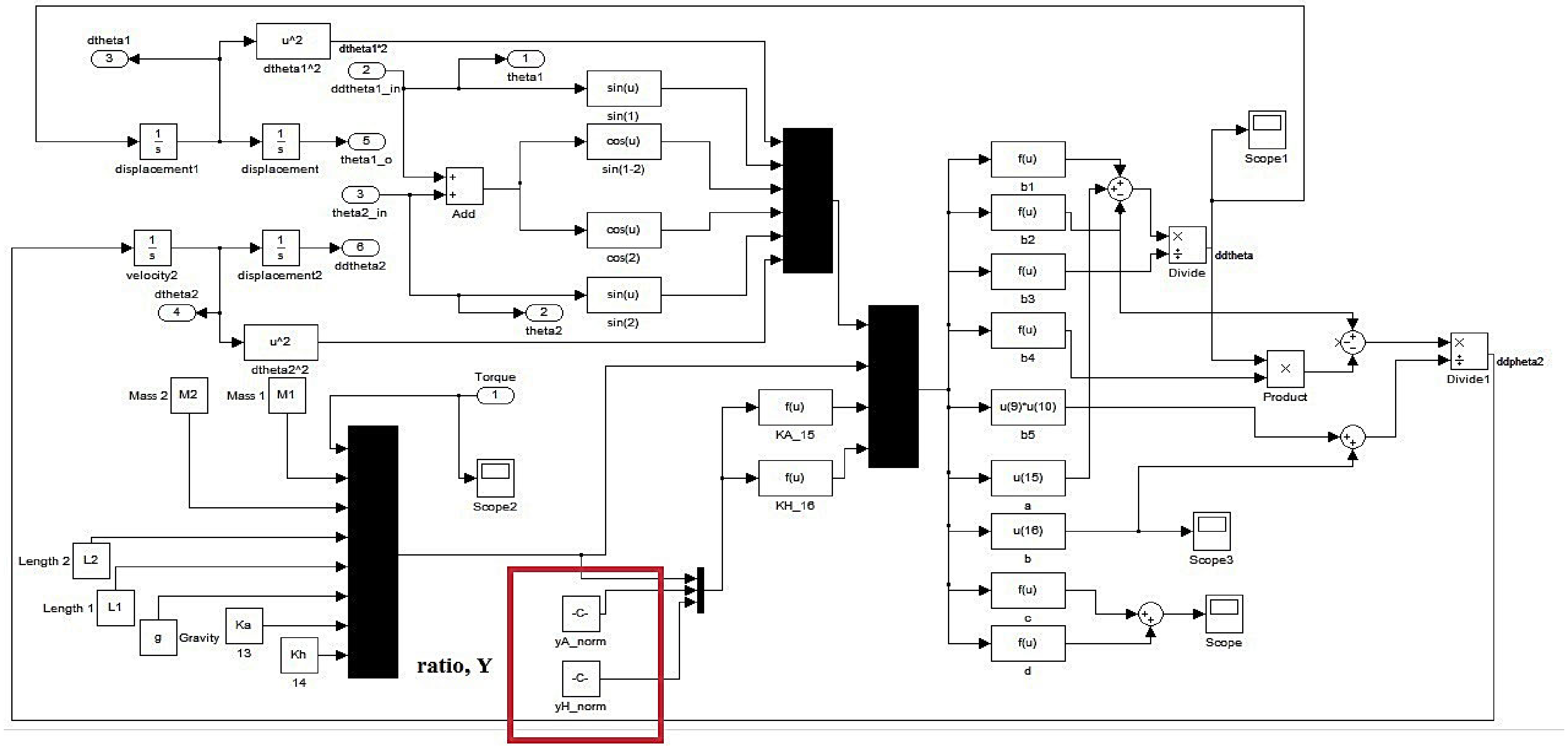


Figure 45 Inverted pendulum subsystem with additional joint stiffness profile.

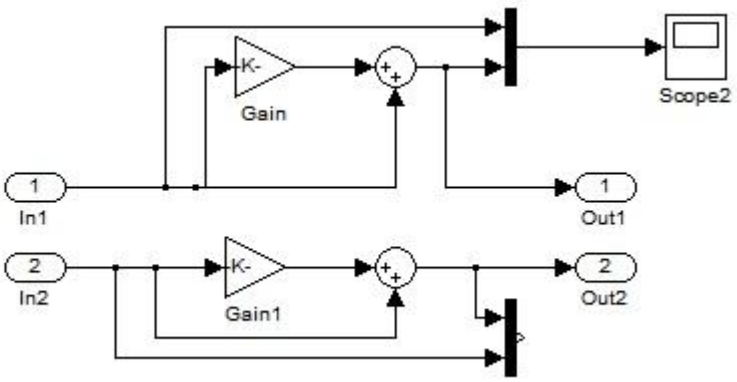


Figure 46 Adaptation model

In order to visualize the simulation result especially the movement of inverted pendulum that would present the existence of posture strategy, a two-link inverted pendulum model were developed as figure below. Average physical details from subjects were used to represent the length and mass of each pendulum joints. Figure 44 below described the animated pendulum diagram.

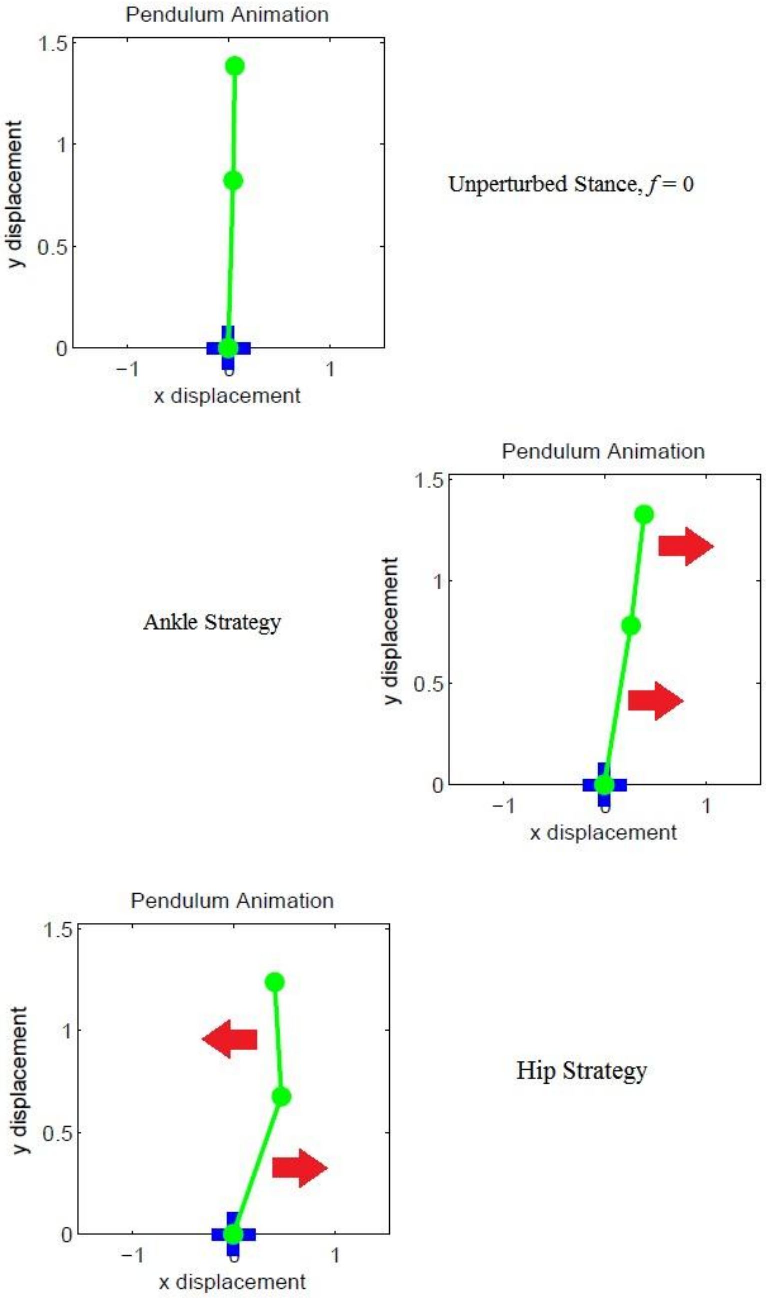


Figure 47 Animated Inverted Pendulum Model

Motion and displacement of each joint were analysed to determine its correlation with the experiment data. Overall, motion pattern of joint at both vision and no vision condition were followed the experiment data. The existence so called ankle and hip strategy can be observed as the direction of joint movement changed at 0.6 Hz and above for vision condition, meanwhile at 0.8 Hz during without vision.

However, amount of displacement is a little bit different. As shown in Table 10 below, displacement difference were recorded at almost double the experiment value especially at 0.8 Hz. By comparing with analysis of absolute percentage difference, most simulated displacement can be assumed to be acceptable since the percentage was observed less than 20% (considered close and near miss) and it is still in the experiment data range. Further improvement is still warranted in order to improve this model so that most accurate outcome can be produced.

Table 11 Comparison between experiment and simulated joint displacement

Displacement	**Frequency (Hz)**			
	0.2	**0.4**	**0.6**	**0.8**
Normal Condition				
Ankle	0.0840*	0.0600*	0.1918	0.3924
Δ	(+0.0232)	(-0.0313)	(+0.0579)	(+0.2216)
(±SD)	0.0241	0.0188	0.0318	0.0313
Hip	0.1095*	0.1000	0.1335*	0.3379
Δ	(+0.0088)	(-0.0524)	(+0.0484)	(+0.0965)
(±SD)	0.0417	0.0551	0.0683	0.1146
Without Vision Input Condition				
Ankle	0.0944*	0.1000*	0.1070	0.2725
Δ	(+0.0224)	(-0.0164)	(-0.0660)	(+0.0655)
(±SD)	0.0235	0.0365	0.0311	0.0715
Hip	0.1140*	0.164*	0.1248	0.1274
Δ	-	(-0.0320)	(-0.0948)	(-0.1265)
(±SD)	0.0656	0.0389	0.0293	0.0702
Without Neck Movement Condition				
Ankle	0.0971*	0.1000*	0.1647*	0.1146
Δ	(-0.0188)	(-0.0211)	(-0.0044)	(-0.0672)
(±SD)	0.0210	0.0247	0.0449	0.0348
Hip	0.1381*	0.1403*	0.1024	0.1009*
Δ	(+0.0294)	(+0.0165)	(-0.0632)	(-0.0959)
(±SD)	0.1087	0.1238	0.1656	0.1968

Value in the bracket is the different with the experiment data where + sign indicates the displacement exceed from the experiment data meanwhile - sign indicated lesser than experiment data. The (*) indicate the simulated data that within the acceptable range based on analysis of percentage of different.

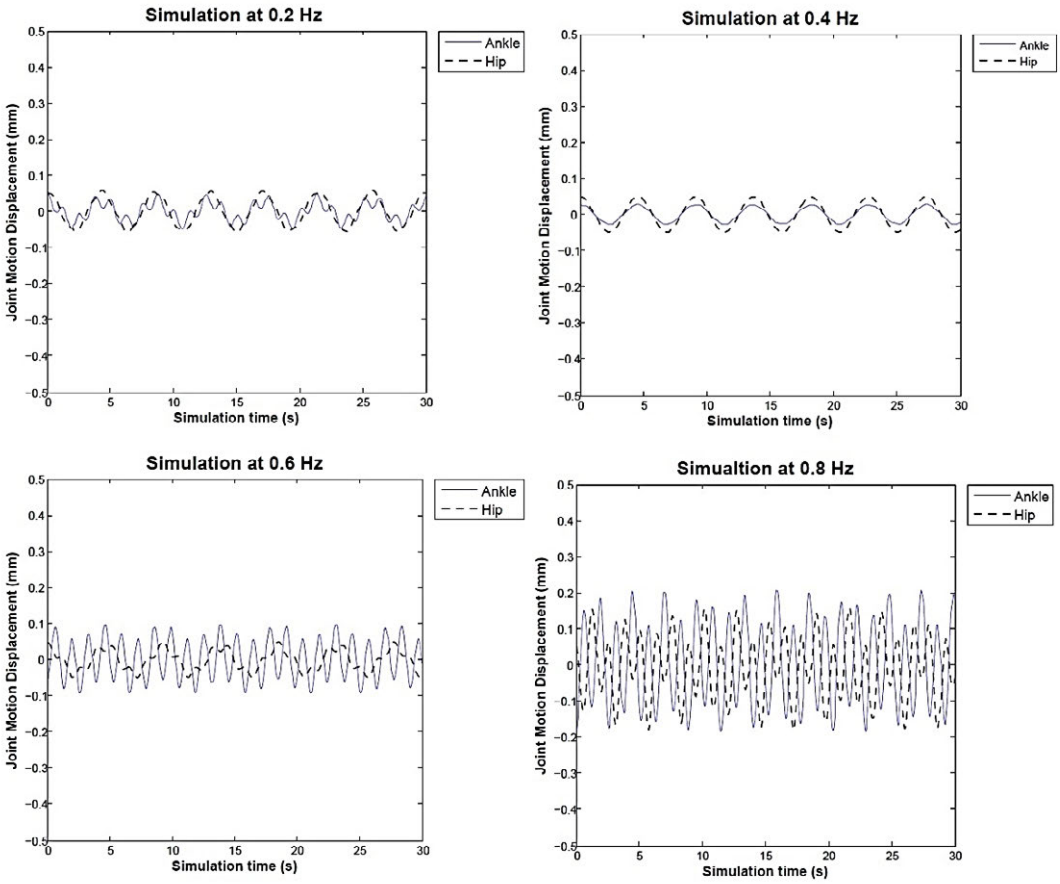


Figure 48 Simulated joint displacement during normal sensory condition (with vision)

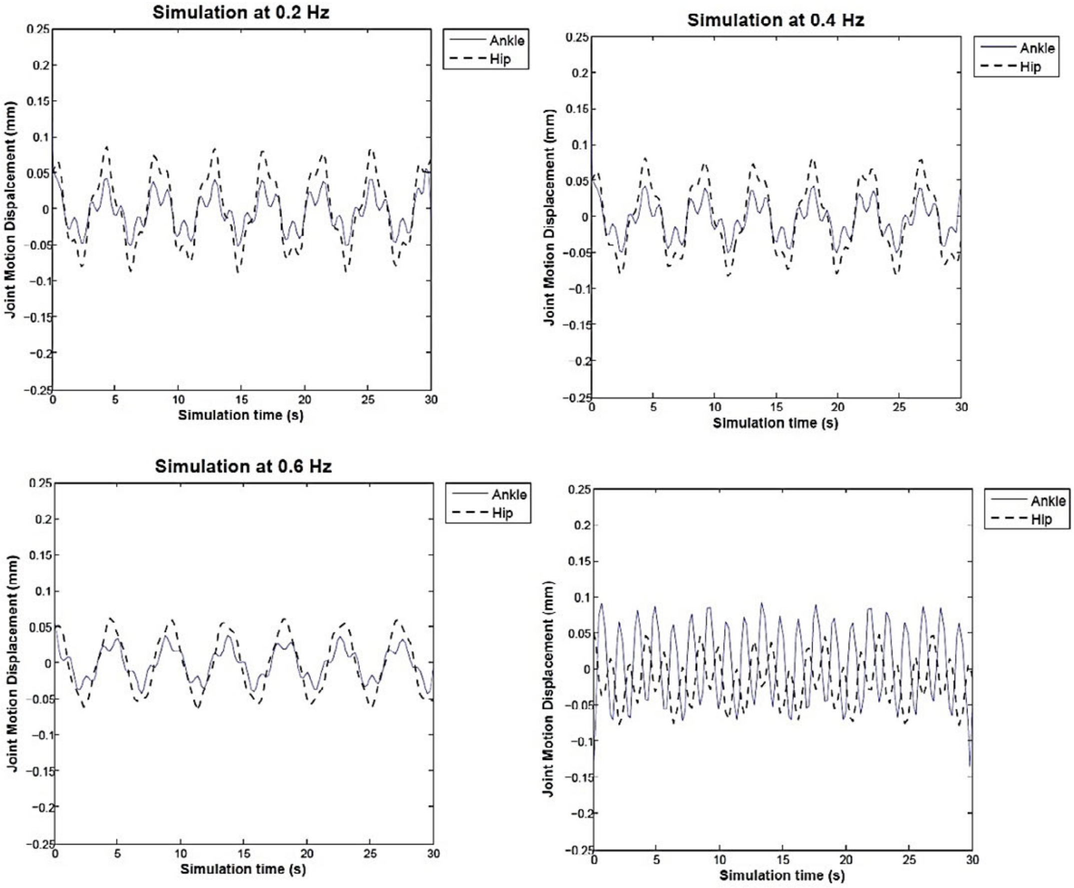


Figure 49 Simulated joint displacement during without vision sensory condition

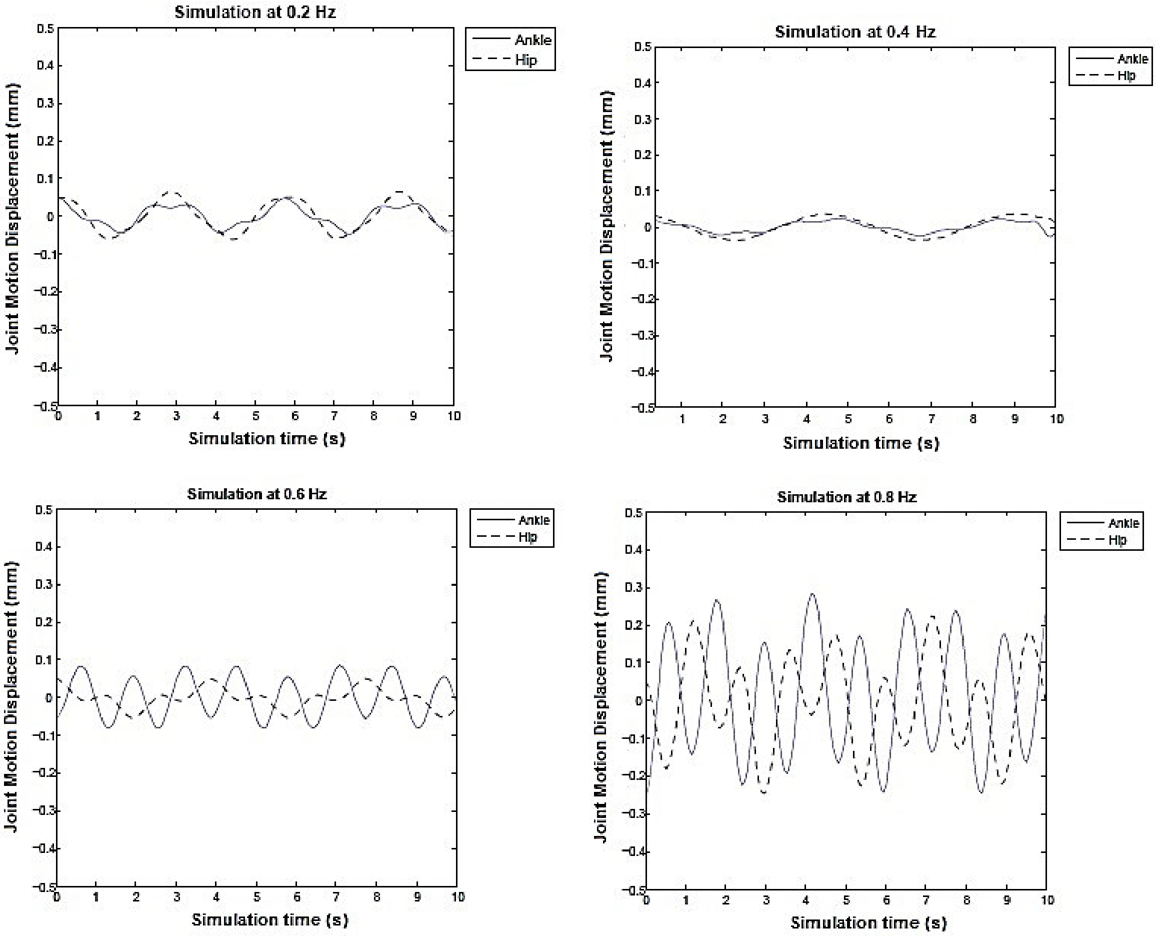


Figure 50 Simulated joint displacement during with neck constraint condition for 10 second.

CHAPTER 5

DISCUSSION

5.1 Stability and Joint Stiffness Response

Based on the results, it was illustrated that joint stiffness responded towards imbalance. Compared to other measurement approaches, joint stiffness provided substance in describing the amount of energy or work that the subject faced to maintain the required position. In previous research, joint stiffness was proclaimed to have a strong correlation with muscle contraction; whereby, the reported CNS tend to apply less energy strategy, resulting in less muscle contraction strategy [53]. As previously mentioned, high stiffness was commonly applied by those who faced movement difficulties due to factors such as disease, ageing, and impairment. It was hypothesised that a person with less balance ability will apply high stiffness at the joints (ankle and hip) in order to maintain a balanced position; based on previous evaluations of the elderly and patients with disease [14, 43, 44, 92]. Thus, the ability to produce less joint stiffness may be strong evidence for substantial balance ability. According to recent results, the amount of joint stiffness was able to distinguish patients according to FRT test scores.

To keep producing high stiffness at each joint is not necessarily good. According to previous research, ankle and hip stiffness were estimated based on theoretical study suggested to be higher than 728 Nm/rad and 179 Nm/rad respectively to maintain posture balance and it is also believed that coaction between ankle and hip is important [47]. However, in case of dynamic perturbation, recent result have shown it is smaller by almost 20 percent. The K_h was observed to be negatively correlated with K_a at normal condition (O) during both dynamic perturbations. However, it was not during less sensory conditions (NO and C) where K_h was higher when K_a was also high. This suggested that the degeneration of vestibular and vision sensory tend to stiffen the body when faced with external perturbation and unable to generate the coaction strategy between the joints.

Analysis of the stability ratio of stiffness provided information on the adequate amount of stiffness required by healthy people to remain in a balanced position during perturbed standing. As mentioned earlier, research by Edwards and Suzuki predicted the required amount of stiffness at both the ankle and hip joints to maintain quiet standing position [93, 94]. Without considering the perturbed situation, their results were small compared to more recent studies. However, by conducting the experiment in a repeated manner with different intensities of perturbation, it was noted that stiffness at both joints was not necessarily higher, in order to remain stable. It was suggested that joint stiffness must generate at a certain range.

With additional sensory input from somatosensory via fingertip touch, joint stiffness also reduced. This finding gave a strong support to confirm that joint stiffness is less during better stability. Analysis of joint moment magnitude and ground reaction force exerted due to perturbation applied have shown that it were also reduced with the existence of touch.

5.2 Weakness in Vestibular and Vision Sensory in response to Perception of Posture Response

Similar to the no vision input (*C*), limitation of the head movement (*NO*) was also recognised to change the posture modulation response; it was shown that during this condition the graviceptor at the head and in the body were distinguished [95]. Limitation of the head movement actually had limited Vestibular-ocular reflex sense [89]. The VOR is a mechanism for triggering eye movement to fix on a desired gaze point when the head was moving. With this mechanism, postural reflex on any changes due to movement can be made quickly and effectively. Projection of the vestibular nuclei regulated the head movement reflex from the neck muscle activation [96]. Simultaneously, the otolith organ which senses any change in gravity and acceleration will then send information (axons) to the spinal cord to influence the excitation of the muscle to maintain posture.

In the above results, weaknesses in the vestibular sense and vision led to different joint stiffness value based on the type of perturbation applied. If other research indicate that the elderly (who normally face degenerative vestibular function and vision) apply the hip strategy (hip sway more and less stiff) when facing external perturbation, then they would differ from recent results which show that the stiffness response was more affected by perturbation manipulation than sensory manipulation. This raised a question regarding the real effect of vestibular and vision sensory weakness towards the perception of posture response since inconsistent and insignificant differences were found. According to Ting et al. (2007), muscle synergy was not affected by deficiency in the sensory, especially the visual and vestibular system [97]. This was also observed in joint stiffness recently. The phenomenon suggested influence from other mechanisms. In a previous research by H. Mittelstaedt (1996), it was proposed that the existence of additional graviceptor outside the labyrinth (mechanoreceptor in joints, skin and muscle) also influenced the posture response [95]. Thus, it was concluded that the manipulation of the vestibular and vision gave less influence to posture modulation; the addictive interaction by somatosensory and graviceptor at other body parts also helped produce the desired counteraction between the lower and upper body since the human posture control system is sensitive to both gravitational and perturbation force. Furthermore, that interaction also depended on individual ability.

5.3 Motor Learning Ability

In this study, perturbations were applied in a repetitive mode. Other than evaluating the joint stiffness response, this approach was performed to observe the adaptation response which may indicate the motor learning ability. The adaptation percentages were further noted to vary according to the perturbation manipulation. Again, the sensory manipulation also provided less influence since no significant difference was found between the sensory manipulation conditions. Based on previous research, the adaptation ability was observed by the decay rate of the exponential curve [60] where a higher rate was noticed only during open eyes. However, the reduction of adaptation due to sensory manipulation was further scrutinised. A consistent average adaptation percentage value between both perturbations was only noticed during the lowest frequency (0.2 Hz) when both perturbations produced almost similar amounts of COP-COM velocity. With the increase in perturbation frequency, adaptation was hard to achieve due to high difficulties. High adaptation values do not necessarily show a better motor learning ability. In Fig. 8(a), *C* and *NO* had a greater percentage compared to *O* during the translation perturbation. In that difficult situation, a greater gap between active and passive components was observed. High active components were required at difficult situations. It was believed that in high difficulty situations (high frequency), adaptation was not a choice. The CNS tend to apply accuracy control to reduce kinematic variability under high speed movement [53] to maintain desired position.

5.4 Light touch improve balance and maintain posture as a single-link pendulum

The results prove with very small amounts of sensory information through fingertip touch young people are able to reduce posture sway and thus improve balance. This is seen through the reduction of joint sway and relative COM displacement at both directions. However, a different response is observed at the head where its displacement increases with touch. Similar to other joints, without visual information head motion becomes more variable [5, 7, 31]. It is well known that visual information is important to stabilise balance as it reduces the variability of both head and trunk motion in space. However, with the existence of additional somatosensory input from touch, head motion becomes more exaggerated. Potentially this takes place due to the effect of damping. Touch limits the motion at ankle and hip which react as inertia absorbers, but it is unable to prevent the same effect at the head. These results provide a clear indication that touch triggers different posture modulation schemes to sustain balance. Strong bottom-up control of lower extremity coordination is provided by touch, warranting a stable base to support global postural patterns, with or without vision.

Earlier research has shown the transformation from single- to multi-segment response in the human body was reported when a patient is faced with higher intensity external perturbation [7]. From our results, we observe changes in posture strategy from ankle to hip occur at 0.6 Hz and above with closed or open eyes and without the influence of hand contact, Even though changes without vision take place later than with vision, normalised MVC at the four muscles is higher with closed eyes, although the difference is not significant. These results concur with previous report, where during a low frequency of~ 0.25 Hz with closed eyes patients tended to stiffen their body more to avoid body segment separation, and muscle activation was also reported to be higher than with open eyes [7]. Their results also support our finding that the potential for change in posture strategy is higher with open eyes than closed eyes, although they also reported the variability of phase is higher with vision than without vision. However, the differences may result from variations in the experiment set-up used.

In the current study, we observe the additional somatosensory input changes the posture strategy, as touch provides much more stability. The small level of sensory information from the fingertip receptor results in avoiding inter-segmental separation of joints. Synchronisation of the movement of ankle, hip, and head, were observed and ankle strategy only occurs when lower frequency perturbation is applied. Our results and other research, show light fingertip touch is able to provide better stability by reducing joint motion, and COM and COP displacement [98-100]. This study provides supplemental information to support where that where there is touch, no multi-segmented separation takes place and only ankle strategy is significant. From our results, we suggest that touch promotes better stability. With a small amount of force at the fingertips, muscle activity reduces by more than thirty per cent (Figure 4). However, no influence is observed at the rectus femoris. Inertia triggered from surface perturbation is absorbed by the hand to maintain posture, especially in the abdominal area, which may also cause increased excitation of the rectus femoris. The effect is a reduction in energy required at the lower extremities to produce sufficient support. With ageing, it is well documented that degeneration affects body functions, and a reduction in muscle strength is common [14]. When facing muscle-weakness problems, especially in the elderly, assistive mechanisms such as canes or handgrips can provide better support. This should encourage society and authorities to provide more handgrips. We also observe that touch almost negates the effect of losing vision input. This validates the use of devices by the visually impaired not only to guide direction, but also to support balance. The current study aimed to imitate conditions inside a moving vehicle in order to understand changes in posture coordinates as a result of touch. Even though some limitations occurred in the set-up, the triggering of postural modulation was satisfactorily observed in the subjects. In conclusion, our research demonstrates that light touch results in better stability, by maintaining posture as a single-link inverted pendulum. The findings expand the current knowledge on posture balance strategies when additional somatosensory input is applied. Furthermore, much of the role of vision in the balancing process can be substituted by touch. We recommend strongly that support mechanisms that involve touch or grasping are used by those with weakened lower extremity function or vision impairment, especially while standing or walking inside a moving vehicle. The results of this study provide important supplementary information to enhance the knowledge on human balance models.

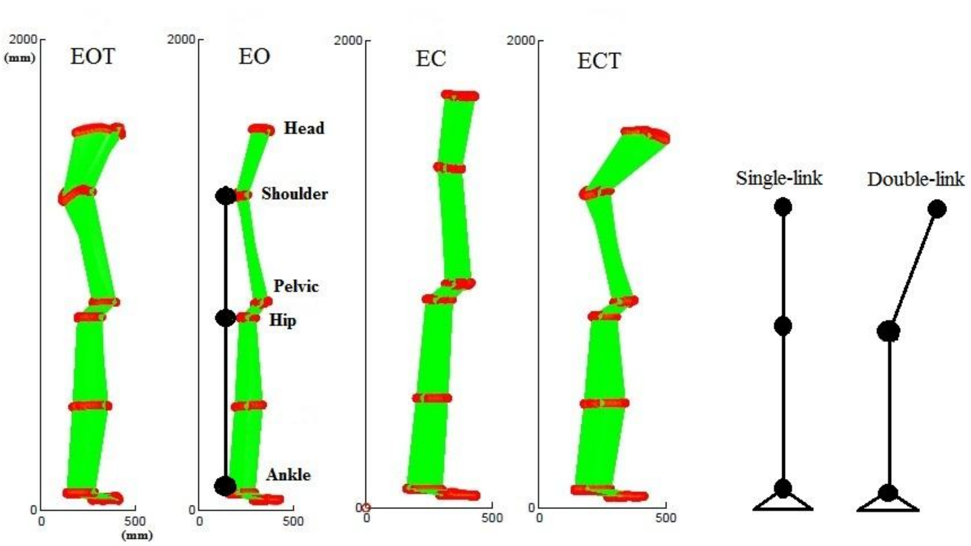


Figure 51 Change in posture modulation of Subject 6 during 0.2 Hz perturbation

5.5 Wrist and finger influenced body motion

It is widely recognised that the existence of additional haptic information from touch allows the hand to move and provide spatial contact information; while also allowing a considerable range of body sway; which in this study, is within the platform displacement range (~70mm for anterior-or direction and ~25% less for posterior direction). However, vertical force generated at the fingertips was shown to be higher during no vision (EC). This indicates that, without vision subjects depended on fingertip receptors to sense and provide information about the body's orientation. In perturbed standing, a force of more than 50g was required to provide significant postural stabilization. This supported by previous research where vertical force of 40 ± 7 g produced during when subject standing heel-to-toe [98] . Furthermore, one interesting finding is about average moment at anterior – posterior's direction. Different directions of moment were recorded between EC and EO conditions; where a significant difference was observed at 0.8 Hz. This indicates the possible existence of different fingertip position preferences; with respect to loss of vision input. The position of fingertip produced during EC was almost similar to a blind person reading Braille (as shown in Figure 6 below); which leads to a moment produced in the posterior direction.

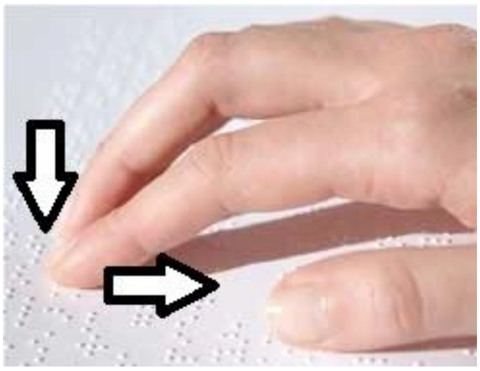


Figure 52 Position of fingertips of a blind person reading Braille

This suggests that the position of fingertip may vary due to sensory loss, in order to gather more sensory information. However, based on the investigation of wrist position, there were no significant different found between different vision inputs. As mentioned in the results section, effect of different vision input influenced fingertip movement but not the wrist. Pressure dependent receptor at fingertip connected directly to ulnar nerve which plays an important role in response to afferent-efferent information transfer in dorsal spinal tract that control the upper part of the body [89].

The wrist flexion and extension degree much influenced by the intensity of perturbation frequency. However, due to limited amount of information due to wrist motion, a detail analysis on the wrist flexion-extension movement cannot be done. The evaluation of wrist motion were depended only on the marker coordinated at ulna bone. For further study on this matter, more market position is suggested in order to gather more information about the wrist movement especially at dorsal surface of hand.

5.6 Simulation of human posture control using joint stiffness profile

Simulation of the new approaches of human posture control have allow a dynamic features of actual posture sway which cannot be fulfilled by one segment inverted pendulum [101]. By using joint stiffness profile gathered from the experiment procedure, this parameter at both ankle and hip joint were included into the double inverted pendulum model. However, some modification to the equation have been made especially to the hip joint. In order to represent the behaviour of hip joint, it must represent in term of damping. Stiffness profile from the experiment were transforms into damping parameter by converted it into torque and then included into double inverter pendulum equation.

Sensory limitation conditions which were normal, loss vision and loss vestibular sense individual have been represented with different stiffness profile. The result have shown that this model correlated with the experiment data in term of displacement and movement pattern that showed the existence of ankle and hip strategy at particular condition. However, this model is still required improvement especially regarding to ripple in the output produced. This model still required filtering in order to produce a smooth response. Although joint motion displacement range produced were still within the experimental data, it is suggested that some improvement is still required. It was observed that some modification is still required. Furthermore, adaptation model also have shown a good correlation with the experiment data while a small adaptation rate is shown, 2 percent.

Compared with existence model by Maurer and Peterka (2005), this new approached have able to present the neuromuscular control model. Not only able to simulate joint motion displacement but the existence of posture strategy due to particular constrains either surface perturbation intensity or sensory weakness. This model provide a decision support to the existence approach in order to improve assessment for balance ability problem.

CHAPTER 6

CONCLUSION

6.1 Conclusion

As stated in the literation review, previous research related on balance ability; from both intrinsic and extrinsic degeneration factor, available assessment, the need of new approach of balance assessment, the use of inverted pendulum model, measurement of joint stiffness in related to neuromuscular response to the introduction of new postural control model were thoroughly reviewed. Even though a considerable amount of research work have been done, it could be said that there are still a space for improvement. Recently, there are increased number of interest in using postural modulation strategy as one of important indicator for balance ability other that both the COP and COM properties. And, it is believed that in this era, the technology or concept will rapidly evolving.

Overall, three main objective of this thesis stated in Chapter 1 have been fulfilled. Joint stiffness characteristic have shown a significant pattern regarding to balance ability, different sensory conditions and external perturbation that applied with the purpose of triggering different postural modulation and mimicking elderly condition. Furthermore, rise of concerns regarding to the effect of additional somatosensory input from the light fingertips touch have encouraged the additional experiment to be conducted. In this experiment, it was observed that touch not only improved stability by reducing postural sway but also reduced stiffness and muscle activation magnitude. This finding reassured that the used of any supporting tool like cane or handrail to be implemented in nursing home or even public area. Analysed parameter such as joint stiffness value were then included into double inverted pendulum model to signify neuromuscular control of human posture control. Each sensory condition were represented with different stiffness gain. Results from the simulated output have shown that this model not only followed the experiment data on the joint displacement range but also able to demonstrate the existence of posture strategy at specific conditions. This model is expected to be a fundamental of development of new assessment approached to support to the existence method. This research have able to proof the reliability of measuring balance ability from the postural modulation strategy which to be more specific, joint stiffness. However, further investigation and study is still warranted to especially to determine the joint stiffness characteristic due to specific disease.

6.2 Future recommendation

This research thesis also had several noteworthy limitations. Some limitation have been stated in the introduction chapter. In this section, some recommendation for improvement of future work as mention below;

a. Clinical data collection

   - Investigate the response of joint stiffness due to specific disease to improve the data reference and its reliability

- The use of knee lock somehow creating unnatural behaviour of posture response. Knee unlock condition data may be considered to improve the reliability of joints stiffness pattern.

b. Simulation model

- Improvement of the model is still needed especially the involvement of PID controller to simulated more precisely the neurological response
- Additional features such as estimation of COP-COM displacement, estimation according to disease and etc.

Last but not least, this thesis will proposed a possible assessment system concept as for future balance ability measurement system as illustrated in Figure 49. According to this concept, the assessment system will be consisted with a platform which required to produce a controlled perturbation. This will allow the users to control amount of intensity and difficulty of perturbation required by each patients according to their situation. In current study, 6-DOF of movable platform by COSMATE Co., Ltd, Japan has provide a sufficient movement to trigger posture change. However, further investigation should be conducted in order to determine a more simple and portable platform. This important consideration should be taken into account when considering it application in clinical environment.

The other necessary system is gesture changes recorder. During the experiment, seven precise infrared camera, (HWK-200RT camera, Motion Analysis, USA) were used. At that stage, a precise camera is required in order to determine as small changes on posture. A small changes should be detected. Besides, seven camera were used in order to provide 3 dimension data recording. However, for the future concept, 2 dimension camera is acceptable. In current market, infrared sensor with built in colour camera such as Kinect by Microsoft has shown a good criteria and should be considered for this system. Besides, it is also reported having an acceptable accuracy when compared with the HWK-200RT camera. Based on the study done by Clark et at, (2012), the Microsoft Kinect and 3D motion analysis systems had comparable inter-trial reliability and excellent concurrent validity when assessing the motion of the sternum, pelvic, knee, ankle the lateral and anterior trunk flexion angle [102]. Besides that, the experiment done by our research member to record posture motion during walking have shown that the Kinect camera have an acceptable level of accuracy for motion capture, low cost, portable and easy to handle. These features have make this camera the best candidate for this future system.

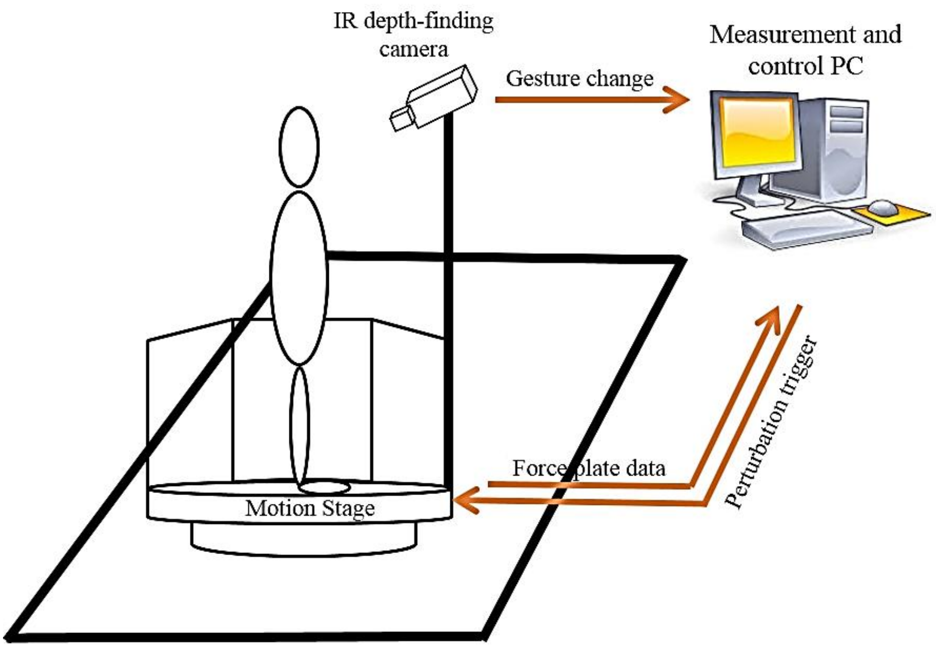


Figure 53 System concept for future development. This illustration describes overall overview of the purpose system structure.

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APPENDIX

(PUBLICATION)

Analysis of joint stiffness of human posture in response to balance ability and limited sensory input during dynamic perturbation

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Abstract: Joint stiffness causes posture movement restriction. However, how joint stiffness responds towards imbalance still remain unclear. The objective was to observe the relationship between the joint stiffness value with balance ability and the efficient amount of stiffness required to maintain posture sway. Moreover, the effects of limited sensory inputs were also discovered. The joint motion at different external perturbations was recorded when different sensory inputs were applied. The results showed that the measurements of joint stiffness displayed imbalance; whereby, less-balanced individuals produced a high stiffness value correlating with the functional reach test (FRT) score. Furthermore, the stiffness value at the joints produced a significant difference with different sensory conditions and when various perturbation frequencies were applied ($p < 0.05$). The stiffness ratio between joints was also obtained. This study had successfully acquired the correlation between joint stiffness with balance ability, sensory inputs and joint synergy which crucial to maintain the posture balance.

Keywords: joint stiffness; sensory input; dynamic perturbation; balance ability; biomechanics.

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1 Introduction

Stability is an important aspect for every moving object. In humans, balance or stability degeneration is due to many factors including ageing, disease and impairment. In clinical applications, a conventional test (a scoring approach performed by physiotherapists) such as the functional reach test (FRT), berg balance scale (BBS) and Tinetti assessment tool (TAT) were popular due to their simple set-up and only required a short time. However, there were some reported drawbacks such as exposure to human error (Jenkins et al., 2010; Tinetti et al., 1986), required continuous evaluation to detect causes of a lack of balance ability (Mao, 2012; Ishimoto et al., 2012) and the singular focus on performance of desired movements (Raîche et al., 2000).

One of the most common clinical tests was the FRT where it is frequently used to measure the balance ability (Jenkins et al., 2010; Maranesi and Fioretti, 2012). compared to other assessments, the FRT test is very simple and only requires patients to stand quietly and reach forward as far as possible. However, the reliability of this test for dynamic stability was questionable, especially when evaluating the balance ability during dynamic conditions. Previous research reported that the reach forward distance was affected by the adopted reach strategy (Liao and Lin, 2008). Thus, they suggested that for dynamic stability, both measurements of reach distance and movement pattern should be considered for more accuracy. On the other hand, increasingly sophisticated systems were also developed such as the computed dynamic posturograph (CDP) and biodex balance system SD; both were based on the centre of pressure (COP) and centre of mass (COM) properties assessment to measure the intensity of body sway. These systems have been commercialised and had shown a high reliability for measuring fall risks and monitoring fall prevention programmes (Parraca et al., 2011). However, systems such as posturography were reported to have poor discriminative abilities, very wide choice of descriptive measures for postural control and lack of ecological validity (complex attachment on patient's body) (Browne et al., 2002). They only depend on the investigation of COP-COM properties and are unable to display any change in postural modulation schemes. According to Horak, the central nervous system's (CNS) ability to maintain a balanced position can be seen once the body faces external perturbation and attributes can be viewed through limb motion (Nashner et al., 1982). Furthermore, the existent methods are sometimes unable to detect changes in balance due to sensory disorder and thus, the assessment method which can represent both kinematics and physiological factors (neuromuscular system) was warranted (Patel et al., 2009). They have opened a wide area of research opportunity, providing an alternative measure in calculating the balance ability.

One of the most promising approaches was measuring joint stiffness. Initiation and limitation of joint motion can be observed through the joint sway; the difficulty of maintaining an appropriate amount of sway was described as joint stiffness. Joint stiffness was observed to have the capability of describing factors that affect the balance ability such as ageing (Lacour et al., 2008), motor learning ability (Azaman and Yamamoto, 2013) and movement performance due to disease or impairment (McGinnis et al., 2013; Tateuchi et al., 2011; Wu et al., 2006). It is also able to address neuromuscular properties such as the reflex response (Fitzpatrick et al., 1992; Kearney et al., 1997), muscle activation (Reynolds, 2010) and the active and passive mechanism of the ankle joint sway (Granata et al., 2004; Cenciarini et al., 2012; Sung et al., 2010). The theoretical study on the effect of joint stiffness by Edwards (2007) illustrated that both joints counteract each other to maintain the balance position (Edwards, 2007). This result was further supported by Suzuki et al. (2011) through the analysis on the double pendulum model during human quiet stance (Suzuki et al., 2011). However, this counteraction strategy between ankle and hip during the perturbed stance still remains unclear. In case of diseases, especially stroke, spinal injury, traumatic brain injury, or even multiple sclerosis where patients normally face spasticity symptoms, they always reported to have a lack of balancing ability due to limited range of motion (ROM) at the joints. Thus, there was an increased amount of stiffness at the affected joints. In such cases, stiffness becomes a complete drawback. For stability, however, information on an appropriate ratio between these joints is believed to be important.

In previous research, there were many types of external perturbation introduced to examine postural modulation changes during the quiet stance such as surface orientation (Buchanan and Horak, 1999; Ishizawa and Yamamoto, 2012; Cenciarini et al., 2012), external force towards the upper body (Fitzpatrick et al., 1992) and varied perturbation frequency (Azaman and Yamamoto, 2013; Buchanan and Horak, 1999). Moreover, those perturbations were found to be suitable in triggering different somatosensory and proprioceptive senses in order to investigate the balance ability under dynamic conditions. According to previous research, posture modulation varied due to perturbation orientation and frequency (Buchanan and Horak, 1999). Furthermore, external perturbation was also used to investigate changes in posture sway due to the degeneration of the posture control function.

Stiffness was specified as the slope of the linear angle-moment curve. In measuring joint stiffness, several assumptions should be considered. Previous research described a crucial requirement for stability in which the sum of effective ankle stiffness (K_a) and gravitational stiffness (K_g) or in other terms the 'load stiffness', must be larger than zero; $K_a + K_g > 0$ (Pinter et al., 2008). This effective stiffness described the intrinsic properties of muscle and reflex response of the mechanism. Stiffness was also described in terms of active and passive behaviour. Passive stiffness arose from mechanical viscoelasticity of the joints; meanwhile, active stiffness was required for stabilisation which was similar to reflex response (Suzuki et al., 2011). The term of active and passive behaviour was also used by Schmid et al. (2011) to describe the ability of the human to adapt to their surroundings when trying to maintain a balanced position (Schmid et al., 2011). Many control models were developed based on the estimation of joint stiffness characteristics to predict stability. However, none have related it with real balance ability and its response due to sensory weakness and manipulation. Thus, investigation on joint stiffness may be a potential measure to determine balance ability. We hypothesised that higher joint stiffness will produced due to imbalance.

The objective of this study was to observe the correlation between the joint stiffness value and balance ability and the efficient amount of stiffness at the ankle and hip joints required to maintain balance. Affects from limited vision and vestibular sensory input to joint stiffness were discovered. The findings from this study will contribute to further investigation on posture sway analysis so that an alternative measurement for balance ability based on joint stiffness can be determined.

2 Methods

2.1 Subject preparation

In this study, nine healthy young male subjects (aged 24.24 ± 2.19 years old) had participated. Each subject was given an explanation regarding the procedure of this experiment and provided with informed written consent prior to participation and comply with declaration of Helsinki. Information on subject's history of falls and physical conditions were recorded as references.

2.2 Experiment set-up

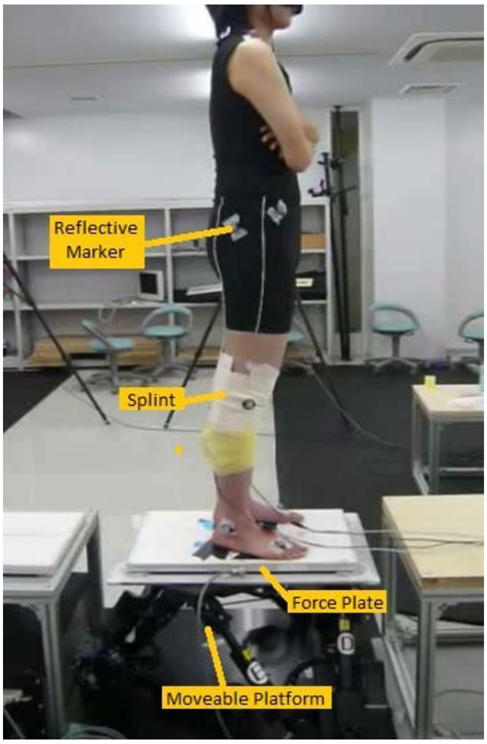
The subjects were exposed to two types of sine wave external perturbation, which were translation (posterior-anterior) and tilt up-tilt down (superior-inferior) at four different frequencies (0.2, 0.4, 0.6 and 0.8 Hz) produced by a movable platform (MB-150, COSMATE, JAPAN). The two varied perturbations were applied to observe how the ankle and hip joints reacted towards different surface perturbation orientation. Displacement for translation perturbation was 70 mm; and 6° (0.104 rad) for the tilt up-tilt down.

Furthermore, the subjects' vision [eyes closed (*C*) and eyes opened (*O*)] and vestibular sense [limited neck movement with eyes opened (*NO*)] were manipulated. The vestibular system input was controlled by constricting the movement of the neck and head using a neck collar (ADFIT collar, ADVAN FIT). This step interfered with the function of proprioceptors in the neck muscle, thus leading to vestibular malfunction (Mann, 1981). Subjects were asked to stand quietly and comfortably and try to maintain their position along the repetitive perturbation movement (for 60s). Joint motion data was collected from 17 reflective markers (placed at 3rd metatarsal, lateral malleolus, lateral condyle, trochanter of the femur, iliac crest, acromion of scapula and top of head) using motion analysis (HWK-200RT camera, Motion Analysis, USA) and force plate (9286A, KISTLER, JAPAN). Each trial (*two types of perturbation;*three type of sensory manipulation; *four different frequencies*) was recorded for 65s with locked knee joints using splint to prevent bias movement from the knee. Before the experiment started, all subjects underwent a FRT to evaluate initial balance scores (Duncan et al., 1990). This assisted in classifying each subject according to initial balance ability for better understanding. The experiment set-up was shown in Figure 1:

Figure 1 Diagram of experiment set-up, (a) the use of neck collar on subject (b) the HWK-200RT camera to capture motion from reflective camera (c) subject preparation (see online version for colours)

(a)

(b)



(c)

2.3 Data measurements

Joint stiffness was measured according to the free body diagram (Figure 2) and equation below. The ground reaction force (F_y), the horizontal component in y-direction direction force (F_y) and force plate moment at the x-axis (M_x), recorded from force plate with 200 Hz sample rate with were filtered with low pass 2nd order Butterworth filter. Joint movement coordinate (x, y, z) obtained from motion analysis system with 1 kHz sample rate were used to measure joint sway angle (θ_a and θ_h) and body segment length (h_A, h_H and h_{seg}) for segmental COM locations. The average joint stiffness was measured from the torque of the ankle joint (τ_{ankle}), torque of the hip (τ_{hip}) along with the period of recording time (T). The COP displacement was determined from (3) below where d_z was the distance from the surface to the platform origin. The COM was obtained from the total segment torque as mention in (3).

Figure 2 (a) An inverted pendulum model of human body (b) Moveable platform movement (c) Force plate diagram (see online version for colours)

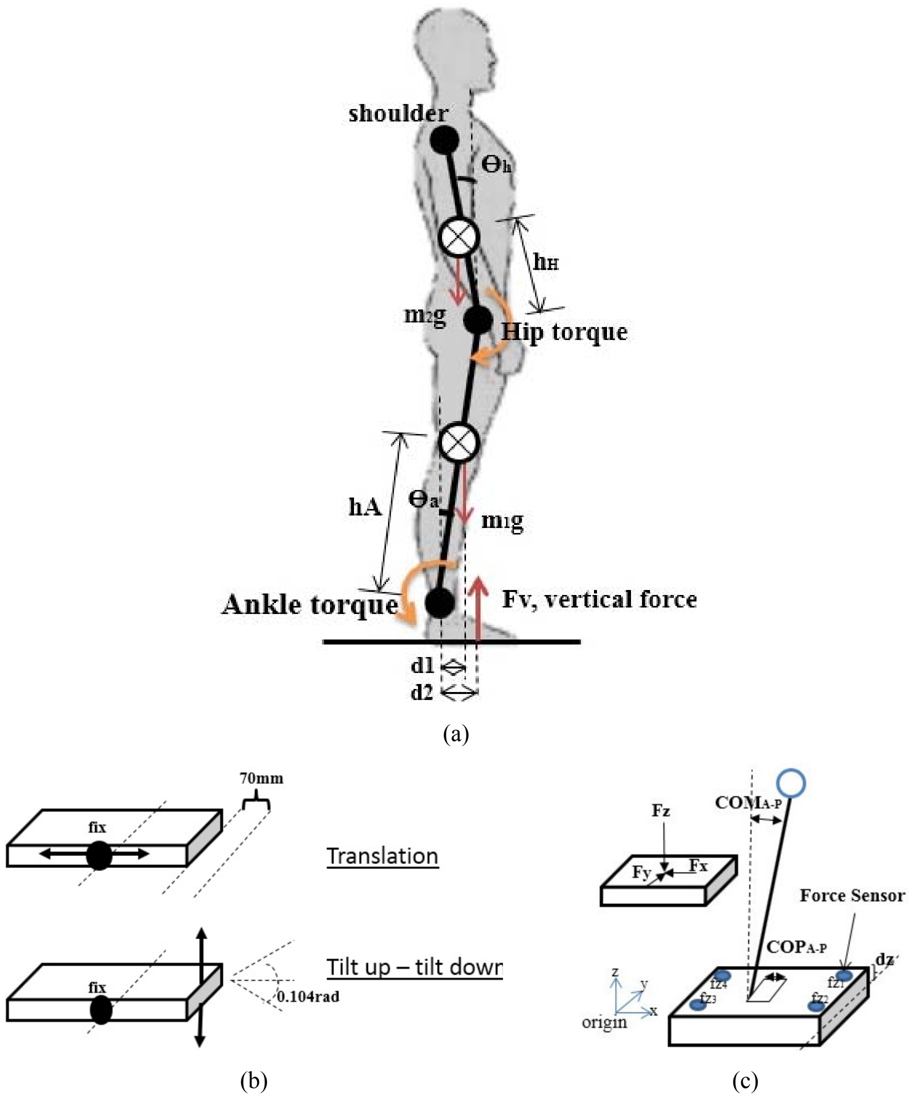
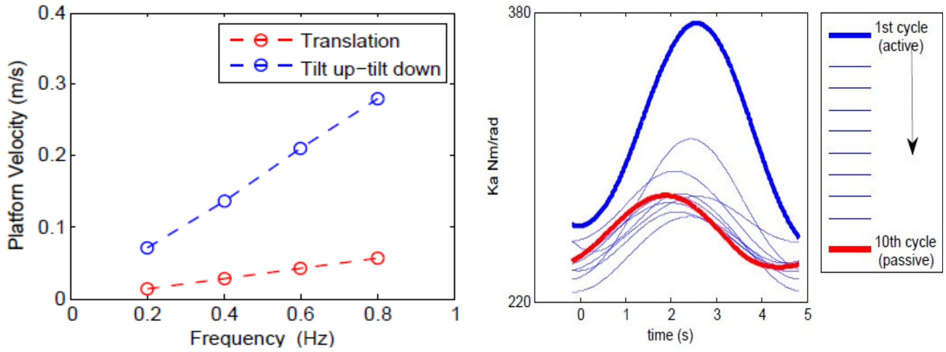


Figure 3 (a) Linear velocity produced by movable platform at four different frequency (b) Example of stiffness pattern at ankle joint during 0.2 Hz of translation for subject 1 (see online version for colours)



(a) (b)

Note: This windowing technique have been used to measure adaptive percentage to detect motor learning ability where first cycle was define as active mechanism meanwhile the rest as passive

The adaptation strategy of CNS towards the joint stiffening response was determined by measuring the area under graph (AUG) using the trapezoidal rule, applying equation (6) and equation (7). $K(t)$ was joint stiffness along the perturbation period, where t was the time for one cycle of perturbation and i was the number of cycles as shown in Figure 3(b).

- *joint torque:*

$$\tau_{ankle} = m_1 g h_A \sin\theta_a, \; \tau_{hip} = m_2 f h_H \sin\theta_h \tag{1}$$

assume that;

$$m_n g \approx \beta_{seg} F_v$$

where β_{seg} is percentage from Plagenhoef's body segment weight data

- *joint stiffness at ankle (Ka) and hip (Kh):*

$$Ka(Nm/rad) = \frac{\tau_{ankle}}{\Delta\theta_a}, \quad Kh(Nm/rad) = \frac{\tau_{hip}}{\Delta\theta_h} \tag{2}$$

- *COP and COM displacement in anterior – posterior direction:*

$$Mx = a(f_{z1} + f_{z2} + f_{z3} + f_{z4}),$$

$$COP_{A-P}(mm) = \frac{M_x(F_y \cdot d_z)}{F_v}, \quad COM_{A-P}(mm) = \frac{\sum(F_v \cdot \beta_{seg}) \cdot h_{seg}}{F_v} \tag{3}$$

where a = sensor offset value, d_z = thickness parameter of force plate

- *COP and COM velocity:*

$$COP_{v,i}\ (\text{mm/s}) = \frac{COP_{A-P,i}(\max) - COP_{A-P,i}(\min)}{t_i},$$
$$COM_{v,i}\ (\text{mm/s}) = \frac{COP_{A-P,i}(\max) - COP_{A-P,i}(\min)}{t_i} \tag{4}$$

- *COP and COM range:*

$$COP_r(mm) = COP_{A-P}(\max) - COP_{A-P}(\min),$$
$$COM_r(mm) = COM_{A-P}(\max) - COM_{A-P}(\min) \tag{5}$$

- *area under graph (AUG):*

$$AUG = \int_1^t K(t)dt \tag{6}$$

- *adaptation percentage (%):*

$$Adaptation_i(\%) = \frac{AUG_i\, AUG_{(i+1)}}{AUG_i} \times 100\%$$
$$i = 1, 2, 3 \ldots n-1 \text{ where } n \text{ is total number of cycle} \tag{7}$$

2.4 Data and statistical analysis

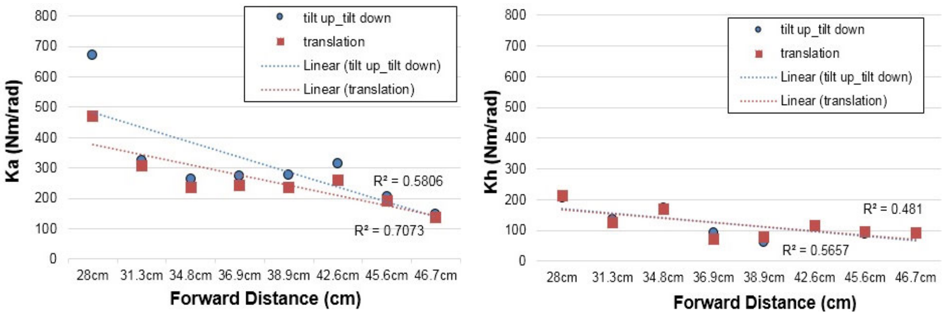
This study focused on measuring the amount of stiffness at both the ankle and hip joints. A comparison of stiffness was made using statistical analysis (mean and standard deviation); and data from each subject was compared using One way ANOVA with Turkey post hoc test at a significant level of $p < 0.05$ (vs. sensory and vs. frequency). Relationship between FRT's score and stiffness was computed using linear regression fit. Correlation analysis was done using the Pearson function. Moreover, to determine the stability region estimation, density plots of *K*/*mgh* ratio which is the comparison between amount of stiffness during perturbed (*K*) and unperturbed standing (*mgh*) using density plot. All measurements of stiffness, COM-COP properties, estimation of stability region, adaptation analysis and statistical analysis were completed using the MATLAB software.

3 Results

3.1 Average joint stiffness vs. FRT scores

Figure 4 indicated regression analysis to observe the relationship between FRT's score and stiffness. Even though all of the subjects' scores were within an acceptable balance range (> 25.4 cm indicating adequate balance ability) (Duncan et al., 1990), correlation made between the average stiffness value at low perturbation frequency (0.2 Hz) and the FRT scores have shown a significant trend to supported the hypothesis. Subjects with high scores produced less joint stiffness compared to low scorers. Based on the results, it was suggested that patients with less balance ability have stiff joints at both the ankle and hip.

Figure 4 Joint stiffness vs. the FRT scores at ankle (Ka) and hip joint (Kh) during lower frequency perturbation (0.2 Hz) (n = 8) (see online version for colours)



3.2 Body sway under various condition

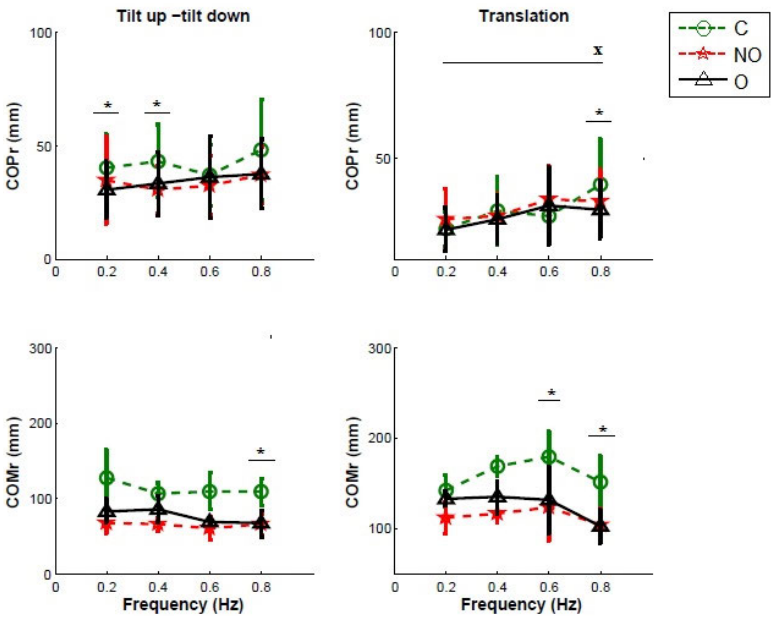
Body sway due to various condition was determined based on COP and COM properties. Figure 5 have shown that effect of varies perturbation type, sensory condition and frequency towards COP-COM displacement range and velocity. By comparing between types of perturbation, COP range (COPr) at tilt up-tilt down was higher than the other perturbation. Meanwhile, the COM range (COMr) was greater during translation perturbation. Between different frequency, the COPr increased with frequency and a significant different between frequency was only observed during translation perturbation. For the COMr, it reduced with the increase of frequency, however, no significant different between frequency ($p > 0.05$) (F (3, 60) =1.19, $p = 0.12$). By comparing between difference sensory condition, significant different (O vs. sensory) was observed only with no vision input (*C*). Furthermore, it is observed that without vision input (*C*), a high COPr and COMr were produced compared to other sensory conditions. Meanwhile, *NO* condition does not differ much from *O*. This finding suggested that without vision, body sway more regardless type of perturbation.

On the other hand, COP-COM velocity were significantly different between different frequencies ($p < 0.05$) (F (7,160) =6.9, $p = 0.004$) and (F (7,160) = 12.25, $p = 0.032$) respectively. However, it were insignificant different between sensory condition. Comparison between these two types of perturbation, COP moved significantly during superior-inferior movement meanwhile during posterior-anterior movement for COM.

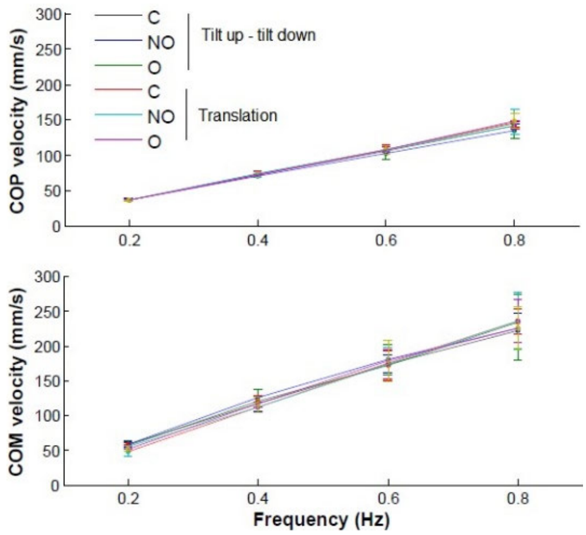
3.3 Joint stiffness at various condition

Based on Figure 6, in general, ankle joint stiffness (K_a) increased with frequency; with correlation $R^2 > 0.5$ as shown in Table 1 during both perturbation. A significant different was found during translational perturbation ($p < 0.05$) (F (3, 36) = 6.59, $p = 0.004$). Meanwhile, hip joint stiffness (K_h) shown a small decreased with frequency with negative correlation for both perturbation (Table 1). During normal condition (*O*), a coaction strategy between both joint was observed as correlation become negatives (Table 1).

Figure 5 (a) COP-COM range at both type of perturbation (b) COP-COM velocity (see online version for colours)



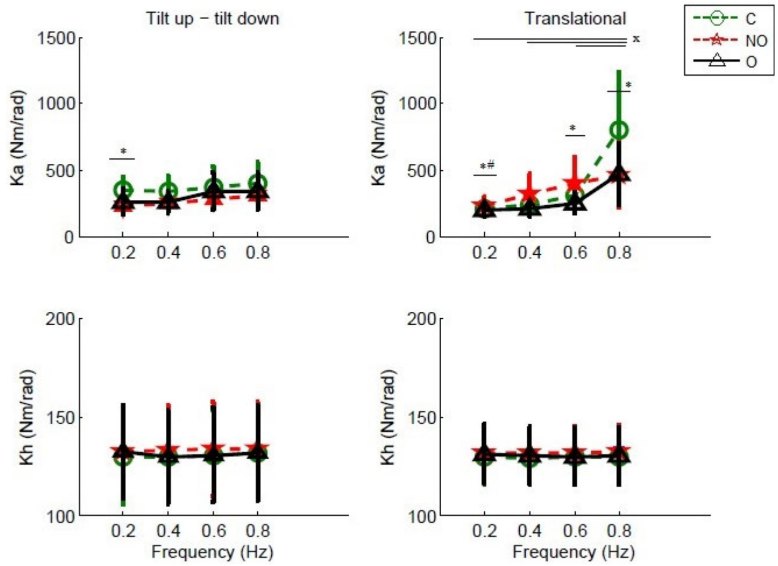
(a)



(b)

Notes: (*) indicated *O* significant different with *C* ($p < 0.05$)
(^{x}) indicated significant with frequency different

Figure 6 Comparison of the average ankle and hip joint stiffness between *C* and *O* conditions at four different frequencies (0.2, 0.4, 0.6 and 0.8 Hz) (see online version for colours)



Notes: (*) indicated *O* significant different with *C* ($p < 0.05$)
(#) indicated *O* significant different with *NO* ($p < 0.05$)
(^{x}) indicated significant with frequency different ($p < 0.05$)

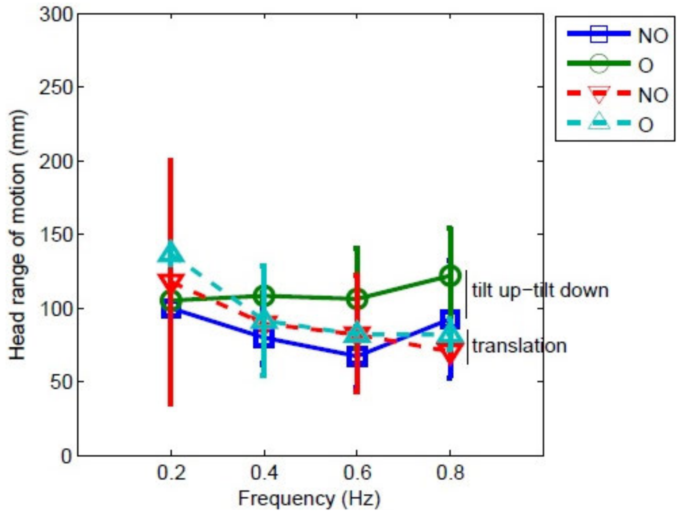
Table 1 Correlation coefficient (R^2) between frequency of perturbation, K_a, K_h, COP and COM parameter

Cond.	*Tilt up–tilt down*						*Translation*					
	K_h	*Freq.*	*COMr*	*COPr*	*COMv*	*COPv*	K_h	*Freq.*	*COMr*	*COPr*	*COMv*	*COPv*
O												
K_a	–0.07	0.88	–0.98	0.91	0.86	0.87	–0.32	0.86	–0.98	0.53	0.82	0.87
K_h		–0.11	–0.23	0.18	–0.13	–0.12		–0.76	0.22	–0.95	–0.80	–0.74
NO												
K_a	0.92	0.95	–0.37	0.47	0.98	0.99	0.84	0.99	–0.22	0.91	0.99	0.98
K_h		0.93	–0.68	0.29	0.94	0.93		0.86	–0.63	0.57	0.86	0.86
C												
K_a	0.95	0.92	–0.37	0.46	0.91	0.92	0.26	0.86	–0.24	0.92	0.85	0.87
K_h		0.90	–0.41	0.71	0.88	0.90		–0.03	–0.68	–0.11	–0.03	–0.02

Notes: The (-ve) value indicated negative correlation

Analysis between sensory manipulation condition have shown that a significant different was only found at K_a during eyes closed (*C*) at certain frequency and perturbation as shown in Figure 6. Without vision sensory input (during *C*), average stiffness at both joints were observed to be higher than the normal condition (*O*). The effect of limited vision input on the produced ankle joint stiffness was not different between the applied variant surface perturbations.

Figure 7 ROM for head (mm) during normal condition O and when neck collar were applied (NO) (see online version for colours)



Note: A significant different with different frequency was found during translation perturbation ($p < 0.05$)

However, mainly no significant different found for *NO*. The use of the neck collar was observed to effectively limit the head movement as the range of head movement was smaller; about 40% than normal conditions ($p < 0.05$) (Figure 7). Furthermore, during normal conditions (*O*), the observed head motion varied according to surface orientation. As perturbation frequency increased, head movement during the tilt up-tilt down increased but it was reduced during translation perturbation. When the neck collar were applied, the head motion was reduced with the increase of perturbation frequency during both perturbations. By analysing stiffness value produced, only the ankle joint was observed to be more stiffened during the *NO* condition; which at 0.4 and 0.6 Hz.

For the hip joint, sensory input manipulation condition (both *C* and *NO*) did not showed any significant different to value of stiffness produced at *O* condition.

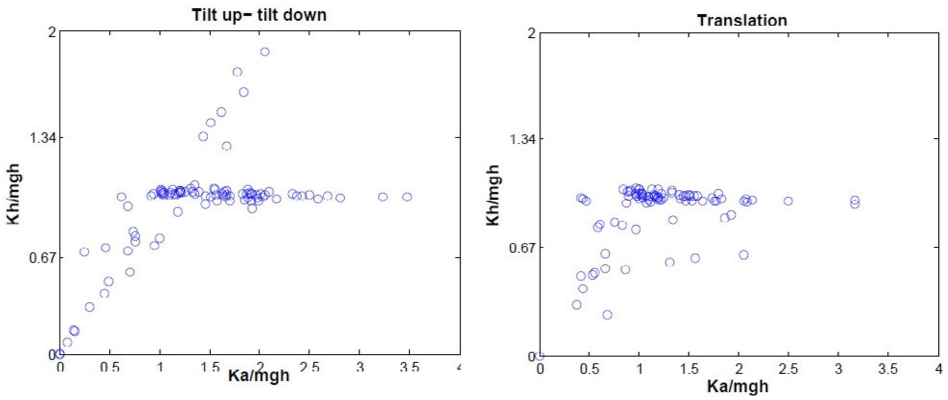
3.4 Correlation between joint stiffness and COP-COM properties

Both perturbations generated various COP-COM properties at the posterior-anterior plane. Joint stiffness also correlated differently with COP-COM properties. Both hip and ankle were observed to negatively correlate with COM range (COMr) at almost all conditions which meant that high stiffness was required to keep COM in a small range. The K_a was noticed to have a positive correlation with COP-COM velocity at all conditions. However, the K_h differed according to the sensory condition and surface perturbation. Stiffness at the hip was negatively correlated with COPv and COMv at *O* during both perturbations. Meanwhile, it was positive at *NO* and *C* during the tilt up-tilt down and *NO* during translation. The results illustrate that joint stiffness not only required to reduce body sway displacement but also to face the increase in sway velocity.

3.5 Joint stiffness and stability

Since the subjects who participated in this study had adequate balance ability, the estimation for both ankle and hip joint stiffness value required dynamic stability, which was determined. The stability area that established the load stiffness ratio at all conditions for the ankle and hip was shown in Figure 8.

Figure 8 Density plot to determine the stability region in the (Ka/mgh, Kh/mgh) plane (see online version for colours)



Note: Blue area (dense area) indicated the amount of stiffness where most of the subjects had applied during all conditions

Based on Figure 8, the concentrated area presented the stability area where it revealed an appropriate amount of load stiffness ratio for the perturbed condition. The load stiffness ration was determined by comparing the amount of joint stiffness during standing with perturbation and without perturbation, $K_{np} = mgh \sin\theta$ where $\sin\theta \approx 1$. Thus, joint stiffness during unperturbed stance can be assumed to be *mgh*. Concerning. Concerning the tilt up-tilt down perturbation, the ratio range was between $1.0 < K_h/mgh < 1.07$ and $1.0 < K_a/mgh < 1.8$. Regarding the translation frequency, it was in the range of $0.9 < K_h/mgh < 1.0$ and $0.8 < K_a/mgh < 1.5$. According to previous research by Suzuki et al. (2011) which performed for quiet standing, the load stiffness ratio was

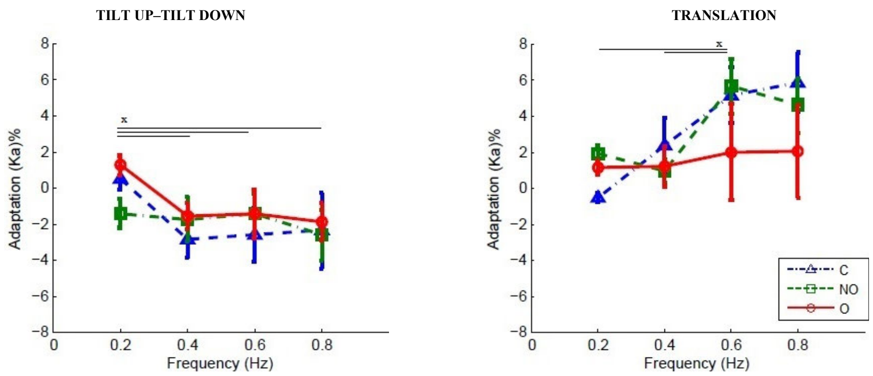
$$\frac{K_h}{mgh} > 0.2 \quad \text{and} \quad \frac{K_a}{mgh} > 1.0$$

which was smaller compared to recent results. It may be due to the external dynamic perturbation that was applied, whereby a higher ratio was required to maintain the desired position. Thus, this result suggested that for perturbation velocity less than 0.3 m/s, $0.95 < K_h/mgh < 1.035$ and $0.90 < K_a/mgh < 1.65$ are required for optimum stability.

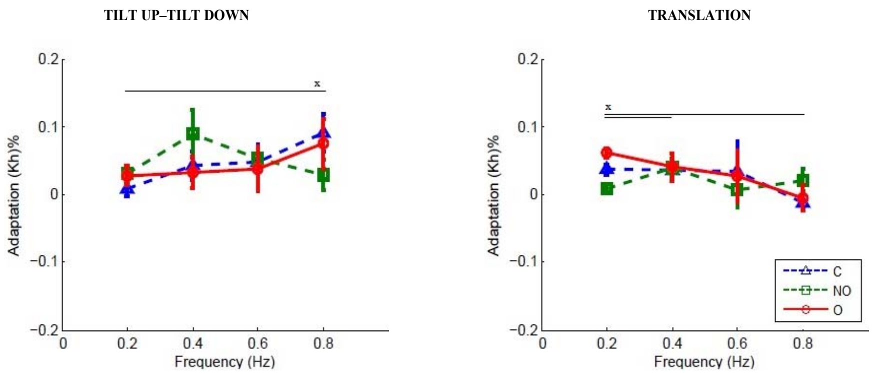
3.6 Adaptation ability during repetitive perturbation

The ability to maintain a balanced position over repetitive external perturbation can indicate the degree of motor learning ability for each individual. As previously mentioned, past studies have shown that the reduction of muscle activation amplitude and postural sway over repetitive perturbation indicated enhanced adaptation ability. In this experiment, the adaptation ability was measured by comparing the area under the graph (AUG) of each cycle of the joint stiffness response. Based on the results in Figure 9, the average adaptation percentage displayed that the normal condition (*O*) displayed better adaptation condition in comparison to the sensory manipulation condition (*NO* and *C*) as the percentage became positive. However, an insignificant difference was found. A significant difference was only discovered between various frequencies ($p < 0.05$) at all perturbations (($F_{ankle,tt}$ (3, 36) = 4.48, p = 0.033), ($F_{ankle,t}$ (3, 36) = 4.06, p = 0.045), ($F_{hip,tt}$ (3, 36) = 8.48, p = 0.023), ($F_{hip,t}$ (3, 36) = 3.39, p = 0.0445)). On average, ankle joint stiffness reduces for about 2% at each cycle; on the other hand, it only reduced about less than 0.1% for hip stiffness at the lowest frequency (0.2 Hz). This applied to both perturbations. However, with the increase of frequency, adaptation was varied; especially at the ankle joint which depended on perturbation. As the frequency increased (0.4 to 0.8 Hz), the adaptation percentage of K_a during the tilt up-tilt down perturbation was reduced and became more negative; however, it increased and became more positive during translational perturbation. These situations addressed the issue that not only sensory weakness affected adaptation ability, but also the frequency and type of perturbation as well.

Figure 9 (a) Adaptation percentage at different frequencies (Hz) for ankle stiffness (b) Hip stiffness (b) (see online version for colours)



(a)



(b)

Note: ([x]) Indicated significant with frequency different ($p < 0.05$)

4 Discussion

4.1 Stability and joint stiffness response

Based on the results, it was illustrated that joint stiffness responded towards imbalance. Compared to other measurement approaches, joint stiffness provided substance in describing the amount of energy or work that the subject faced to maintain the required position. In previous research, joint stiffness was proclaimed to have a strong correlation with muscle contraction; whereby, the reported CNS tend to apply less energy strategy, resulting in less muscle contraction strategy (Missenard and Fernandez, 2011). As previously mentioned, high stiffness was commonly applied by those who faced movement difficulties due to factors such as disease, ageing and impairment. It was hypothesised that a person with less balance ability will apply high stiffness at the joints (ankle and hip) in order to maintain a balanced position; based on previous evaluations of the elderly and patients with disease (Lacour, et al., 2008; McGinnis et al., 2013; Tateuchi et al., 2011; Wu et al., 2006). Thus, the ability to produce less joint stiffness may be strong evidence for substantial balance ability. According to recent results, the amount of joint stiffness was able to distinguish patients according to FRT test scores.

To keep producing high stiffness at each joint is not necessarily good. According to previous research, ankle and hip stiffness were estimated based on theoretical study suggested to be higher than 728 Nm/rad and 179 Nm/rad respectively to maintain posture balance and it is also believed that coaction between ankle and hip is important (Edwards, 2007). However, in case of dynamic perturbation, recent result have shown it is smaller by almost 20%. The K_h was observed to be negatively correlated with K_a at normal condition (*O*) during both dynamic perturbations. However, it was not during less sensory conditions (*NO* and *C*) where K_h was higher when K_a was also high. This suggested that the degeneration of vestibular and vision sensory tend to stiffen the body when faced with external perturbation and unable to generate the coaction strategy between the joints.

Analysis of the stability ratio of stiffness provided information on the adequate amount of stiffness required by healthy people to remain in a balanced position during perturbed standing. As mentioned earlier, research by Edwards and Suzuki predicted the required amount of stiffness at both the ankle and hip joints to maintain quiet standing position (Edwards, 2007; Suzuki et al., 2011). Without considering the perturbed situation, their results were small compared to more recent studies. However, by conducting the experiment in a repeated manner with different intensities of perturbation, it was noted that stiffness at both joints was not necessarily higher, in order to remain stable. It was suggested that joint stiffness must generate at a certain range.

4.2 Weakness in vestibular and vision sensory in response to perception of posture response

Similar to the no vision input (*C*), limitation of the head movement (*NO*) was also recognised to change the posture modulation response; it was shown that during this condition the graviceptor at the head and in the body were distinguished (Mittelstaedt, 1996). Limitation of the head movement actually had limited vestibulo-ocular reflex (VOR) senses. The VOR is a mechanism for triggering eye movement to fix on a desired gaze point when the head was moving. With this mechanism, postural reflex on any changes due to movement can be made quickly and effectively. Projection of the vestibular nuclei regulated the head movement reflex from the neck muscle activation (Purves et al., 2001). Simultaneously, the otolith organ which senses any change in gravity and acceleration will then send information (axons) to the spinal cord to influence the excitation of the muscle to maintain posture.

In the above results, weaknesses in the vestibular sense and vision led to different joint stiffness value based on the type of perturbation applied. If other research indicate that the elderly (who normally face degenerative vestibular function and vision) apply the hip strategy (hip sway more and less stiff) when facing external perturbation, then they would differ from recent results which show that the stiffness response was more affected by perturbation manipulation than sensory manipulation. This raised a question regarding the real effect of vestibular and vision sensory weakness towards the perception of posture response since inconsistent and insignificant differences were found. According to previous research, muscle synergy was not affected by deficiency in the sensory, especially the visual and vestibular system (Ting and McKay, 2007). This was also observed in joint stiffness recently. The phenomenon suggested influence from other mechanisms. In a previous research by Mittelstaedt (1996), it was proposed that the existence of additional graviceptor outside the labyrinth (mechanoreceptor in joints, skin and muscle) also influenced the posture response (Mittelstaedt, 1996). Thus, it was concluded that the manipulation of the vestibular and vision gave less influence to posture modulation; the addictive interaction by somatosensory and graviceptor at other body parts also helped produce the desired counteraction between the lower and upper body since the human posture control system is sensitive to both gravitational and perturbation force. Furthermore, that interaction also depended on individual ability.

4.3 Motor learning ability

In this study, perturbations were applied in a repetitive mode. Other than evaluating the joint stiffness response, this approach was performed to observe the adaptation response which may indicate the motor learning ability. The adaptation percentages were further noted to vary according to the perturbation manipulation. Again, the sensory manipulation also provided less influence since no significant difference was found between the sensory manipulation conditions. Based on previous research, the adaptation ability was observed by the decay rate of the exponential curve (Schmid et al., 2011) where a higher rate was noticed only during open eyes. However, the reduction of adaptation due to sensory manipulation was further scrutinised. A consistent average adaptation percentage value between both perturbations was only noticed during the lowest frequency (0.2 Hz) when both perturbations produced almost similar amounts of COP-COM velocity. With the increase in perturbation frequency, adaptation was hard to achieve due to high difficulties. High adaptation values do not necessarily show a better motor learning ability. In Figure 8(a), *C* and *NO* had a greater percentage compared to *O* during the translation perturbation. In that difficult situation, a greater gap between active and passive components was observed. High active components were required at difficult situations. It was believed that in high difficulty situations (high frequency), adaptation was not a choice. The CNS tend to apply accuracy control to reduce kinematic variability under high speed movement (Missenard and Fernandez, 2011) to maintain desired position.

5 Conclusions

This study found the correlation between joint stiffness responses with balance ability. Without vision input, the results have shown that there was a significant difference in stiffness value between sensory conditions. The negative correlation between the ankle and hip stiffness was common for balanced individuals. Moreover, the synergy between these two joints is important in order to maintain a balanced position under repetitive perturbation. The ratio of load stiffness provided information on the adequate amount of stiffness required to maintain a balanced position. Further investigation will be necessary in order to predict balancing strategy based on both kinetic and kinematics data together with intrinsic factor such as muscle activity.

5.1 Conflict of interest

The authors declare that there are no conflict of interest.

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Ankle Joint Stiffness and Damping Pattern under Different Frequency of Translation Perturbation

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Keywords: Posture control, ankle joint, frequency of perturbation.

Abstract. The change of effective stiffness and damping characteristic of ankle joint are able to indicate degeneration of balance ability due to ageing effect. This paper will discuss the ankle joint stiffness and damping pattern along repeated translation perturbation. Six young healthy subjects were exposed to five trials of five different frequencies of perturbation (quiet standing, 0.2 Hz, 0.4 Hz, 0.6 Hz and 0.8 Hz). The result showed that the mean of effective stiffness was reduced with the increase of frequency applied; meanwhile the mean of damping value increased with increasing frequency. Additionally, a cubic polynomial curve (u-shape) was estimated to represent stiffness pattern when using curve fitting method with correlation $R^2>0.5$. These estimations also suggested that ankle joint does not oscillate like spring-damper system which is based on inverted pendulum model; however, it applied a different strategy to maintain balance, in particular during initiation, middle and termination of perturbation. These also indicate the influence of sensory processing and adaptation to maintain balance under a long period of disturbance. On the other hand, damping pattern seems to be similar over different frequencies and under repeated perturbation. Besides, the change of stiffness pattern at higher frequency of perturbation (0.8 Hz) recommends the change in posture strategy from ankle to hip strategy. These findings indicated that stiffness and damping are able to describe adaptation of human posture strategy to keep balance and motor learning under repeated perturbation.

Introduction

Quiet standing approach is the simplest approach to analyze balance impairment in a patient. Under different external perturbation, changes of center of mass (COM) and center of pressure (COP) can give a hint on weakness of posture control system. Furthermore, dominance of hip or ankle strategy can be observed through this approach. The elderly was reported to have higher COP component than young subject during quiet standing and relied more on hip strategy to keep balance [1]. However, there is still limited information regarding the central of nervous system (CNS) adaptation to maintain balance under different intensities of perturbation and changes due to ageing. One of the reason of balance impairment among elderly was suggested to be caused by the lack of cognitive system which can be observed by measuring joint stiffness [2]. According to Horak, CNS adaptation can be seen once the body faces the perturbation and this adaptation can be seen through the effect of muscle co-contraction which is attributed to limb stiffness [3]. Even though muscle contraction is meant to preserve accuracy of movement under high speed movement, it is less preferred by the nervous system due to high metabolic energy consumption [4]. Besides, in previous research done by M.Casadio et.al, they ignored the damping factor by stating that it does not represent any physiological meaning [5]. However, Ishida and Agarwal was against that assumption in which they suggested that stiffness and damping properties can describe passive properties of muscle fibers and tendons [6, 7].

Ankle joint stiffness and damping were reported to be increased with ageing. Ho and Bendrups have shown that older adults tend to have higher stiffness than young adults and the elderly fallers have larger stiffness compared with the non-fallers under unnoted medial-lateral perturbation [8].

This is similar with damping parameter where elderly is much higher than young subjects [9]. This previous research shows that stiffness has been one of the mechanisms for elderly to maintain balance when exposed to different sensory information [9].

Stiffness strategy to face initiation and termination of perturbation can be a good component to express balance ability and thus, the effect of ageing in posture control system can be observed. Under different frequency intensities, stiffness tends to reduces over frequency increment [10]. However, the characteristic along repeated perturbation especially effect of different stages of perturbation (initiation, middle, termination) still remain unknown. Our early hypothesis assumed that stiffness and damping pattern are similar to spring-damper system under repeatable perturbation. This is due to characteristic of the CNS that tends to apply less energy. For this study, inverted pendulum model is adapted in order to determine the value of ankle joint effective stiffness and damping as mentioned below;

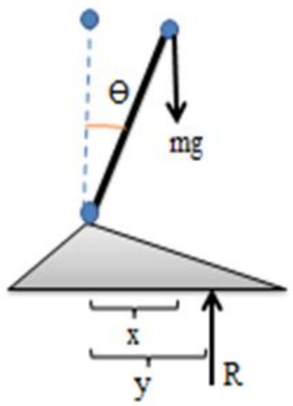

Fig.1. Inverted pendulum model for ankle joint.

Stiffness at ankle joint;

$$K_a = \frac{\tau_a}{\theta} \tag{1}$$

where τ_a is torque of ankle and θ is angle of ankle sway. The oscillation of the model is describes as below;

$$I\ddot{\theta} + B_a\dot{\theta} + K_a\theta = mgh\ sin\theta \tag{2}$$

where I is moment of inertia, B_a is damping at ankle joint, $\ddot{\theta}$ is angular acceleration, $\dot{\theta}$ is angular velocity, m is mass, g is gravitational acceleration and h is a distance of COM from ankle joint.

Based on Figure 1, at equilibrium state, total of ankle joint torque should follow the equation below;

$$mgx + Ry \approx 0 \tag{3}$$

where R is a vertical component of ground reaction force. From eq. (1) and (2),angle of ankle sway is important to determine stiffness and damping parameter. The change of angle over time will determine the way of ankle joint counteract against the oscillation of perturbation which is may results from evoking the intrinsic mechanical properties of ankle joint, and muscle contractile elements which triggered by the CNS.

Methodology

Subject. Six healthy subjects (aged 22.3±0.8 years; height 169±2.5; weight 61±3.1 kg) participated in this study.

Experiment Protocol. Subjects were asked to stand quietly on a moveable platform. Postural sway was induced by six axis motion control base (MB-150, COSMATE, JAPAN). This control base platform created a translation movement of 80 mm at four different frequencies (0.2, 0.4 0.6 and 0.8 Hz). Ground reaction force at anterior posterior direction was calculated from force platform (9286A, KISTLER, JAPAN) data. Ankle joint angles were determined and calculated from the coordinates of reflective markers captured by motion analysis system (HWK-200RT, Motion Analysis, USA) with sampling frequency of 200 Hz. These markers were attached at these following locations of lateral aspect of bilateral limb: metatarsal bone, external condyle, lateral condyle, great trochanter, pelvis, and acromion. Each subject underwent five sessions of experiment including quiet standing measurement and four different frequencies of perturbations. Each session was conducted in 90s and rest times were provided in between of each session. Figure 2 below shows the experiment set up for this study.

Data Analysis. Kinetic and kinematic data obtained from both motion analysis system and force plate were used to analyze ankle joint stiffness and damping characteristic based on eq. (1), (2) and (3) as mentioned above. Figure 3 below shows a portion of a trial session (90s) in which the ankle joint stiffness are displayed together with a sinusoidal like perturbation applied at different frequency. Data at the first cycle of each perturbation need to be eliminated for next calculation to avoid movement bias due to introduction of perturbation. Then, by using curve fitting method, a best polynomial curve line to represent a stiffness pattern was obtained. Experiment data were compared with simulation data obtained based on eq. (2) which was developed by using Simulink.

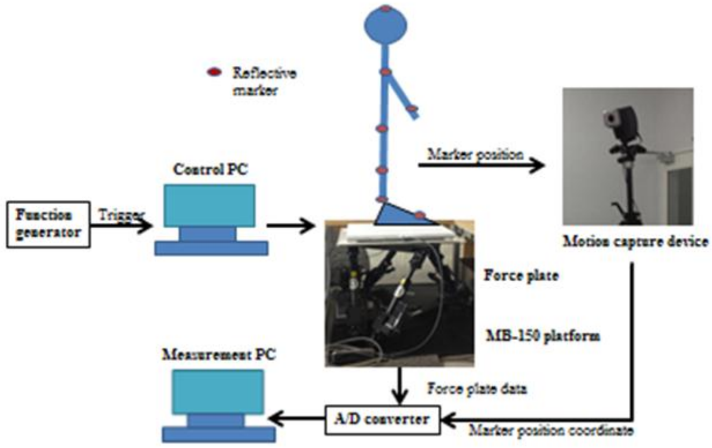

Fig. 2. Experiment set up.

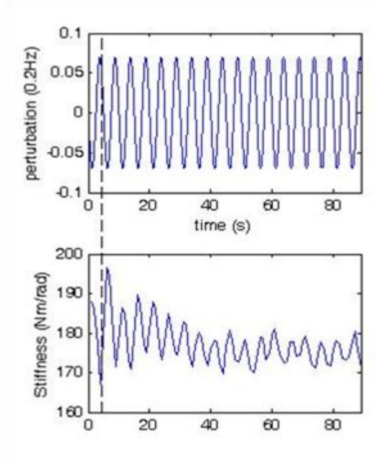

Fig. 3. Elimination of first cycle data (dash line) to avoid movement bias and passive movement.

Result and Discussion

The simulation result of ankle joint stiffness based on mathematical model of inverted pendulum was described as shown in figure 4 below;

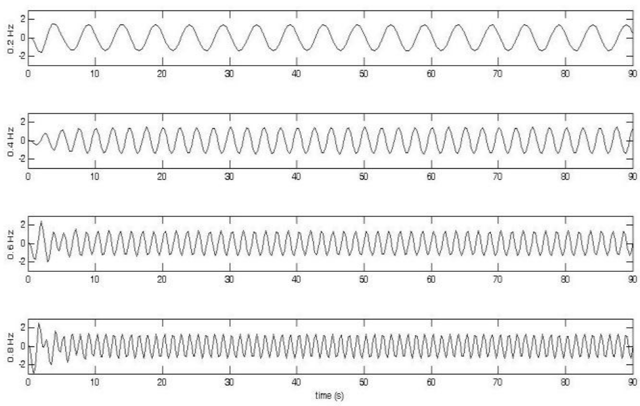

Fig. 4(a). Estimation ankle joint stiffness at four different frequency of perturbation.

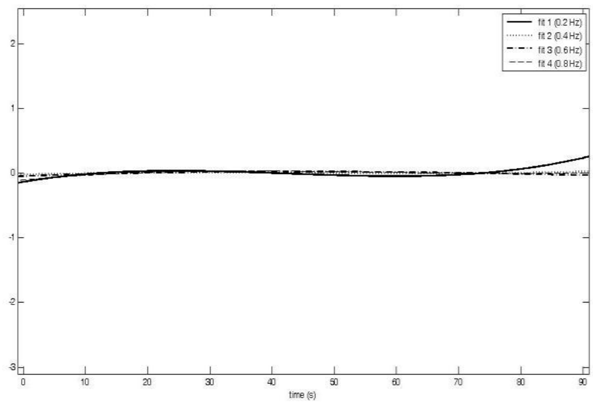

Fig. 4(b). Estimation pattern of ankle joint stiffness along 90 seconds.

Estimation pattern of ankle joint stiffness along 90 seconds of different translation perturbation frequency is also shown in figure 4(b) . Based on kinetic and kinematic data obtained from the experiment, ankle joint stiffness and damping pattern over time were observed as shown in figure below;

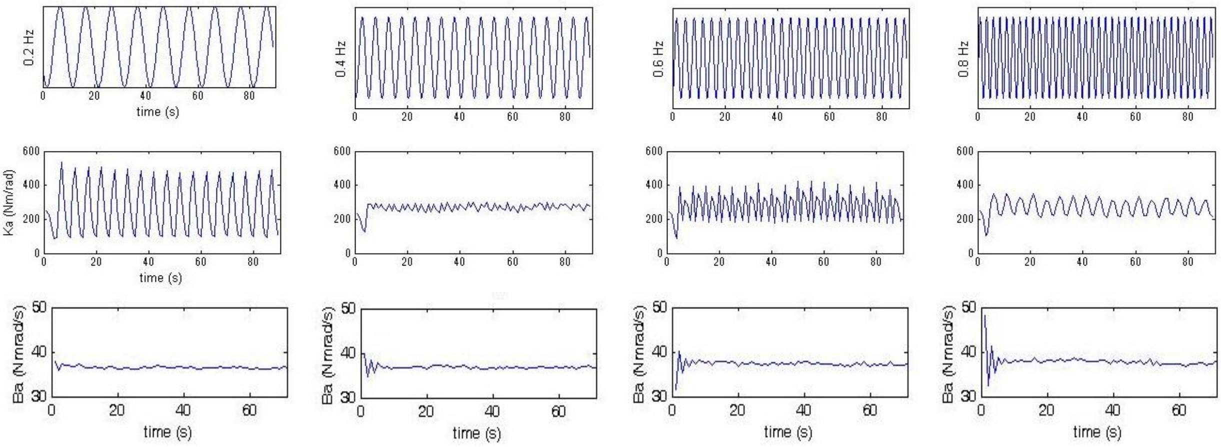

Fig. 5. Ankle joint stiffness (Ka) and damping (Ba) pattern of six subjects . Top column represent the 4 different frequencies of perturbation.

Based on figures 4 and 5 above, there are differences between both simulation and experiment results. In Figure 5, effective stiffness pattern over time is changed. At the initial state of perturbation (first 10s) , ankle joint stiffness were slightly high due to stiffening strategy applied to reduce body sway but then reduced and fluctuated over time. These reduction might be caused by the increase in ankle sway which is due to adaptation of postural control system in order to reduce energy consumption, or loss of sensory awareness due to rapid change of sensory information and tiredness.

Furthermore, simulation results show that frequency component of ankle joint stiffness followed the perturbation frequency. However, the experiment results indicates a different condition. Here, the frequency component of ankle joint stiffness is somehow higher than perturbation applied at three frequencies (0.2 Hz, 0.4 Hz and 0.8 Hz). But at 0.8 Hz, frequency component of effective stiffness is smaller than the perturbation frequency applied. Figure 6 below explains the frequency component of stiffness based on frequency spectrum analysis. At frequency 0.2 Hz , 0.4 Hz and 0.6 Hz ankle joint oscillated more faster than the frequency of perturbation applied. However at 0.8 Hz of perturbation, the oscillation frequency reduced and no longer followed the perturbation oscillation. This might be due to change of posture strategy from ankle strategy to hip strategy. This proves similar to results as reported by Buchanan et al, that healthy human tend to change posture strategy from ankle to hip strategy under perturbation at frequency above 0.5 Hz [11].This observation might be an improvement to Buchanan's finding where changes of posture strategy happen in between 0.6 Hz and 0.8 Hz.

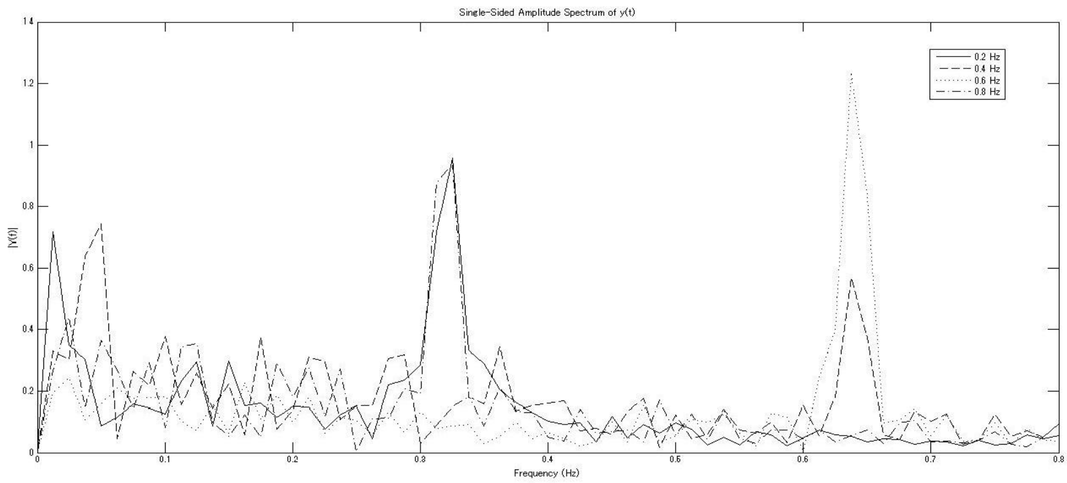

Fig. 6. Frequency spectrum analysis of ankle stiffness pattern.

Under different frequencies of perturbation, mean values of ankle joint stiffness were reduced as the frequency increased. The mean value of ankle joint stiffness along the trial time for frequencies of 0.2 Hz, 0.4 Hz, 0.6 Hz and 0.8 Hz were shown in Figure 7 below. Overall, the amplitude of ankle stiffness increased with increasing perturbation frequency. In previous research by Ishizawa et. al., ankle joint stiffness was reported to decrease with the increase of rotation perturbation [10]. Types of external disturbance somehow maybe affect the stiffness produced by the ankle joint.

As mentioned previously, damping parameter is described as the ability of joint to absorb shock. According to Figure 5, high value of mean damping is triggered at the early stage of perturbation and almost remains unchanged over time. The highest triggered value is shown at frequency of 0.8 Hz with 48.00 Nmrad/s. The mean value of damping is increased as frequency of perturbation increases as shown in Figure 8. This result shows that higher frequency of perturbation gave higher amount of shock towards the joint. This consideration is important especially when dealing with elderly subjects.

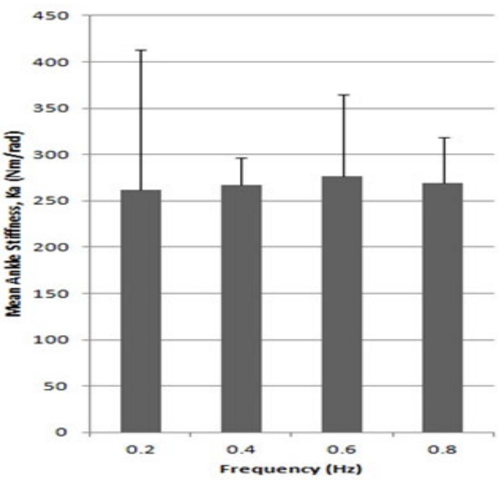

Fig. 7. Mean of stiffness at different frequency of perturbation.

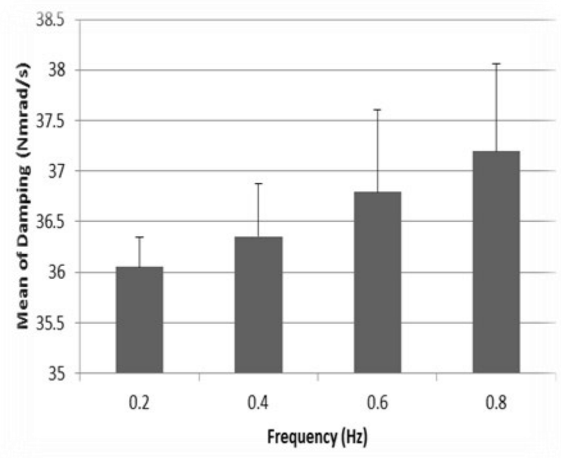

Fig. 8. Mean of damping at different frequency of perturbation.

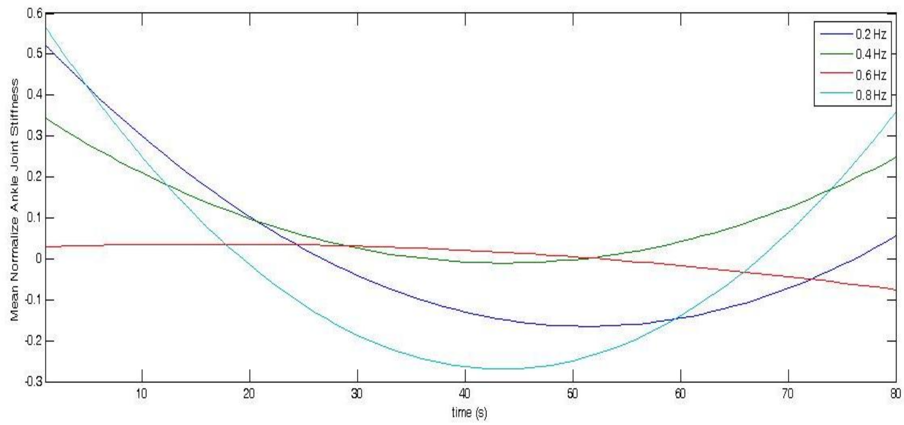

Fig. 9. Mean Estimation of Stiffness Pattern

Effective stiffness value of each subject were then calculated to determined the best curve over time by using curve fitting approach. Based on this estimation, almost all subjects produced a u-shaped curve. Figure 8 explains the estimation pattern of effective stiffness pattern. These estimation curves showed that stiffness pattern at 0.2 Hz, 0.4 Hz and 0.8 Hz had almost similar patterns except at 0.6 Hz. Meanwhile, at 0.4 and 0.6 Hz, the curve slope was very small which may indicate small reduction rate of stiffness. Higher slope was observed at frequency of 0.8 Hz that explained stiffness reduces slightly faster here than other frequencies. Reduction of ankle joint stiffness is expected to be covered by stiffness at other joints (hip and knee).Besides, the u-shaped curve has shown that ankle joint applied different stiffness during initiation, middle and termination stage of perturbation. At the initiation and termination stage, stiffness becomes higher due to unpredictable introduction and stop of perturbation. Meanwhile, at the middle stage, subjects are able to maintain balance and adapt with the repeatable perturbation and therefore applied less stiffness. Contrastingly, the simulation model is unable to explain these adaptation strategies which play an important role in explaining posture control system under repeated translation perturbation.

Conclusion

The estimation pattern of stiffness has indicated that human applied different ankle joint stiffness under different frequencies of perturbation and also along the time of repeatable perturbation. The increase of perturbation's frequency not only changed the value of stiffness but also the frequency component. The results also suggested that healthy young subjects shift from ankle strategy to hip strategy at frequency of 0.6 Hz. Effective stiffness and damping value are expected to be higher for elderly subject due to degeneration of muscle and joint function. In addition, additional mathematical equation is required to represent posture control adaptation strategy, so that the simulation model will be more reliable. Further investigations in determining estimation of stiffness pattern of hip joint and also muscle properties, in particular under different frequencies of translation perturbation are warranted in order to develop a reliable measurement system to measure balance ability and motor learning.

Acknowledgement

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ROLE OF HAND IN POSTURE BALANCE CONTROL

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Abstract: The aims of this study is to investigate the influence of fingertip touch to center of pressure (COP) displacement so that contribution of hand role in balance process can be determine. External translational platform perturbation at anterior-posterior direction was applied to the subject at five different frequency. Subjects were asked to maintain their position when fingertip touch and vision input were manipulated. Touch force, wrist motion and the COP displacement were measured and compared. The results have shown that a small amount of touch allowed subject to increase based of support (COP) and significantly improved stability. Horizontal fingertip touch force shows a significant change with vision input. However, wrist movement and fingertip did not show any significant correlation. Thus, touch significantly improve stability, however, uncorrelated motion between wrist and fingertip still leave an empty space for further investigation and discussion.

1. INTRODUCTION

At late 1990s, there were several research done in order to determine the effect of touch towards improvement of balance ability. Individual with specific disease such as vestibular loss patient (Lackner, Rabin, & DiZio, 2001), diabetes patient (Dickstein, Shupert, & Horak, 2001), peripheral neuropathy patient (Dickstein & Laufer, 2004) and also elderly people (Tremblay, Mireault, Dessureault, Manning, & Sveistrup, 2004) took more benefits of touch information when compared with healthy individuals and they was reported applied more touch force than healthy person. They were reported able to reduce their postural sway when asked to maintain stance position during particular feet position (Baldan, Alouche, Araujo, & Freitas, 2014). Previous study also have reported that, with amount of less than 1N able to reduce postural sway by providing neurological support. Those finding have raise a question on the effect of touch especially when individual facing a massive external disturbance. Does touch will still able to maintain posture balance and provide sufficient support to individual?

These study aims to investigate the influence of fingertip touch to the COP displacement so that contribution of hand role in balance process can be determine. Thus, a clear understanding on the effect of haptic information from touch especially improvement of balance condition during perturbed stance can be achieved.

2. EXPERIMENT

In this study, 11 healthy young subjects (aged 24.24± 2.19 years old) participated. Each subject was fully briefed regarding any possible risk and provided informed written consent prior to participation. Subjects were exposed to external platform perturbation (anterior-posterior direction) at five different frequencies (0 (quiet standing (QS)), 0.2, 0.4,0, 6, and 0.8 Hz) by a movable platform (MB-150, COSMATE, JAPAN) with a translational displacement of 70mm. Furthermore, subject's vision (eyes-closed (**EC**) and eyes-opened (**EO**)) was manipulated. For additional somatosensory input, subjects were asked to maintain their standing position with their right fingertip touching (**T**) on a 3-axis force sensor with a built in amplifier (MFS20-010, LINIAX, Japan) at a waist level.

Kinematic data was collected using motion analysis with 7 high precision infrared cameras (HWK-200RT camera, Motion Analysis, USA) at a sampling frequency of 200 Hz using 17 reflective markers. Meanwhile, a force plate (9286A, KISTLER, JAPAN) and a force sensor recorded at a sampling frequency of 1 kHz. Each trial was recorded for 40s with knee joints locked using splint to prevent bias movement from knees. The experimental set-up is shown in Fig. 1. The coordinates of each marker were then analysed to determine the position of each joint. Force data from the force sensor was low pass filtered at a 60Hz cut-off. Paired t-test and Pearson’s correlation coefficient (r) were used to determine significant levels of force at different directions. Furthermore, multi-way ANOVA was used to determine a significance level of $p<0.05$. All measurements were made using MATLAB software.

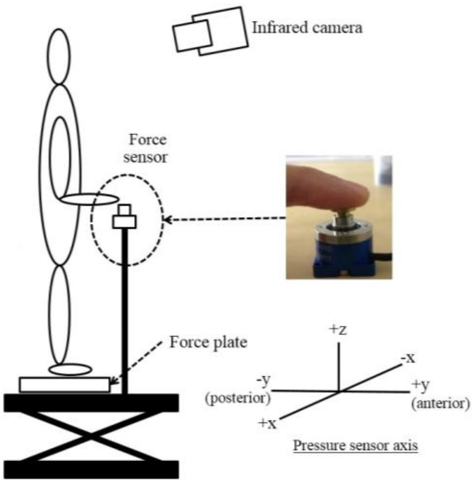

Fig. 1 Experiment Set up

3. RESULTS

The results in Fig. 2 (Top) illustrates the displacement of COP at both anterior and posterior direction under the perturbed stance. The average COP was observed increased with frequency with significant different found at all sensory conditions ($p<0.05$). By comparing between sensory condition, with existence of touch, COP displacement was increased and significant different was found at all perturbation frequency ($p<0.05$).

Based on results in Fig. 3, fingertip force (z-direction) significantly increases with frequency [$F(4, 79) = 5.9$, $p<0.05$]. The force increases up 2 N from 0.2 Hz to 0.8 Hz. When comparing the different vision conditions, force is higher with closed eyes compared to open eyes at all frequencies but insignificant difference ($p>0.05$). Furthermore, horizontal force at y and x directions was found to be insignificantly different between frequencies with $F(4, 79) =0.18$, $p=0.94$ and $F(4, 79) =2.11$, $p=0.097$, respectively. And also, insignificantly with vision condition except at 0.8 Hz. For force at the y-direction, difference force direction pattern was observed between EO and EC. This preference is unique and may indicated the change of hand motion due to loss of sensory input.

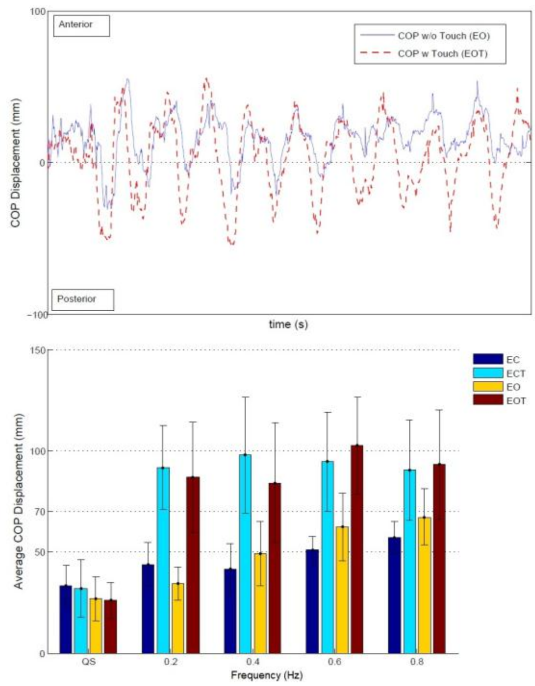

Fig. 2 (Top) Illustration of COP displacement during perturbed stance. (Bottom) The average COP displacement.

These have encourage for further investigation on wrist flexion and extension movement (z-direction) which may contributed to this even to be happen according to previous research (Su, Chou, Yang, Lin, & An, 2005) .

Fig. 4 below shows that, wrist at superior and inferior direction increased with the increase of frequency with significant different found only at inferior direction ($p<0.05$). Based on the statistical analysis of superior direction, there is no significant different found ($F(1, 47) =0.05$; $p=0.8271$). A similar results also recorded on inferior direction ($F(1, 47) =0.27$; $p =0.6078$). These insignificant results have indicated that the wrist movement might not correlated with vestibular dysfunction or more specifically vision input.

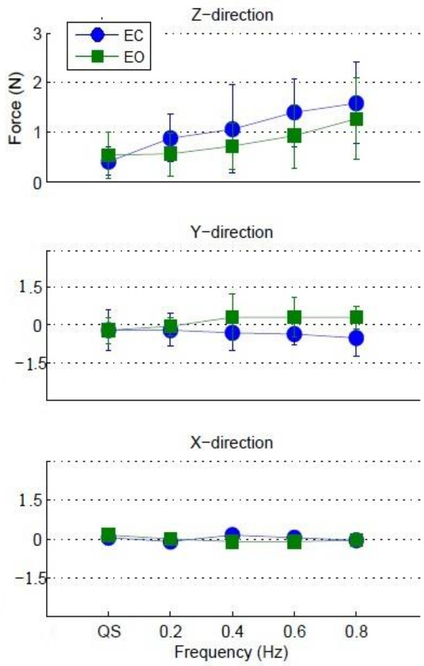

Fig. 3 Average vertical force (z-direction) and horizontal force (y-direction and x-direction) produced by fingertip during EO and EC (M±SE).

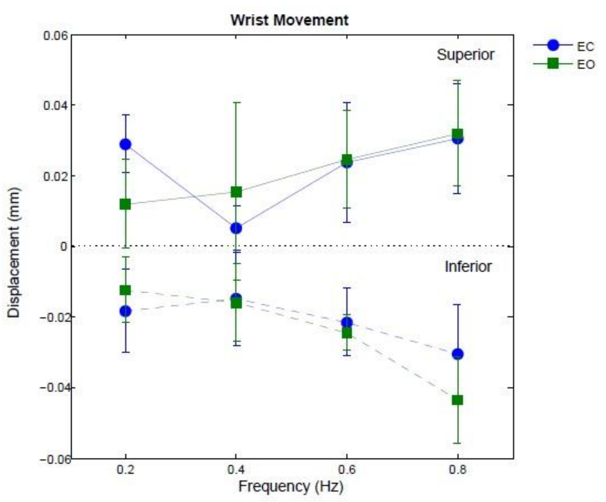

Fig. 4 Wrist movement displacement at both superior-inferior direction during EC and EO

4. DISCUSSION

It is widely recognised that the existence of additional haptic information from touch allows the hand to move and provide spatial contact information; while also allowing a considerable range of body sway; which in this study, is exceed almost 40% from the platform displacement range. This followed the concept of human balance described by Pollock (2000), where a better sensory inputs, individual able to provide a bigger based of support to always make sure the COM lays within the COP range. Previous research by (Gatev, Thomas, Kepple, & Hallett, 1999) reported that, absence of vison have increase the body sway included COP. Furthermore, additional sensory information like fingertip touch, was observed reduced postural sway when comparison make especially between balance and unbalance individual. When compared with absence of vision, many research reported that COP displacement during EC higher than EO. This only can be seen at QS and 0.2 Hz of recent study.

Vertical force generated at the fingertips was shown to be higher during no vision (EC). This indicates that, without vision subjects depended on fingertip receptors to sense and provide information about the body’s orientation. In perturbed standing, a force of more than 50g (~0.5N) was required to provide significant postural stabilization. This supported by previous research where vertical force of 40 ± 7 g produced during when subject standing heel-to-toe (Lackner, et al., 2001) . Furthermore, one interesting finding is about horizontal force at anterior – posterior’s direction. Different directions of force were recorded between EC and EO conditions; where a significant difference was observed at 0.8 Hz. This indicates the possible existence of different fingertip position preferences; with respect to loss of vision input. According to F.-C. Su et al, (2005), motion of wrist reported influenced the fingertip motion. They have determine negative slope from regression analysis that demonstrated the so called ‘‘reciprocal’’ nature of joint motion (Su, Chou, Yang, Lin, & An, 2005). For example, during wrist extension, passive finger joint flexion was induced and, alternatively, during wrist flexion full finger joint extension was induced. However, based on the investigation of wrist position, there were no significant different found between different vision inputs. As mentioned in the results section, effect of different vision input influenced fingertip movement but not the wrist. Force dependent receptor at fingertip connected directly to ulnar nerve which plays an important role in response to afferent-efferent information transfer in dorsal spinal tract that control the upper part of the body (Proske & Gandevia, 2012).

The wrist flexion and extension degree much influenced by the intensity of perturbation frequency. However, due to limited amount of information due to wrist motion, a detail analysis on the wrist flexion-extension movement cannot be done. The evaluation of wrist motion were depended only on the marker coordinated at ulna bone. For further study on this matter, more market position is suggested in order to gather more information about the wrist movement especially at dorsal surface of hand.

CONCLUSION

In conclusion, this study able to demonstrated the COP response with existence of additional somatosensory and different vision input to posture stability during perturbed stance. A small amount of touch allow sufficient neurological support for posture control, which then increase the COP magnitude. A larger base of support increases stability. However, insignificant between hand motion and stability still leave an empty space for further investigation and discussion.

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Fingertip touch adjust postural orientation during perturbed stance

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Abstract - **Additional sensory information; especially from touch, was suggested to improve stability by reducing body sway. However, it is less known about the effect of touch on the body's joint movement during perturbed standing; which is commonly experienced by public transport users. In this study, subjects were asked to try to maintain their standing position with their fingertips on a rigid surface, while surface perturbation was applied at four different perturbation frequencies (0.2, 0.4, 0.6, and 0.6 Hz) and different vision input. Motion of joint (ankle, hip and head) and relative centre of mass (COM) were recorded and analysed. The results show that fingertip pressure was higher without vision. Furthermore, different fingertip moment directions were recorded between with vision and with no vision. This possibly indicates a preferred fingertip position that can provide better sensory information to replace sensory loss; especially vision. The range of motion of joints also decreased with fingertip touch - except for head motion. Furthermore, even though there were no significant differences observed between with touch and without touch, the relative COM displacement was less with touch. Thus, even with a very light touch, subjects were able to reduce body sway even in a perturbed stance. Further investigation is needed to determine the changes in centre of pressure (COP) and significant position of fingertip, which can enhance stability.**

Keywords - **Joint, Touch, Vision, COM, Perturbed stance.**

I. INTRODUCTION

Touch or hand grasping is a common strategy used by passengers of public transportation, like trains and buses, and even while standing or walking, in order to maintain a balanced position. However, it is less known how touch effects posture change. In general, touch and hand grasping was observed to affect the COP's sway during both quiet standing and perturbed standing [1-4]. It also suggests that a light touch force, as small as 1N, is able to preserve stability during quiet standing [1, 3]. Research has been conducted to determine the effect of touch or grasping position on stability. According to Babic et al. (2014), the position of handrail and the location of hand contact produced no effect on COP displacement during perturbed stance [5]. However, this was different to the results obtained by Sarrat et al. (2014), where they found that handrail position affected the normalized value of peak COP [6]. They also suggested that hand grasping did not influence COP displacement during sideways perturbation. Curiosity of how touch mechanisms can lead to improved stability is still increasing. Recent research has been conducted on the mechanisms of handgrips on postural stability. According to Vanderhill et al. (2014), imagery and actual hand gripping gained similar amounts of postural stabilization [7]. They also suggested that the mechanism of handgrip could be attributed to extra sensorimotor activity; triggered by hand contact or cognitive effort. This was supported by Vuillerme et al. (2005), who suggest that touch makes the regulation of postural sway more cognitively dependent, where it increased time reaction in order to initiate the balance strategy [8]. Based on the previous research mentioned above, we already know that touch improves stabilization by reducing the COP displacement. However, how the force generated from a small finger contact with a surface triggers the motion of a body's joint is still unknown. In this study, we will discuss the amount of force and the moment produced from a light finger touch during perturbed stance, and how it influences a range of motion of ankle, hip and head, which leads to relative COM displacement. Furthermore, the effect of vision on touch response will also be investigated. Thus, the effect of touch to benefit posture stabilization will be determined.

II. METHODOLOGY

In this study, 11 healthy young subjects (aged 24.24± 2.19 years old) participated. Each subject was fully briefed regarding any possible risk and provided informed written consent prior to participation. Subjects were exposed to external perturbation (anterior-posterior direction) at four different frequencies (0.2, 0.4,0, 6, and 0.8 Hz) by a movable platform (MB-150, COSMATE, JAPAN) with a displacement of 70mm. Furthermore, subject's vision (eyes-closed (**EC**) and eyes-opened (**EO**)) was manipulated. For additional somatosensory input, subjects were asked to maintain their standing position with their right fingertip touching (**T**) on a 3-axis pressure sensor with a built in amplifier (MFS20-010, LINIAX, Japan). In this study, a pressure sensor was mounted on a pole on a moveable platform so that it swayed simultaneously with the platform to create a situation similar to that of inside a moving vehicle. Kinematic data was collected using motion analysis with 7

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IFMBE Proceedings 51, DOI: 10.1007/978-3-319-19387-8_82

high precision infrared cameras (HWK-200RT camera, Motion Analysis, USA) at a sampling frequency of 200 Hz using 17 reflective markers. Meanwhile, a force plate (9286A, KISTLER, JAPAN) and a pressure sensor recorded at a sampling frequency of 1 kHz. Each trial was recorded for 40s with knee joints locked using splint to prevent bias movement from knees. The experimental set-up is shown in Figure 1. The coordinates of each marker were then analysed to determine the position of each joint. Pressure data from the pressure sensor was low pass filtered at a 60Hz cut-off. Paired t-test and Pearson's correlation coefficient (r) were used to determine significant levels of pressure moments at different directions. Furthermore, one way ANOVA was used to determine a significance level of $p<0.05$. All measurements were made using MATLAB software.

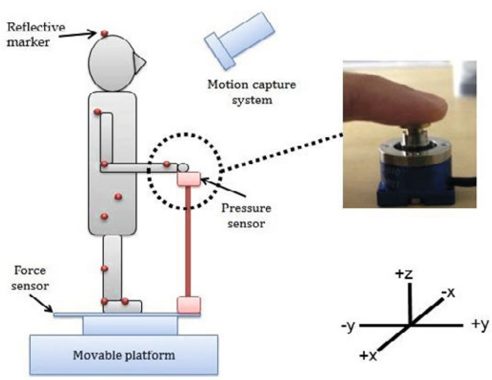

Fig. 1 Experiment set-up

III. RESULTS

A. Force and moment produced by fingertip touch

Based on the results, vertical force (z-direction) was significantly increased with frequency ($F(4, 79) = 5.9$, $p=0.0007$). Perturbation frequency, from 0.2 Hz up to 0.8 Hz, increased force from 0.5N to almost 2N. By comparing the different vision conditions, force during eyes-closed (EC) was higher than eyes open (EO) at all frequencies. However, no significant difference was found (as shown in Table 1). Furthermore, the amount of moment experienced at y and x directions was found to be insignificantly different between frequencies with $F(4, 79) = 0.18$, $p=0.94$ and $F(4, 79) = 2.11$, $p=0.097$, respectively. Almost all moments in the x and y directions also found insignificant differences during difference vision conditions (e.g., EC vs. EO) except at 0.8 Hz (as shown in Table 1). More interestingly, for moments in the y-direction, difference in direction was observed between EO and EC. During eyes-opened, fingertip generated more moment in the anterior direction; and in the posterior direction during eyes-closed. This can be seen by the weak correlation coefficient ($r<0.5$) between EC and EO. These preferences are unique and give information about the possibility of different posture leanings during touch for both with and without vision.

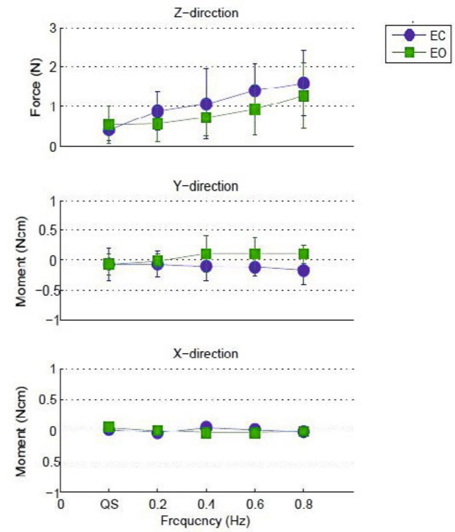

Fig. 2 Mean vertical force (z-direction) and moment (y-direction and x-direction) produced by fingertip during EO and EC (M±SE).

Table 1 Comparison of force and moment recorded between eyes-opened (EO) and eyes-closed (EC) using correlation coeff. (*r*) and paired t-test (*p*).

Force			**Frequency (Hz)**				
	df		**QS**	**0.2**	**0.4**	**0.6**	**0.8**
Z-direction	22	*p*	0.52	0.12	0.30	0.25	0.25
		r	0.53	0.89	0.71	0.96	0.19
Y-direction		*p*	0.91	0.58	0.14	0.25	0.01*
		r	0.94	0.25^{x}	0.05^{x}	-0.19^{x}	0.24^{x}
X-direction		*p*	0.36	0.38	0.35	0.38	0.86
		r	0.33	0.07^{x}	0.51	0.19^{x}	0.61

The (*) indicates significant difference between EO and EC with $p<0.05$ and (^{x}) indicates the correlation between EO and EC with $r<0.3$.

B. Range of motion

As mentioned in previous research, touch affects body sway. It was reported to reduce postural sway during quiet standing; and thus, enhance stability. In a recent study, the results followed previous studies; whereby the range of motion at ankle and hip joints was reduced with touch. However, this did not occur for head motion; and the head swayed more during touch conditions. Overall, ankle and hip did not shown any significant changes due to different sensory inputs *(control vs. sensory)* (F_{ankle} (3,119) =0.82, p=0.82 and F_{hip} (3,119) =0.96, p=0.42), but head movement was significantly different with sensory conditions, with F_{head} (3,119) =9.38, p<0.001. By comparing the effects of sensory input at each frequency, hip and head joints showed a significant difference with the existence of touch sensory (as shown in Figure 3). With the increase of frequency, the ROM of ankle and hip was also increased and a significant difference was found between frequency with F_{ankle} (3,119) = 46.28, p<0.001, and F_{hip} (3,119) =17.59, p<0.001, respectively. However, no significant difference was found at head with F_{head} (3,119) =1.76, p=0.15.

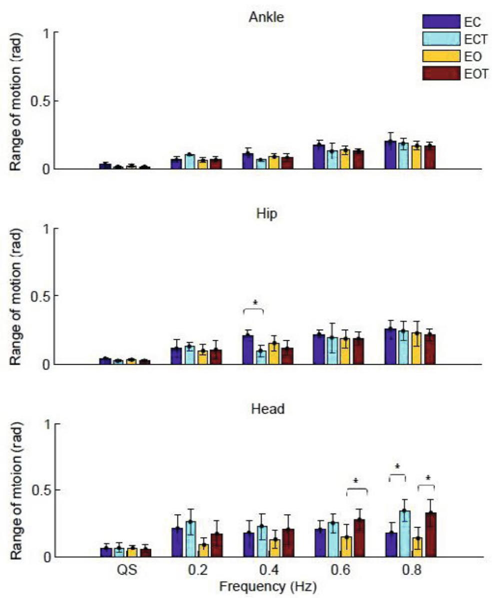

Fig. 3 Range of motion based on angular displacement (ROM) of ankle, hip and head during three sensory conditions, such as eyes-opened (EO) *(control),* eyes-opened with touch (EOT), and eyes-closed with touch (ECT). The (*) indicates a significant difference at different sensory conditions with p<0.05.

C. Relative COM displacement range.

As expected, for relative COM displacement based on the results on joint movements mentioned previously, COM displacement during touch was lower than no touch. By comparing sensory conditions (i.e., *control vs. sensory*), a significant difference was found only in the posterior direction with (F (3, 95) =9.31, p<0.01). However, no significant difference was found in the anterior direction (F (3,127) =1.83, p=0.14)). Furthermore, the relative COM displacement in the anterior direction was insignificantly different with a frequency of (F (3,103) =2.1, p=0.11). Meanwhile, a significant difference was found in the posterior direction (F (3,103) =3.43, p=0.019).

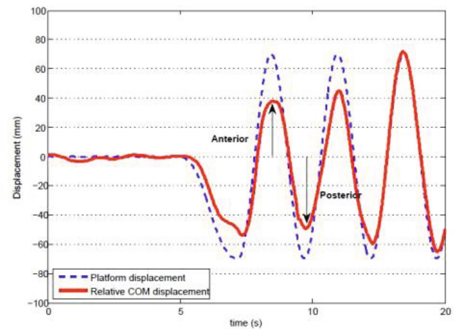

Fig. 4 Determination of relative centre of mass (COM) displacement (in mm) in both anterior and posterior directions.

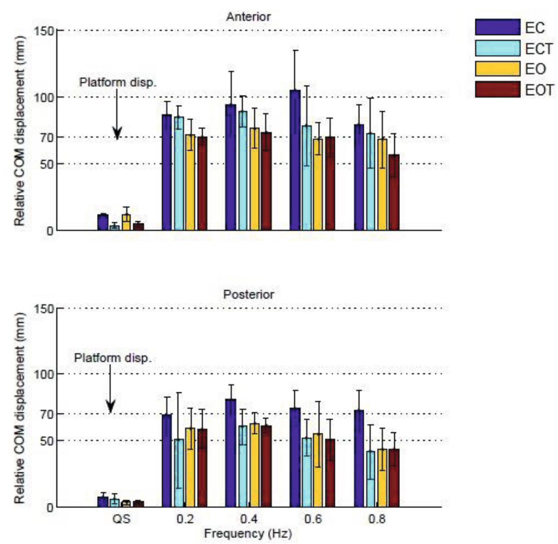

Fig. 5 Mean of relative centre of mass (COM) during 4 sensory conditions i.e., eyes-opened (EO) *(control)*, eyes-closed (EC), eyes-opened with touch (EOT), and eyes-closed with touch (ECT) in posterior and anterior directions.

IV. DISCUSSION

Light touch is expected to improve stability by avoiding body to sway. Research by Lackner et al. (2001), suggests that finger force (of around 2g) on a rigid surface is adequate to allow some postural stabilization to be elicited [9]. Furthermore, as mentioned previously, finger touch has been seen to reduce sway during quiet standing. Recent results from previous studies were followed. However, different responses were produced by head movement, whereby it was increased with touch. This raises a question about the existence of other strategies imposed by finger touch to improve stability during perturbed standing. However, this statement is still speculative. In a recent study, the average relative COM displacement was observed to decrease with additional somatosensory information from light finger touch. Even though it was insignificant, it still showed a reduction. These results followed a study by Hausbeck et al., (2009) [10]. It is widely recognised that the existence of additional haptic information from touch allows the hand to move and provide spatial contact information; while also allowing a considerable range of body sway; which in this study, is within the platform displacement range (~70mm for anterior direction and ~25% less for posterior direction). However, vertical pressure generated at the fingertips was shown to be higher during no vision (EC). This indicates that, without vision subjects depended on fingertip receptors to sense and provide information about the body's orientation. In perturbed standing, a pressure of more than 50g was required to provide significant postural stabilization. Furthermore, one interesting finding is about average moment at anterior – posterior's direction. Different directions of moment were recorded between EC and EO conditions; where a significant difference was observed at 0.8 Hz. This indicates the possible existence of different fingertip position preferences; with respect to loss of vision input. The position of fingertip produced during EC was almost similar to a blind person reading Braille (as shown in Figure 6 below); which leads to a moment produced in the posterior direction. This suggests that the position of fingertip may vary due to sensory loss, in order to gather more sensory information. However, further investigation is still needed.

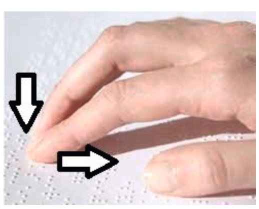

Fig. 6 Position of fingertips of a blind person reading Braille.

V. CONCLUSION

In this study, light fingertip touch was observed to allow the body to sway in a considerable range during perturbed standing; more than without touch. This suggests that additional somatosensory information allows the CNS to predict and tolerate possible sway caused by the moving platform; as long as stability can still be achieved. However, without vision, more pressure is required to trigger the force dependent receptors at the fingertips to sense changes in body position. Furthermore, different moment directions between EC and EO at posterior-anterior direction may give a clue to the preference of fingertip position to provide better sensory information to replace sensory loss; especially vision. Further investigation is needed to determine the changes in COP and the significance of fingertip position to enhance stability.

ACKNOWLEDGMENT

We would like to thank all participants who involved in this study.

CONFLICT OF INTEREST

The authors declare that they have no conflict of interest.

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Balance Process during Repeated Surface Perturbation: Adaptation Response of Joint Stiffness and Muscle Activation

Aizreena Azaman and Shin-ichiroh Yamamoto

***Abstract*—It is believed that humans are keen to learn and initiate more efficient and less energy consumption strategies, especially when they desire repetitive work or motion. However, in human's balancing process, the ability to adapt a repeated surface movement and its response towards imbalance, due to less sensory input, is still unclear. In this study, adaptation behaviours of joint stiffness pattern and muscle activation were observed during limited sensory inputs. Seven young subjects participated in this study. Two different surface perturbations (tilt up-tilt (TT) down and translation (T)) at four different sensory manipulation conditions (include vision and vestibular system) were introduced to the subject. Then, they were asked to maintain their position as long as possible. The results have shown that amplitude of joint stiffness decreased by almost 1.2 percent at the ankle over 10 cycles. However, there is almost no adaptation at the hip. Even though average the adaptation percentage increased as sensory inputs became better (r^2>0.3), no significant difference between sensory conditions was recorded (p>0.05). Meanwhile, different adaptation patterns were observed among five different muscles at both types of perturbation, with adaptation at almost 1 percent on average. The findings have shown that adaptation behaviour is able to describe motor learning functions of the balancing process in humans. It helps to enhance human posture control model and muscle dynamic model especially related to continuous repeated motion or force applied to the system.**

I. Introduction

Initiation and co-ordination of human movement when facing any perturbation much depends on sensory input from vestibular, vision and somatosensory receptor [1]. Decline in function of sensory inputs will lead to declination in balance ability. Human posture control tends to initiate and constraint joint movement so that the Centre Of Mass (COM) and Centre Of Pressure (COP) are in an equilibrium state [2].This would lead to an existence of stiffness at the joint to create a desire for movement. Stiffness is believed to be a product of neuromuscular activity. In previous research, stiffness was observed and significantly correlated with muscle activation [3-5].

Earlier, researches of the human balancing process were focused on reaction of posture modulation through sensory input manipulations, and intensity of perturbation, balance during gait performance, etc. Integration between sensory information and the Central Nervous System (CNS) has defined the human posture control system responses. It is known that the CNS makes use of two types of control responses, such as feedback control for reflective movement and feed forward control for specific command response like kicking a ball [5]. Thus, in order to maintain the desired position, through compensation of feed forward control that involves both memory and motor learning ability, it is observed that the individual needs to be able to reduce the magnitude of action required (i.e.; muscle activation and movement) [6]. However, whether these reductions will still occur, even though continuous surface movement is applied to maintain a balanced position, remains unclear.

In previous study by Schmid et al., (2011) is shown that Tibia Anterior (TA) and soleus (SOL) muscles indicated a larger decay under 3-minutes of backward and forward surface movement through exponential estimation [7]. On the other hand, by comparing age status, it is observed that elderly people cannot fully compensate adaptation control, where posture adaptation and stimulus adaptation percentage were reported higher than middle-aged people during vibration stimulus of the legs [6]. Those have indicated that adaptation analysis of the balancing process can distinguish disability due to different neuromuscular and physiological ability. This has raised concern for the development of the new approach in assessing balance ability and enhancement of human posture model. This is because neuromuscular-based response was suggested to be combined with mechanics parameter so that a more reliable assessment and human model can be produced [8].

Thus, in this study, we aim to investigate the reaction of joint stiffness and muscle activation in response to repeated surface perturbation and how adaptation control of the CNS effects under limited sensory information.

II. Methodology

In this study, seven healthy young male subjects (aged 24.24± 2.19 years old) with acceptable vision ability were selected to participate. Each subject provided informed written consent prior to participation and was fully explained as to any possible risks. Information regarding the subject's history of falls and physical condition were recorded as reference.

A. Experiment Set-up

Subjects were exposed to two types of external perturbation (translation (***T***) and tilt up-tilt down (***TT***)) at 0.2 Hz produced by movable platform (MB-150, COSMATE, JAPAN) with displacement of 70mm and 6 degree respectively. These two different perturbations were applied to observe how the ankle and hip joint react towards different surface orientations. Furthermore, subject's vision (eyes-closed(***C***) and eyes-opened (***O***)) and vestibular sensory

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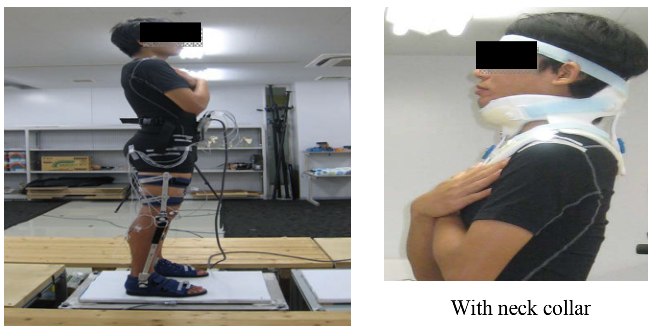
system (*N*) were manipulated. Vestibular system was manipulated by limiting the movement of neck and head using a neck collar (ADFIT collar, ADVAN FIT). This step will interfere with the function of proprioceptors in the neck muscle, thus will lead to vestibular malfunction [9]. Subjects were asked to stand quietly and comfortably and tried to maintain their position along the trial periods.

Kinematics data was collected using motion analysis (HWK-200RT camera, Motion Analysis, USA) at sampling frequency of 200 Hz using 17 reflective markers. Meanwhile, force plate (9286A, KISTLER, JAPAN) and EMG data of tibia anterior (***TA***), soleus (***SOL***), medial gastrocnemius (***MGAS***), rectus femoris (***RF***) and bicep femoris (***BF***) were recorded at sampling frequency of 1kHz. Each trial was recorded for 60s with knee joint was locked using a splint to prevent bias movement from the knee. Experiment set up as shown in Fig. 1. Sensory manipulation combinations are described in Table 1.



With neck collar

Fig. 1. Subject's preparation for experiment.

Table 1: Sensory manipulation condition combination

Combination	Sensory Input Condition
NC	Eye closed with neck collar
C	Eye closed only
NO	Eye opened with neck collar
O	Eye closed only (*normal*)

B. Data Analysis

Joint stiffness at ankle and hip joint were measured based on free body diagram (Fig. 2) and Eq. (1) below;

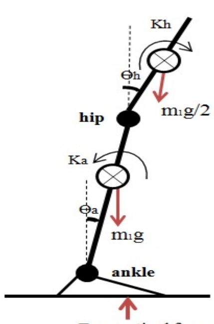


Fig. 2. Inverted pendulum free body diagram

Stiffness at joint can be defined as;

$$K=\tau/\Theta \qquad (1)$$

Where τ is torque of ankle and Θ is angle of joint sway. The F_v which is a vertical ground force gather from force plate data is assume to be as follow;

$$F_v \approx m_1 g \qquad (2)$$

Both EMG and force plate data were filtered with second order Butterworth filter. Joint stiffness pattern and EMG data were then cut according to cycle of perturbation (1 cycle of platform movement = 5 seconds). Adaptation ability is measured by comparing the area under the graph (AUG) of cycle side-by-side and presented in term of percentage. Result of adaptation percentage is made using statistical analysis (mean and standard error) and differences between sensory conditions were compared using Two Way ANOVA with a significant level of $p<0.05$. Association between parameter and condition were tested using correlation coefficient. All measurement was done using MATLAB software.

Percentage of Adaptation

$$= \frac{AUG_n - AUG_{n+i}}{AUG_n} \times 100\% \qquad (3)$$

where n= 1,2,3…..10
i= 1

III. Result

A. Joint stiffness adaptation

In our early analysis, we have found that over 10 cycles of repeated perturbation, joint stiffness at the ankle and hip have reduced at certain amount. This finding is similar to our previous study in analysing joint stiffness pattern at translation perturbation which described using curve fitting methods [10]. However, it required high orders polynomial equation for less residual.

In order to determine the actual amount of amplitude reduction of stiffness response, the analysis area under the curve was done by comparing each cycle side by side.

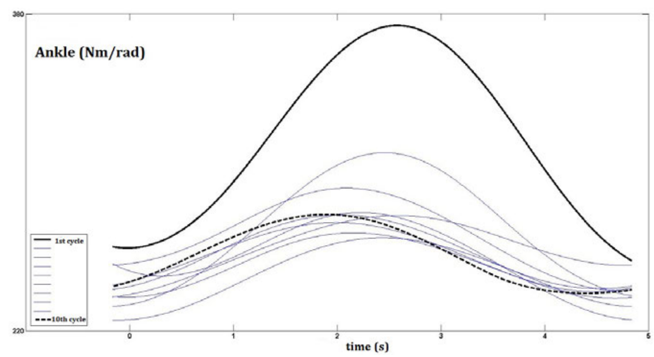


Fig 3. Ankle joint stiffness response during TT and comparison between 1st (solid line) and 10th (dash line) cycle.

Overall, during normal condition (O), subjects have reduced stiffness for about 1.2 percent at the ankle and almost 0.06 percent at the hip. The percentage for the hip is very small, with almost no reduction in stiffness. With an increase of sensory input (NC to O), adaptation percentage at both joints were also improved ($r^2>0.3$). However, there was no significant difference recorded between different sensory limitation conditions. By comparing adaptation between types of perturbation, no correlation was recorded ($r^2<0.1$).

Based on Table 2, adaptation of the ankle can describe disability due to sensory input limitation and changes of surface perturbation. During TT, negative adaptation (no consistent reduction in stiffness's amplitude) was recorded when head movement was limited (NC and NO). Meanwhile, during T, negative adaptation was observed when no visual input was provided (NC and C).

Table 2: Mean and standard error of mean (±SE) of joint stiffness's adaptation percentage (%) under different sensory limitation.

Perturb.		Sensory Limitation NC	C	NO	O	Coeff. p	r^2
TT	ankle	-2.90 (1.20)	0.51 (0.61)	-1.44 (0.80)	1.24 (0.55)	0.53	0.25
	hip	0.02 (0.01)	0.01 (0.01)	0.03 (0.01)	0.03 (0.01)	0.96	0.91
T	ankle	-0.06 (0.19)	-0.53 (0.25)	1.89 (0.49)	1.12 (0.38)	0.25	0.81
	hip	0.03 (0.02)	0.04 (0.01)	0.01 (0.01)	0.06 (0.01)	0.18	0.34
T x TT (O)	ankle					0.33	0.02
	hip					0.41	-0.05

*p is significant value between all sensory condition and r^2 coefficient compare sensory input condition with average stiffness value.

These indicate that human rely on different sensory input combinations when facing different perturbation, which also affected motor learning skill in balancing process.

B. Muscle activation adaptation

Analysis of this adaptation phenomenon was continued with muscle activation. As mentioned before, five main muscle activations (TA, MGAS, SOL, RF, and BR) were analysed. Based on Fig. 4, activation pattern and reduction of activation amplitude between muscles was observed. Muscle activation at each cycle of perturbation of each cycle was again compared side by side and was presented in terms of adaptation percentage based on Eq. (3).

In general, all muscles reduced their activation by 1 percent from one cycle to another. Adaptation percentage of muscle activation was improved (become more positive) as sensory inputs improved (in Fig.5). However, no significant difference was found between sensory condition ($p>0.05$).

Based on results in Fig. 5, there were muscles that were unable to reduce it activation over the trials. During TT, four muscles (SOL, TA RF, and BF) were shown to increase in adaptation percentage when sensory input increased from N to O. However, only MGAS and TA were increased during T perturbation. These results also explained patterns of energy consumption. During TT, less energy was consumed compared to T. This was due to a larger number of muscles that were able to adapt when sensory input improved. Besides, TA adaptation patent are similar during T and TT and suggested that this muscle plays a main role in the balancing process and, it was correlated closely with movement of ankle joint.

Based on correlation analysis, SOL and TA shows a strong correlation with ankle joint stiffness adaptation pattern under different sensory inputs ($r^2>0.5$) during TT, meanwhile TA and MGAS during T. This might indicated the dominant muscle that is active during each perturbation.

Besides, for RF and BF, a strong correlation was found only during TT perturbation but not during T. This might indicate that during posterior-anterior type disturbance (during T); hip joint adaptation was influenced by massive trunk or spine movement that might trigger fluctuation in RF and BF activation thus unable to show adaptation.

IV. DISCUSSION

Based on the findings mentioned above, both joint stiffness response and muscle activation adaptation have shown a strong correlation with the increase of sensory inputs. The results are supported by a previous study by M. Schmidt et al. (2011) on the reduction in muscle activity along repeated external perturbation especially TA [7]. However, for SOL, there is a difference. In previous research, SOL also improved adaptation from C to O, but in our result, it was reversed during T. It might be due to experiment set up, such as the displacement of perturbation and measurement used to determine adaptation. Besides, results in Fig. 5 indicates main muscles which play a dominant role in reducing stiffness at each joint as different muscular chains are applied under different perturbation.

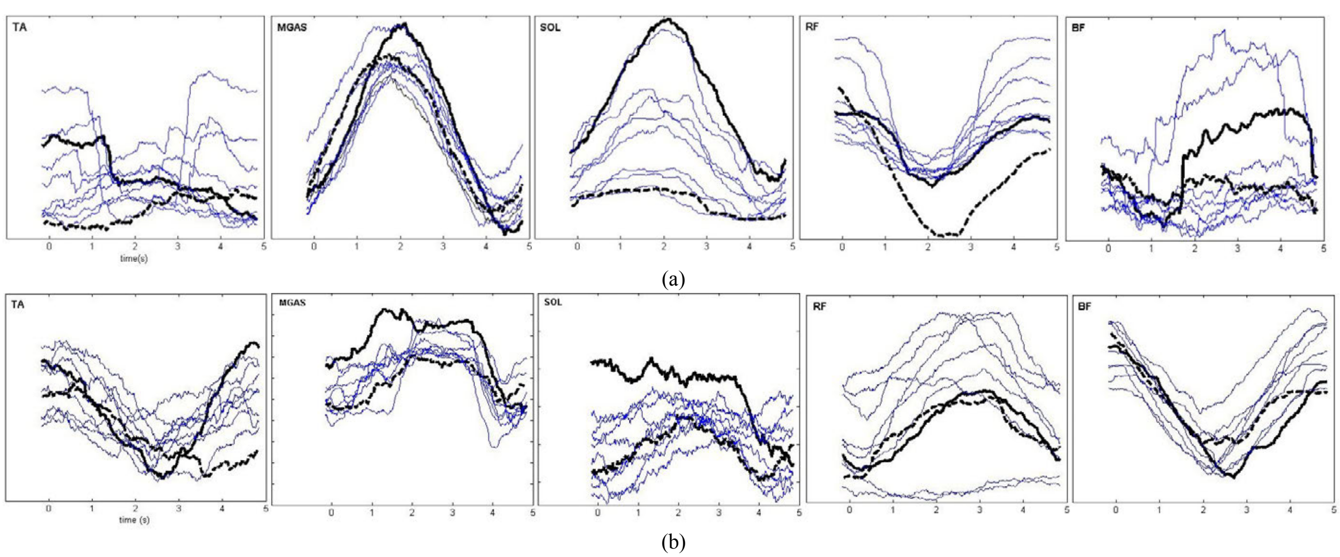


Fig. 4. Moving average of muscle activation response at each cycle during (a) TT and (b) T of one subject. This plot has shown a reduction in muscle activation along 10 cycles. Lines (-) indicates EMG of 1st cycle and (- -) 10th cycles of trial. (Note: This plot not under a same axis scale).

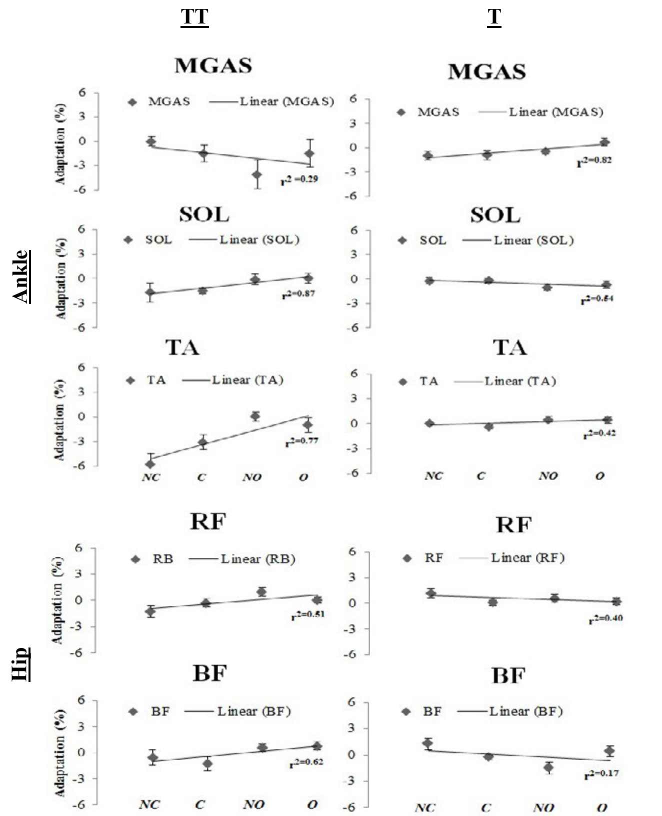


Fig. 5. Mean adaptation percentage of five muscles (TA, SOL, MGAS, BF and RF) during (a) tilt up-tilt down and (b) translation.

Previous research has shown an inverse relationship between antagonist and agonist [6, 7]. Combination of agonist and antagonist muscle's role (MGAS-TA and SOL-TA for TT and T respectively) at each perturbation indicated muscles that were more involved in control of stability during both surface orientations.

Regarding to stiffness pattern, the role of head motion in influencing ankle stiffness adaptation is not clear. It observed to improve adaptation during T but not during TT. We suggest that it might be due to different anterior-posterior velocity between those perturbations. On the other hand, previous research has suggested that reduction of postural sway (which generates high posture stiffness) during quiet standing showed improvement in balance position [11]. However, in this research, stiffness was observed consistently reduced (positive adaptation percentage) as sensory input got better. However, joint stiffness adaptation response was less discussed. Thus, improve of joint stiffness adaptation especially at the ankle in correlation with better balance ability is promising.

In this research, analysis of adaptation can also help to determine existence of active and passive behaviour of stiffness pattern. If before, stiffness pattern was represented as a passive component through linear regression estimation [3,12,13], and now by measuring it along repeated perturbation, that behaviour can be observed. Shifting of behaviour from active to passive can be a good judgment of motor learning ability [7]. On the other hand, adaptation percentage during repeated perturbation can be used to enhance the human posture model (inverted pendulum) and the muscle co-contraction model [14, 15].

V. CONCLUSION

In this study, adaptation analyses of joint stiffness and muscle activation have shown to improve as sensory inputs increased. These can represent motor learning ability of the balancing process. However, reduction in stiffness amplitude and muscle activation to indicate better balance ability still needs to be investigated. Through this study, we were also able to identify active-passive behaviour of joint stiffness, which can identify a reflex passive response as the subject continues riding the platform perturbation. The findings help to enhance human posture control and the co-contraction model interpretation, especially related to continuous repeated motion or force applied to the system.

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Effect of Sensory Manipulations on Human Joint Stiffness Strategy and Its Adaptation for Human Dynamic Stability

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***Abstract*—**Sensory input plays an important role to human posture control system to initiate strategy in order to counterpart any unbalance condition and thus, prevent fall. In previous study, joint stiffness was observed able to describe certain issues regarding to movement performance. But, correlation between balance ability and joint stiffness is still remains unknown. In this study, joint stiffening strategy at ankle and hip were observed under different sensory manipulations and its correlation with conventional clinical test (Functional Reach Test) for balance ability was investigated. In order to create unstable condition, two different surface perturbations (tilt up-tilt (TT) down and forward-backward (FB)) at four different frequencies (0.2, 0.4, 0.6 and 0.8 Hz) were introduced. Furthermore, four different sensory manipulation conditions (include vision and vestibular system) were applied to the subject and they were asked to maintain their position as possible. The results suggested that joint stiffness were high during difficult balance situation. Less balance people generated high average joint stiffness compared to balance people. Besides, adaptation of posture control system under repetitive external perturbation also suggested less during sensory limited condition. Overall, analysis of joint stiffening response possible to predict unbalance situation faced by human.

***Keywords*—**Balance ability, joint stiffness, sensory, adaptation, dynamic.

I. Introduction

WEAKENED in sensation of lower extremities, visual acuity and vestibular response are not uncommon among elderly, and it may increase a risk of fall which sometimes can lead to death [1], [2]. Sensory inputs play an important role for human posture control system to initiate strategy for counterpart any unbalance condition and then, stop us from fall. Perturbation or disturbance senses by sensory system lead central nervous system (CNS) to decide an appropriate balancing strategy neither limits nor initiates movement at any parts of the body where it may be seen through joint stiffness.

In previous research, investigations on joint stiffness characteristic have shown its response towards some movement performance issues. Joint stiffness strategy would act to correct center of pressure (COP) to move in the same direction as center of mass (COM) to maintain in balance position [3]. Besides, the study of gait performance on osteoarthritis's patient suggested that defected joint is stiffer than other parts [4]. Furthermore, research by Fitzpatrick et.al (1992) concluded that posture sway confine when reflex response was higher especially during sudden disturbance, which lead body part to stiff [5]. Thus, those findings indicate that imbalance makes patient generate high joint stiffness and have shown it is relevance to be used detect weakness in stability. However, a focus studies on joint stiffness properties to define imbalance is still less.

Furthermore, human balance characteristic or human posture strategy usually represent as an inverted pendulum model. Inverted pendulum model has been beneficial to describe postural sway and it is used widely in analysis of posture control system. Joint stiffness has become one of the important parameter in the model as a feedback in most of them [6], [7]. Estimation of joint stiffness amount using linear regression of moment-angle was reported to give a limited input on active component of stiffness characteristic, thus needs further investigation [8]. It is believed that active - passive component of balancing behavior can be observed under repetitive work [9]. Besides, characteristic of joint stiffness under different type of perturbation and sensory input condition are still less discussed.

Thus, this study aims to identify effects of sensory manipulations and different type of perturbation on posture control system especially, joint stiffness and its adaptation over repeatable perturbation. Based on these, the relationship between joint stiffness response and balance ability can be defined. Thus, a reliable model to present posture control system can be built.

II. Procedure

A. Subjects

In this study, seven healthy young subjects (aged 24.24± 2.19 years old) were participated. Each subject provided informed written consent prior participation.. Information regarding to subject's history of falls and physical condition was recorded as reference.

B. Experiment Set Up

Subject was exposed with two type of external perturbation which were forward-backward and tilt up-tilt down with displacement of 70mm and 6º respectively at four different frequency (0.2, 0.4, 0.6 and 0.8 Hz). These perturbations were produced by movable platform (MB-150, COSMATE,

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JAPAN) while their vision (eyes-closed (C) and eyes-opened (O)) and head movement (N) were manipulated. By fixing the movement of neck and head to body using neck collar (ADFIT collar, ADVAN FIT), it is believed that vestibular system will not able to sense changes in surface orientation precisely [10]. Both kinetic and kinematics data was collected using motion analysis (HWK-200RT, Motion Analysis, USA) and force plate (9286A, KISTLER, JAPAN). Each trial (*two type of perturbation*four type of sensory manipulation combination *four different frequency*) was recorded for 60s with a locked knee joint (using splint) to prevent bias movement from knee. Before the experiment started, all subjects were undergo Functional Reach Test (FRT) to evaluate initial balance score [11].

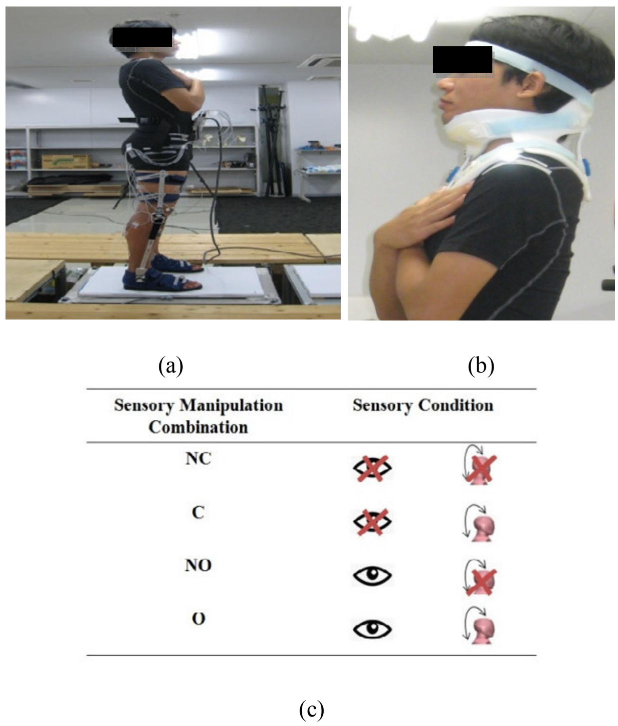


Fig. 1 (a) Subject preparation for experiment (b) The use of neck collar to limit head movement (c) Explanation on sensory manipulation combination

C. Statistical Analysis

All data were analyzed using mean and standard deviation. All results were described in average of 60s. Comparisons between conditions were done using Two Way ANOVA and Pearson correlation coefficient.

D. Measurement of Joint Stiffness

Joint stiffness was measured based on inverted pendulum model as shown below. Stiffness (K) at joint can be defined as (1) below;

$$K = \frac{\tau}{\theta} \tag{1}$$

where τ is torque of ankle and θ is angle of joint sway. The Fv which is a vertical ground force gather from force plate data is assuming to be as follow;

$$Fv \approx m_1 g \tag{2}$$

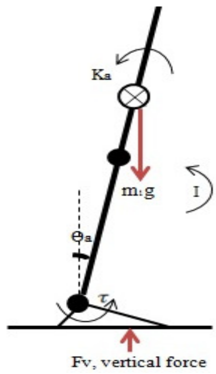


Fig. 2 Free body diagram to measure ankle stiffness based on inverted pendulum

III. Results

A. Joint Stiffness under Different Condition of Perturbation and Sensory Manipulation

As mentioned, before the experiment, subject was asked to perform a conventional clinical test which is the FRT to evaluate their actual balance ability and also their physical conditions. The details are shown in Table I. The FRT's score was observed to be negatively correlated with joint stiffness value (r^2>-0.2) as joint stiffness increase for subject with the lowest score (Table I). This indicates that people with low balance ability tends to stiff their joint more to maintain balance position under dynamic perturbation condition. However, these correlations were weak. It might be due to the nature of the FRT test where it was conducted during quiet stance without any external factors meanwhile joint stiffness analysis was examined during perturbed standing.

TABLE I
COMPARISON BETWEEN AVERAGE JOINT STIFFNESS AT 0.2 HZ OF PERTURBATION WITH FRT'S SCORE

Subject No.	FRT's score (cm)	Tilt up-tilt down (Nm/rad)		Forward-Backward (Nm/rad)	
		ankle	hip	ankle	hip
1	28	672.78	206.15	472.10	212.51
2	30	132.58	105.41	65.68	104.58
3	34.9	74.42	71.16	91.66	69.43
4	38.9	277.83	31.23	236.93	79.23
5	42.63	312.68	211.14	258.89	217.33
6	45.6	206.36	88.97	193.34	94.04
7	46.73	147.67	88.93	138.07	91.00
Correlation coeff. (r^2)		-0.41	-0.25	-0.26	-0.24

The comparison was done between FRT's score and joint stiffness by using Pearson Test.

By manipulating the type of perturbation, it is believed that it is able to manipulate the proprioception system. Tilt up-tilt down and forward-backward type of perturbation did triggered different joint stiffness strategy. Based on Table I, ankle joint is more stiff during tilt up-tilt down while hip joint is stiffer during forward-backward movement. However, no significant different found ($p>0.05$).

According to Fig. 3, limitation of vision sensory (NC and C) produced highest ankle joint stiffness especially during tilt up-tilt down. Meanwhile, for the forward-backward perturbation, the highest ankle stiffness was observed when head movement was being constraint (NC and NO). Overall, hip joint stiffness is higher during head motion constraint condition during both types of perturbation (NC and NO). Besides, the increase of perturbation frequency did increased joint stiffness especially at ankle ($r^2>0.5$). But no significant different was observed between sensory manipulation conditions at ankle joint.

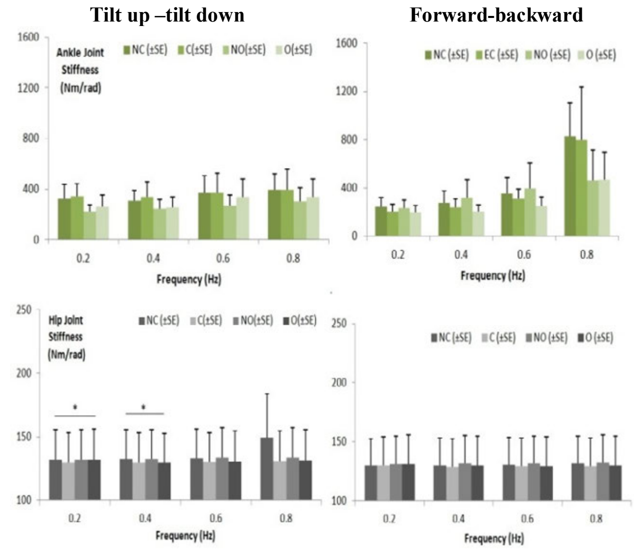


Fig. 3 Joint stiffness under four different frequency of perturbation (±SE)(*first row*: ankle, *second row*: hip) with four different sensory limitations (NC=eyes closed with neck collar, C=eyes closed, NO=eyes-opened with neck collar, O=eyes opened)(*: $p<0.05$)

B. Adaptation of Joint Stiffness over Repeated Perturbation and Limited Sensory Conditions

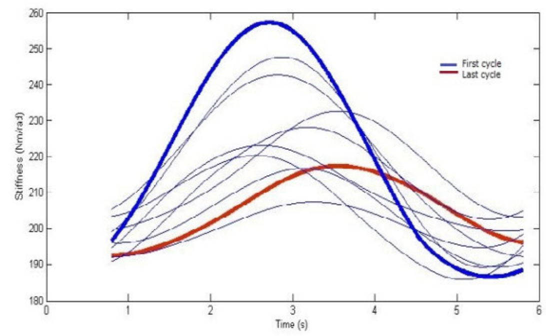
In order to evaluate the adaptation of CNS towards stiffening strategy over repeated perturbation, we compared stiffness response of each cycle of 0.2 Hz trial. During this low intensity of perturbation, it is easy to determine the effect of sensory input modification towards stiffening adaptation strategy. Adaptation of CNS towards the joint stiffening response is determined by measuring area under graph (AUG) using trapezoidal rule using (3) and (4).

$$AUG = \int_1^t K(t)\ dx \tag{3}$$

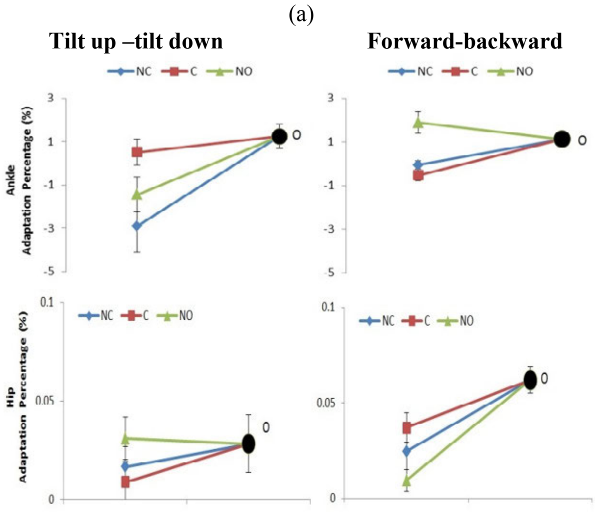
$$Adaptation\ (\%) = \frac{AUG_i - AUG_{(i+1)}}{AUG_i} x\ 100\% \qquad i = 1,2,3 \ldots. \tag{4}$$

where K (t) is joint stiffness along perturbation period, t is time for one cycle of perturbation, and i is number of cycle

In this analysis, each cycle of joint stiffness was compared with cycle before it. Based on Fig. 4, in average, adaptation of the CNS through ankle joint stiffness was almost 1.5% (±SE) which means that during normal condition (O), healthy young subject reduces joint stiffness by 1.5% at each cycle of repeated movement. But under weak sensory input condition, subjects almost unable to adapt and joint stiffness was keep increases in order to remain balance (less or negative (-ve) adaptation percentage). Meanwhile, for hip joint, adaptation percentage was smaller than ankle but it was still able to indicate that limitations of sensory input also reduced the percentage of adaptation. However, there were no significant different found between different sensory manipulation condition ($p>0.05$).



(a)



(b)

Fig. 4*Top*: (a) Ankle joint stiffness at each cycle during 0.2 Hz. *Bottom*: (b) Comparison of average adaptation percentage of joint stiffness between normal (O) and sensory condition (±SE)

IV. Discussion

The results have shown that joints stiffness at both ankle and hip joints were able to define the balance ability of human. Where, higher value of stiffness was required by less balance people and while faced a difficult conditions. On the other hand, different types of perturbation and sensory limitations will also generate different joint stiffening strategy as expected.

Visual input plays an important role in balancing process where ankle joint was measured to be higher during eyes closed (C) at both perturbation and all frequencies. However, ankle joint stiffness was higher during NO than O when forward-backward perturbation was applied. This suggested that head and neck segment is important during high acceleration of posterior and anterior movements. During forward-backward perturbation, the COM and COP of human body were moved intensively forward and backward as the body was observed to sway more. Without movement at the head and neck segment, balance condition will become worst. This is because an otolith organ which important to detect change in acceleration is being disturbed. Moreover, the results show tilt up-tilt down perturbation did not cause large body sway than forward-backward. However, limited head motion seems to improved balance during different surface's level as subjects decreased their joint stiffness.

In determining the CNS adaptation, it is believed that weakened in sensory inputs did affect the motor learning process where, subject faced difficulty to maintain their position. They need to continuously generated force to produce stiffness by increasing muscle effort to maintain their balance along trial period. Analysis of stiffness adaptation over different types of perturbation also can detect the situation where subjects felt less balance and how ankle and hip joints working with each other to create synergy strategy between them. On the other hand, analysis on adaptation percentage has shown that joint stiffness was also an active component at initial stage of perturbation. Then, it shifts to passive behavior following the platform and was altered according to information received by the subject. Less sensory information due to certain factors (i.e., impairment, disease, ageing and etc.) will lead to reduce in adaptation.

This study has faced some limitations, firstly, the ability of the FRT test to relate with balance ability under perturbed stance. In general, the FRT describes balance ability through capability of a person to reach forward distance as far as possible which represents by COM maximum displacement. However, result from this test is very limited. Furthermore, effect of the use of neck collar to limit the head movement and thus, disturbed vestibular function is still not evident. It was observed to influence more on vision input as it permit a limited visual space (subject reported unable to see anything at below).But, the result of average stiffness have shown that it able to distinguish between sensory limitation condition.

V. Conclusion

In this study, the effects of vestibular system's and vision's input limitations did affect the joint stiffening strategy. Besides, it is acceptable to said that people with less balance ability tends to have high stiffness at both hip and ankle joints. Adaptation percentage of the CNS over repeated perturbation shows that healthy people were able to adapt much better compared to those who faced weakness in their sensory inputs. Further analysis especially related muscle activation and posture control system synergy will be proposed to determine their response under unbalance condition.

Acknowledgment

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EFFECT OF HUMAN POSTURAL CONTROL FOR DIFFERENT PERTURBATION DURING WALKING

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ABSTRACT

We examined the effect of unexpected backward perturbation (BP) and dorsiflexion perturbation (DP) on postural control strategy during walking in humans. We measured both kinetics; the vertical component of ground reaction forces (VGRF) from reaction force, and kinematics data; lower limb joint angle and center of mass (COM), together with electromyography activities (EMG) of lower limb muscles. The BP and DP were occurred by six-axis parallel link mechanism platform at heel contact. Results indicated that medial gastrocnemius (MGAS) of the perturbed leg during BP and DP indicated larger response than normal walking. The latency of this response was longer than 100[ms]. Therefore, it suggested that the MGAS response might be not short-latency reflex response for perturbation, but might be long-latency reflex response. The stimulation intensity of BP for ankle joint was smaller than the DP, however the BP COM forward displacement was larger than the DP from 0 to 300 ms after MB onset. The voluntary responses of human posture control of BP are depended on the COM position.

1. INTRODUCTION

Even if humans receive unexpected perturbation during walking, human will involuntary responsive to the perturbation to prevent a fall down. This response to recover postural stability is reflective. The response senses perturbation from visual, vestibular and somatic senses input and prevents fall down by functional adjustment of lower leg muscle based on these information. Stretch reflex is considered as one of the representative reflex of somatosensory system. This reflex can support the sudden posture drifting to go through spinal cord without centrum and contributes much to the fall prevention. However, the detail of postural reflex is still remaining unclear. Research of a compensation movement and stability of body under perturbation is important to understand a fall prevention strategy of elderly and a person with disturbance of motor function and, thus, effective walk rehabilitation method can be produced. The human postural control for strategy under extemal perturbation has been studied till now. Previous studies such as the experiment of different input perturbation during standing position [1], a false step perturbation in swing phase during treadmill walking [2], the natural slip during walking by support surface painted with oil [3], and a sudden drop during walking [4] have shown a different posture control responses in order to prevent fall. According to Nashner et al. (1976), the role of stretch reflex response in posture control was intentionally adjusted under a kind of extemal perturbation during standing [1]. However, the study on the effect of postural control under different unidirectional perturbation during walking is still less.

Therefore, in this study, we examined the effect of different perturbation on the postural control during walking.

2. METHODS

Six healthy young adults (mean age: 21.5±0.5 year) participated in the study. No subject had a prior history of neurologic disease at the time of testing that may have affected the ability to perform the experiment. Unexpected perturbations were induced by the motion base (MB-150, COSMATE, JAPAN) as subjects walked along 5 m walkway. Force plate (9286A, KISTLER, JAPAN) was placed on the motion base (MB) as shown in Fig.1. When pressure-sensitive sensor on the motion base (MB) was tripped, the MB moved posteriorly (backward perturbation: BP, distance: 40 mm, velocity: 200 mm/s) or rotate superiorly (dorsiflexion perturbation: DP, distance: 4 deg, velocity: 20 deg/s) upon heel contact (HC) of the right lower limb. Non-perturbations (control), BP, and DP of each 15 trials were performed in random order. The subject walked to the metronome which we set

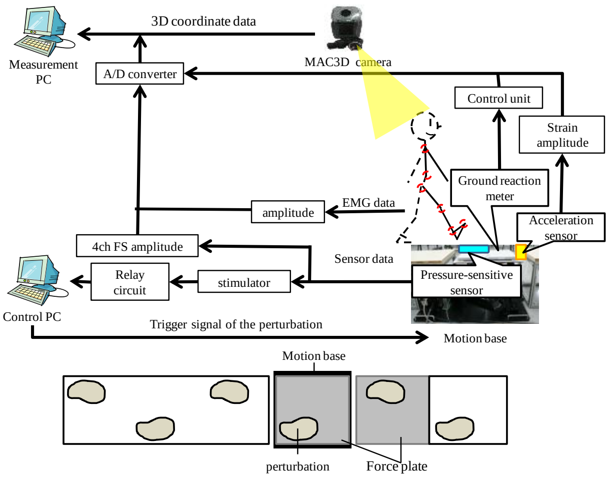

Fig. 1 System configuration

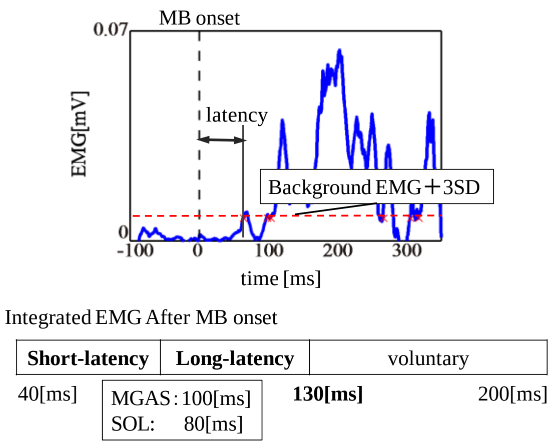

Fig. 2 Definition of EMG latency and

to the speed that was easy to walk. Electromyography (EMG) signals of medial gastrocnemius (MGAS), soleus (SOL) and tibialis anterior (TA) of both lower limbs were recorded at sampling frequency of 1000 Hz. The beginning of the EMG response was defined as the times when EMG activities reached levels higher than thrice the standard deviation of the background activity. The latency of response was defined as the duration from the impact of the perturbation to the EMG onset. The integrated EMG of short-latency, long-latency, and voluntary was calculated by the period as shown in fig.2. The vertical component of ground reaction forces (VGRF) of both lower limbs were calculated from two force plates data recorded at 1000 Hz. Kinematics data was recorded at 200 Hz by six cameras (HWK-200RT, Motion Analysis, USA) that were positioned around the subject's walkway. The center of mass (COM), hip and ankle joint angles were calculated from the kinematics data. Twelve kinematic reflective markers were placed on the skin overlying the base of the third metatarsal, lateral malleolus, lateral condyle of the femur, greater trochanter of the femur, and acromion process of the scapula.

To test statistically the difference between BP and DP, Wilcoxon signed-rank test with repeated measures was used. $P < 0.05$ was defined as a level of significance.

3. RESULTS AND DISCUSSION

A typical example of the ensemble-averaged waves of each parameter from the lower limbs during perturbed and unperturbed walking is demonstrated in Fig.3. Dashed vertical line is mean of MB onset, black line is control, blue line is BP, and red line is DP. Right VGRF and hip joint angles wave form during the BP generally paralleled the control condition wave form from MB onset. However, the right BP MGAS response was large compared to the control condition from 130 to 350 ms after onset of perturbation. The right DP MGAS response was also similar result with the right BP MGAS response. The right BP ankle joint was dorsiflexor compared to the control condition until 500 ms after onset of perturbation. The right DP ankle joint was also similar the result with the right BP ankle joint.

The latencies of the responses recorded in right MGAS and SOL are shown in Table.1. The onset of the EMG responses latency in MGAS and SOL in the present study was longer than the results of previous study. There were no significant difference between the BP and DP. Nakazawa et al. (2003) reported that the stretch reflex response (the latency is 40-70 ms) was observed in MGAS and SOL of the perturbed side by dropping perturbation during walking [4].

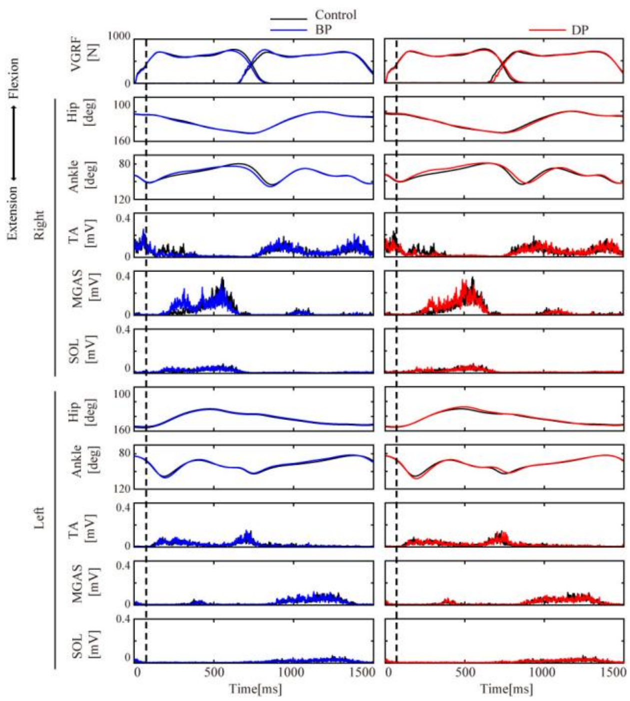

Fig. 3 Typical example of each parameter from the lower limbs during perturbed and unperturbed walking

Table.1 EMG response latency in muscles (n=6)

Perturbation	MGAS [ms]	SOL [ms]
BP	94.83±26.98	85.83±20.11
DP	108.83±34.03	75.33±16.08
Nakazawa et al.(2003)	41.27±1.88	41.67±1.51

An integrated EMG of the each right EMG latency is shown in Fig.4. Both right MGAS and SOL involved in the range of the long-latency response. These results suggest that the EMG responses of perturbed walking might not be short latency response, however a long-latency polysynaptic reflex pathway might be involved in these responses.

The BP MGAS voluntary response was significantly larger than that of DP. The BP SOL long-latency and voluntary response were significantly larger than those of DP.

An average angular velocity of ankle joint is shown in Fig.5. The BP was significantly smaller than the DP. The results suggest that the stimulation intensity of BP for ankle joint was smaller than the DP.

Fig.6 shows the typical example of anterior-posterior COM during perturbed walking. The BP COM forward displacement was larger than the DP from 0 to 300 ms after MB onset. This result agrees with Nashner et al. [1].

In conclusion, the stimulation intensity of BP for ankle joint was smaller than the DP, however the BP COM forward displacement was larger than the DP from 0 to 300 ms after MB onset. We suggest that the BP might cause postural modulation and the perturbed MGAS, produces the driving force at the late stance phase, and performs to recover balance.

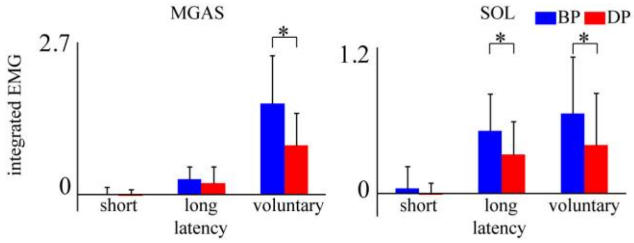

Fig. 4 Integrated EMG of the each latency. *P <0.05.

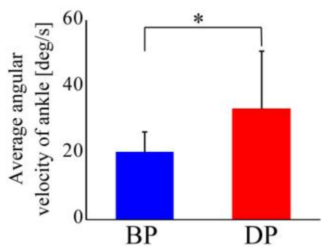

Fig. 5 Average angular velocity of ankle. *P <0.05.

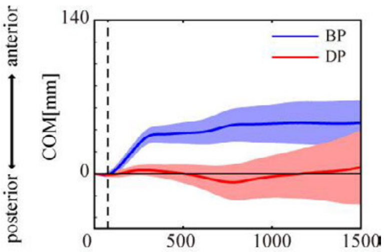

Fig. 6 Anterior-posterior COM
The colored area is ±SD.

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Estimation of stiffening strategy of ankle and hip based on joint sway

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Abstract— **Stiffening strategy is a posture strategy to maintain a desired position of the body. This strategy is generated by the Central Nervous System (CNS), which creates muscle forces around a specific joint; and thus, affects joint movement. In this study, stiffening strategy at the ankle and hip joints of five young healthy subjects was observed based on joint sway. Based on stiffness pattern, both joints were seen to oscillate slower from external perturbation sway, when exposed to a high frequency of translation perturbation. Furthermore, stiffness ratio between ankle and hip joints, at four different frequencies of translation perturbation (0.2, 0.4, 0.6, and 0.8 Hz), indicated the transition of posture strategy from ankle to hip strategy when responding to a high intensity of external disturbance which useful to determine adaptation to instability.**

Besides, additional parameters are suggested to represent the change of sway pattern of the ankle and hip joint model, which was built based on an inverted pendulum model. Estimated sway pattern produced a high correlation (r^2>0.5) with actual data. In conclusion, stiffening strategy can be seen through the change of sway pattern and the value of stiffness at the joint. Therefore, development of a control model, according to the improvement suggested for both joints is warranted, in order to develop a reliable simulation model for a posture control measurement system.

I. Introduction

During unpredictable external perturbation, the nervous system applies reflective or voluntary movements to maintain balance or desire movement. Input signals detected from the muscle spindle will then transmit to the nervous system. Both reflective and voluntary commands will transmit to muscles to produce required movement, and depend on additional input from vision and vestibular systems, which give a sense to detect perturbation; and thus, initiate an efficient strategy based on experience. Muscle force produced to create or maintain movement can be interpreted as stiffness at the joint [1]. Horak and M. Casadio agree that central nervous system (CNS) adaptation towards external perturbation can be seen through limb stiffness[2, 3].

Furthermore, joint stiffness is described as the change of torque over the change of joint rotation. Stiffness was reported to be decreased at one joint as it increased at other joints [4]. Besides, Edwards estimated that ankle and hip stiffness's had to be larger than 728 Nm/rad and 179 Nm/rad, respectively; in order to be stable, as based on theoretical study [3].

Moreover, selection of strategy depends on many factors that include age, health condition, and lifestyle. In previous research, Ho and Bendrups showed that elderly subjects had higher stiffness compared their younger counterparts [5]. Meanwhile, Horak reported that the elderly used hip joints more than younger subjects [2]. Somehow, these two studies gave us a clue to the relationship between the selection of posture strategy and joint stiffening. This study was conducted to observed ankle and hip joint stiffness and sway behaviour, in response to external perturbation, in order to develop a simulation model. In this paper, stiffening patterns at both ankle and hip joints were analysed, based on an inverted pendulum model.

A. *Ankle and hip joint's model.*

Both ankle and hip joints can be modelled as a single link inverted pendulum model. The oscillation of the model can be described by the equation below:

$$I\, d\Theta^2 t/dt^2 + B\, d\Theta t/dt + K\Theta = mgh \sin\Theta \qquad (1)$$

Where, $\Theta = A \sin(\omega t)$ represents the angle of joint sway, I is the moment of inertia, B and K are damping and stiffness at the joint, respectively. Stiffness, K, can be described as torque over Θ. Furthermore, $d\Theta^2 t/dt^2$ is defined as angular acceleration, $d\Theta t/dt$ is angular velocity, m is mass, g is gravitational acceleration ($9.81 ms^{-2}$), and h is a distance of COM from the joint. Previous research by K.P. Granata et al., introduced effective system stiffness parameters to represent intrinsic muscle behaviour and contributions from reflex movement when the ankle was exposed to load in Eq. (1). However, it was unable to explain the musculoskeletal dynamics in a variable environment. According to Buchanan, postural sway switches from the single inverted pendulum model to a multi segmented type of sway at a high frequency of perturbation [3]. In this study, the same pattern was expected at both ankle and hip joints (Fig. 1).

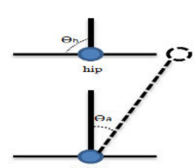

Fig. 1. Multi segmented pendulum model for ankle and hip joints.

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The sway angle at ankle and hip can be described as follows:

$$\Theta = \begin{bmatrix} \Theta a \\ \Theta h \end{bmatrix}$$

II. METHODOLOGY

B. Experiment setup

A dynamic experiment was conducted to observe both hip and ankle joint stiffness's and adaptation towards repeatable translation perturbation. In this experiment, five healthy young subjects were involved (aged 22.3 ±0.8 years; height 169 ±2.5; weight 63.9 ±1.85kg). The subjects were asked to maintain a quiet standing position when exposed to external translation perturbation (displacement: 70mm) at four different perturbation frequencies (0.2, 0.4, 0.6, and 0.8Hz). This external perturbation was induced by a six motion control base (MB-150, COSMATE, JAPAN). Motion data was recorded by a motion analysis system (HWK-200RT, Motion Analysis, USA) and a force platform (9286A, KISTLER, JAPAN) (Fig. 2). Each subject underwent four experimental sessions (at 0.2, 0.4, 0.6, and 0.8Hz). Each session was recorded for 90s with a rest time between each session.

C. Data Analysis

The motion data obtained from both the motion analysis system and the force plate was used to analyse ankle and hip joint stiffness, as per the equation above. Next, spectrum analysis was used to determine the change of postural sway for both ankle and hip joints to possibly indicate postural control and adaptation strategy. A statistical analysis, using cross-correlation, was used to observe the response time for both joints when exposed to perturbation. This analysis was performed using MATLAB software.

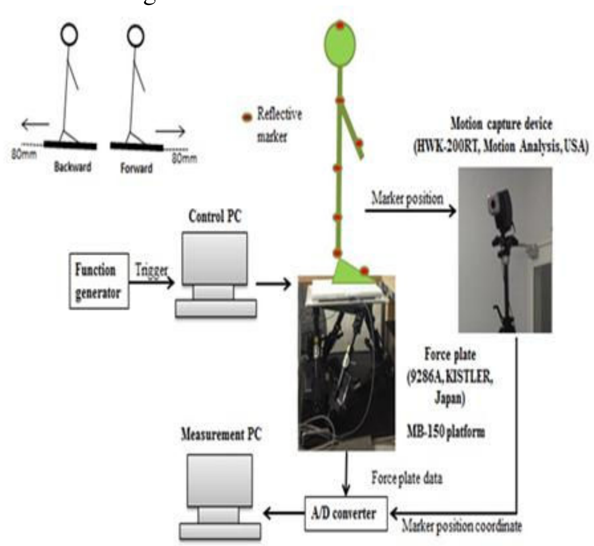


Fig. 2. Experiment setup.

III. RESULTS

A. Stiffness pattern

Based on the recorded data, both ankle and hip joint's stiffness was calculated by measuring the torque of the joint over the sway angle. Changes, in term of stiffness pattern under different frequencies of perturbation were observed (Fig. 3). Both joints were seen to follow the perturbation sway at a frequency of 0.2 Hz. However, at a frequency higher than 0.2 Hz, there were only slight changes to the joint's sway.

In order to identify the change of sway frequency, the results from the spectrum analysis were analysed (Fig. 4). The results showed that ankle joint sway followed the perturbation frequency at 0.2 and 0.4Hz. However, the sway frequency was reduced as the hip joint swayed at 0.4Hz, when the external perturbation frequency was at 0.6Hz and swayed at 0.2Hz when the external perturbation was at 0.8 Hz. On average, the frequency of ankle sway reduced from 33 to 75% of perturbation frequency. These conditions applied almost similarly for hip joints, but occurred at a much shorter frequency, to almost 0Hz when perturbation was > 0.4Hz.

Furthermore, the ratio of ankle joint stiffness against hip joint stiffness has indicated that there was change in the use of joints. Table 1 shows the ratio of ankle to hip joint stiffness.

Table 1: Ratio of ankle joint stiffness vs. hip joint stiffness.

Ratio of ankle joint stiffness vs. hip joint stiffness					
Subject	**Gender**	**Frequency**			
		0.2 Hz	**0.4 Hz**	**0.6 Hz**	**0.8 Hz**
S1	f	0.189	0.565	1.980	3.258
S2	m	1.160	1.160	0.985	0.989
S3	m	0.546	0.419	3.473	1.218
S4	m	0.865	0.742	1.096	2.855
S5	m	0.399	0.221	0.264	0.514

With no significant change between subject ($p>0.05$)

Based on the table above, ankle joints become stiffer with the increment of perturbation frequency, as the ratio increase. This result indicates a shifting of posture strategy from ankle to hip joint strategy. Hip joints are stiff at low frequency as the ankle joint strategy is dominance and able to act better to maintain a balanced position.

The amplitude of ankle and hip joint stiffness, which is normally represented as a gain factor for a controller system, can be interpreted through the values shown in Table 2. These results show that hip joint stiffness reduces with increase of perturbation frequency; meanwhile ankle joint stiffness is increase. On the other hand, average angle of sway shows the

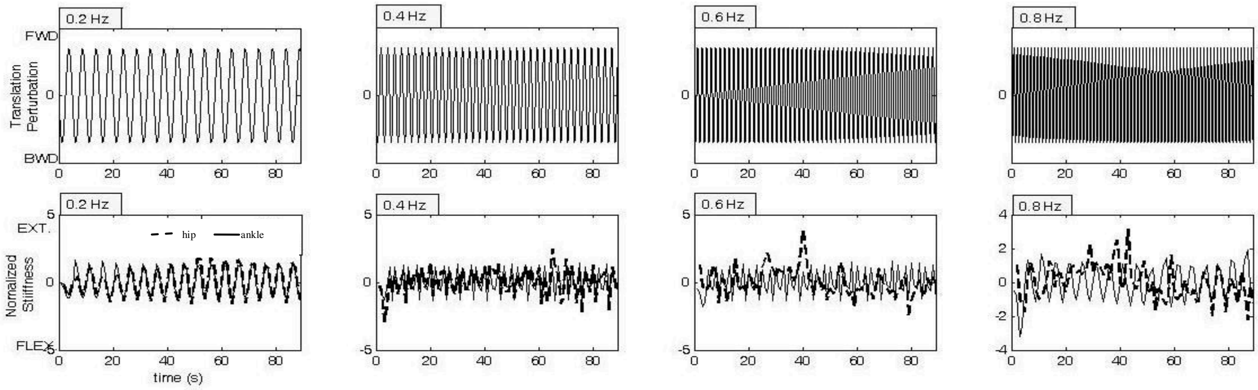
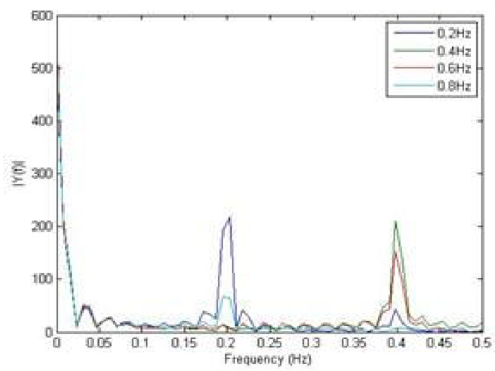
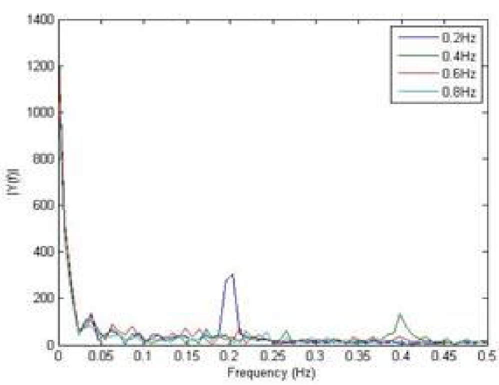


Fig. 3. Ankle and hip joint stiffness's at four different frequencies.



(a)



(b)

Fig. 4. Spectrum analysis for (a) ankle and (b) hip stiffness pattern.

opposite, as we already expected that the hip joint angle of sway will increase drastically as the disturbance become harder. Moreover, response time between both joint also seen to be decreased as the frequency of perturbation increased. This suggests that the nervous system responded faster to improve stability, due to an increase in perturbation intensity. However, the hip joint responded delay than the ankle at 0.2 Hz, as the perturbation did not really cause unbalance. Thus, shows hip joint is less dominance at low intensity of perturbation.

Table 2. Average value of stiffness at 4 different perturbation frequencies

Parameter	**Frequency**			
	0.2 Hz	**0.4 Hz**	**0.6 Hz**	**0.8 Hz**
Stiffness (Nm/rad) (±SE)				
hip	4460.05 (±981.33)	3761.35 (±624.94)	3307.18 (±798.91)	3312.39 (±144.81)
ankle	2594.07 (±9.42)	2957.06 (±10.36)	2910.53 (±20.88)	2968.70 (±15.00)
Sway (rad) (±SE)				
hip	0.042 (±0.0010)	0.041 (±0.0012)	0.066 (±0.0019)	0.091 (±0.0020)
ankle	0.237 (±0.0004)	0.241 (±0.0006)	0.241 (±0.0013)	0.245 (±0.009)
Time response (s) ankle vs. hip sway (±SD)				
	0.567 (±0.003)	-0.890 (±1.412)	-0.035 (±0.007)	-0.373 (±0.641)
Time response (s) perturbation activation vs. joint sway (±SD)				
hip	0.510 (±1.02)	-3.745 (±3.25)	-1.515 (±2.35)	-0.953 (±1.19)
ankle	0	0	0	0

With no significant change between subjects ($p>0.05$)

B. Additional parameters for the ankle and hip joint pendulum model.

Based on the experiment results obtained, several additional parameters should be considered to represent the actual model, for both ankle and hip joint stiffness and sway. Previous studies identified time response to external perturbation through electromyography (EMG) analysis [6]. Cross correlation analysis between platform movement perturbation and angle of joint sway, indicated a response time; as shown in Table 2. In order to represent the change in sway pattern, f_{stiff} was introduced. Thus, modification towards an oscillation equation was performed to include this stiffening strategy element, as shown in the equation below.

Angle of sway is represented by Eq. (2); where, ø is a phase delay.

$$\Theta = A \sin(\omega t + \phi) \quad (2)$$

Where, $\omega = 2\pi\,(f + f_{stiff})$ indicates the stiffening effect. Thus, Eq. (2) will be as follows:

$$\Theta = A \sin(2\pi\,(f + f_{stiff})\,t + \phi) \quad (3)$$

Furthermore, f_{stiff} must follow certain conditions, as determined from the experiment result for the ankle joint, as follows:

$$f_{stiff} = \begin{cases} 0 & ;\ \text{for } f \leq 0.4 \text{ Hz} \\ f - (f - 0.33) & ;\ \text{for } 0.4 < f \leq 0.6 \text{ Hz} \\ f - (f - 0.75) & ;\ \text{for } 0.6 < f \leq \infty \text{ Hz} \end{cases} \quad (4)$$

Meanwhile, for the hip joint, the f_{stiff} condition was as follows:

$$f_{stiff} = \begin{cases} 0 & ;\ \text{for } f \leq 0.4 \text{ Hz} \\ f & ;\ \text{for } 0.4 < f \leq \infty \text{ Hz} \end{cases} \quad (5)$$

The equation above was evaluated by comparing it with actual experiment data. The inverted pendulum model based on K.P. Granata et al model and external functions that include Eqs. (3), (4), and (5), was built to produce estimation data (Fig. 5). The estimated result shows that the equations above are reliable to present the stiffening strategy, when correlation analysis between actual and estimated results for both joints was $r^2 > 0.5$ (Table 3). However, some improvements needed to be made, in order to reduce lags and optimize the model.

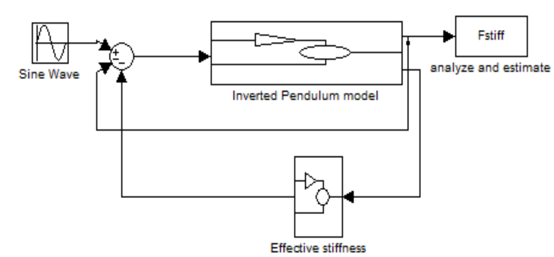


Fig. 5. A block diagram of single joint model with external function of Eqs. (3), (4), and (5) to produce an estimated data.

Table 3. Cross correlation analysis between actual and estimated stiffness pattern

Corr. coeff. (lags)	**Frequency**			
	0.2 Hz	**0.4 Hz**	**0.6 Hz**	**0.8 Hz**
Hip	0.95(8)	0.81(15)	0.95(7)	0.97(8)
Ankle	0.84(8)	0.82(14)	0.94(13)	0.97(4)

IV. Discussion

The results mentioned above give additional information to describe the actual behaviour of joints in reflex to repeatable external perturbation. However, a small number of subjects cannot represent the entire population.

Even though all subjects were free from any neurological diseases, none applied the same pattern of stiffness to each other. This made data analysis interpretation very complicated. However, somehow, the use of spectrum analysis allowed us to see that they applied almost the same stiffening strategy at different perturbation frequencies; when their joints swayed at almost the same range of frequencies. At low frequencies, ankles and hips swayed as one pendulum model, whilst at faster frequencies, they were separated as a multi-segmented type of sway (Fig. 3). This finding agrees with previous research by Buchanan [7]. However, the stiffness amplitude at both ankle and hip can explain the dominant joint used. The ratio of hip-ankle stiffness showed that subjects applied various posture strategies, depending on the perturbation. This explains why ankle strategy is relevant at a low intensity of perturbation, whilst a hip strategy are required at a higher intensity.

The use of the inverted pendulum model to represent the segmented part of posture was really beneficial. However, there is still room for improvement of this model, for it to represent not only physical movement, but also some neurological effects. We frequently assumed that posture sway was according to perturbation sway; as mentioned in Eq. (1). In contrast, experimental results show that the nervous system applied different strategies to maintain balance. Besides, the introduction of f_{stiff} in the equation was expected to represent the stiffening element for this model - and it worked well when the correlation with the actual experimental data was $r^2 > 0.5$ at constant stiffness gain is 1. Since the current model for both joint and external stiffness function was built separately, further analysis and development is required.

V. Conclusion

In conclusion, by manipulating the intensity of perturbation, the stiffening strategy used at ankle and hip joints can be observed through the change of sway pattern and the ratio of ankle-hip stiffness. Furthermore, the introduction of f_{stiff} was able to symbolize the joint stiffening strategy in terms of joint sway. However, further analysis should be done to analyse its relationship with muscle activity.

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